THE EFFECTS OF PROSTHETIC ALIGNMENT ON THE STABILITY OF THE KNEE IN ABOVE KNEE AMPUTEES

By

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DEDICATION

This thesis is dedicated to my parents,

my wife and my daughter Lama.

ABSTRACT

The components and alignment of a prosthesis have a large influence on the gait of an above knee amputee. The present criteria for determining the optimum alignment are mainly subjective, based on visual observation of the amputee's gait and by considering his/her comments. These comments however, are not always helpful. It has been reported that a range of alignments is acceptable to the patient and to the prosthetist, and it is believed that the optimum alignment can be selected from the range of acceptable alignments using biomechanical analysis.

The socket is an important component in an above knee prosthesis and can affect the gait of the amputee. While the conventional socket is the quadrilateral, several problems have been reported with this type of socket. These include instability in the coronal plane, discomfort and restriction of the stump muscles causing limitations in the function of the prosthesis. In an attempt to overcome these problems the ischial containment (IC) socket was introduced in 1985. The biomechanical characteristics of the IC socket have, however, not been objectively assessed and compared with those of the quadrilateral socket.

In this study, the effect of alignment adjustments on the gait variables for eight above knee amputees wearing quadrilateral sockets was investigated. Three of these amputees were also tested wearing IC sockets. The alignment of the prosthesis was systematically changed at the ankle, knee and socket. The primary aim of this project was therefore, to systematically vary the alignment of the prosthesis and to study the effects on the gait, and to compare the performance of amputees wearing IC and quadrilateral sockets. The ultimate goal of this research work is to provide a method for the determination of the optimum alignment from a range of acceptable alignments.

A socket axis locator and a coordinate measuring system were used for measuring the prosthetic alignment accurately. Three TV cameras and two Kistler force plates were operated simultaneously and synchronously at a rate

of 50 Hz to acquire the displacements of the body segments and the ground reaction forces. All angular movements at the joints, moments and the temporal-distance parameters were calculated for both the prosthetic and the sound legs of the amputees, and for the right and left legs of ten normal subjects. The movement of the upper body was also recorded. Computer programs were developed to calculate and graphically present the above parameters in three dimensions.

It was found that alignment changes affected the gait parameters of the whole body. At the prosthetic joints, certain changes in the alignment of the prosthesis resulted in specific alterations in the gait pattern. These effects were repeatable. The anterior-posterior (AP) joints moments and the fore-and-aft ground reaction force were found to be the most sensitive variables to alignment changes. The trunk rotations in the AP and medio-lateral planes, and the torso rotation in the transverse plane were also found to be sensitive to alignment changes, and the trunk was the main compensating element for any misalignment. At the sound side, the alignment changes resulted in noticeable changes in the AP joints moments and fore-and-aft ground reaction force. In the coronal and transverse planes, changes in the gait patterns that were analysed were not always consistent. These changes mainly depend on the method of compensation which is adopted by the patient. The IC socket showed improvements in the patient's performance in terms of higher speed of walking, comfort, improved symmetry in the two legs and the gait parameters are more comparable with those of normals.

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CHAPTER ONE

INTRODUCTION

The gait of above knee amputees can be influenced by a number of factors, such as the type of components, alignment of the prosthesis and fit of the socket. Selection of the most appropriate components of the prosthesis (foot, knee mechanism, socket type) can affect the gait of the amputee especially if the patient does not have the ability to compensate. Alignment of a prosthesis can be defined as the relative position and orientation of its components. To achieve a satisfactory performance, the prosthesis must have an acceptable alignment otherwise it may not provide the required support and function. Gait deviations may occur directly as a result of inadequate alignment or indirectly from pain, discomfort or tissue breakdown caused by inappropriate alignment (Zahedi et al 1986). Thus, an unsatisfactory alignment will lead to a limitation of the function of the prosthesis and as a consequence the patient may reject it. Furthermore, of paramount importance to the patient is the display of an aesthetically pleasing gait, which may be compromised if the alignment is unsatisfactory.

Radcliffe (1969 and 1977) and Foort (1979c) stated the principles for aligning a prosthesis. The method of achieving dynamic alignment is dependent upon observations made during gait assessment and on feedback provided by the patient. Alignment of the leg should be accomplished within a reasonably short time as too many adjustments may confuse the patient. The present criteria for determining the best alignment are still mainly subjective and rely on the skill and judgement of the prosthetist with no clear definition of the optimum alignment. The visual observations cannot be accurate and the amputee's comments are not always helpful because of misunderstanding, inability to comment effectively or reluctance to express faults with a prosthesis or prosthetist in a bid to comply with the treatment.

Gait optimization can be considered under three headings which are the assessment methods, the prosthesis and the amputee. The visual observation assessment method alone is not accurate enough to achieve optimum alignment and provide the most comfortable fit. Saleh (1988) reported that these subjective methods of assessment are not adequate and the amputee's responses to any misalignment are not always detectable by eye and would require objective analysis.

Solomonidis (1975 and 1980) defined six and eleven alignment parameters for BK and AK prostheses respectively, and he used them in order to compare different types of modular prostheses. Solomonidis found that a range of dynamic alignment was acceptable for any particular patient, and suggested that a relationship may exist between the various alignment parameters. Zahedi et al (1985 and 1986) also reported that a range of alignment is acceptable to the patient and to the prosthetist. It was found that the prosthetist was unable to reproduce a certain alignment configuration when the prosthesis was aligned and realigned. It was suggested that by using biomechanical analysis, it would be possible to determine the optimum alignment from the range of the acceptable alignments.

Hannah et al (1984) and Mizrahi et al (1986) studied the effect of alignment changes on the symmetry between the gait parameters for the prosthetic and sound legs. Hannah et al investigated the effect of alignment changes on the symmetry of kinematic variables of the lower limbs of BK amputees, and reported that the symmetry is maximum at the optimum alignment. Mizrahi et al reported (on one AK amputee) that the differences in the pattern of ground reaction forces at the sound and prosthetic sides can be eliminated by achieving optimum alignment, and maximum symmetry coincided with optimum alignment. It is however doubtful that the amputee who is already an unsymmetrical system may achieve a symmetrical gait pattern, and the optimum alignment is most unlikely to coincide with the most symmetrical gait pattern. This was confirmed by Winter and Sienko (1988) who found that modified motor patterns existed depending on the residual muscles of the amputated leg.

Appoldt et al (1968) studied the effect of alignment changes on the local pressure between stump and socket wall of two AK amputees. The aim was to convert a knowledge of desirable pressure values into an alignment tool. The results were not so conclusive because the alignment changes had no effect on the pressure values.

Pearson et al (1973) and Winarski and Pearson (1987) also studied the effect of alignment changes on the resulting pressure between the stump and socket of BK amputees. However, these studies have considered the normal pressure only and only at certain areas of the stump without looking at the gait variables of the whole body. Morimoto et al (1987) measured the effect of alignment changes on the gait parameters by using a pylon and flexible electrogoniometer. Most of the work discussed above was for BK amputees. However, Solomonidis (1980), Zahedi et al (1986) and Yang Lang (1988) reported work on the alignment of AK prostheses. Zahedi et al (1989) confirmed that kinematic and kinetic data are necessary to understand the locomotion of the amputee.

Objective assessment of the function of a prosthesis can be carried out by means of biomechanical analysis which can assist the prosthetist in determining an acceptable static and dynamic alignment. Yang Lang (1988) carried out the most comprehensive study yet on the effect of alignment changes on the AK amputee's gait. The kinetic and kinematic data for the whole body were obtained and subjected to a biomechanical analysis. Although, the effect of the alignment changes was noticeable on the prosthetic side, the conclusions were limited by the fact that only two subjects were completed the full series of test and it was suggested that the test population be increased in order to establish a normative data base. Yang Lang has only studied the effect of AP alignment changes, and it was suggested that the effect of ML alignment changes on the gait variables should also be investigated.

Following these recommendations, Marmar and Solomonidis (1989) used three TV cameras and two force plates to study the effect of alignment changes on the gait parameters of the whole body of AK amputees. It was found the AP knee moment and the fore-and-aft ground reaction force were the most sensitive parameters to alignment changes.

As mentioned above, the gait of an amputee is influenced by the components of the prosthesis. One of the most sensitive elements in the prosthesis is the socket, which should be carefully chosen and fabricated. The conventional socket for the above knee amputee is the quadrilateral socket which has been in use since 1964 (Schuch 1988a). In this type of socket, the ischial tuberosity is sitting on the posterior brim of the socket and supports the load. Lehneis (1985) has questioned the suitability of the quadrilateral socket and reported that it may be the reason for discomfort expressed by amputees. Long (1985) also reported that the quadrilateral socket is uncomfortable, producing an abducted gait, poor stability in the coronal plane, and its narrow AP dimension restricts the activities of the stump muscles. Long (1985) and Sabolich (1985) introduced a new socket design for above knee amputees. It is the ischial containment (IC) socket in which the ischial tuberosity is contained inside the socket and is aligned so as to maintain the femur in adduction and it is claimed to be more comfortable and to provide a more natural gait than the quadrilateral socket. This new design of socket has never been subjected to an evaluation programme independent of the development groups (Schuch 1988a) and the underlying biomechanics are not yet fully understood. Flandry et al (1989) stated that the IC socket showed superiority over the quadrilateral socket in four subjects out of five. It was found that the femur was adducted on average of 6.5 degrees, the lateral trunk lean disappeared, and the energy expenditure was reduced by 50%. In Flandry's study, only three subjects were assessed completely and the evaluation was largely subjective.

In order to follow the work discussed above and fill the gaps which were found in it, this study was planned and its aims were:

1- To study the gait patterns of AK amputees and their differences for amputees wearing IC and quadrilateral sockets. 2- To study the effect of alignment changes on the gait parameters of AK amputees and on the stability of the joints. 3- To identify the alignment parameters having the most influence on the gait variables and explore its repeatability. 4- To study the patient response to any alignment changes with a view to determine the method adopted for compensation. 5- To identify the optimum alignment and the values of its parameters, and the differences between the IC and quadrilateral sockets, and determine whether this is the alignment which was set subjectively as the optimum. 6- To carry out a systematic study of the data obtained to provide an objective method for assisting the prosthetist in fitting and achieving the optimum alignment of any prosthesis. 7- To compare the kinetic and kinematic parameters of the whole body for subjects wearing IC and quadrilateral sockets in order to find out which socket has advantages over the other.

It was planned, that a socket axis locator (designed by Szulc, 1983) and a coordinate measuring system (designed in Strathclyde University) were to be used for quick and accurate measurement of the eleven alignment parameters (defined by Solomonidis 1980) of AK prostheses. Two Kistler force plates were to be used to measure the ground reaction forces on the foot, and three TV cameras to determine the position of each segment of the body in space. The force plates and the TV cameras were to be operated at a rate of 50 Hz and the data were to be acquired simultaneously and synchronously by a PDP-11 minicomputer. Thus, the forces and moments acting at all the joints of both limbs and various other parameters of gait can be computed. This will produce a set of results corresponding to a number of limb alignment configurations which can then be compared to study the effect of alignment changes on the gait parameters and establish the optimum alignment. This set of results can also be used to study the function and stability of the joints, and establish a gait pattern for AK amputees. Comparing the result sets of the IC and quadrilateral sockets will provide an objective evaluation of the IC socket, and can determine which socket functions better. It was decided to study the effect of AP and ML alignment changes, but the need for a high number of subjects which requires a great deal of work and time, has forced the author to leave the ML study to be carried out by a colleague in this University.

It was planned that eight patients were to participate in this study, this was influenced by the long time which will be needed in order to complete the test of one patient. All of them were to be tested wearing the quadrilateral and the IC sockets, and were to be supplied with Otto Bock prostheses because the alignment unit of the Otto Bock system facilitates the aim of this study in changing the alignment of the prosthesis. Systematic changes in the socket flexion/extension angle, knee forwards/backwards position relative to the hip-ankle line and foot dorsi/plantar flexion angle were to be made. All the angular movements of the body segments were to be recorded together with the force reactions acting on the feet. All the temporal-distance parameters were to be calculated, as were the moments at the various joints. It was also planned, that before testing any patient, the above testing procedure was to be applied to ten normal subjects, in order to compare the data of normal subjects with that of the amputees.

Some limitations were foreseen in this work: 1- Because so many markers were to be used and only three cameras were installed in the laboratory, it was decided that the motion of the upper limbs was not to be recorded, and some markers (heel and tail) will be only seen by one camera. 2- The use of only two force plates will limit the gait monitoring to only one stride. 3- The long time which is needed for the data reduction and analysis, may dictate a reduction in the number of the tested patients.

This thesis contained seven chapters. chapter two deals with the literature review on body modelling, body segments parameters, methods and systems of measuring the kinetic and kinematic data, and the last section presents a representative review on the gait pattern of normals. Chapter three discusses the lower limb prosthetics, amputation surgery, AK sockets and their biomechanics, knee mechanisms, ankle/foot assembly and the modular systems. Alignment studies and gait patterns representative of AK amputees are also presented. The experimental and theoretical aspects of the work are presented in chapters four and five respectively. In chapter six, all the results are presented and discussed for the normals and the amputees. The conclusions and recommendations for future work are stated in chapter seven. Finally, four appendices are included to give further details of some calculations, and to present more results to the reader.

CHAPTER TWO

LITERATURE REVIEW ON HUMAN LOCOMOTION AND GAIT ANALYSIS

2.1 Introduction

In this chapter a review of published work in human locomotion and gait analysis is presented. Special emphasis is given to the literature relating to the work presented in this thesis. This will enable the author and the reader to compare this work with the work of previous investigators.

The review starts with definitions of human locomotion and gait analysis, and goes on to deal with the gait parameters and methods for their acquisition. The literature on normal gait given in this chapter is used later in the assessment of pathological gait. Literature on amputee gait will be presented in chapter 3.

2.2 Human Locomotion

Movement of the body from one place to another, requires movement of all the major segments of the human body. This movement of the human body and its segments is called " Human Locomotion ". The movement from one place to another is usually achieved by what is called "walking". Walking is a very complicated process, as it involves a three dimensional motion of a multiple linkage system which is controlled by the central nervous system. The quantitative description of the mechanical aspects of walking, and relating the motion of the human body, to the forces which actually produce the motion is referred to as " Gait Analysis ", and is the basis of understanding human locomotion.

Gait analysis and gait evaluation can be used in the laboratory or at a clinic, however, collaboration between the two makes the best use of them. By determining the parameters of normal gait in the laboratory, and comparing them with those of pathological gait in the clinic, the degree of disability of the patient's locomotor system can be assessed. Considering the whole human body as one object, the measurement of gait parameters can be limited to the measurement of the time over a fixed distance and the measurement of the three dimensional movement of the body centre of mass. This provides information to calculate the potential and kinetic energy of the human body. However, when considering the two legs separately, many more parameters would be measured such as the duration of the stance and swing phase for each leg, the three dimensional position of the joint centres in space, the three dimensional angles of each segment in space and in relation to the adjacent segment, the trunk extension/flexion, tilt and axial rotation, and the angle of arm movements. Many instruments have been developed to measure and analyse the gait parameters. Video recorders and cine photography could be used for visualising and tracking the pattern of the movement. Television cameras and associated equipment (computers, monitors, AD converters, ...etc.) are the basis of an automated system for the registration and measurement of the three dimensional movement of a marker which has been fixed on a body segment. The movement between the skin and the underlying bone causes inaccuracies which are estimated to be 10 to 20 mm (Morris 1991). Other techniques of measuring movement utilise devices that can be fixed on the body segment such as goniometers. This technique has other sources of inaccuracies such as the requirement for accurate alignment of the device with the anatomical joint axis and the change of the knee instantaneous axis position with increased flexion.

A complete gait analysis system should consider the measurement of the ground reaction forces acting on the feet. Most researchers use a force platform which measures six signals representing the ground forces on the foot in the direction of progression, laterally and upwards. These signals also present the moments of the resultant force about the three orthogonal axes (vertical, forwards and lateral) with origin at the centre of the force plate. With such platforms it is always difficult to ensure that the test subject lands completely on one platform with one foot only during normal walking. Thus, using a force platform for measuring the three ground reaction force components and the three moments, and a system to measure the three dimensional position of the body segments in space, it is possible to analyse the loads transmitted between adjacent body segments at the joints.

Human locomotion has been a subject of interest for many years. Although objective measurement systems which quantify locomotion have been in use for about a 100 years (Braune and Fischer 1895), it was not until World War II , when thousands of men returned home to the United States with amputations, that technology was really applied to the understanding of normal and prosthetic gait. A biomechanics laboratory was founded at the University of California to study and establish the fundamental principles of human gait, especially in relation to problems faced by lower limb amputees. The biomechanics laboratory's team used interrupted light photography to study the motion of the body segments, a force plate to measure the ground reaction forces acting on the feet, and electromyography to detect the activity of the muscles. Their aim was to provide fundamental data for the design of prosthetic limbs.

A comprehensive study of human locomotion should make use of three major fields of work. Firstly; obtaining a model for the human body and studying the body segment parameters for this model. Secondly; measuring the kinematic parameters which provide information about the motion of the body segments, and finally; measuring the kinetic parameters which provide information about the forces that produce the motion.

2.3 Human Model and Body Segment Parameters

Conducting any biomechanical analysis of human gait should start by deriving an adequate mechanical model of the human body and determining the physical properties of the segments of this model. These properties are important when one is performing a kinetic analysis.

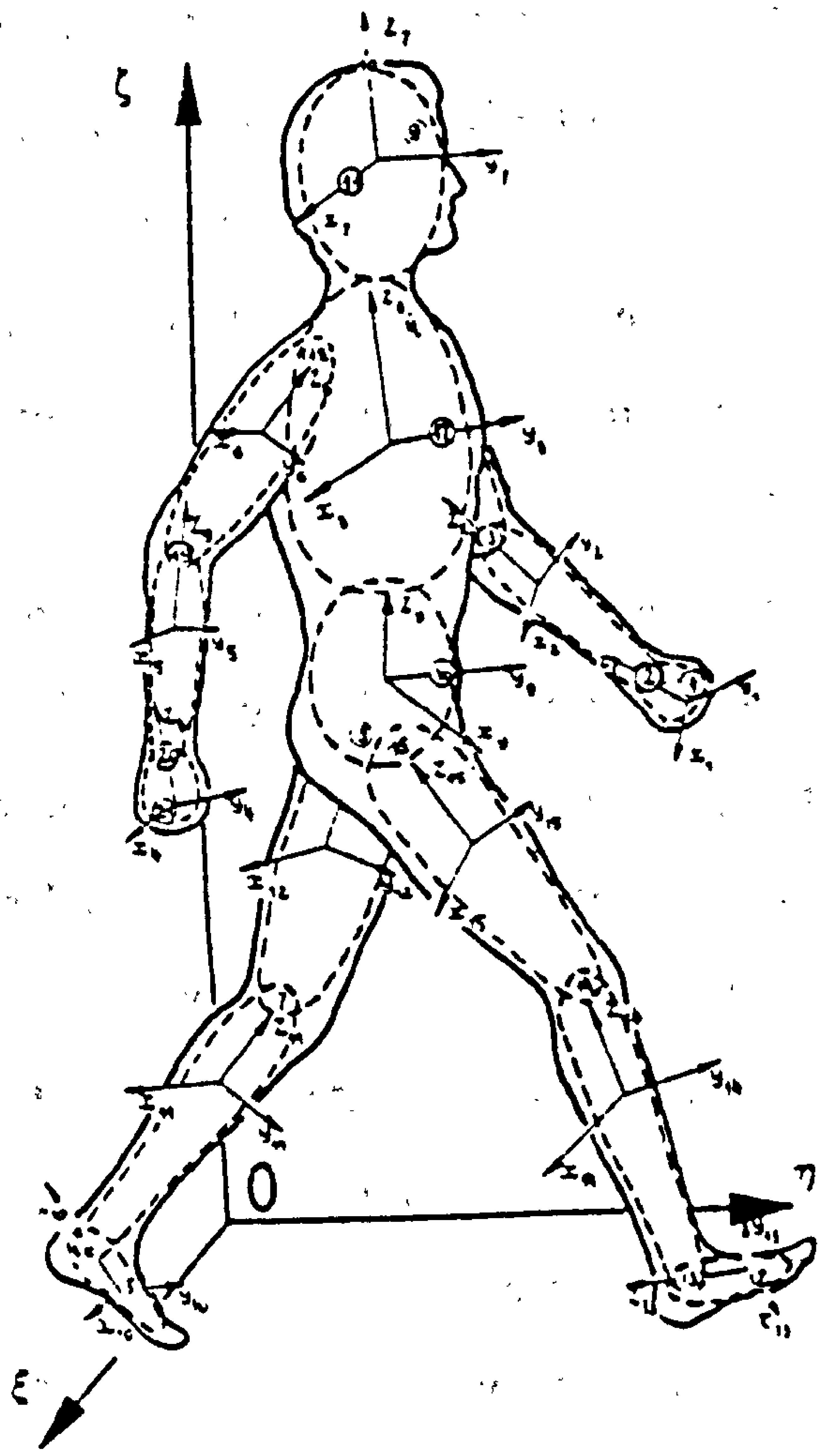


Figure 2.1 Model of the human body.
(from Aleshinsky and Zatsiorsky, 1978)

2.3.1 Human Model

The human body consists of several segments linked to each other to give the unique shape of Man. Each segment is joined to the adjacent segment by what is called a synovial joint, found where two or more bones are articulated.

For reasons of simplicity, it has been assumed that the segments of the human body are solid with uniform density and connected to each other by mechanical joints. The type of each mechanical joint is influenced by the joint investigated and by the accuracy needed in the results of the investigation. In general a synovial joint has six degrees of freedom, three translations and three rotations.

Clauser et al (1969) divided the human body into 14 segments in order to study the inertia properties.

Aleshinsky and Zatsiorsky (1978) introduced a 15-link chain model of the human body and carried out an analysis to determine the moment patterns arising at the major joints of the human skeleton during three dimensional motion. Figure 2.1 shows the human body model used by Aleshinsky and Zatsiorsky with the anatomical frames of reference and the fixed ground frame of reference. Hatze (1980) presented a human body model consisting of 17 segments and reported that a model with not less than 10 segments was essential for a realistic simulation of gross body motion. Figure 2.2 shows the 17 segments model of Hatze (1980). Goh (1982) used a model consisting of 8 segments. As he was studying the lower limbs, he ignored the head and the upper limbs and assumed their effect is reflected on the movement of the upper trunk . Knowing that Elftman (1939a) showed the effect of the arms on human gait, Yang Lang (1988) used the same model as Goh and achieved a result comparable with the results of previous investigators. The work presented in this thesis also used an 8 segment model, because it provides perfectly acceptable results for the purposes of this study which concentrates on the

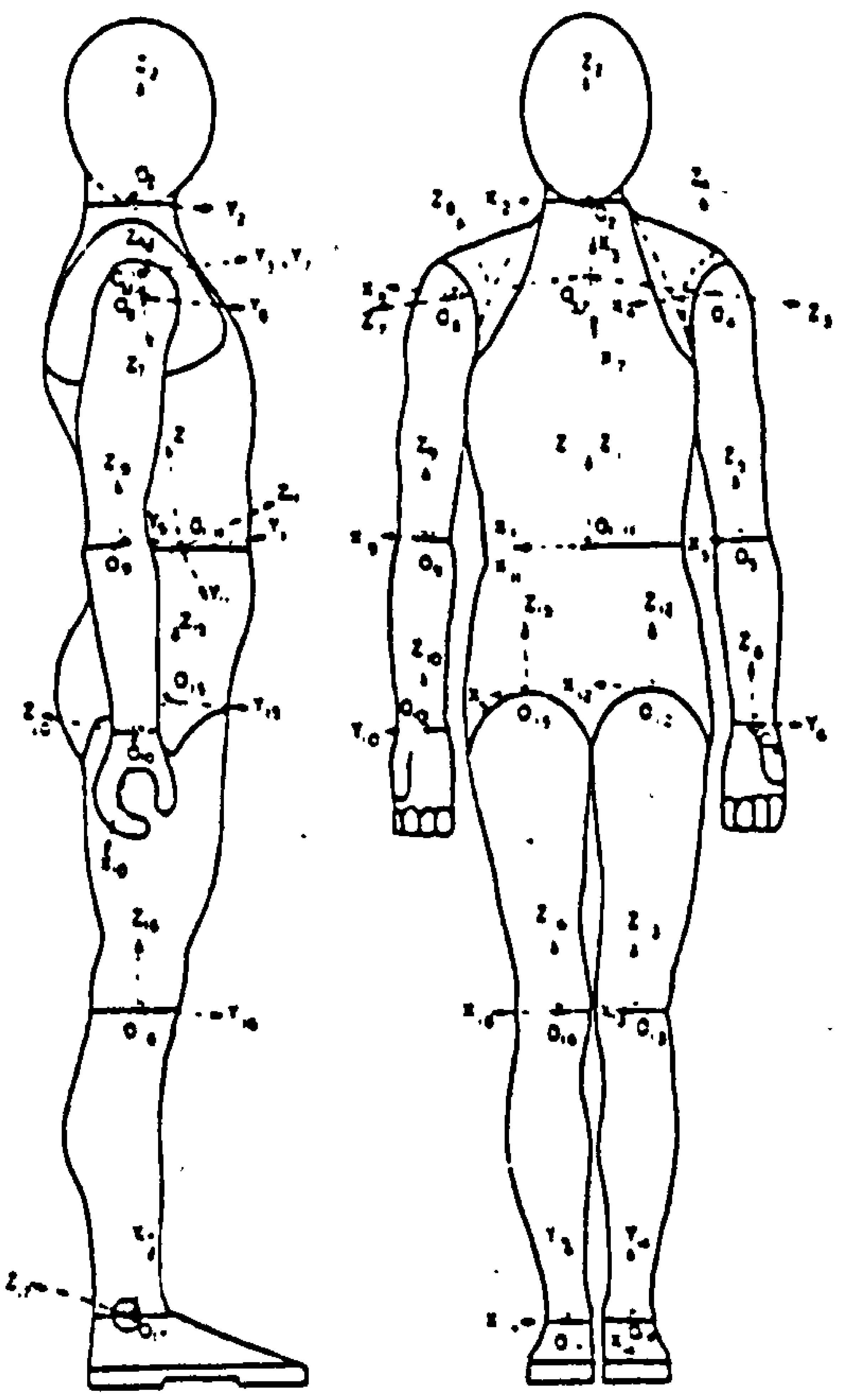


Figure 2.2 A 17 segments anthropomorphic model.
(from Hatze, 1980)

lower limbs. Inclusion of the upper limbs would have created some problems in the analysis, due to the intersection between markers, which would have confused the computer and given some errors, especially with the system used for data collection and analysis (see chapter four of this thesis).

2.3.2 Determination of Mass Properties of the Body Segments

As mentioned above, the mass properties of the body segments have a great effect on the locomotion of the human body. These properties are the mass, the location of the mass centre and the mass moment of inertia. For homogeneous material, the centre of mass of a body is generally at the centre of the volume of this body, but this is not the case in the human body segments.

The location of the centre of gravity (CG) for the whole body is affected by the location of the body segments and changes with body configuration. Paul and Barbenel (1974) stated that " Generally, in normal standing, the CG is situated at about 55% of the body height above the heel and 2 cm anterior to the lumbar vertebrae ". Finding the location of the centre of gravity for the body segments is much more difficult than that for the whole body and different methods have been used to determine the location of the mass centre and the other properties of the body segments.

Regression equations derived from data which were obtained from living subjects and cadavers may be used to determine body segment parameters. The coefficients of these equations are defined as follows:

$$C1=\frac{\textit{Segment Mass}}{\textit{Whole Body Mass}}$$

$$C2=\frac{\textit{Distance Of Mass Centre From Proximal Joint}}{\textit{Segment Length}}$$

$$C3 = \frac{\textit{Radius Of Gyration About An axis Through The CG}}{\textit{Segment Length}}$$

There are various methods to determine these coefficients and which have been reported as by many researchers, Dempster (1955), Drillis and Contini (1966), Clauser et al (1969), Contini (1972), Chandler (1975) and Hatze (1980). It has been reported that Harless (1860) was the first to introduce the method of coefficients. He used cadavers and presented data for C1 and C2 only.

Demspter (1955) produced relatively accurate values for C1, C2 and C3. The data have been taken from eight male cadavers and have been widely used. Drillis and Contini (1966) used twelve living subjects to measure the body segment parameters. They found that the mass of live subjects is not constant and that the volume and the volume distribution of the segments can vary with time. They also analysed the data from previous investigators and found that body parameters are different from one country to another.

Clauser et al (1969) measured body segments parameters for 13 adult male cadavers. Clauser et al found that more than one segment and more than one anthropometric measurment should be used in the regression equation to determine any body parameter for any segment, and the more segments involved the higher the accuracy obtained. These data were difficult to use, because the centres of body mass were related to certain bony landmarks and not to the joint centres. Therefore, Hinrichs (1990) adjusted these data to be related to the centre of joints. Contini (1972) presented data for 29 live normal subjects, and 19 live subjects with hemiplegia or amputation. The experiments on the normal subjects were conducted in two stages, the first based on 12 male subjects while the second was on 9 male and 8 female subjects. The results were found to be comparable within the whole subjects since the differences in mean values are of the order of 1% to 10%, and Contini stated that the mean values were useful in general computation. Because of the lack

Table 2.1 Body segment parameters, averaged for data from several investigators. (from Goh 1982).

Body Segment	C1	SD	C2	SD	C3	SD
Foot	0.0152	0.0019	0.5000	___	0.4750	___
Shank	0.0453	0.0035	0.4030	0.0260	0.2890	0.0134
Shank&Foot	0.0632	0.0036	0.4673	0.0455	0.3467	0.0639
Thigh	0.1027	0.0119	0.4169	0.0325	0.2912	0.0384

of a satisfactory method to find the centre of mass, it was assumed that the centre of mass is coincident with the centre of volume. Although it is a doubtful assumption the results were comparable with the average result of some other researchers.

Hatze (1980) considered a very complicated method to find the body segment parameters, based on 242 measurements taken directly from the subject. He represented the human body as a model of 17 segments and gave each segment a specific geometrical shape to find the mass properties. He reported that by means of a computer program, he could obtain the segment parameters with an overall accuracy of 3% and maximum error of 5% for each of the 17 segments in less than 80 minutes.

Goh (1982) carried out an extensive search on body segment parameters. He averaged the data of most of the researchers from Harless (1860) until Chandler (1975) and obtained practical values for C1, C2 and C3 as shown in table 2.1.

Kaleps et al (1984) used a combined stereophotometric and anthropometric method to determine the volume, centre of volume, principal moments and axes of inertia. Thereafter, a multiple regression equation was created to calculate the above parameters. The technique was used on living subjects, but only the head data was presented.

Schneider and Zernicke (1992) have studied the body segment parameters of infants aged 0.04 to 1.5 years, the segmental centre of mass location was found to have no systematic change with age.

2.4 Kinematic Measurement

As mentioned above kinematics is one of the ways to describe human locomotion, and can be measured by either direct or indirect methods.

2.4.1 Direct Measurement Technique

The direct measurement technique fixes a transducer to body parts to measure their relative movement and does not define the space position of the

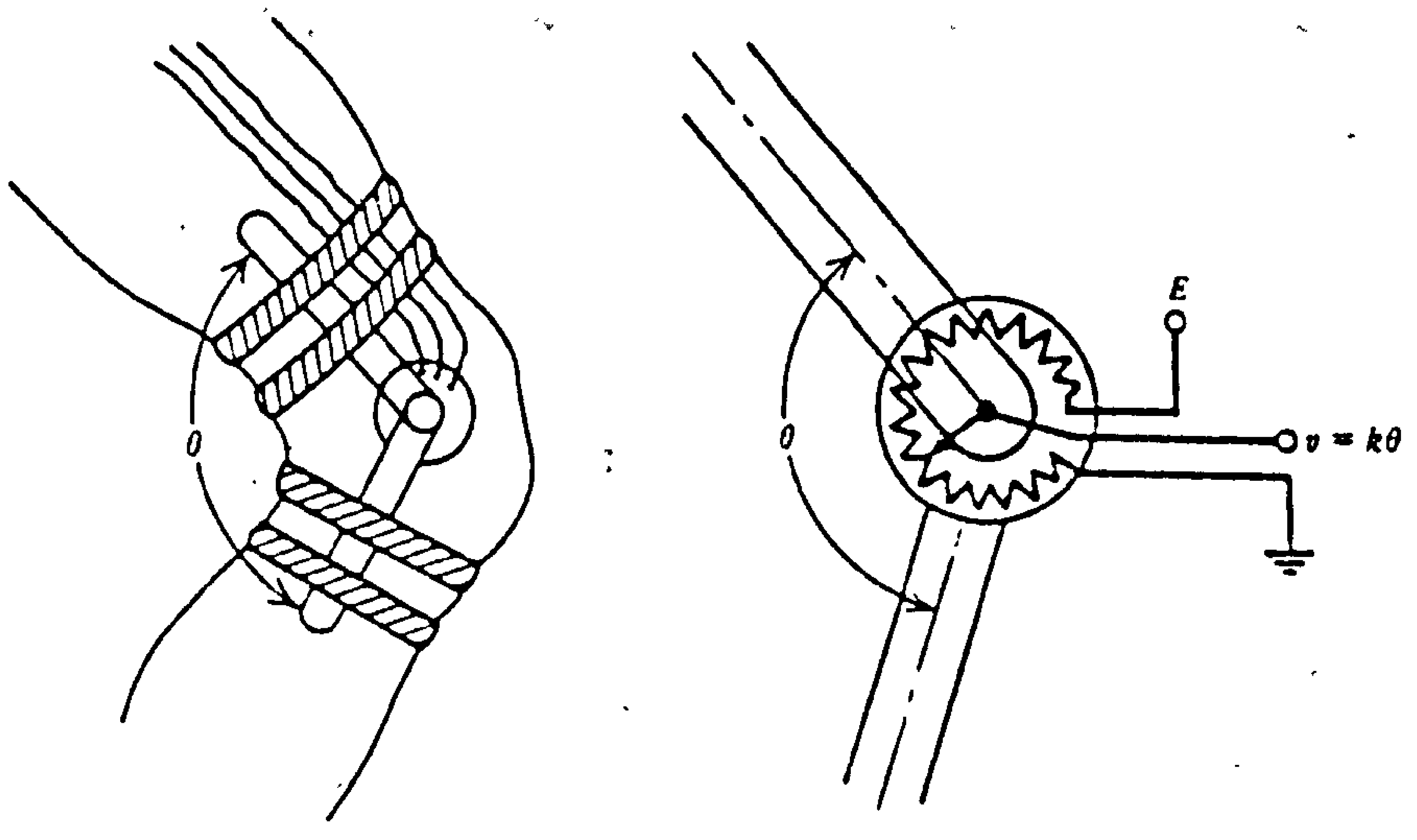


Figure 2.3 Mechanical and electrical arrangement of a goniometer located at the knee joint. Voltage output is proportional to the joint angle. (from Winter, 1979).

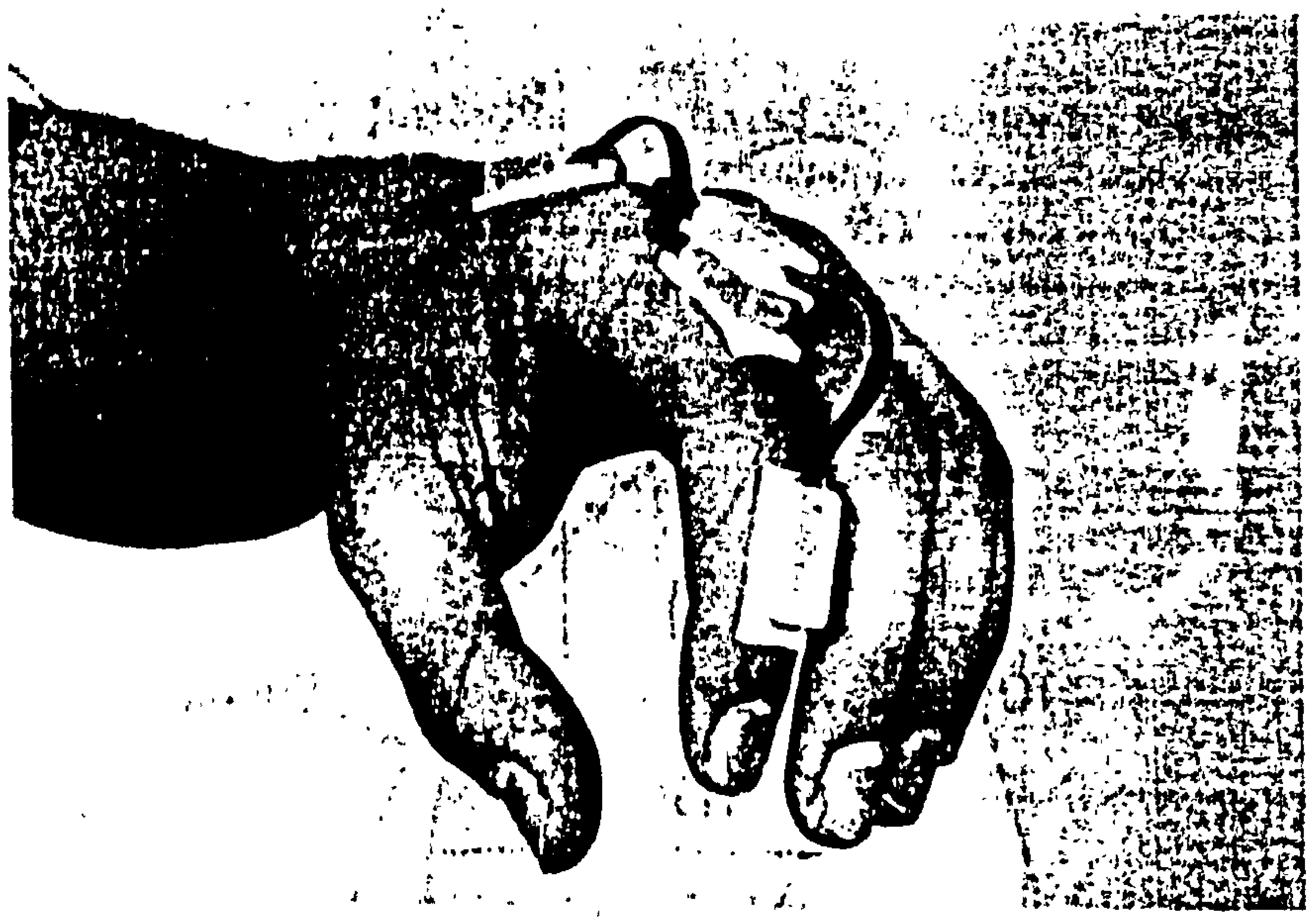


Figure 2.5 Flexible goniometer. (from Nicol, 1987)

body element. The signals from such transducers depend critically on their fixation relative to the instantaneous joint axes. The most widely used devices in the direct measurement technique are goniometers and accelerometers and the following is a brief historical review of the development and use of the two kind of devices.

2.4.1.1 Goniometers

One type of goniometer is an electrical potentiometer which can be attached to a mechanical or anatomical joint to measure the joint angle. Figure 2.3 shows a goniometer fitted to a knee joint, and the electric circuit is also drawn. The output voltage of the circuit always corresponds to the measured angle depending on the potentiometer and on the electrical circuit.

Johnston and Smidt (1969) used three electrogoniometers to measure the hip joint motion for thirty three normal subjects. Three potentiometers were oriented in each of the three primary axial planes of the hip joint. One was oriented in the sagittal plane and measured the hip flexion/extension angle, the second potentiometer was oriented in the coronal plane and measured the hip abduction/adduction angle. The third potentiometer was oriented in the transverse plane and measured the hip rotation angle. The electrogoniometric assembly was attached to a leather belt and laterally secured to the subject's pelvis, and was supported distally with firm elastic straps. The reliability of the system was tested by the use of a test-retest technique, and a range of 2 to 4 degrees of difference was obtained.

Lamoreux (1971), Kinzel et al (1972), Mitchelson (1975), Townsend et al (1977) and Chao (1980) have developed different types of goniometers. Chao (1980) designed a triaxial goniometer based on the gyroscopic concept and used three potentiometers, to measure the three dimensional angular motion of a joint (fig. 2.4). Each potentiometer was fixed according to the general gyroscopic mechanism to measure a single rotation angle. The rotational axis of the flexion/extension angle was fixed to a rigid bar which was attached to

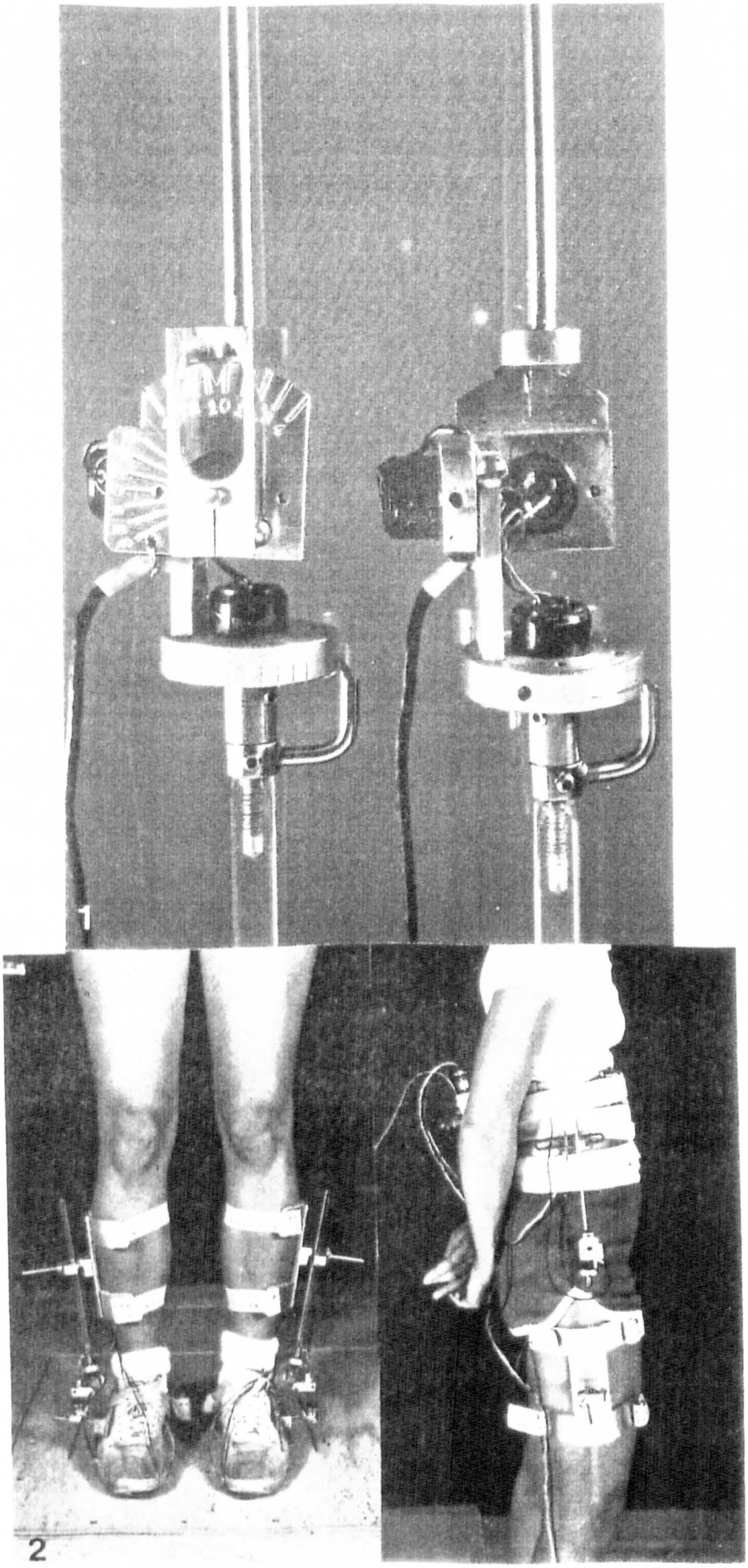


Figure 2.4 Triaxial goniometer for the measurement of the three-dimensional angular motion of joints. (from Chao 1980).

the proximal limb segment of the joint. The axis of the axial rotation angle was fixed to the distal joint segment by a telescoping rectangular rod through a yoke, this arrangement was to keep the axis of the goniometer parallel to the rotation axis of the joint throughout the range of the motion. The axis of the abduction/adduction angle was perpendicular to the axial rotation and the flexion/extension axes. Since the goniometer assembly was attached to the side of the joint the axes of the goniometer were not coincident with the joint axes of rotation and did not intersect at the centre of the joint to be measured. This would cause considerable error in the measurement, however, Chao called this error "cross talk" and he reported that this error was corrected. This system seems to match the requirements of both mechanics and functional anatomy, but the system has some disadvantages and its overall accuracy was not determined. However, it was reported that the system has an error of 2 degrees for all angular components which were estimated by placing the system between two plastic cylindrical bodies and then the angulations of these two bodies were measured by the system and by X-rays. It would have been more interesting if the error was estimated under subject testing conditions. Aligning this goniometer was quite difficult and failure to achieve proper alignment would cause a considerable error in the measurement. Also the movement between the device and the underlying tissues causes errors which must be kept to a minimum. The weight, size and wiring of this goniometer may disturb the subject and force him to alter his gait pattern.

Johnson et al (1981), Morimoto and Tsuchiya (1985) and Nicol (1987) have separately developed flexible goniometers. As a result of the flexibility, the goniometers have self adjustment features and they can be fixed very easily to the joint to be measured. Johnson et al reported a mercury-in-rubber strain gauge for measuring the movement of the joints. The gauge was made from a 200 mm length of 0.2 mm bore, 0.5 mm outside diameter rubber tubing which was filled with clean mercury. It was reported that 50 patients were tested, however, the accuracy of the device was not estimated. Morimoto et al (1987) used the goniometer which was presented by Morimoto and Tsuchiya in 1985 (flexible electrogoniometer made of silicon rubber column), and presented a comparison graph for the knee angle measured by that flexible goniometer and by a movie camera. The data from the movie camera and from the goniometer coincided. Nicol (1987) studied the mercury-filled tube goniometers and reported that the flexibility of this goniometer, was undermined by the instability and potential hazard of the mercury. He therefore introduced a flexible single-axis electrogoniometer which is very similar to that presented by Morimoto and Tsuchiya. It uses a narrow steel foil fitted with long strain gauges. The electrogoniometer has a resolution of 0.02 degrees, and it has a very useful feature in that the measured angle is independent of the shape of the bend along the length of the foil. Moreover, it will measure the angle of joint movement regardless of skin stretch or poor alignment with the axis of rotation. Figure 2.5 shows the device measuring the motion of a finger joint. This goniometer is now manufactured by the company "Penny and Giles", to measure two dimensional joint rotations with infinite resolution. Flexible goniometers were found to be useful and reliable, and a "Penny and Giles G180" flexible goniometer was used for assessment of the accuracy of uniaxial accelerometers by Willemsen et al (1990).

Hull et al (1990) used an electromechanical goniometer to measure the hip motion in cycling while "standing" in a stationary exercise machine. It has a linear resolution of 0.2 mm, rotation resolution of 0.15 degrees, and an angular error of 0.19 degrees. For more details the reader is referred to the original paper.

2.4.1.2 Accelerometry

An accelerometer is a device which measures the reaction force resulting from a linear acceleration. The transducer is usually either of the strain gauge, or piezoresistive type. Inside the accelerometer there is a known mass which

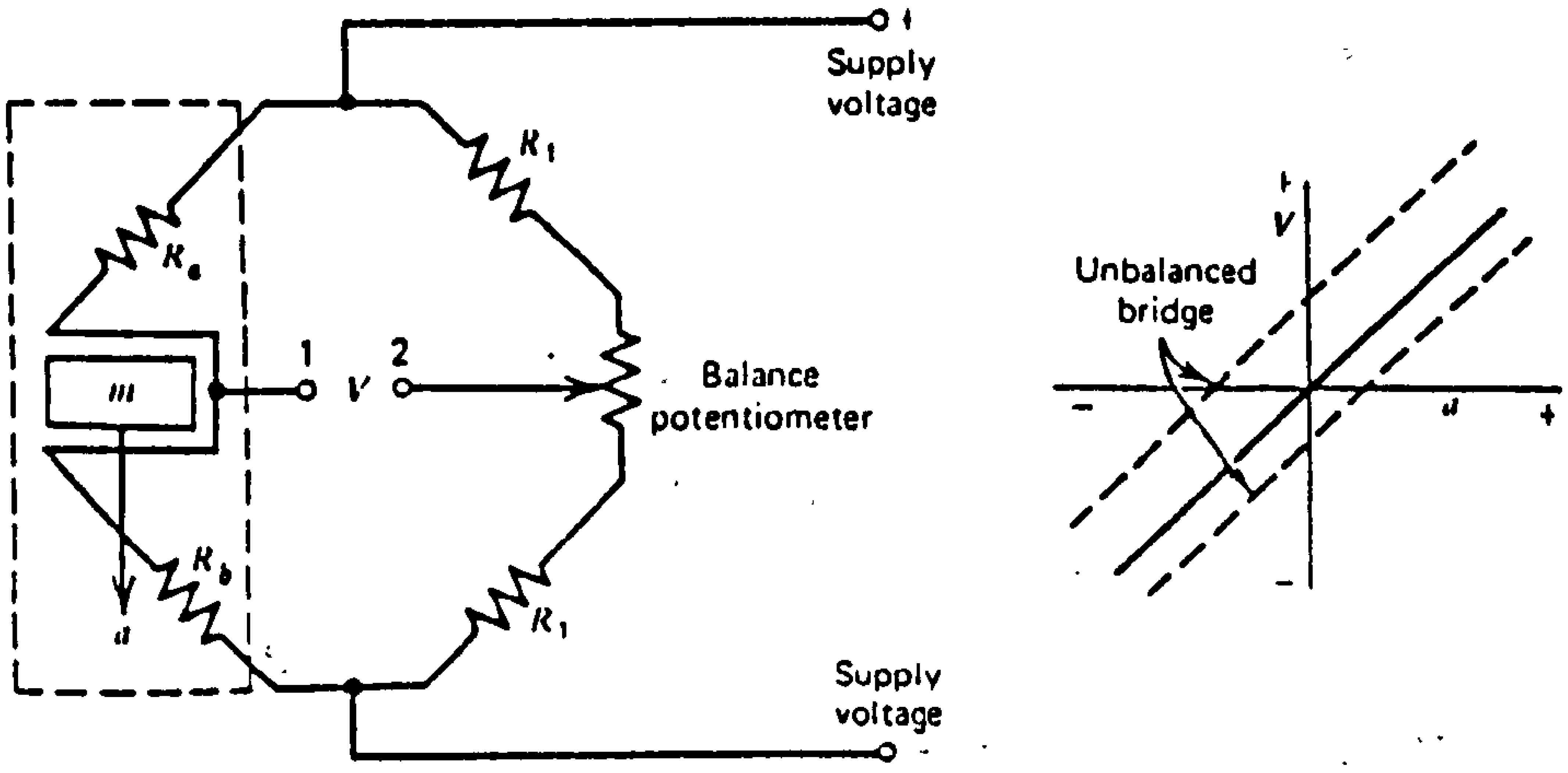


Figure 2.6 Electrical bridge circuit used in most accelerometers.
m is a mass producing a force which proportional to the acceleration " a ". This force is measured by the force transducer. (adopted from Winter, 1979)

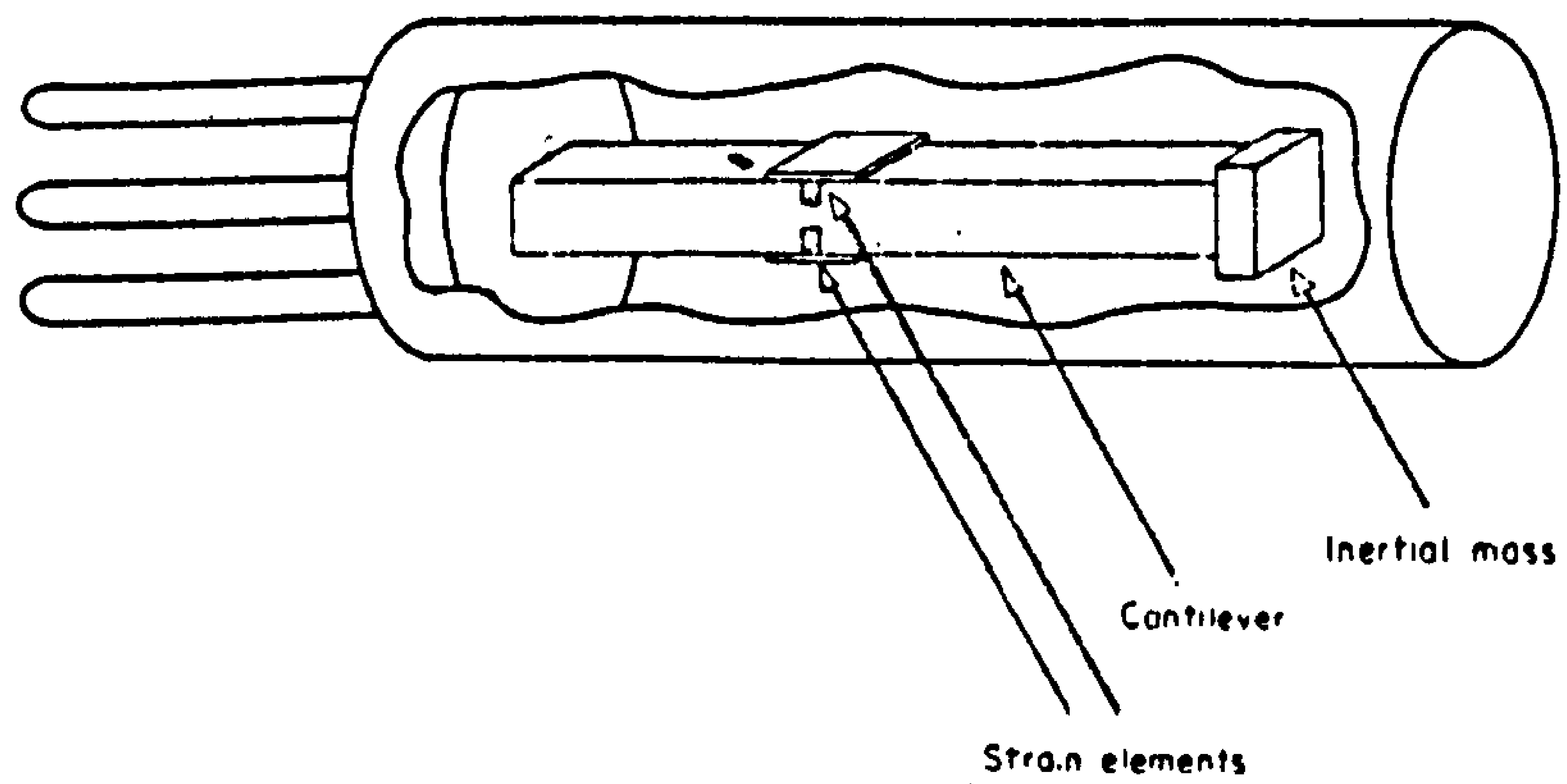


Figure 2.7 An accelerometer. (from Morris, 1973)

is usually accelerated against a force transducer. The transducer will generate a voltage signal which is proportional to the force resulting from the acceleration. To measure the three components of the acceleration, a triaxial accelerometer should be used. It consists of three individual accelerometers mounted at right angles to each other. Figure 2.6 shows the electrical bridge circuit used in accelerometers, and figure 2.7 shows an accelerometer. The advantage of this method of measurement is that the output is available to be recorded or displayed. There are some disadvantages in this method: 1- the acceleration measured is at the fixation point in the limb. 2- the device is sensitive to shocks and can easily break down. 3- if a large number of accelerometers is used they could interfere with the gait and would be expensive. 4- after measurement of the acceleration, the angular displacement is found by integrating the angular acceleration, this could generate drift and distort the result.

Morris (1973) and Smidt et al (1977) used accelerometers in gait analysis. Willemsen et al (1990) showed that by using pairs of accelerometers, relative angles (ankle, knee and hip angles) can be calculated without integration. Thus, the error of integration drift which is normally associated with accelerometry, can be avoided.

2.4.2 Imaging Methods

The imaging measurement technique is one of the most successful methods of recording and studying animal locomotion, and in particular human motion. The technique depends on recording the image of body segments, or the image of a few landmarks which are identified on the body segment. Then, by using mathematical calculation and the imaging laws, the position of the body segment in space can be determined. A continuous recording of the body segment's position in space, together with the time intervals between the successive positions of this segment are the information required to obtain all the kinematic variables. By combining the kinematic data with the ground

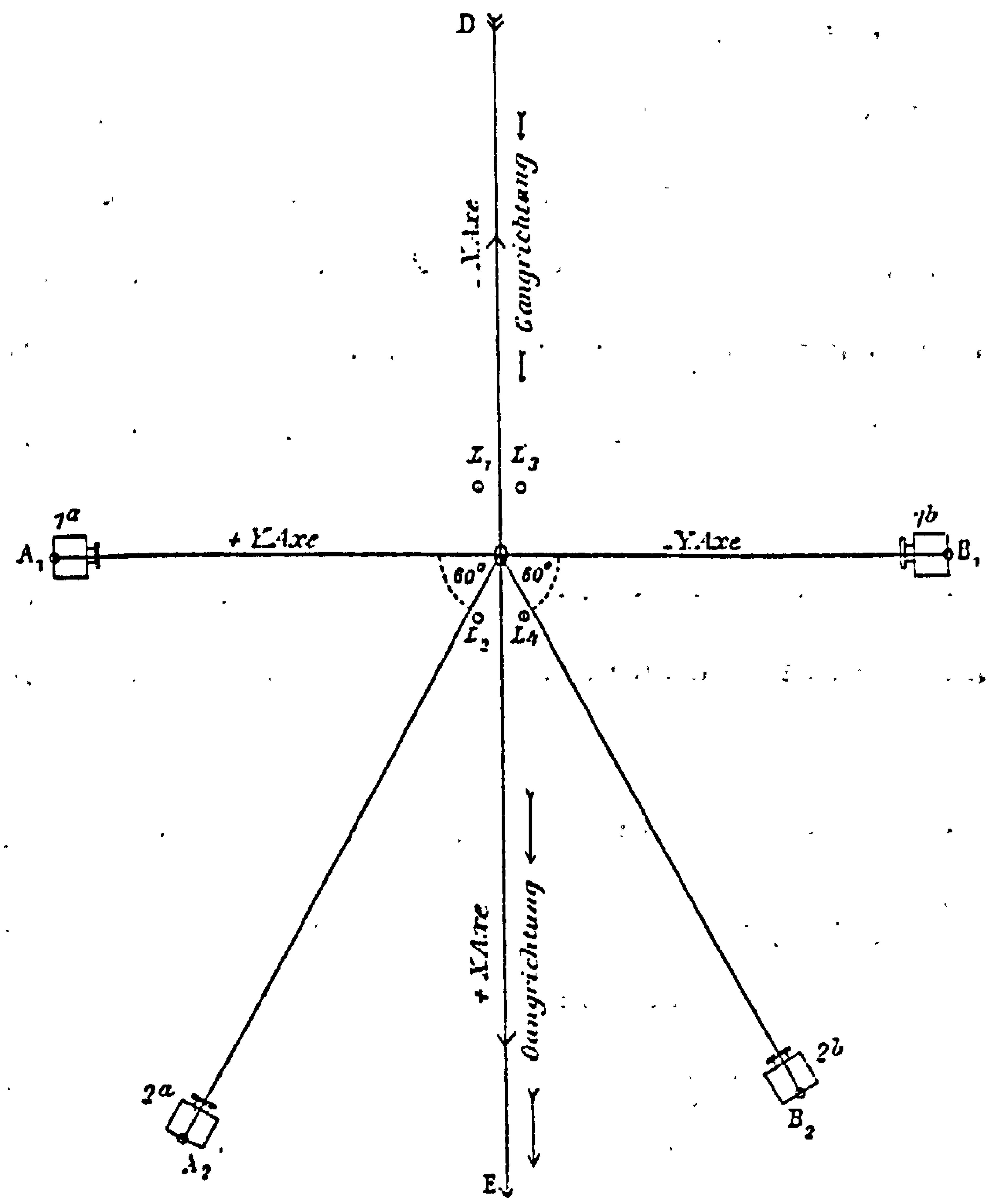


Figure 2.8 The layout of the four cameras in Braune and Fischer's experiments. (from Braune and Fischer, 1895)

reaction forces data, a comprehensive analysis of the gait can be obtained.

Imaging measurement techniques, can be classified in two main categories: the photographic methods which have been in use since the late nineteenth century, and the optoelectronic methods which have been developed and used since the 1960s.

2.4.2.1 Photographic Methods

One of the most economical and simplest imaging systems is the photographic based, which includes chronophotography which records the whole movement of the body segment on a single photographic plate, and cinematography which records the instantaneous movements of the body on separate frames with a time interval between successive frames.

Chronophotography

Although the use of photography as a scientific research tool is credited to Marey of Paris (1882), Muybridge of San Francisco, at about the same time, was the first to use chronophotography to determine the phases of movement (Braune and Fischer, 1895). Muybridge set up a series of cameras close to one another and opened the shutters one after the other. The purpose was to record the successive movement phases of a horse. Later, after the discovery of the sensitive bromine-silver-gelatine plate, Muybridge obtained instantaneous pictures of the locomotion of human and various animals with great accuracy. Some of the animals were photographed from two sides.

Marey (1882) used an exposure of 1/700 s and a plate at the back wall of the camera rotating about its axis in 1 s, so that a dozen pictures per second was obtained. This was a great improvement in photography, and was the birth of the cinematographic technique.

Braune and Fischer (1895) used four cameras located as seen in figure 2.8 to photograph their subject, and obtain the three dimensional trajectory of each joint. The subject was dressed in a black jersey suit and the various parts of his body were illuminated by " Geissler " tubes which emit light intermittently at a frequency of 26.09 flashes per second.

Eberhart (1947) used the interrupted light technique, the field of view of the camera was interrupted by a rotating disc with an 18 degree opening in front of the lens. The disc rotated at a rate of 30 revolution per second to give an exposure time of 1/600 of a second. Later (in 1954) Eberhart et al concluded that the interrupted light technique was only useful for obtaining simple stick diagrams.

Murray et al (1964) used interrupted light photography to record the displacements associated with locomotion. A Speed Graphic camera was used to photograph subjects who were appropriately marked with reflective targets and walked in a hallway in semi-darkness at a distance of 16 feet (4.88 m) from the camera. A source of interrupted light flashing at a rate of twenty times per second, was used, together with a mirror which was mounted over the walking area. By this means overhead projections and sagittal displacements were recorded. The mirror and camera were tipped upward from the horizontal at an angle of 25 and 15 degrees respectively. It was reported that the errors resulting from this photographic method were systematically determined by photographing grids located parallel to the plane of progression, parallel to the horizontal and at a variety of angles from the horizontal. It was found that correction of the results was only necessary for the lateral oscillation parameters. However, the method of using data from the grids in order to estimate the errors, was not discussed. Also, the accuracy and resolution of the camera were not given, and the error in the derived data was not stated.

Cappozzo (1982) used chronophotography. Using four ordinary open shutter cameras placed in stereoscopic pairs, he obtained 3D quantitative information of the linear displacement of the head and trunk during the process of walking in a straight line at different speeds. The targets on the subject were light emitting diodes (LEDs) which were driven at a frequency range of 30 to 60 impulses per second. The laboratory was illuminated with a green

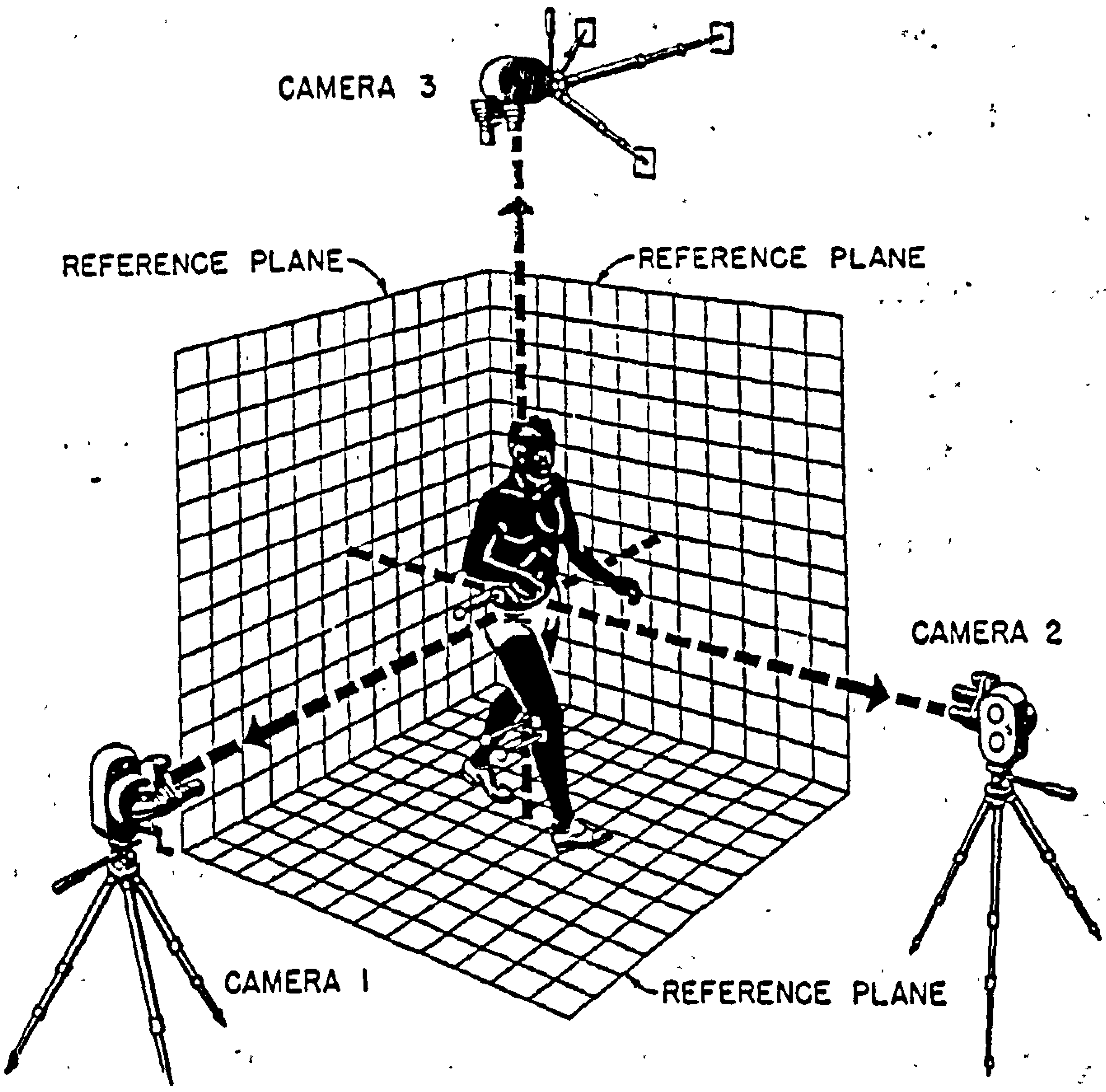


Figure 2.9 Arrangement for recording the "pin-study" data.
(from Eberhart, 1947)

light to improve the contrast of the yellow light of the LEDs relative to the surrounding. Residual systematic errors of 2.5 mm on the X and Z coordinates, and 8 mm on the Y coordinate were obtained.

Chronophotography is a simple, inexpensive, and accurate technique for studying human locomotion. Some disadvantages have restricted the applications of this technique. One of these disadvantages is the overlapping of images on one another which may give a misleading picture, especially when the point is not moving or when the point is moving very slowly. The overlapping markers merge and may make the identification process difficult. Another disadvantage of this technique is that the darkened room, greenlighting, and flashing strobe light can disturb the subject and may influence the gait.

Cinematography

As mentioned above, the birth of cinematography was in (1882) when Marey used a camera with a rotating plate behind it, and operated at a rate of 12 frames per second. Elftman (1934) used a cine-camera running at a rate of 72 frames per second to study the distribution of pressure under the human foot. This was by placing a rubber mat studded with pyramidical projection upon the heavy glass plate, then noting the increase in area of each pyramid in contact with the glass as the subject walked on the rubber mat. Eberhart (1947), Bresler and Frankle (1950) and Eberhart et al (1954) used cinematography. Eberhart employed three 35 mm motion-picture cameras operating at 48 frames per second, to determine the magnitude of transverse rotations of the segments of the lower extremity. The cameras were located as shown in figure 2.9 and 3D coordinates were obtained for the wooden targets. The target was fixed to a stainless steel pin of 2.5 mm in diameter, and the pin was inserted into the bone under local anaesthesia. Although, the parallax errors were corrected, the accuracy of obtained data was still doubtful because of the painful method of pin fixation which certainly must have been disturbing to the subject. This method can also be costly because the 35 mm film is very expensive.

Paul (1967), Goh (1982), Winter (1984) and Stuberg et al (1988) have used cine-cameras for gait analysis. Paul used two 16 mm cine cameras operated at a rate of 50 frames per second to record the 3D positions of the leg in space, together with a force platform to measure the resultant force acting between the floor and the foot. His aim was to obtain the forces transmitted between the loaded surfaces of the human hip joint, in order to assist the design of implants for fractured bone or for joint replacement. Parallax errors were corrected and good accuracy was obtained in the results, but the use of only two cameras limited the photographic procedure to one side only.

Cinematography is an accurate technique for a comprehensive study of human motion. It does overcome the problem of image overlapping in chronophotography, but it is an expensive technique and data reduction is time consuming.

2.4.2.2 Optoelectric Methods

As mentioned above, processing of the cine data is time consuming, and the quality of the data acquired cannot be known until data reduction is carried out. For these reasons, scientists in the late 1960s started to develop new techniques for three dimensional analysis of human locomotion that could be used where the cine technique has failed. The result of the scientists' efforts to fill the gap in the cine technique was the introduction of optoelectric techniques. A major advantage of these techniques over the cine techniques is the capability of instant replay of the collected data, this serving both as a quality control check and as an initial quantitative assessment of the acquired data. There are several kinds of optoelectric techniques and the following is a brief summary of them and their development stages.

Television-Computer System

Since the 1960s, the television-computer system has developed rapidly

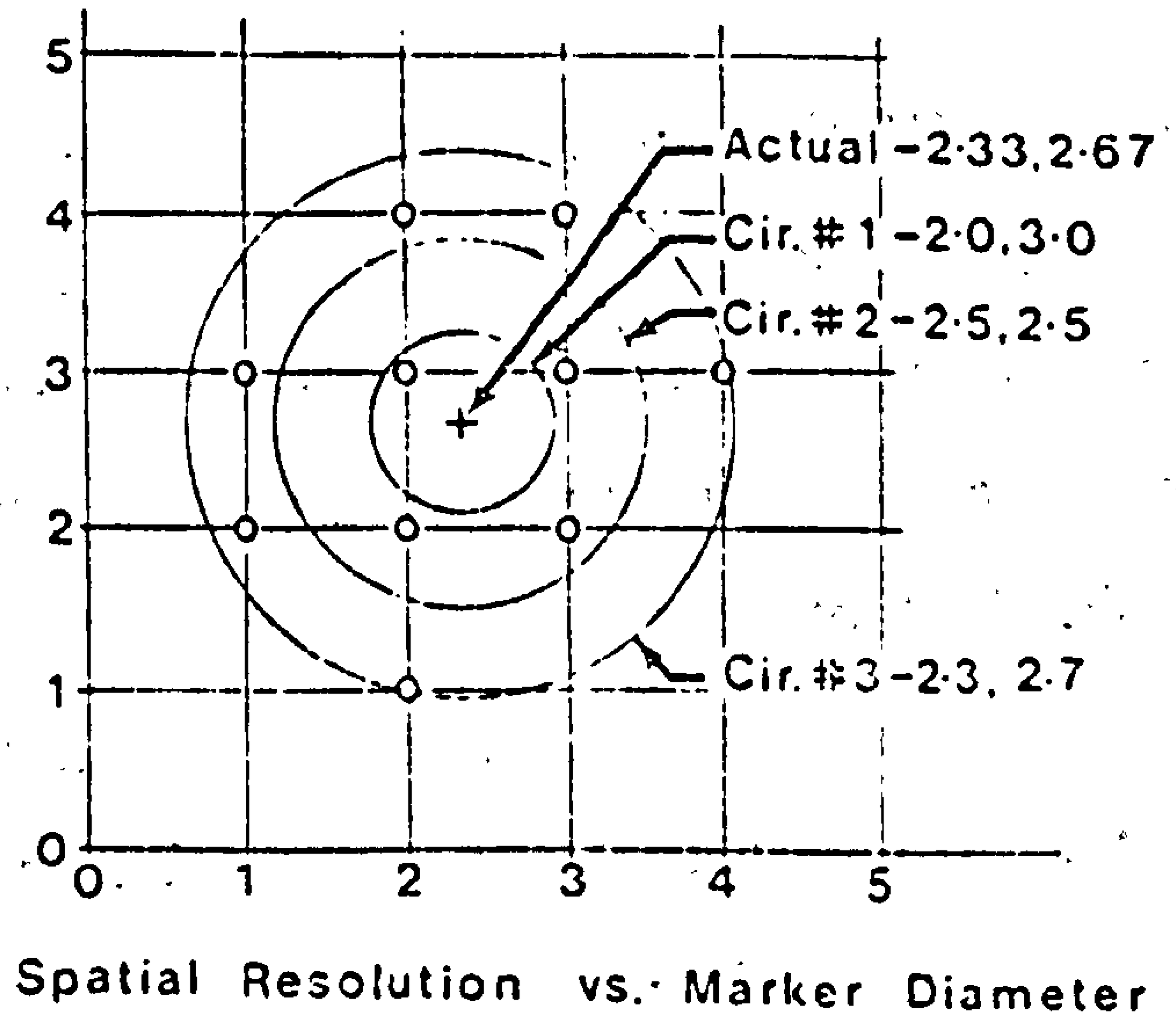


Figure 2.10 The influence of marker size on the accuracy of calculating the coordinates of its centre.
(from Winter et al 1972)

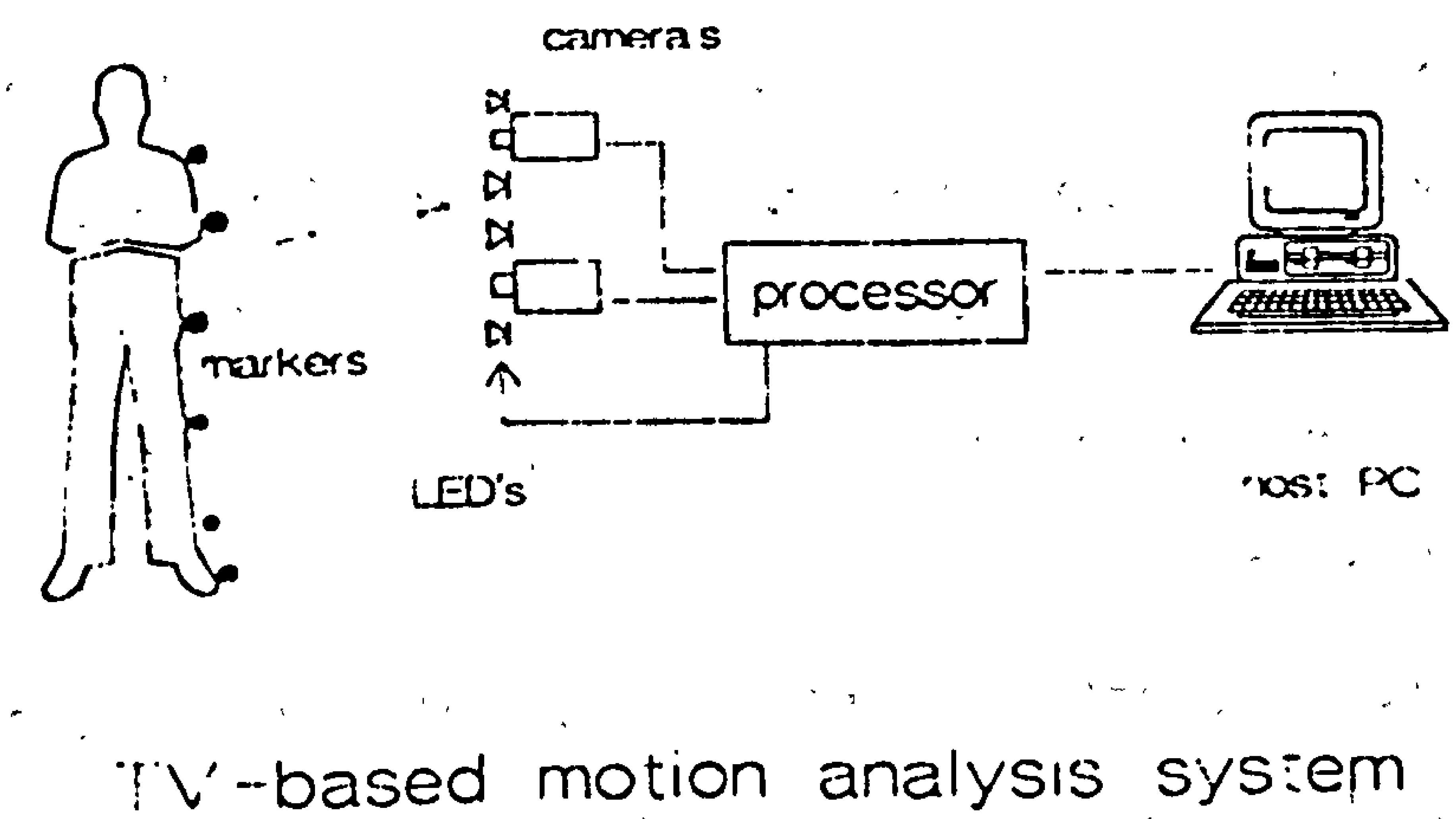


Figure 2.11 PRIMAS, TV based motion analysis system.
(from Furnee, 1988)

and several systems based on this technique have been established. This technique is probably the most popular of all the optoelectric techniques. All the television-computer systems share the same principle. The spatial coordinates of the marker are obtained by sampling and registering the marker image for each TV field scan. When a marker is detected, a digital counter will register the sampling of the horizontal and vertical lines of the marker's position. This will be stored in a buffer to be transferred later to the computer's memory. The size of the marker can influence the accuracy of its calculated centre, this influence can be seen in figure 2.10 which also gives an idea of the TV sampling system. If the marker is small, there will be only one sample point in the image circle of this marker and the computer will consider the centre of this marker to be at that sample point. But if the marker is larger, there will be a number of sample points in the image circle and the computer will calculate the average of them. If the marker is large enough, the desired spatial resolution can be obtained without the need of sampling every TV line and thus sampling every 2nd line will be sufficient (Winter et al 1972). This will reduce the data rate and volume. The interface allows several synchronised TV camera outputs to be processed, so that a 3D analysis of the motion can be obtained.

Furnee (1967) described the first TV/computer system for the study of arm movement. Active light markers were attached to certain landmarks on the body segment which were under investigation. A TV camera was employed to pick up the marker's positions. The TV signal was fed to a video-digital data processor which produced digitally the coordinates of the marker as seen by the camera. The digital position of the marker could then be stored in a computer. The sampling rate was 50 Hz which is the television frame frequency. No data were presented and the accuracy of the system was not mentioned, but the author made claims for the feasibility, reliability, and convenience in the setting up of the system. Only one camera was employed so that only 2D analysis could be obtained, but it was reported that the system could incorporate up to 5 cameras. Furnee (1988) presented the latest version of this TV/Computer based system which was developed over a long period of time, and it is now commercially known under the name 'PRIMAS'. It is a multi-camera system for 3D motion analysis, operating at standard rates of 50 and 100 Hz and can also be operated at a rate of 200 Hz. It uses up to 100 passive retro-reflective markers and can obtain a precision of 1:18000 in the horizontal lines and 1:14000 in the vertical lines. Figure 2.11 shows a schematic diagram of the PRIMAS. The PRIMAS system including the TV cameras, is manufactured by High Technology Holland BV (HTH) in Eindhoven in the Netherlands.

Winter et al (1972) introduced another TV/Computer system, in which a TV camera was placed at about 3 m from the subject, and operated at a rate of 60 Hz to pick up the position of the body markers which were low inertia hemispherical reflectors. The camera was able to monitor the subject for up to 5 strides and the data were stored in a computer. The absolute coordinates of each marker were calculated using reference markers which were of a different size from the body markers and located at a known distance from each other in the subject's background. It has been stated that the X and Y coordinates of the centre of a marker can be calculated to a spatial resolution of 1 mm. This system requires a large computer memory as it stores the whole television image and later a computer program is used to cluster the marker points and calculate the coordinates of their centres. Because of the use of only one camera, this system is only able to acquire 2D data.

Jarrett (1976), working at Strathclyde University, developed a new TV/Computer system. Passive markers were used and the coordinates of the circumference of the marker's image circle were recorded unlike Winter et al (1972) who sampled continuously. Although, only two cameras were installed, the system could employ up to six cameras. A digital counter system was used

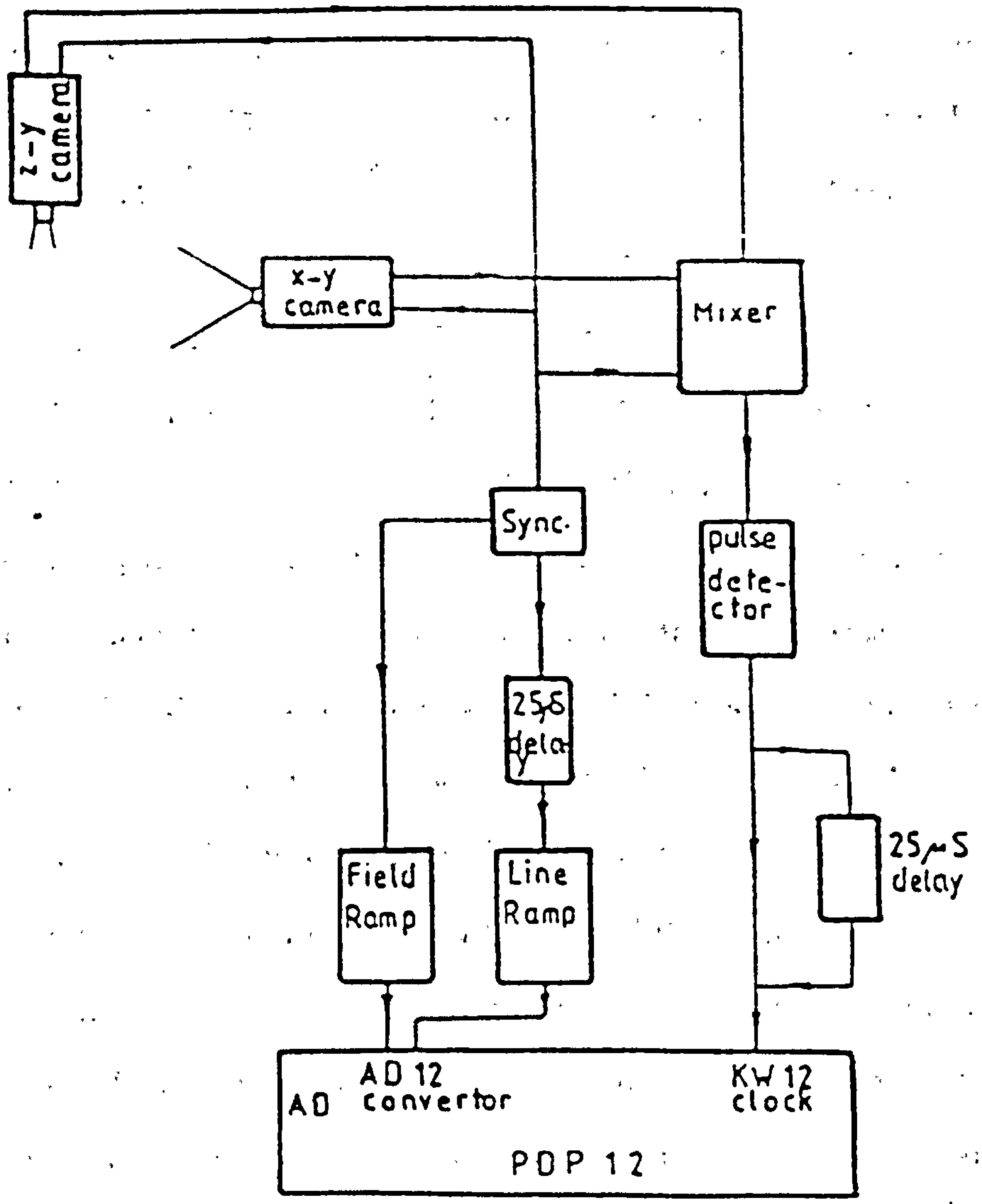


Figure 2.12 Block diagram of the TV/Computer system presented by Jarrett. (from Jarrett, 1976)

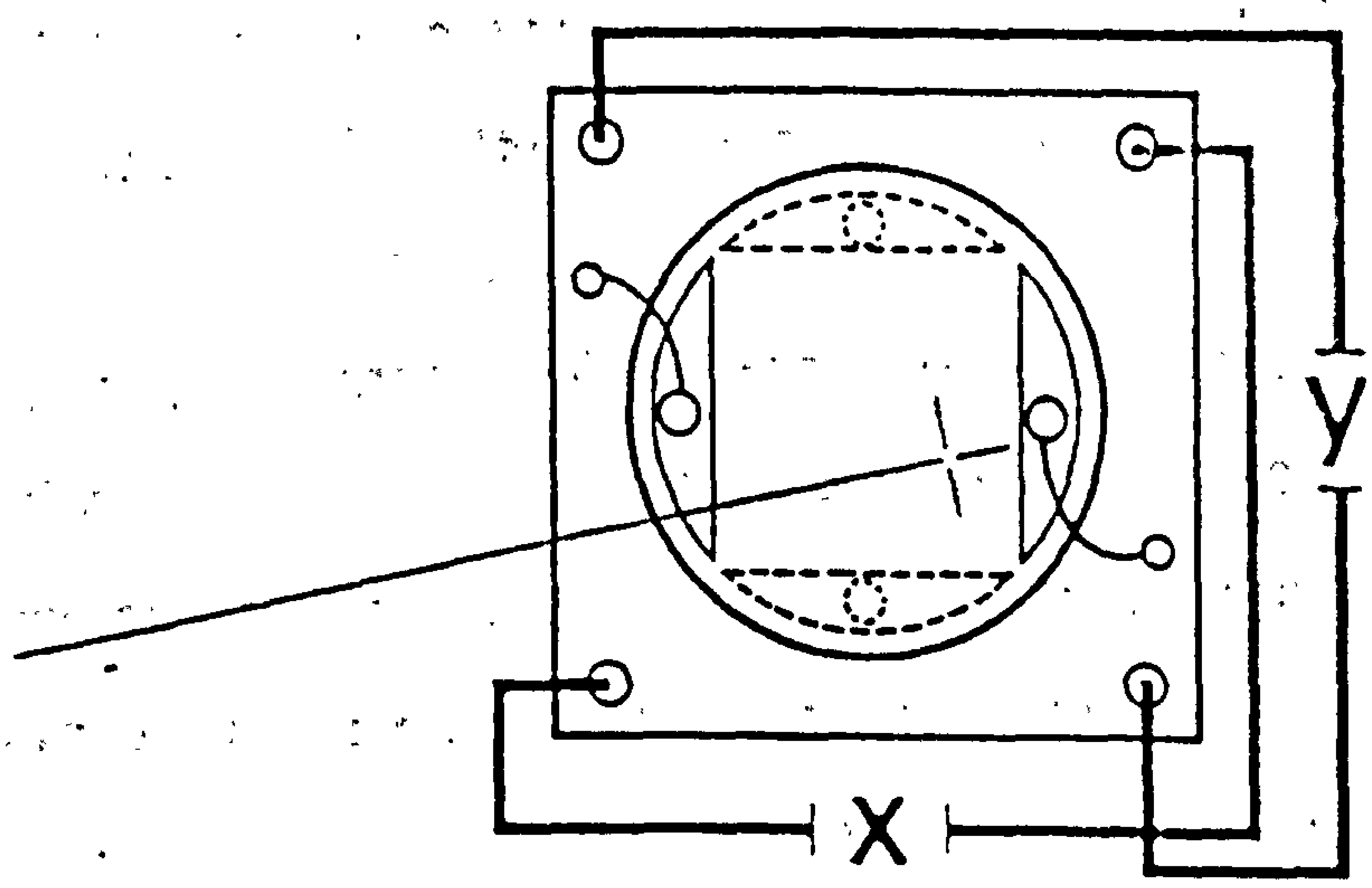
and the target's coordinates were stored temporarily in buffer memory to be transferred later to a PDP12 minicomputer. It was stated that this system had a resolution of 0.1% for the horizontal coordinate and 0.3% for the vertical coordinate. The cameras were located as seen in figure 2.12 and the sampling rate was 50 Hz. To iluminate the passive retro reflective markers, a tungsten halogen light source was placed close to the camera, however, this proved to be disturbing for the subject. Moreover, installing, two cameras only limited study, by this system, to only one side of the subject.

Andrews et al (1981) were concerned with improving the accuracy of the Strathclyde TV/Computer system and with making it easier to use and less disturbing to the subject. To avoid the use of visible light, infra-red light was employed together with rotary shutters. The shutter was placed between the camera's lens and the vidicon faceplate, and its aperture synchronised with the TV field scan. The shutter opening of each camera can be offset from the others, so that the camera will not be blinded by the light of the opposite one. With the above development, in addition to employing a third camera and a new PDP11 minicomputer, the Strathclyde system was able to perform 3D analysis of the whole body. This system has been in successful use since 1981, and was used for the analysis of the work which is presented in this thesis. More details of this system will be discussed in chapter four.

Oxford Metrics Limited (1980), produced the Strathclyde system commercially under the name 'VICON'. It is a three dimensional TV/Computer system for motion analysis with an accuracy of 0.1 percent of the field of view. This means that an accuracy of 2 mm can be obtained in a space of measurement of 2 metres. Lightweight reflective markers are used, and up to seven cameras can be used at a time. The cameras are operated at a frequency of 50 Hz and can be run at 200 Hz according to the camera type. During data collection the data are stored in an 'Etherbox' and are transferred to a computer at the end of the test. The Etherbox is an intelligent stand-alone system for real time video coordinate generation and storage and can take up to seven cameras plus digitised signals from up to 64 analog channels.

Whittle (1982) conducted an evaluation of the stability and the accuracy of the VICON system. The frame to frame accuracy was tested and found to have a standard deviation of 2 mm, but the interface introduced an additional error of 3 mm. The standard deviation of relative measurements was found to be 2.6 mm (static measurement), and the standard deviation under operational conditions was between 4 and 5 mm (absolute accuracy for a moving object) at a camera-to-subject distance of 4 m. For long term stability, it was found that a warm up period of between one to two hours was necessary before using the system for data acquisition. It should be noted that attempting to use VICON for data acquisition before a minimum of one hour of warming up will give very misleading data.

Oxford Metric Ltd proposed a standard test protocol, for the assessment of a 3D kinematic system accuracy. Measurement volume, marker size, sensor position and the parameters derived from the measured 3D positions of the marker, were taken into consideration when the protocol was proposed. Using this test protocol Morris (1991) at Oxford Metrics Ltd conducted a test on the stability of the VICON system and found that over 50 samples of 20 markers, the standard deviation was 0.262 mm for the X, 0.345 mm for the Y, and 0.303 mm for the Z direction. The static accuracy was estimated by measuring the coordinates of 20 markers which were located at known positions. For over 100 samples, a mean of 1.67 mm error was reported. For the small displacement test (dynamic test), Morris measured the length of three pendulae as these oscillated with an excursion of the lower end of each pendulum equal to about 40 cm. The standard deviation was found to be 0.579 mm for the first length, 0.496 mm for the second length and 0.543 mm for the third length. However, the error of the measurement was not reported for this displacement test and the length of each pendulum was not clearly stated. A large



The unique photodetector facilitates accurate, two-dimensional positioning. (3D-positioning is made in the computer analysis).

Figure 2.13 The SELSPOT photodetector unit. (from Selcom, 1988) displacement test was also conducted by placing two markers at each end of a rod of 978 mm length, the rod was moved in four different directions and its length was measured. The results had a mean and standard deviation of 979.38 ± 0.56 mm, 978.40 ± 0.65 mm, 979.07 ± 0.74 mm and 978.79 ± 0.57 mm for each of the four directions.

The VICON system has wide acceptance in the world of motion analysis, and this system was installed in Strathclyde University and has been in use since April 1990.

Ferrigno and Pedotti (1985) presented another TV/Computer system which is commercially available under the name " ELITE ". It was developed at the bioengineering centre in Milan in 1983. It is based on real-time processing of the TV images to obtain the 3D coordinates of passive markers. The system is operated at a sampling rate of 50 Hz independent of the number of markers, and provides a resolution of one part in 2500.

SELSPOT System

SELSPOT is a non-contact optoelectric measurement system. The original concept was born in the early 1970s in the field of medical technology, and it was made commercially available by Selcom AB in Sweden (1975). Selcom (1988) provided the latest description of their product under the name SELSPOT II. The system can use up to 120 markers which are active LEDs each of which emits light in a predetermined sequence. This is governed by a control unit which is attached to the test subject and can control up to 8 LEDs. This can be disturbing for the subject if human locomotion is monitored. The system can use up to 16 cameras at different rates of sampling frequency depending on the number of markers. Despite that, the system is able to sample at a rate of 200 Hz if 50 markers are in use. The camera is a photodetector unit which registers only the light pulses from the LEDs. The photodetector is a flat semi-conductor disc, each side of which is coated with a light sensitive silicon. When light from the LEDs strikes a point on the disc,

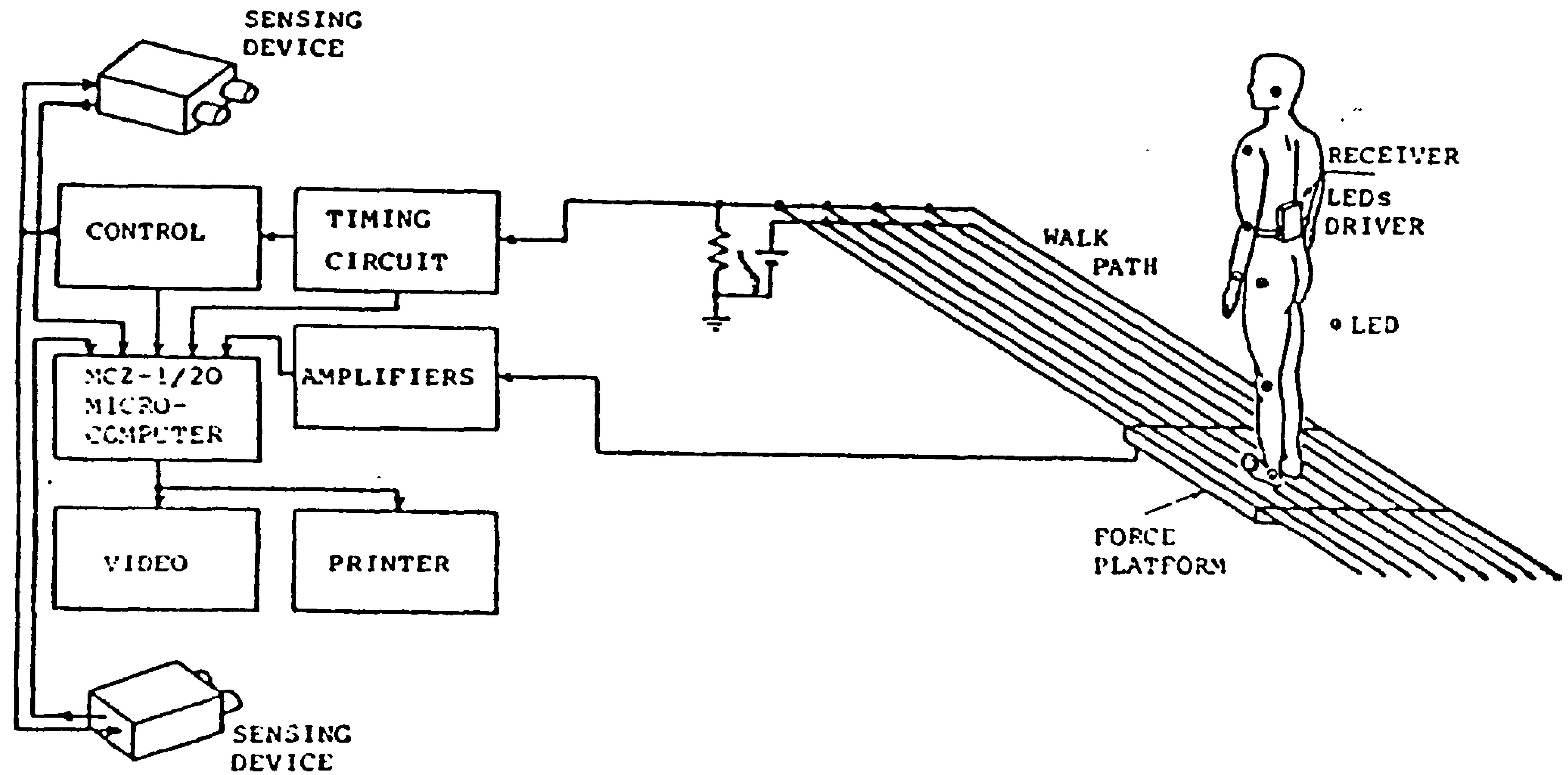


Figure 2.14 Data acquisition system. (from Leo and Macellari, 1981)

the electronics of the camera will register X and Y coordinates which will enable the system to determine the point from which the light pulse was emitted (fig. 2.13). The data are stored in a computer for data reduction and 3D production. It has been reported that the system has a resolution of 0.025% and an accuracy of 0.5% of the measurement range.

Paul and Nicol (1981) examined the useability of the SELSPOT system in biomechanical data acquisition. They concluded that it is by no means a simple task to obtain accurate and reliable data. Any reflected stray light will be recorded along with the actual image on the detector, and the output will be their average. Another area of error in the SELSPOT system is electronic noise, especially if long range measurements are required. Selcom (1988) stated that all these error sources no longer exist in the latest version of SELSPOT. Stokes (1984) tested the relative accuracy of SELSPOT and found that the static error ($|\Delta x|$, $|\Delta y|$, $|\Delta z|$) is 0.9, 0.6 and 0.5 cm in the x, y and z directions respectively. The dynamic standard deviations were 0.06, 0.04 and 0.04 cm for the x,y and z directions respectively, but the errors were not reported.

Leo and Macellari (1981) developed an on-line microcomputer system for gait analysis based on the SELSPOT optoelectric system. It is a 3D measuring system to study whole body movement. The targets are active infrared LEDs which fired sequentially from top to bottom through a telecontrol system. Thus, no wiring was needed and the subject walked freely. The system can use up to 8 markers on each side of the subject at a sampling rate of 359 Hz. The sensing device consists of two identical transducers , each of which is an optoelectric camera detecting one coordinate of the position of a landmark. Figure 2.14 shows the data acquisition system with the relative subsystem. It should be noted that the system in its present state has only one sensing device, therefore the overall accuracy can not be determined but it was stated by Leo and Macellari that each device has a resolution of 1.6 mm.

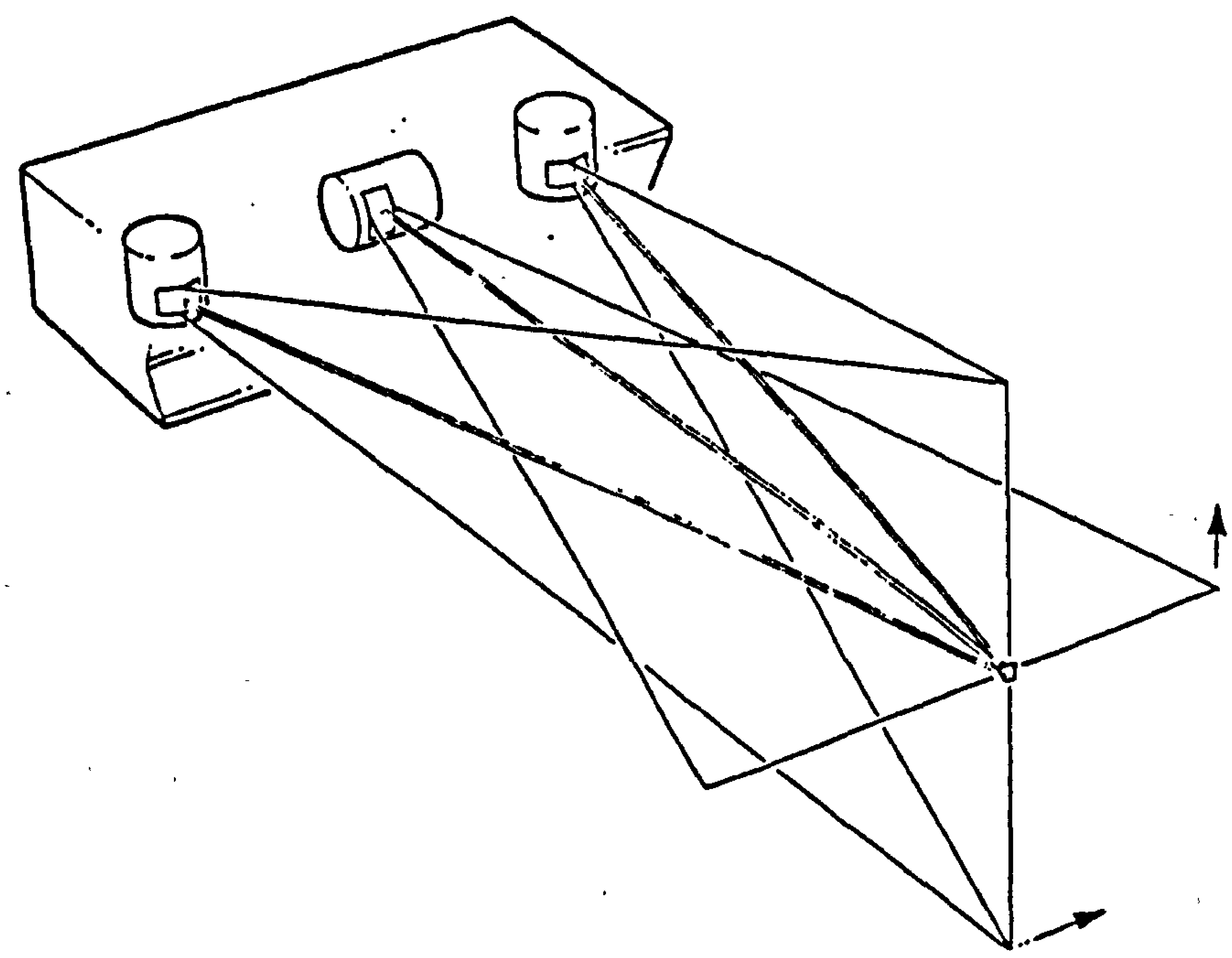


Figure 2.15 Relative position of the scanners and the scanning light beams in Coda-3 system. (from Mitchelson 1988)

The CODA-3 System

CODA-3 is an advanced remote movement monitoring system. It can track the three dimensional positions of 12 passive markers which have the shape of a pyramid, in real time and at a sampling frequency of 300 Hz. The system is portable and could use any computer for data reduction and storage. Its development started in the 1970s at Loughborough University and was continued from 1980 by Movement Techniques Limited. Mitchelson (1988) presented the system in its latest specification. The detecting unit has three optical scanners mounted as seen in figure 2.15, and each of them has a multi-faceted mirror illuminated by a powerful arc lamp. When light is reflected from a marker, it will be received by the scanner which determines the angular position of the marker in the scanner's field of view. The two scanners, which are mounted at each end of the unit, determine the horizontal coordinates X and Z. The third scanner, which is in the midpoint of the unit, determines the vertical coordinate Y. Mitchelson stated that the standard deviation of a stationary marker was 0.2 mm in the X and Y axes, and 1.4 mm in the Z axis when the distance between the marker and the scanning unit was 4 metres, and the sampling frequency was 300 Hz. O'Brien and Jenkinson (1991) have used the CODA-3 system to analyze the gait pattern of twenty normal children. No data were presented, but they stated that the CODA-3 system is simple, accurate, and provides the measurement with the minimum of interference to the child being tested. It would be useful to have the accuracy of the system stated, if it was obtained.

The WATSMART System

The Canadian company, Northern Digital Inc. (1983) developed a three dimensional motion analysis system known as WATSMART. It uses up to 4 cameras and can track up to 64 markers and calculates their 3D coordinates in real time with an accuracy of 1 to 3 mm at a distance between the camera and the marker of from 1.1 to 8 meters. The markers are active infra-red light emitting diodes which are individually activated in sequence at a very high frequency. The sampling rate is 4700 markers per second independent of the number of cameras (i.e. 4700 marker per second per camera). It should be mentioned, that the wiring of the markers can disturb the subject and could alter the gait. Northern Digital Inc. (1989) presented the latest version of WATSMART under the name OPTOTRAK. It can use up to 24 cameras and may track up to 256 markers. It is not affected by infrared reflections and its 3D accuracy is 0.1 mm and 0.61 mm, for distances between the camera and the marker of 0.914 and 4.877 m respectively.

Video Based System

A three dimensional measuring system, based on a video camera and video processors, has been developed and has been in use for quite a long time. The Motion Analysis Corporation of California developed two systems based on video analysis. The systems are Flextrak 3-D and ExpertVision 3-D. They are very similar in their performances and specifications. For this reason only one of them (ExpertVision) will be discussed. Motion Analysis corporation (1990) reported that ExpertVision was a 3D system providing up to six cameras real-time input at a rate of 200 HZ via videotape, based on a VP320 video processor. It can track up to 30 markers which are active LEDs. The camera is a shuttered video camera with a variable speed of 60 to 200 HZ. It is reported that the system is highly accurate, but no figures have been given.

The optoelectric method has the advantage of providing instant replay which serves as a quality control check of the data. It can also automatically manipulate raw data. However, it has some disadvantages over the photographic type such as the resolution and high cost of installation, and some of the optoeletric systems can be disturbing (if light was used) to the subject if human motion is being monitored.

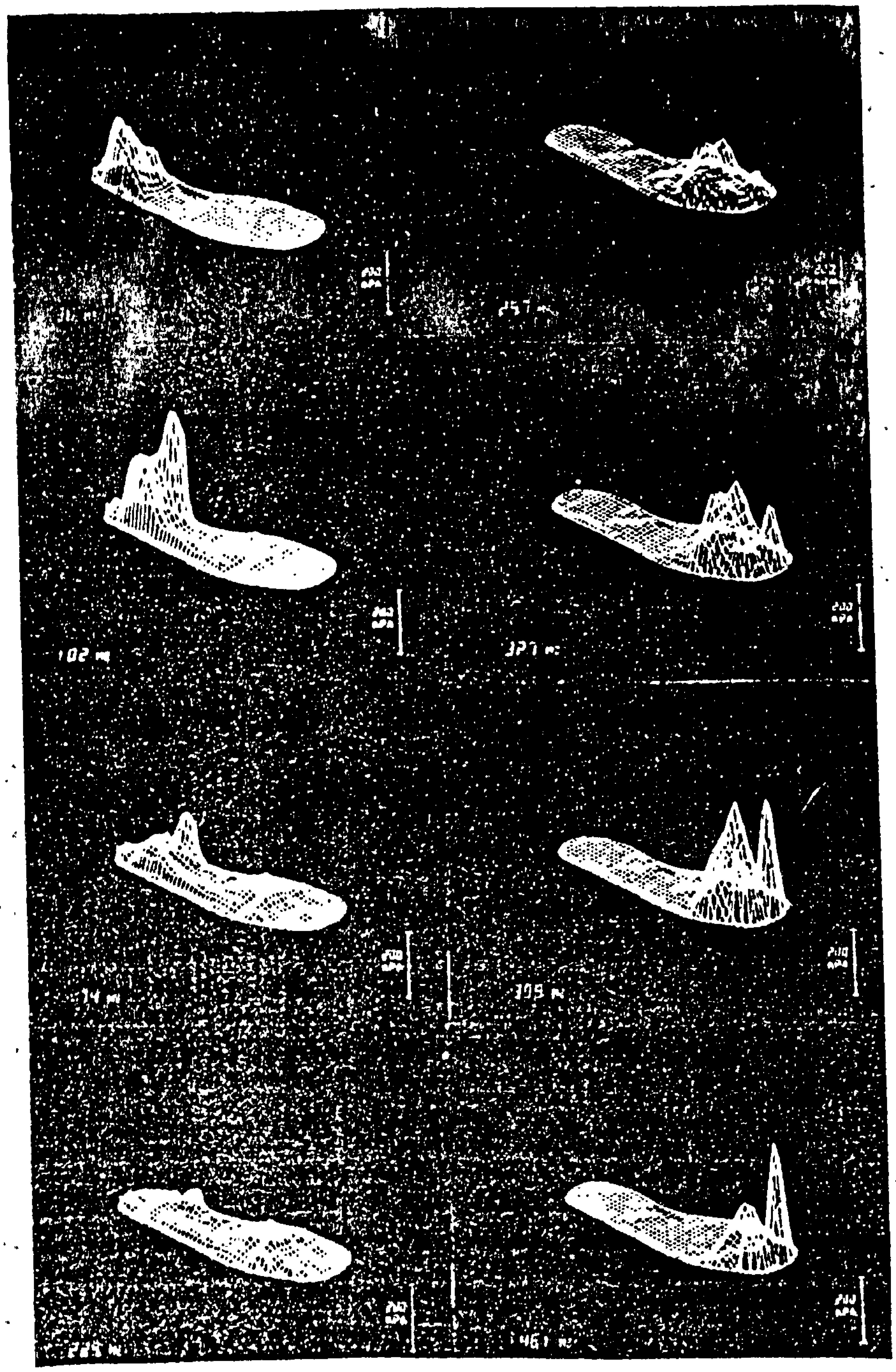


Figure 2.16 Sequence showing pressure distribution under the foot during walking. Units: the pressure unit is kPa (Pa=N/m^2), and length of line shown corresponds to 200 kPa , and the time unit is ms. (from Cavanagh et al 1983)

2.5 Methods for Measuring Kinetic Parameters

In the science of human movement analysis, kinetics is the study of the forces which cause movement. The muscles generate external forces which act on certain areas of the foot, these are the ground reaction forces. These forces overcome gravity and friction forces and cause body movement at a certain velocity and acceleration. The ground reaction force can always be presented by a three dimensional vector acting at the centre of pressure. There are several methods for the measurement of the ground to foot forces; most experimenters use a force platform to record information enabling the calculation of the reaction forces and moments about the reference axes and the required coordinates of the centre of pressure. Some experimenters use pressure devices which measure the pressure distribution on the foot.

2.5.1 Pressure Distribution Measurement

Several researchers have measured and studied the pressure distribution under the foot and it is still the subject of interest for many researchers. Elftman (1934), Nicol and Hennig (1976), Chodera and Lord (1978), and Hennig et al (1983) used various methods for such a study. The work of Hennig et al has been chosen as an example to be discussed since it is the most recent and was interesting. They described a piezoelectric device for studying the distribution of the vertical contact stresses between the foot and the shoe insole. The device is a " pressure-sensitive shoe insole " and has 499 lead Zirconate titanate transducers embedded in a layer of silicon rubber. Each transducer is a part of a simple electronic circuit which can be used as a charge amplifier. The amplifier is installed in a small box of 2.9 kg weight which can be carried on the subject's back. The output voltage of each transducer is directly proportional to the charge generated by the transducer. As the transducer is made of an oriented piece of material, it can be mounted in such a way that the generated charge is proportional to the vertical force and independent of the horizontal shear. The data can be stored in a PC computer,

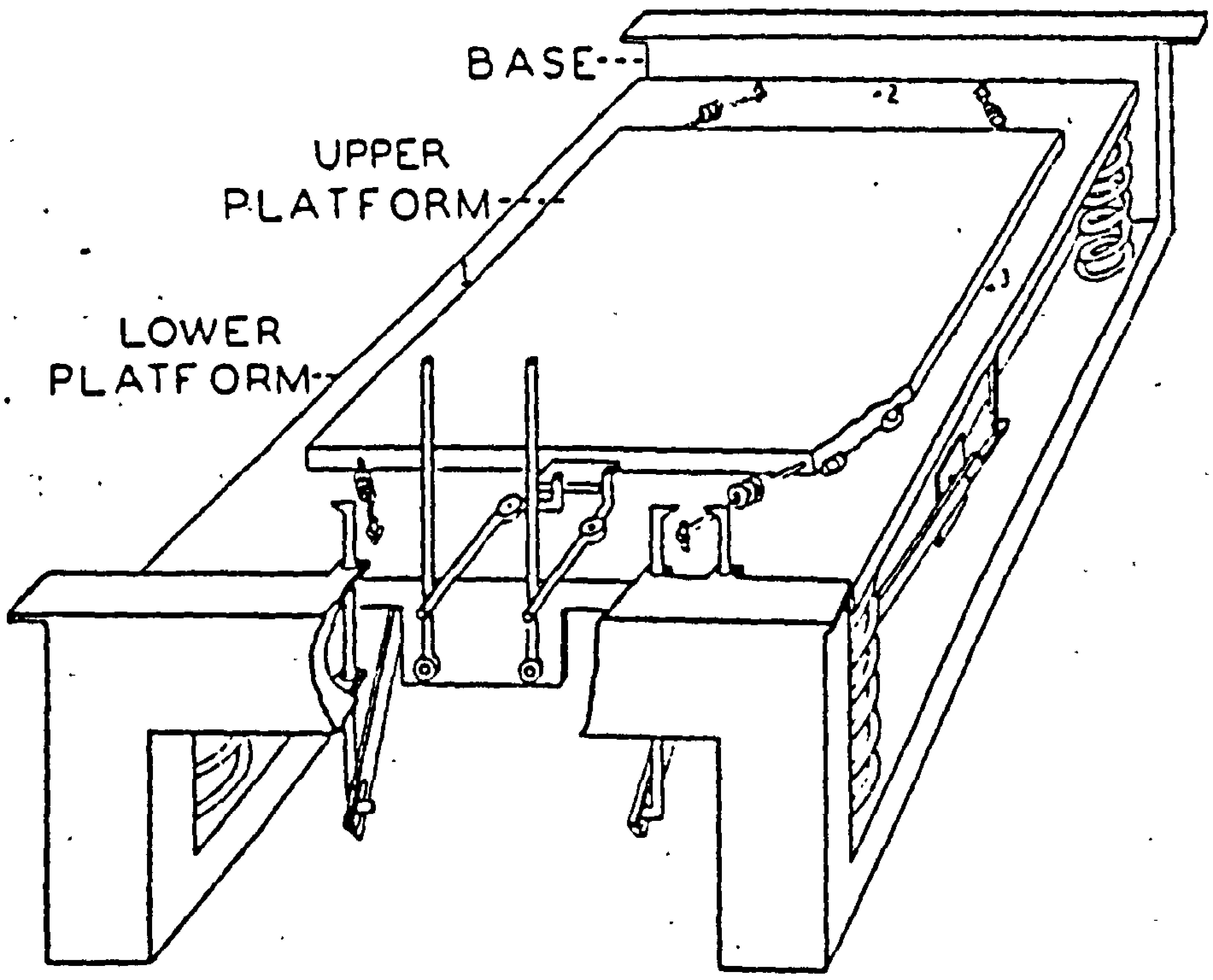


Figure 2.17 Force plate developed by Elftman. (from Elftman, 1938)

and therefore the system can be used in the laboratory and in the clinic. It was stated that the individual transducers used in this device are capable of following 1 Hz to 1 kHz compressive stress pulses, reaching peak stresses as high as 1500 kPa with an accuracy of a few percent (no figure was given). Cavanagh et al (1983) used the above "pressure sensitive shoe insole", and figure 2.16 shows the development of the pressure distribution on the foot for a complete stance phase. It is recommended that since the system has a high frequency response, and is therefore accurate, it should not be limited to the study of human locomotion only.

2.5.2 The Force Plates

The concept of the force plate was introduced by Elftman (1938) when he designed a 2D force plate consisting of a base and two plates mounted on this base (fig 2.17). Elftman (1939b) developed a mathematical method to determine the point of the force application on the force plate. No report is available concerning the accuracy of this force plate. It should be mentioned that in this force plate, the measurement of the moment exerted between the foot and the ground about the vertical axis was not considered at all.

The first comprehensive force plate was designed by Cunningham and Brown of California (1952). Two rectangular platforms are mounted on each other, the top one being an aluminium plate mounted on the lower plate by means of a tubular column at each corner. Each column has a combination of strain gauges which are connected in Wheatstone bridge circuits to produce three components of the resultant force and three components of the resultant moment about perpendicular axes through the centre of the top platform. This force plate became the basis of many force plate systems which are in use now all over the world.

Kistler (1975) marketed a force plate using piezoelectric transducers. The piezoelectric effect was discovered by the Curie brothers in 1880 and showed that if a mechanical stress is applied to the surface of certain types of

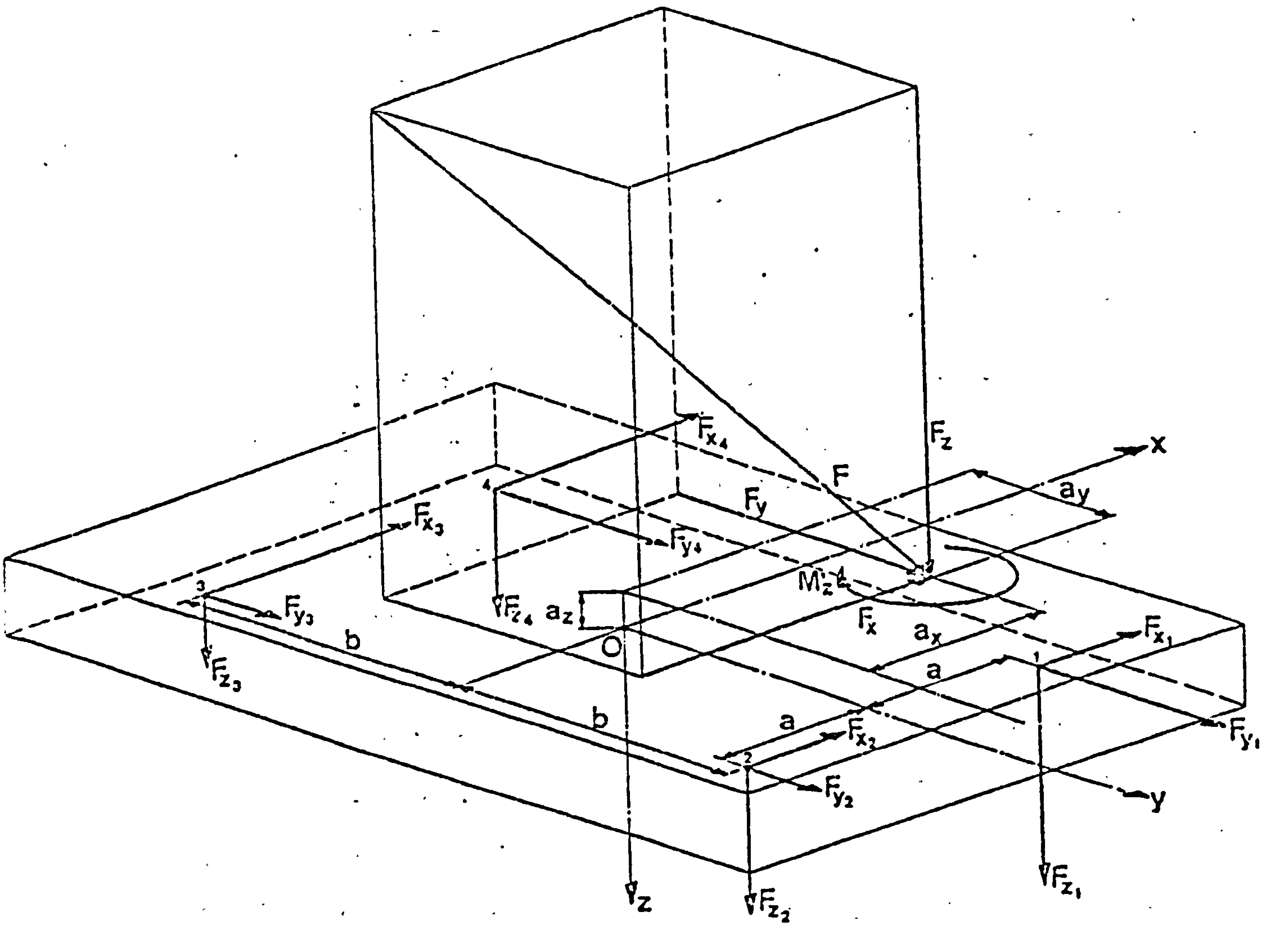


Figure 2.18 Definition of Kistler (±) coordinate system and of the measured parameters. (from Kistler 1975)

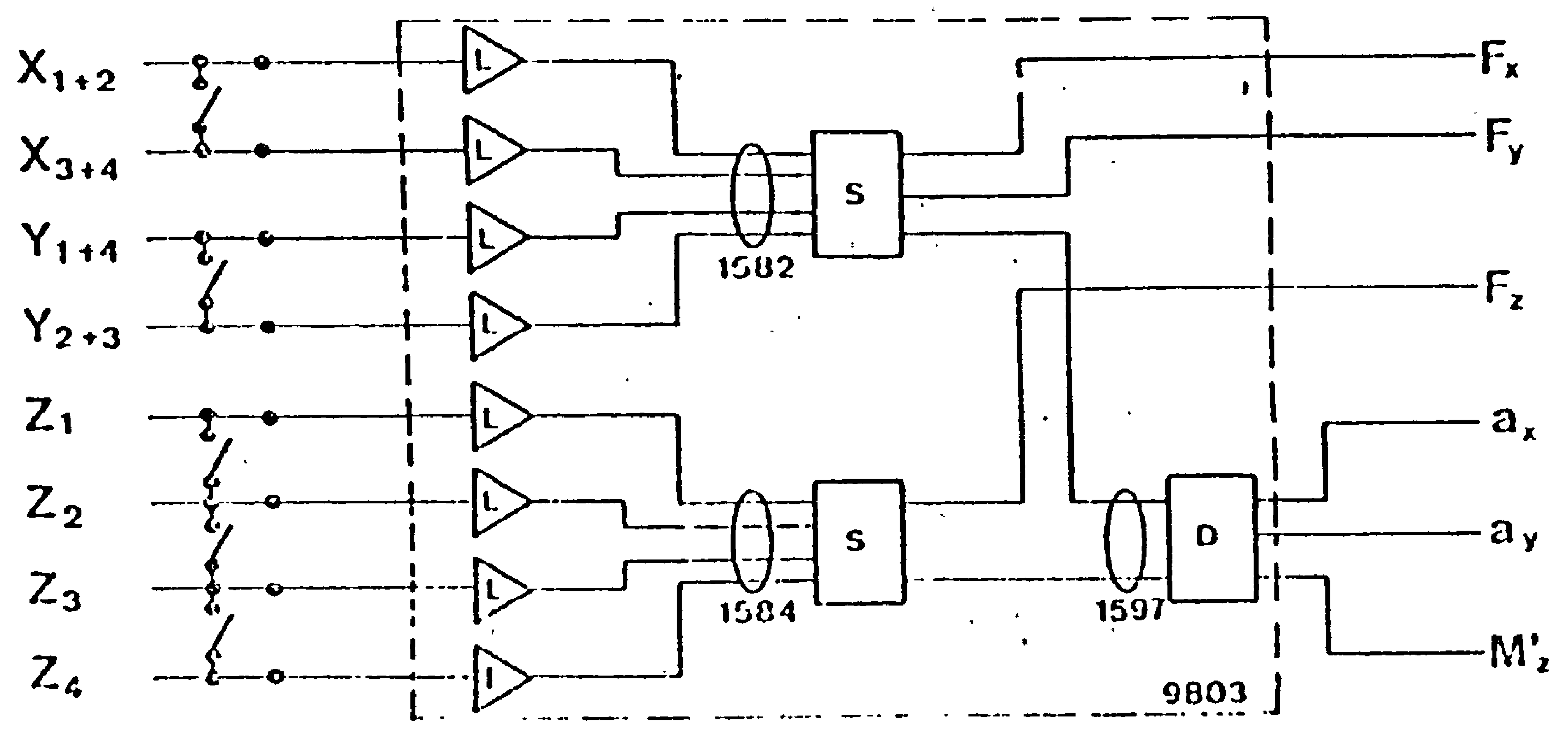


Figure 2.19 Circuit diagram for the external equipment of the force plate.
Decoding: 9803 is the electronic unit number, the inputs of this unit are the 8 outputs of the force plate, and its outputs are 3 force components, 2 coordinates of the centre of pressure and the friction moment. S and D, are the summing and dividing amplifiers respectively. L is the charge amplifier for converting the electrical charge into proportional voltage. 1582, 1584 and 1597 are special cables. (from Kistler 1975).

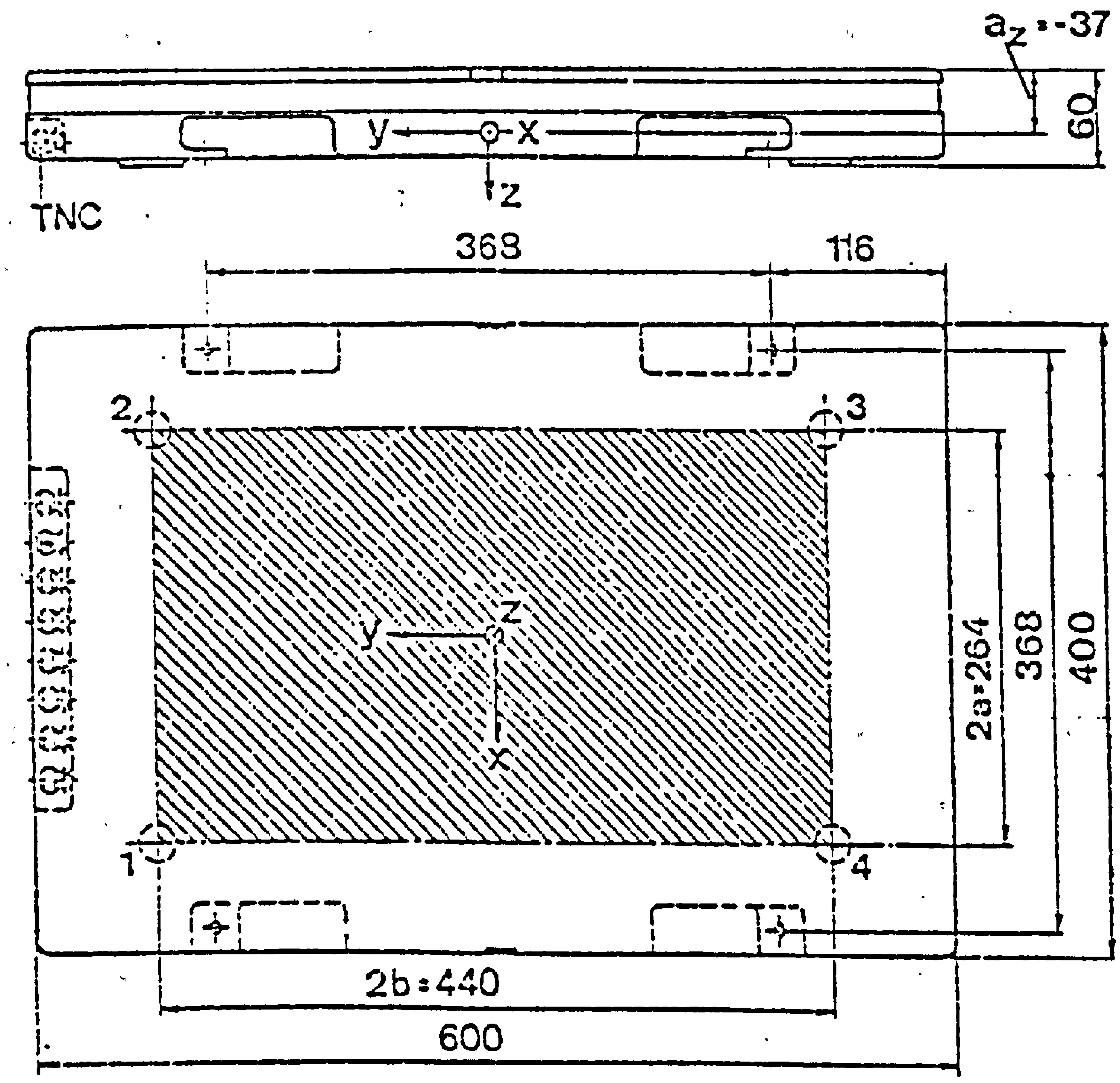


Figure 2.20 Dimensions of the measuring platform. Measuring surface is hatched: 440 x 264 mm . (from Kistler 1975)

crystals, an electrical charge will be produced on that surface. Quartz, which is a very good piezoelectric material has been used in the Kistler force plate. Two rectangular plates are mounted one above the other, the top plate can be made of aluminium or glass to match the purpose of the study. The top plate is supported on the lower plate by four multicomponent piezoelectric transducers which are fitted at the corners of the lower plate. Each can measure the forces in three directions as defined in figure 2.18. Each force plate has eight outputs for the external connections of the platform; these outputs are connected to 8 charge amplifiers for conversion of the electrical charges produced in the piezoelectric force transducers into proportional voltages, and then through two summing amplifiers which enable the components of the resulting force vector and the resulting moment vector referred to the force plate coordinate system to be calculated. Thus, signals are taken through a dividing amplifier which calculates the coordinates of the force application point and the free moment about the vertical axis. If the dividing amplifier is not installed (as in the case of Strathclyde University) the output variables will be the three components of the resultant force and the three components of the resultant moment about the centre of the force plate. These output variables can be used to calculate the two coordinates of the centre of pressure and the friction moment between the foot and the ground about a vertical axis passing through the centre of pressure. Figure 2.19 shows the electronic diagram for the external facilities of the force plate, and figure 2.20 shows the physical dimension of the force plate, the hatched area indicating the effective measuring surface. Kistler reported a sampling frequency of data acquisition of up to 20000 measuring values per second with an error less than 0.05 percent. Also a cross talk of less than ± 2% between the two horizontal forces (FX and FY), less than ± 3% to the vertical force (FZ) from FX and FY, and less than ± 1% to FX and FY from FZ were reported. Free of hysteresis, and a natural frequency of more than 200 Hz which gives the device an excellent frequency response were also reported.

Bobbert and Schamhardt (1990) examined the accuracy in the determination of the point of force application with a Kistler force plate. 117 points were checked in the range from 0 to 2000 N of force. The errors in determining the x and y, coordinates of the point of application ranged from -20 mm to +20 mm in the x and in the y directions (x is the short axis and y is the long axis of the FP). The average absolute[1] error over 117 points was 3.5 mm for the x and 6.3 mm for the y component of the centre of pressure. A correction algorithm was presented in their paper based on the measured errors, and it has been reported that the absolute errors can be reduced to 1.3 mm for x and 1.6 mm for y if the correction algorithm is used.

The Kistler force plate is an accurate device for measuring the 3D forces on the foot but it is expensive, relatively small and can be used only in the laboratory. It is difficult to disguise from the subject and misleading measurements may be obtained as the subject certainly attempts to strike the plate.

2.6 Analysis of Normal Gait

Scientists have studied normal gait for some time, and its basic pattern is fully understood. However, as it is impossible to find two persons with exactly the same pattern of walk, investigations in this field cannot be considered complete. The actual research in gait analysis was begun at the end of the nineteenth century by Braune and Fischer (1895), and it has been of interest to many scientists over the twentieth century. Now it is possible to find a new report on gait analysis every other day. For this reason, only a few samples of each aspect of gait will be reviewed here representing the various

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[1] The absolute error was calculated as:

$$\lceil \bar{x} - x \rceil \ , \ \lceil \bar{y} - y \rceil$$

in x and y coordinates respectively.

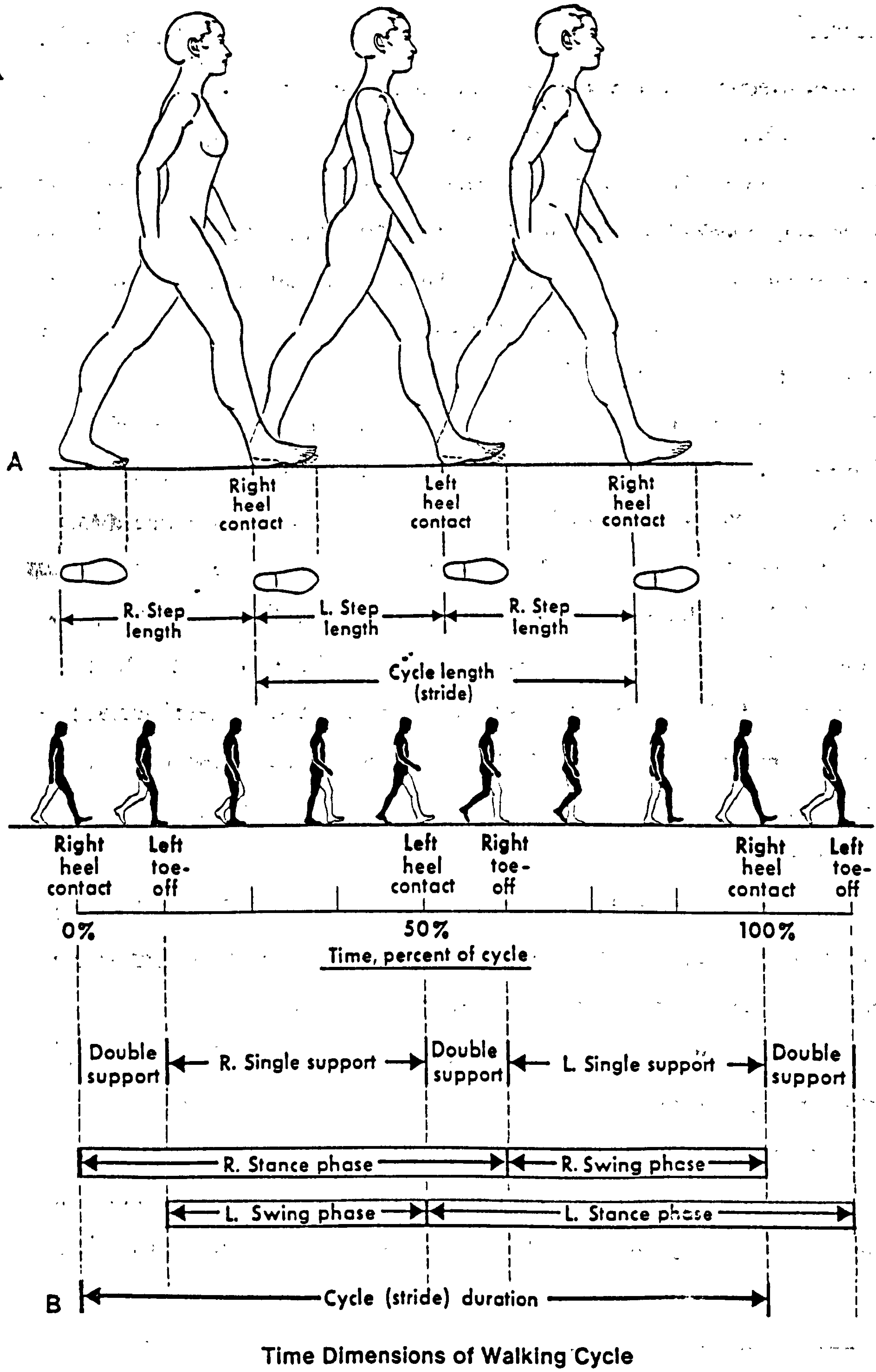


Figure 2.21. Definition of the temporal-distance parameters. (from Inman et al 1981)

patterns that have been studied. This will basically cover the temporal-distance parameters, kinetic, and kinematic of normal gait.

2.6.1 The Temporal-Distance Parameters

Temporal-distance parameters determine the time and the distance for certain events that happen between the foot and the floor during walking. Human motion has been defined as a number of events which occur in a repetitive pattern. The period between two successive heel contacts of the same foot with the ground is called the gait cycle and the distance achieved during one gait cycle is call the stride length. Each gait cycle has two phases; stance phase when the foot is in contact with the ground, accounting for about 60% of the cycle for the normal person at normal speed of walking, and swing phase when the foot is swinging in the air prior to being placed on the ground, accounting for about 40% of the cycle for the normal person at normal speed of walking. The period when the two feet are on the ground is defined as the double support period and this period increases as the speed of walking decreases. Figure 2.21 presents the temporal-distance parameters as defined by Inman et al (1981).

Grieve (1968), Lamoreux (1971), Andriacchi et al (1977), Inman et al (1981) and Shiavi et al (1988) have studied the temporal distance parameters and their relationship with speed of walking.

Grieve (1968) found that the relationship between stride frequency and speed of walking for adults is:

$$f = a V^{b}$$

where f is the number of strides per minute, V is the speed expressed in the number of statures per second and a and b are constants varying individually from person to person.

Chao et al (1983) studied the temporal-distance parameters for 110 normal adults during level walking. It was stated that the step width and length were measured with an instrument of ± 1 cm resolution. The subjects were put into

Table 2.2 Temporal-distance parameters for 110 normal subjects during level walking. Group I (ages 32-85) and group II (ages 19-32). (from Chao et al 1983)

Parameter	Men ($n = 53$) I ($n = 32$)	Men ($n = 53$) II ($n = 21$)	Men ($n = 53$) Total	Women ($n = 57$) I ($n = 37$)	Women ($n = 57$) II ($n = 20$)	Women ($n = 57$) Total
Cadence (stride min^{-1})	52 ± 5	50 ± 8	51 ± 6	56 ± 5	51 ± 5	54 ± 5
Stride Length/LEL*	1.56 ± 0.15	1.48 ± 0.17	1.53 ± 0.16	1.40 ± 0.14	1.39 ± 0.17	1.40 ± 0.15
Speed ($m\ min^{-1}$)*	76.1 ± 12.5	71.9 ± 18.3	74.4 ± 15.1	69.4 ± 11.0	63.9 ± 11.1	67.5 ± 11.2
Step length, right (cm)*	73 ± 8	69 ± 9	71 ± 8	61 ± 8	60 ± 7	61 ± 7
Step length ratio (small/large)	†	1.0 ± 0	†	†	1.0 ± 0	†
Step length/LEL, right	0.78 ± 0.07	0.73 ± 0.08	0.76 ± 0.08	0.70 ± 0.07	0.69 ± 0.08	0.70 ± 0.07
Right stance (% G.C.)	59 ± 2	60 ± 2	59 ± 2	60 ± 2	59 ± 2	59 ± 2
Left stance (% G.C.)	†	60 ± 3	†	†	59 ± 3	†
Single stance, right (% G.C.)	41 ± 2	40 ± 2	41 ± 2	40 ± 2	41 ± 2	41 ± 2
Double stance, right (% G.C.)	8.8 ± 1.9	10.2 ± 2.6	9.4 ± 2.3	10.0 ± 5.5	8.9 ± 2.0	9.6 ± 4.6
Double stance ratio (small/large)	†	0.8 ± 0.1	†	†	0.9 ± 0.1	†

% G.C. = Percent of Gait Cycle

*Significantly corrected with 10 or more parameters ($p < 0.01$).

†The stride characteristics were assumed symmetric in this group.

Table 2.3 Temporal-distance parameters for normal adults. CV is the coefficient of variation defined as the ratio of SD to the mean value. (from Kadaba et al 1989)

Parameter	Mean ± SD gait parameters	Mean ± SD intrasubject CV% within day	Mean ± SD intrasubject CV% between days
Cadence (steps/min)	111.6 ± 8.3	1.9 ± 0.8	3.4 ± 1.8
Velocity (m/s)	1.306 ± 0.170	2.9 ± 3.3	6.1 ± 7.1
Swing to stance ratio	0.64 ± 0.05	3.8 ± 1.4	6.1 ± 2.7
Stride length (m)	1.361 ± 0.12	1.7 ± 0.6	3.0 ± 1.3

two groups, according to their age, group I had the older subjects (age 32 - 85) and group II the younger subjects (age 19 - 32). Each group had males and females and thus the study also took into account the sex. In order to make the data comparable, the stride length and the step length were normalised to the lower extremity length (LEL). Table 2.2 shows a table with the values of the temporal distance parameters for the above two groups with sex classification. It was found that the cadence (stride/min) is the only temporal-distance parameter that has significant difference between the two groups and was present only among the females. However, stride length/LEL, step length and cadence were found to be different between males and females. Regarding the relationship of the distance parameters with age, the above results are different from those of Murray et al (1964). Murray et al studied 60 adults (aged 20 to 65 years) in five groups of age, each group having three different groups of height. It was found that step and stride lengths have significant differences between the oldest and the youngest groups, and the tall subject took longer steps and strides than the short subject. This indicates that the age groups of Chao et al should be broken down to groups with smaller ranges.

Kadaba et al (1989) studied the temporal-distance parameters of 40 adults. The repeatability of the data was checked for each subject by testing within a day and on three different days. Table 2.3 shows the mean of the temporal distance parameters and the coefficient of variation (CV) for tests within a day and between days. It was found that the repeatability of the temporal parameters acquired during tests within a day is higher than that obtained in various days.

2.6.2 Kinematics and Kinetics of Normal Gait

Most researchers study the kinematic and the kinetic parameters simultaneously. This is for several reasons such as: firstly, a comprehensive analysis of the gait should consider the kinetic and the kinematic data, because any change in the kinetic data will be reflected in the kinematic data and vice

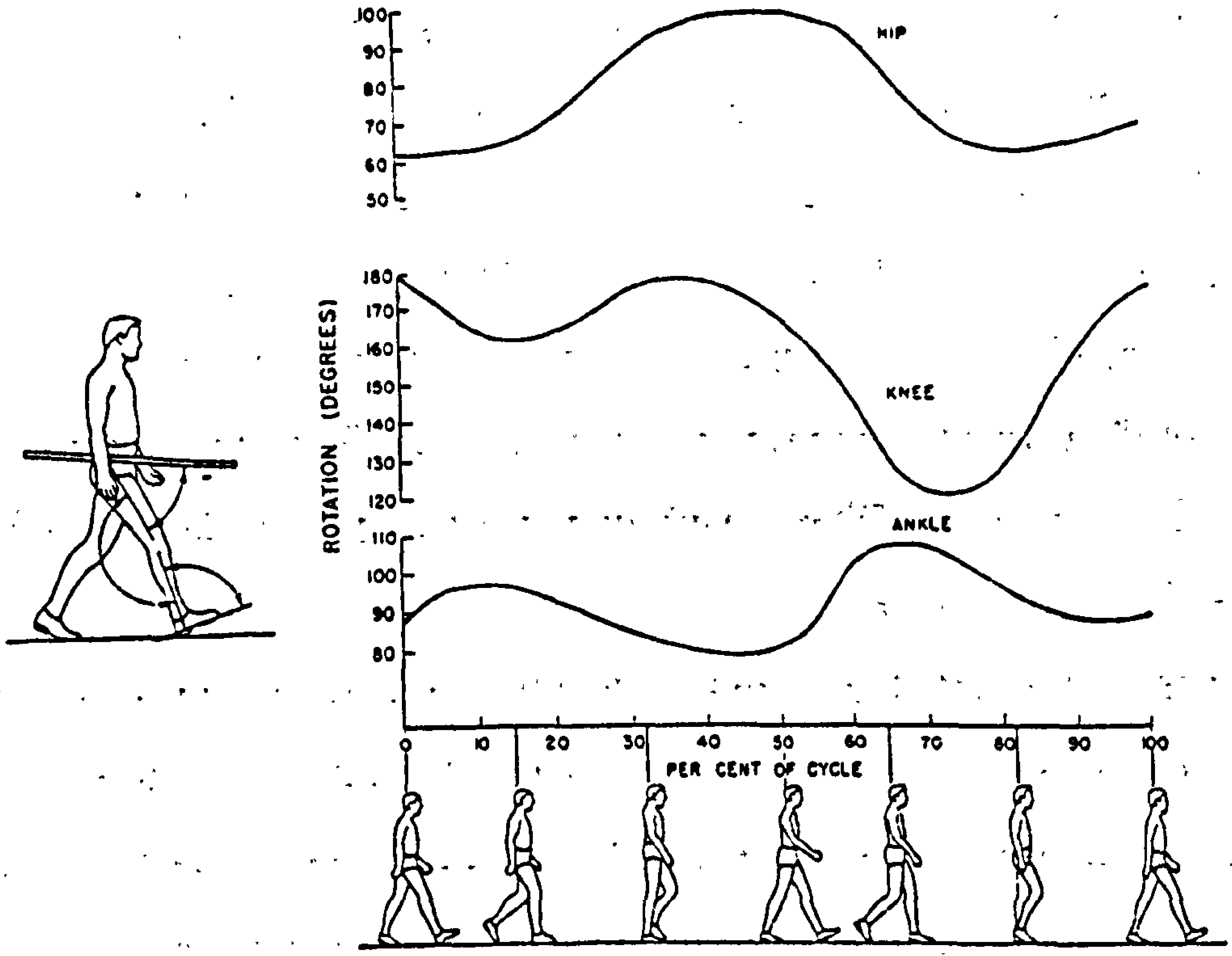


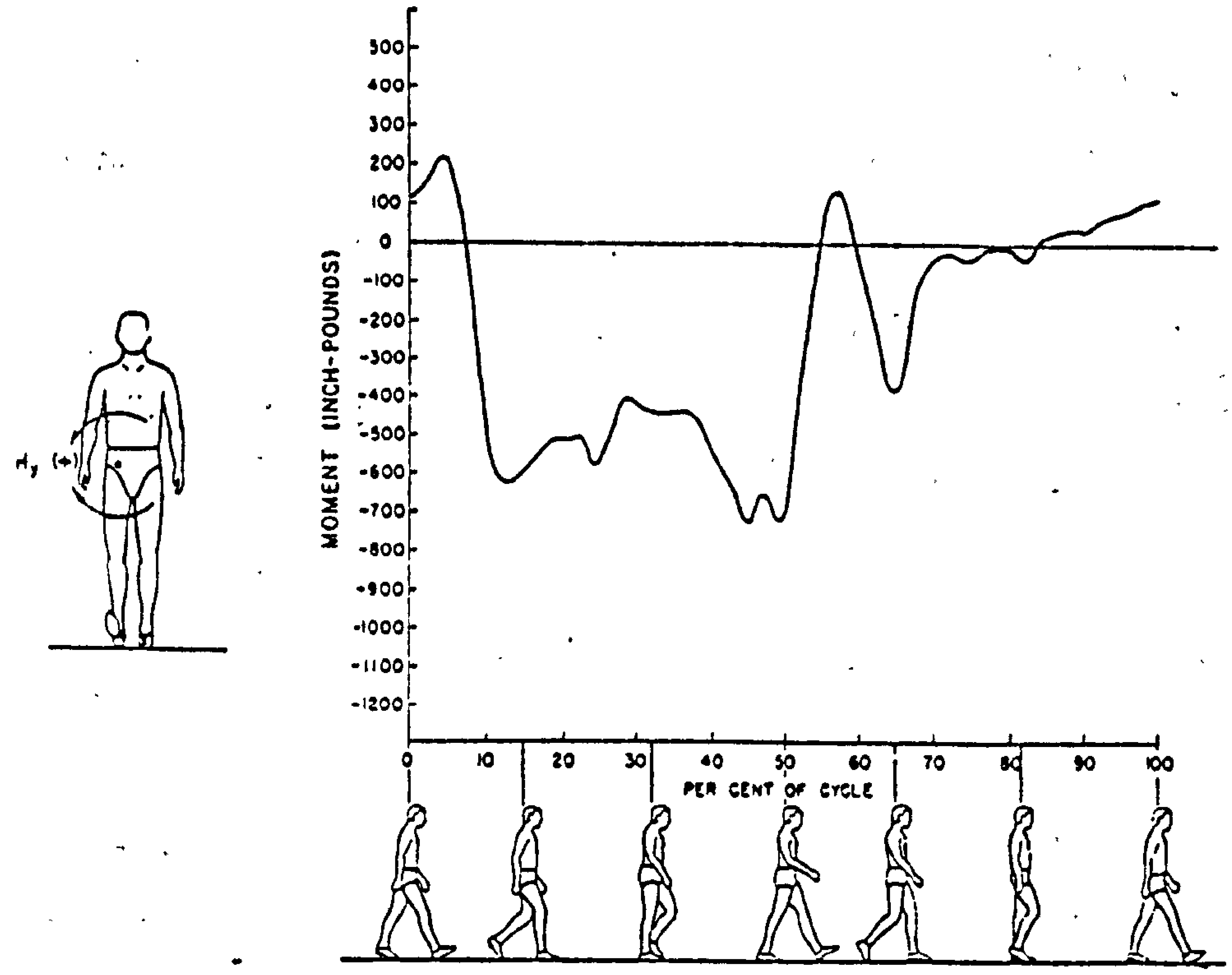
Figure 2.22 Typical joint rotation in the plane of progression for 8 normal subjects. (from Eberhart et al, 1954)

versa. Secondly, the intersegmental moments are obtained from the kinematic data and the forces acting on the body segments. Finally, it is more convenient to study the kinematic and the kinetic parameters together, because this will save considerable time during data acquisition. For the above reasons, the author decided to present the review of the kinematics and the kinetics together under the same subtitle. This will enable the reader to relate them to each other even if they belong to more than one researcher.

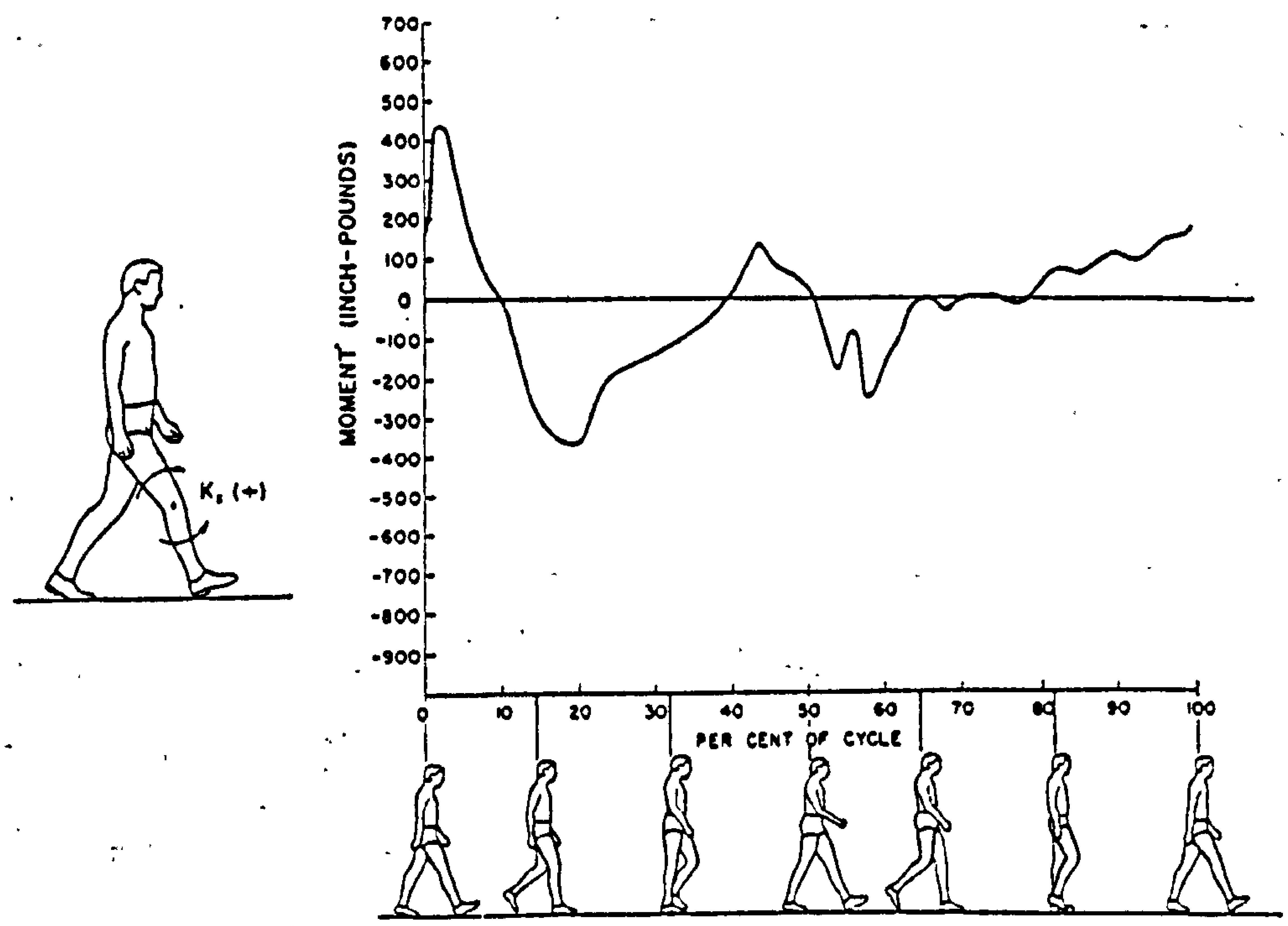
Elftman (1939b) measured the variations in the force exerted by the ground on the body during the stance phase. The vertical force was found to have the two typical peaks and was divided into two parts. The first part being the gravity reaction which is equal to the body weight. The second part being the effective force resulting from the effect of the vertical acceleration on the centre of gravity of the body.

Bresler and Frankel (1950) studied the three dimensional components of the forces and the moments at the lower extremity joints for four normal subjects. The knee sagittal moment was found to have the typical " double-locking " action (the knee locked under flexing moment) which allows the subject to move forward with minimum vertical travel of the centre of gravity, resulting in minimum energy expenditure of body. The hip and ankle moments were also presented and discussed.

Eberhart et al (1954) at the University of California presented a complete set of kinetic and kinematic data for 8 normal subjects. The interrupted light technique was used with two 35 mm cameras to obtain the location of the targets. It is stated that the location of the targets was obtained with a maximum error of 0.5 inch (1.23 cm), and angles were measured to an accuracy of 1 deg. Figure 2.22 shows the rotation of the leg joints at the anterior-posterior plane. During the stance phase, the knee has shown the " double-locking " action (they are during the knee flexion periods after heel strike and prior to swing phase) which gives the gait its smooth movement.

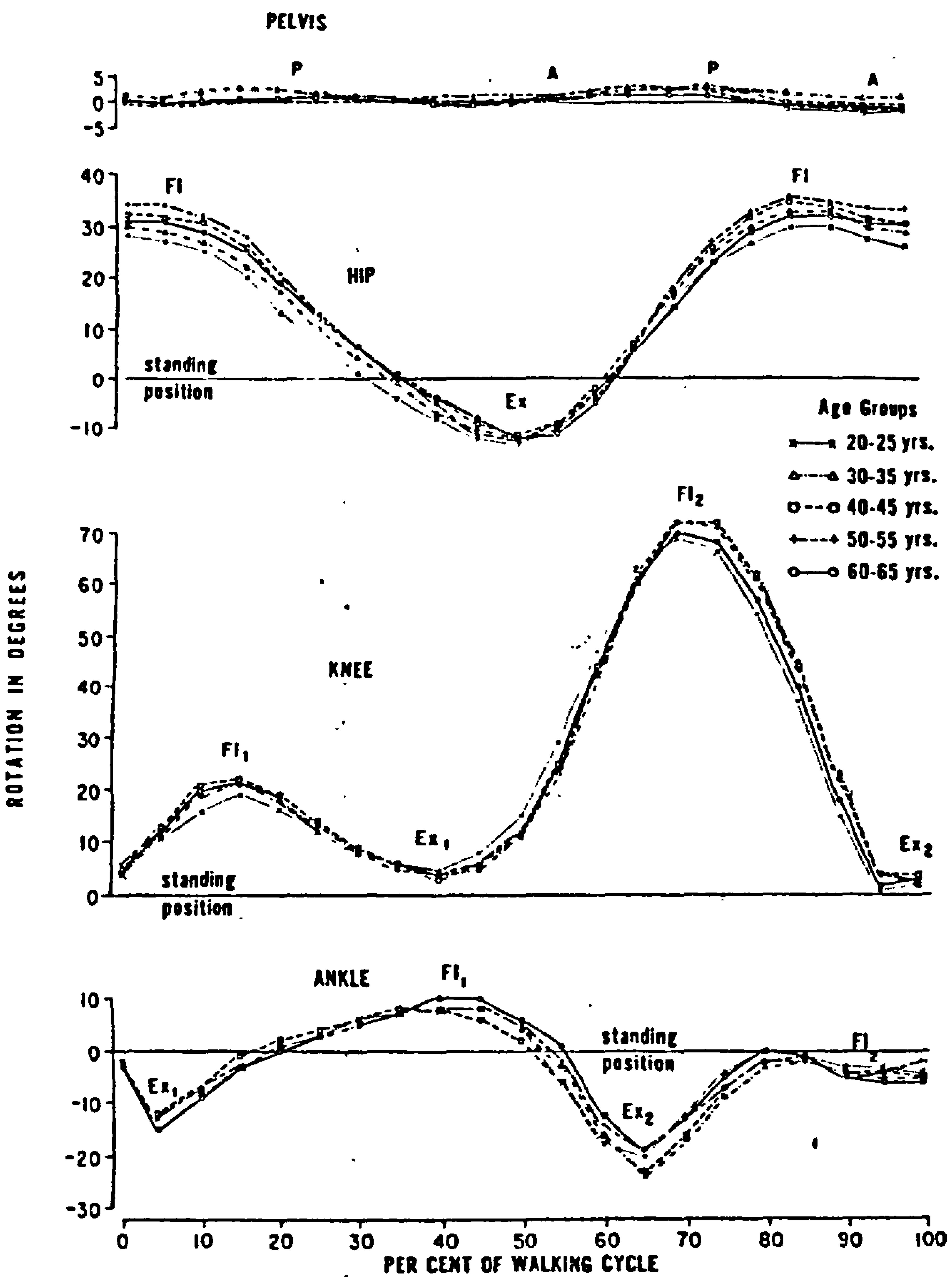


Lateral hip moment. Force studies, typical subject.

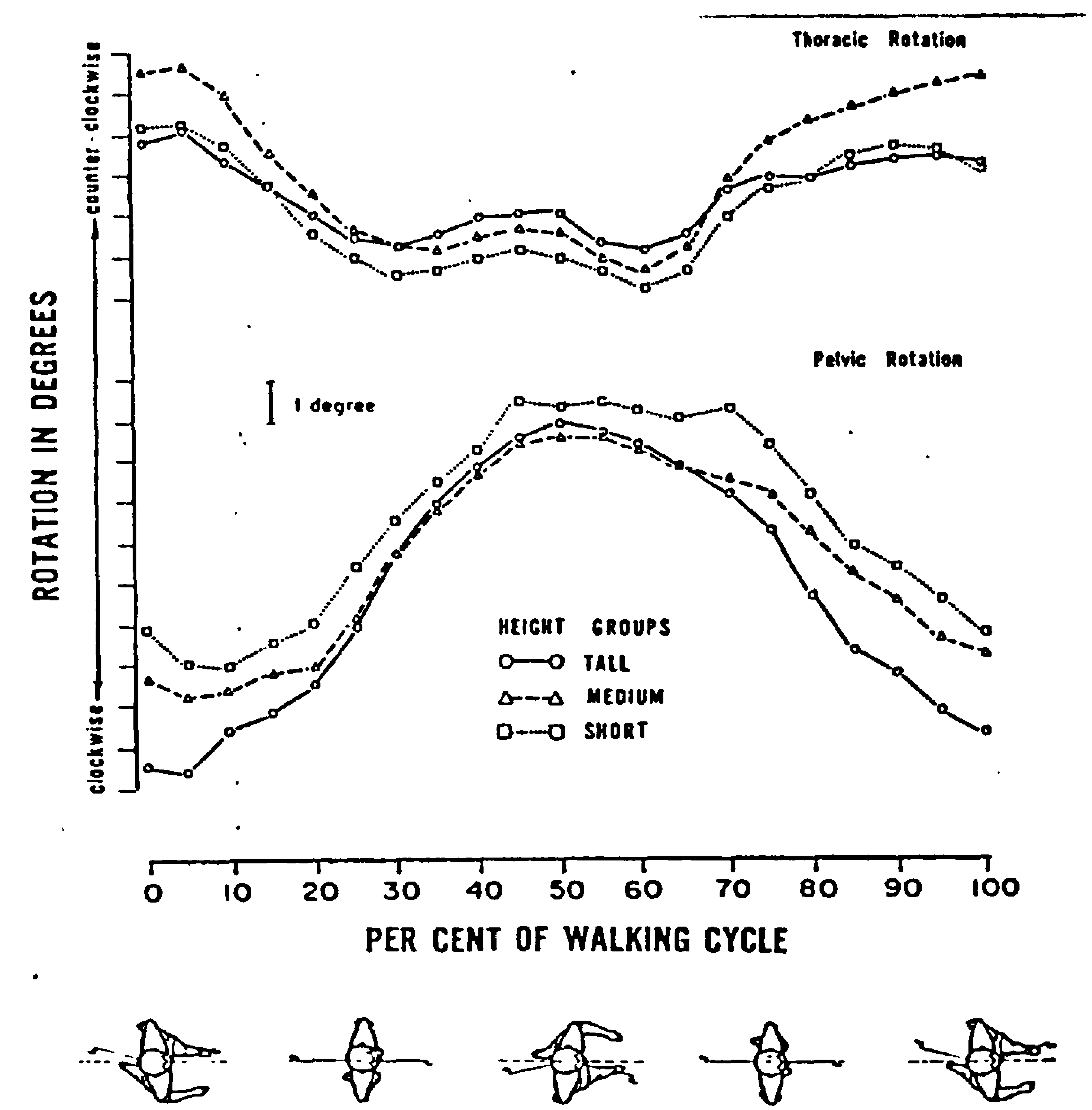


Fore-and-aft knee moment. Force studies, typical subject.

Figure 2.23 Hip lateral and knee sagittal moments for a normal subject. (from Eberhart et al, 1954)



Mean patterns of sagittal rotation for the five age groups, twelve men in each group, two trials for each man.



Mean patterns of transverse rotation of the thorax and pelvis for the three height groups, twenty men in each group, two trials for each man.

Figure 2.24 Kinematics of the leg (left graph) and thorax and pelvis (right graph) during level walking. (from Murray et al 1964).

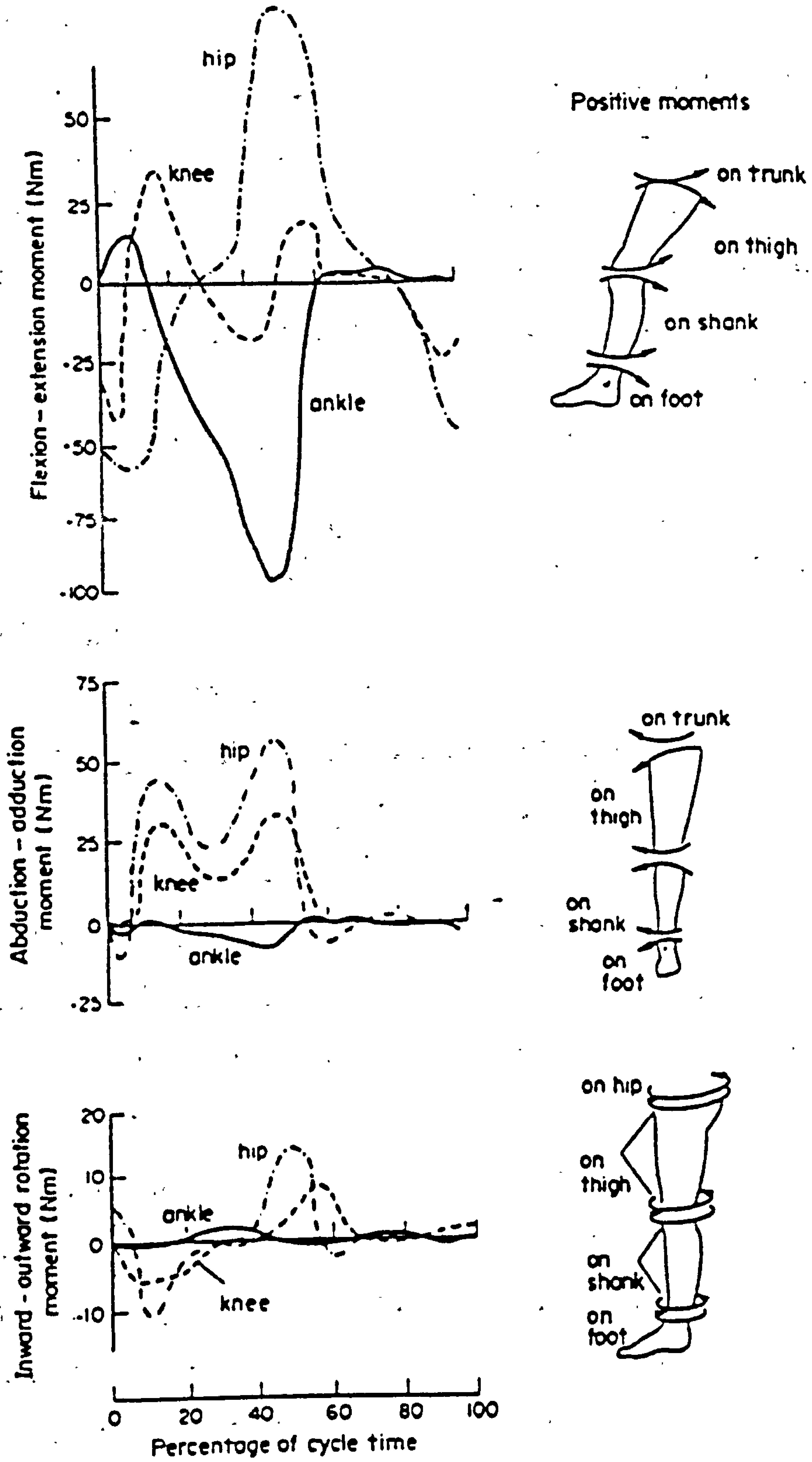


Figure 2.25 Variation with time of the moments transmitted between the segments of the leg during the walking cycle. (from Paul, 1986)

Figure 2.23 shows the resulting knee moment in the sagittal plane, and the hip moment in the medio-lateral plane, the hip medio-lateral moment is achieving body stability during the shifting of the body mass from one leg to the other. In this study, pins were inserted into the subject's bone to provide rigid fixation for the targets; this, however could be painful and may alter the gait and give a misleading result.

Murray et al (1964) carried out extensive kinematic measurements on 60 normal adults. The sagittal rotation of the pelvis, hip, knee, and ankle were measured, together with the trunk vertical, lateral, and forward displacements during level walking. The transverse rotation of the pelvis and thorax during free cadence was also measured. All the results were classified in age and in height groups. Figure 2.24 shows the sagittal rotations of the pelvis and the lower leg and the transverse rotation for the pelvis and the thorax. No significant differences were found between either the age groups or the height groups. The sagittal rotation patterns were found to be in agreement with work of previous researchers. The pelvis and thorax rotations were found to be out of phase; this is reducing the trunk rotation and producing smoothness in the gait.

Paul (1967) used cine cameras to record displacements of the lower limb, and by means of a force plate, measured the reaction force which is acting on the foot. From these data the joint moments were calculated for the hip, knee, and ankle. Fourteen normal subjects were tested, three of them females. Paul (1971 and 1986) showed the joint moments results and compared them with data taken from Bresler and Frankel. The results were comparable in pattern but some of Bresler's data had a higher magnitude, this being related to the different speed and body weight of the subjects. Paul (1986) presented the average moments transmitted between the segments of the lower limb in three dimentioms as seen in figure 2.25.

Lamoreux (1971) recorded three dimensional kinematic data of the

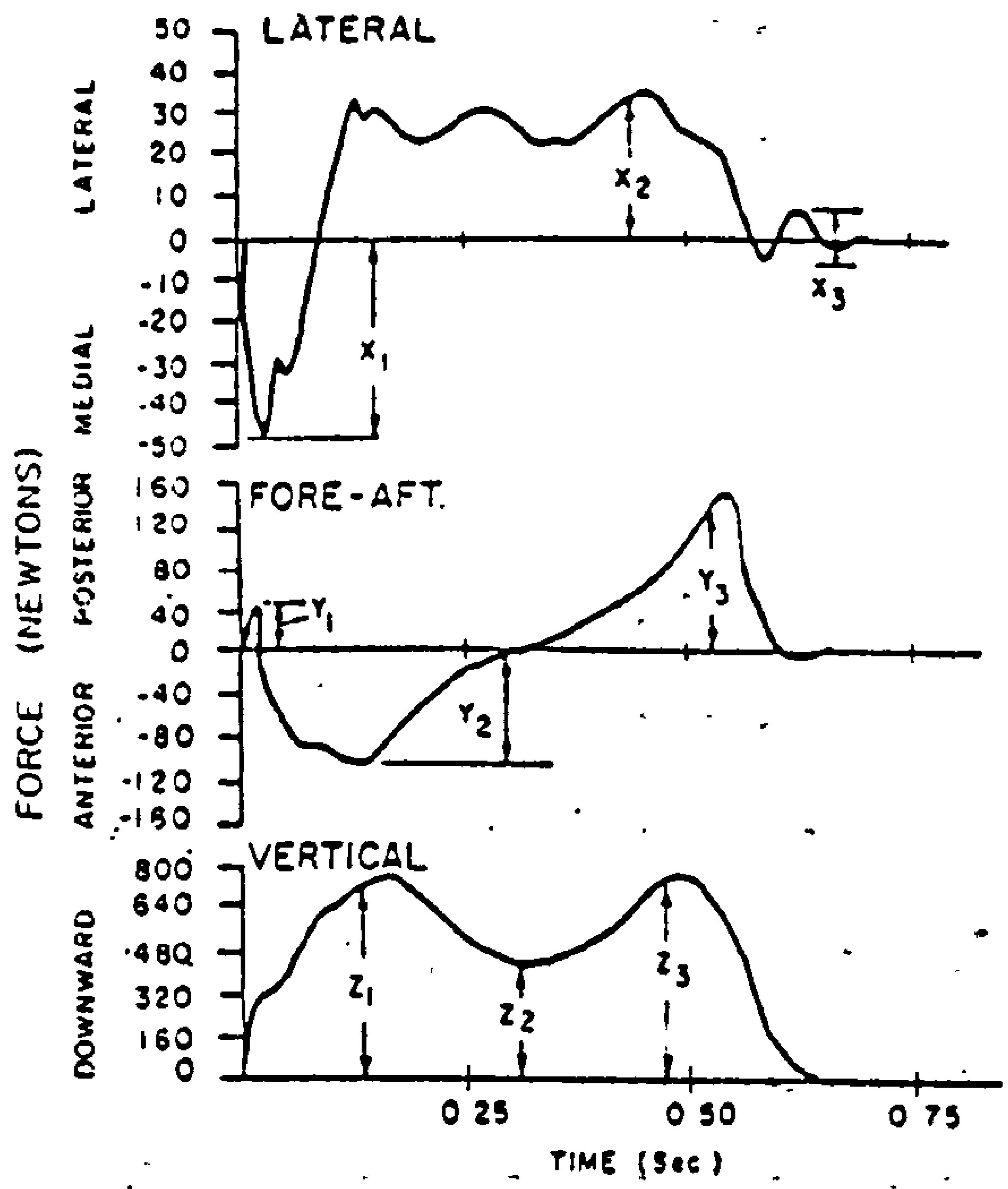
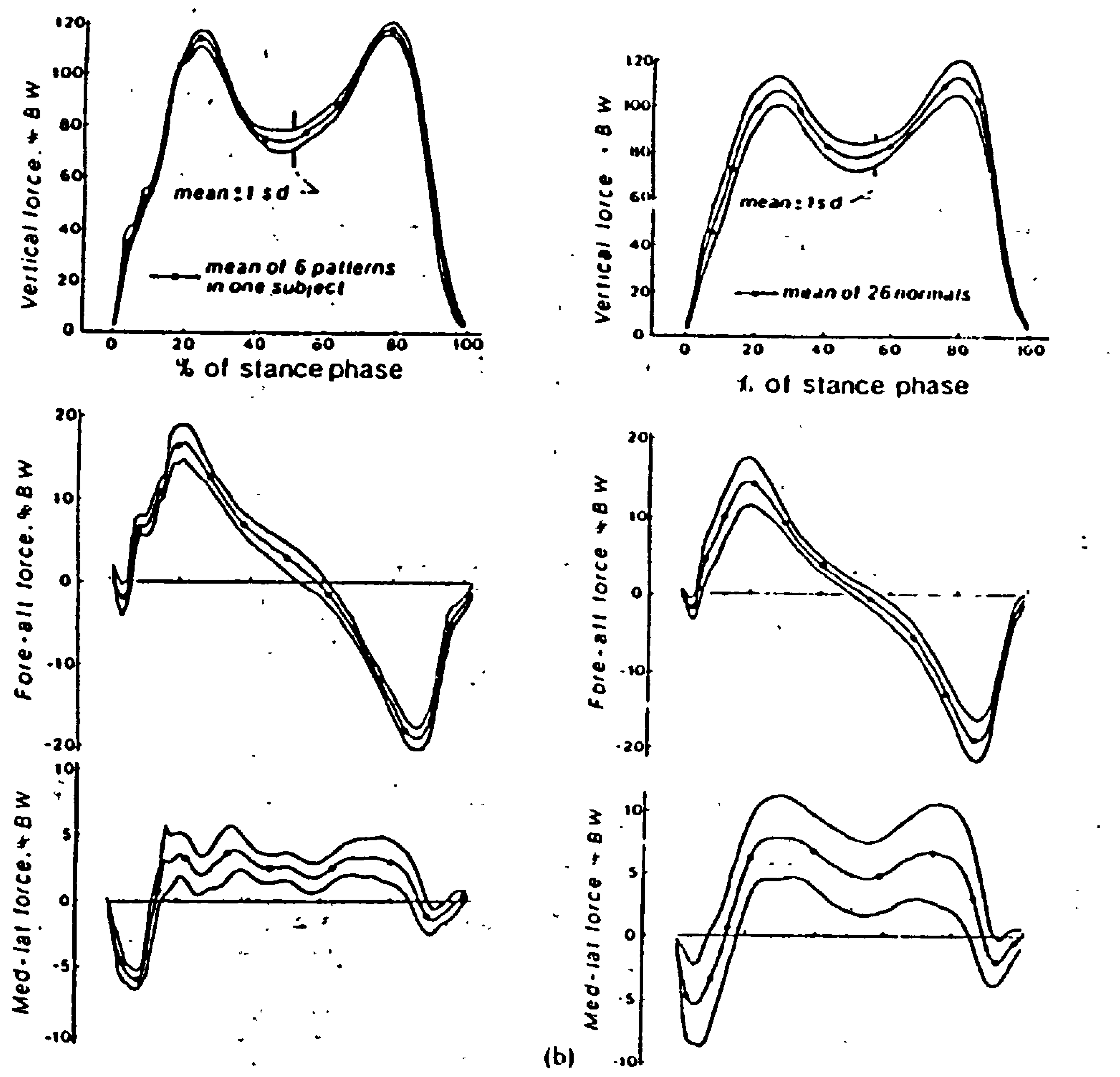


Figure 2.26 Patterns of the ground reaction force acting on the foot, (force plate data). (from Andriacchi et al, 1977).

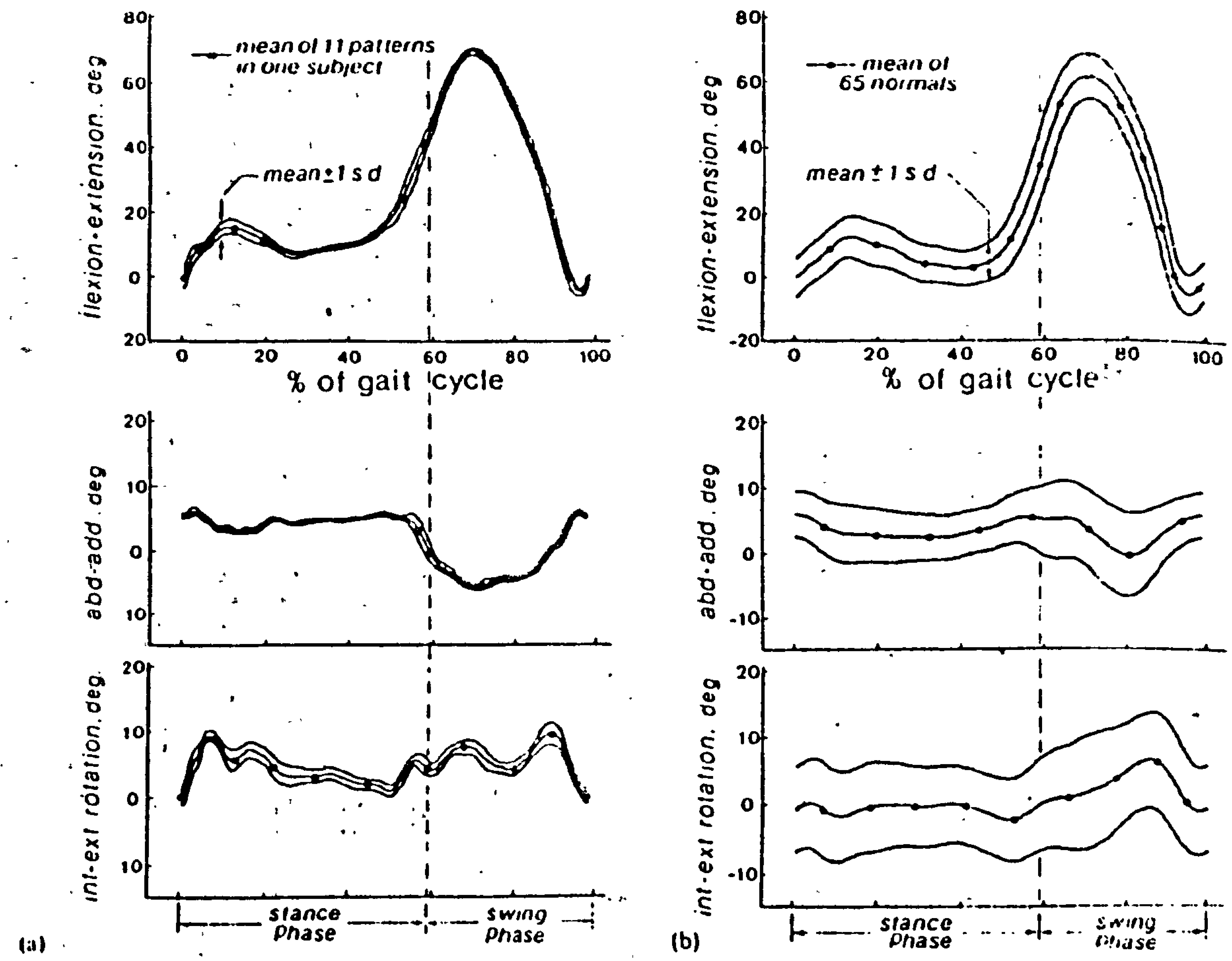
pelvis and lower joints of the right leg of a single subject at six different speeds of walking. It was found that the magnitude of the kinematic variables was increased when the walking speed increased. It was estimated that the measurement of the knee relative rotation is obtained with precision of 0.5 degrees, and peak flexion of about 13 and 65 degrees were obtained for the knee during stance and swing phase respectively.

Andriacchi et al (1977) studied the variations of the ground reaction force components which are acting on the foot with changes in walking speed. A piezoelectric force plate was used to measure the forces at a rate of 200 Hz. Seventeen normal adults were tested and the waveforms of the ground reaction forces were characterised into 9 amplitudes (x1,x2,x3,y1,y2,y3,z1,z2,z3). Figure 2.26 shows a typical pattern of the results obtained together with the 9 amplitudes. Linear relationships were found between the 9 amplitudes and the speed of walking. The accuracy of the measuring device was not discussed.

Chao et al (1983) studied the kinematics of the knee and the ground reaction force variations with age and sex of normal adults. 110 subjects aged between 19 and 85 years were classified into 3 groups of age and tested according to their sex and age. The data were characterised to 41 parameters which were then subjected to linear correlation matrix analysis. Only ten of these parameters, (none of them related to the knee joint motion), were found to have a significant correlation with ten or more of the remaining parameters. No significant difference was found between the two sexes or between the age groups. Figure 2.27 shows results of the ground reaction forces and knee rotation patterns. Fourier analysis was used to remove any data distortion during the averaging process. Although the data presented in this work seem to be comparable with those of other researchers, the goniometer, which was used for joint angle measurement, is not the best method for motion measurements because of the alignment errors, and the accuracy of the devices was not given. The age effect on the data should be studied in smaller groups,



Normal ground reaction force patterns. (a) The mean and 1 standard deviation envelope of the 'typical' pattern of one normal subject based on 6 individual patterns. (b) The mean and 1 standard deviation envelope of the 'general' pattern of 26 normal subjects



Knee joint rotation patterns. (a) The mean and 1 standard deviation envelope of the 'typical' pattern of one normal subject based on 11 individual patterns. (b) The mean and 1 standard deviation envelope of the 'general' pattern in 65 normals

Figure 2.27 Ground reaction forces (left) and knee joint rotation (right) patterns. (from Chao et al, 1983).

and whole body analysis should be considered.

Many researchers have used Fourier analysis techniques in gait analysis. Fourier analysis is a mathematical procedure which describes periodic signals in terms of harmonic coefficients. It is an efficient method to use for data storage and for averaging the gait parameters over several cycles. The essential number of harmonics depends on the sampling rate of the raw data and on the accuracy required in the reconstructed signal. Schneider and Chao (1983) used 26 normal subjects and 10 with knee diseases to justify the ideal definition for ground reaction force periodicity and to determine the essential number of Fourier coefficients required to study the significance of normal and abnormal ground reaction force patterns. By using a simple method depending on the achievement of a certain accuracy in the reconstructed signal of the force reaction patterns, Schneider and Chao found that the stance phase is the best period to use the application of Fourier analysis. However, only the first two to four harmonics and the constant term were found to be essential coefficients in describing each pattern of the ground reaction components. This was found for data collected at a rate of 100 Hz; thus the application of the above finding is limited to data collected at a frequency of 100 Hz only.

Cappozzo (1984) studied the symmetry and maintenance of balance in human movement. The data were presented with respect to the anatomical coordinates as "Lissajous's figures" and not in the form of time displacement. He suggested that this form of presentation can identify symmetry with simplicity, and the effectiveness of maintaining the balance of the body during walking. This method can be effective, but all gait parameters which were presented by Cappozzo using the above method can be clearly identified using the three dimensional time-displacement form of data presentation.

Winter (1980, 1983 and 1984) investigated the variation of the kinetic and the kinematic patterns in normal human gait at various cadences. He introduced a coefficient of variation (CV) which reflected the average standard

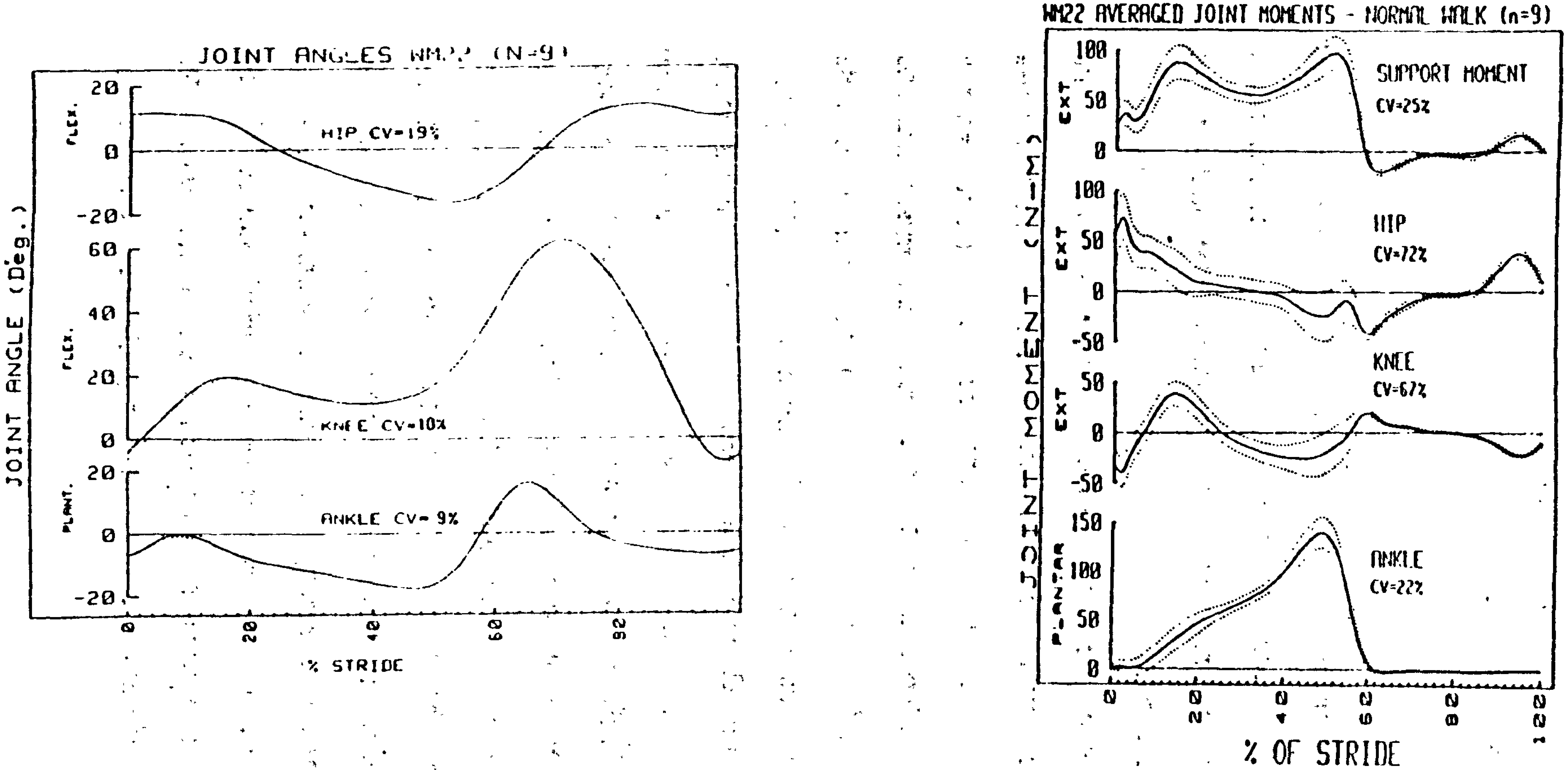


Figure 2.28 Kinematics and moments of the lower limb for a normal subject. Data were averaged from nine trials obtained over three days at level walking. (from Winter, 1984)

deviation over the stride period as a percentage of the mean curve and was calculated as:

$$CV = \frac{\sqrt{\frac{1}{N}\sum_{i=1}^{N} \sigma_i^2}}{\frac{1}{N}\sum_{i=1}^{N} | M_i |}$$

Where: N is the number of intervals over the stride, M_i is the amplitude of the normalised moment of force (Nm/kg) at the ith interval and σ_i is the standard deviation of M_i at the ith interval.

Winter also introduced the support moment at the joint which is the algebraic sum of the three joint moments (hip, knee, and ankle) and represents the total extensor/flexor pattern of the lower limb. Figures 2.28 and 29 show the vertical and horizontal forces at the foot, and also show the joint angles and moments in the sagittal plane resulting from nine repeated trials of the same normal subject at level walking. It was found that the shapes of the moment patterns were the same for all the speeds, but the magnitudes were higher for faster speeds. The hip and knee moments show less variability as the speed increases and the ankle moment shows low and constant variability with cadence changes. The support moment has low variations at all cadences. Although Winter was making use of the coefficient of variation and the support moment, the coefficient of variation does not offer information additional to the standard deviation. In fact, it may limit the usefulness of the standard deviation which varied along the gait cycle and cannot be presented by one value. The concept of the support moment is rather strange and not beneficial. It is true that the leg's joints compensate for each other, in fact it is always possible for the right and left legs to compensate for each other. However, in

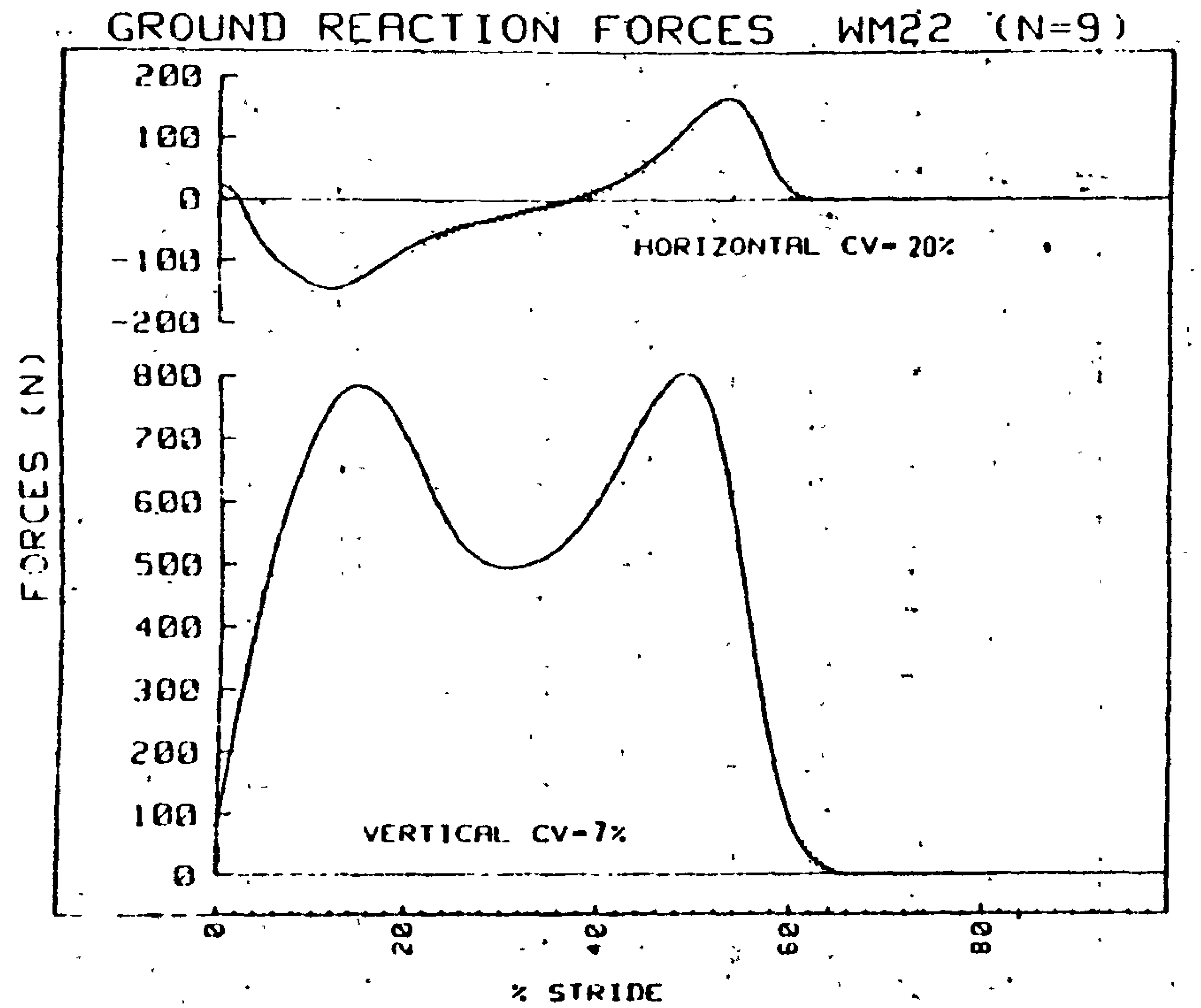


Figure 2.29 Ground reaction forces acting on the foot of a normal subject. Data were averaged from nine trials obtained over three days at level walking. (from Winter 1984).

the design of a mechanical structure each joint is considered separately from the others and forces and moments of its connecting bars only, affect its design. Only one camera was used, thus measurements were 2D only. The camera was calibrated by having reference markers in the background and the parallax error was corrected. The accuracy in obtaining the body markers' coordinates was not given but it was reported that the noise in these coordinates was mainly due to the digitising process, and was calculated to have an error of 2 mm.

Andriacchi and Strickland (1985) studied the moment variations at the joints of 29 normal subjects over a range of walking speeds. The hip flexion/extension moment was found to have the same pattern for males and females and there was no change in the pattern with changes in walking speed. The knee and ankle sagittal moments however were found to have more than one pattern over the range of speeds. No explanation was offered for these. The maximum amplitudes of the flexion/extension moments at the hip, knee, and ankle joints were influenced by the speed of walking. The magnitude of these moments varies between males and females, but this variation disappeared after normalising the data to a percentage of the body weight times height.

Marmar and Solomonidis (1989) at Strathclyde University, used three TV cameras and two Kistler force plates, running simultaneously and synchronously at a rate of 50 Hz, to acquire kinetic and kinematic data of both sides of the body for normal and pathological gait. In order to optimise the alignment of above-knee prostheses, extensive experiments were carried out to find the effect of alignment changes on the amputee's gait. To compare the amputees data with normal walkers, ten normal subjects were tested and the forces and moments acting at all the joints acquired. The data of the normal walkers were found to be comparable with previous work, Paul (1986) and Winter (1984). The results will be presented later in the results chapter of this thesis with more details and discussion.

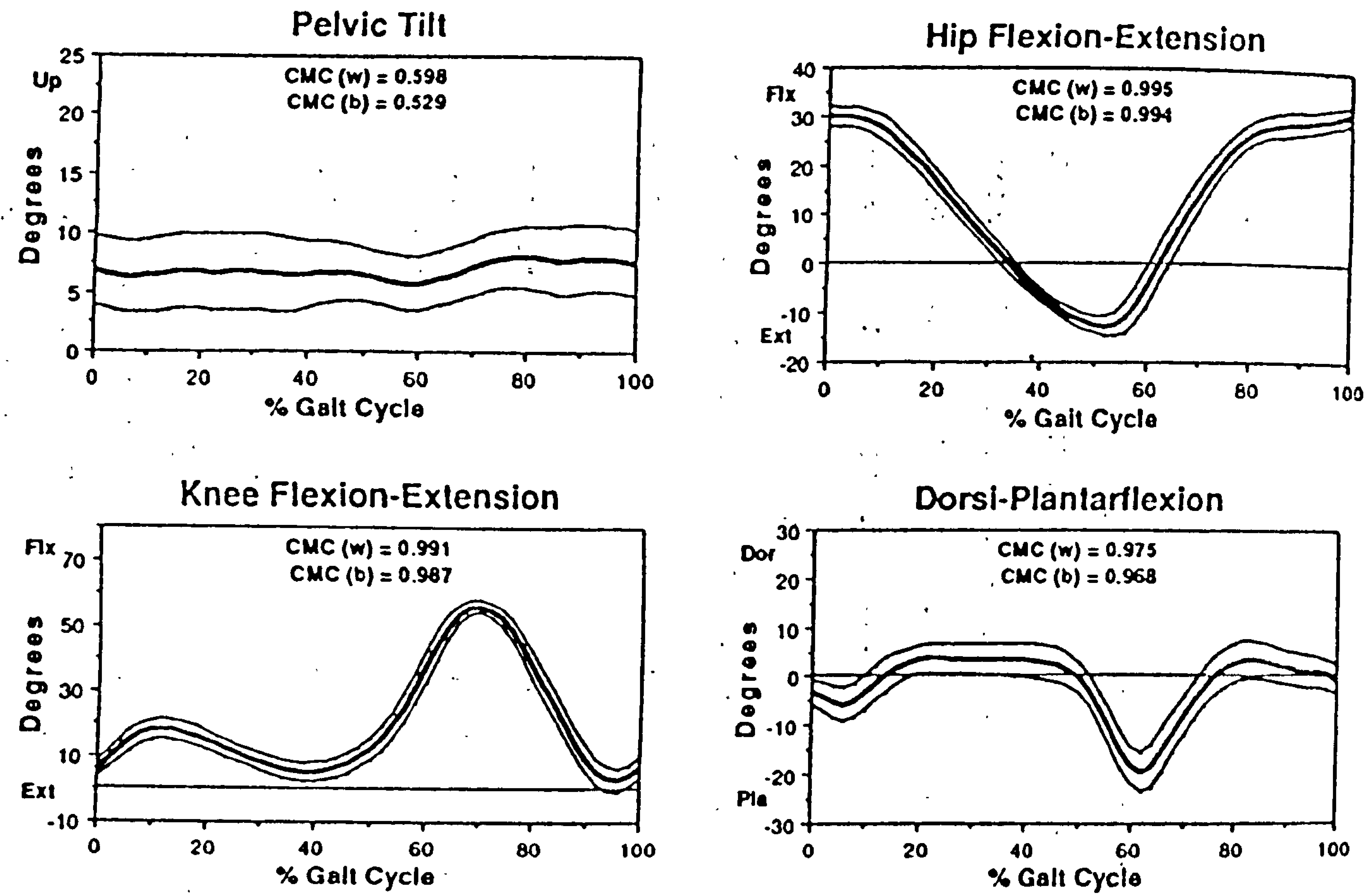


Figure 2.30 Joint angle motion in the sagittal plane. Average of nine cycles collected on three different days for a normal subject. (CMC) is the coefficient of multiple correlations, it reflects the repeatability of the waveforms. (w) within day, and (b) between days. (from Kadaba et al 1989)

Kadaba et al (1989) investigated the repeatability of the kinetic and kinematic data of 40 normal subjects at their normal speed. A repeatability measure was introduced to evaluate the repeatability of the kinetic and the kinematic data. The measure was called the coefficient of multiple correlation (CMC) and reflected the repeatability of the data. Figure 2.30 shows the mean and standard deviation of joint angle motion in the sagittal plane for a representative subject over three different days of testing. The repeatability of the kinematic data was high in the sagittal plane for the tests done both within a day and on three different days. However, the repeatability of the joint kinematics in the frontal and transverse planes was high only for the tests done within a day and poor for the tests done on three different days. This was attributed to errors resulting from the reapplication of the markers. The pattern of the ground reaction forces was repeatable for all type of tests, especially for the vertical force and for the anterior-posterior force. Figure 2.31 shows the joint moment patterns in the sagittal plane for one subject tested on three different days. The repeatability of the moments in the sagittal plane was higher than that in the frontal and transverse planes. This was explained by the fact that a higher level of control is exerted by the neuromuscular system since the direction of progression is along the sagittal plane. Although five cameras were used in this work, the two sides of the subject were not monitored simultaneously.

Kadaba et al (1990) described a system of measuring the three dimensional angular motion of the pelvis, thigh, shank and foot, based on a four-segment rigid body model of the lower extremity and recommended the use of that system as a uniform method for data acquisition so that results can be compared. Normative data based on 40 normal subjects were presented and were found to be very similar to the data presented in Kadaba et al (1989).

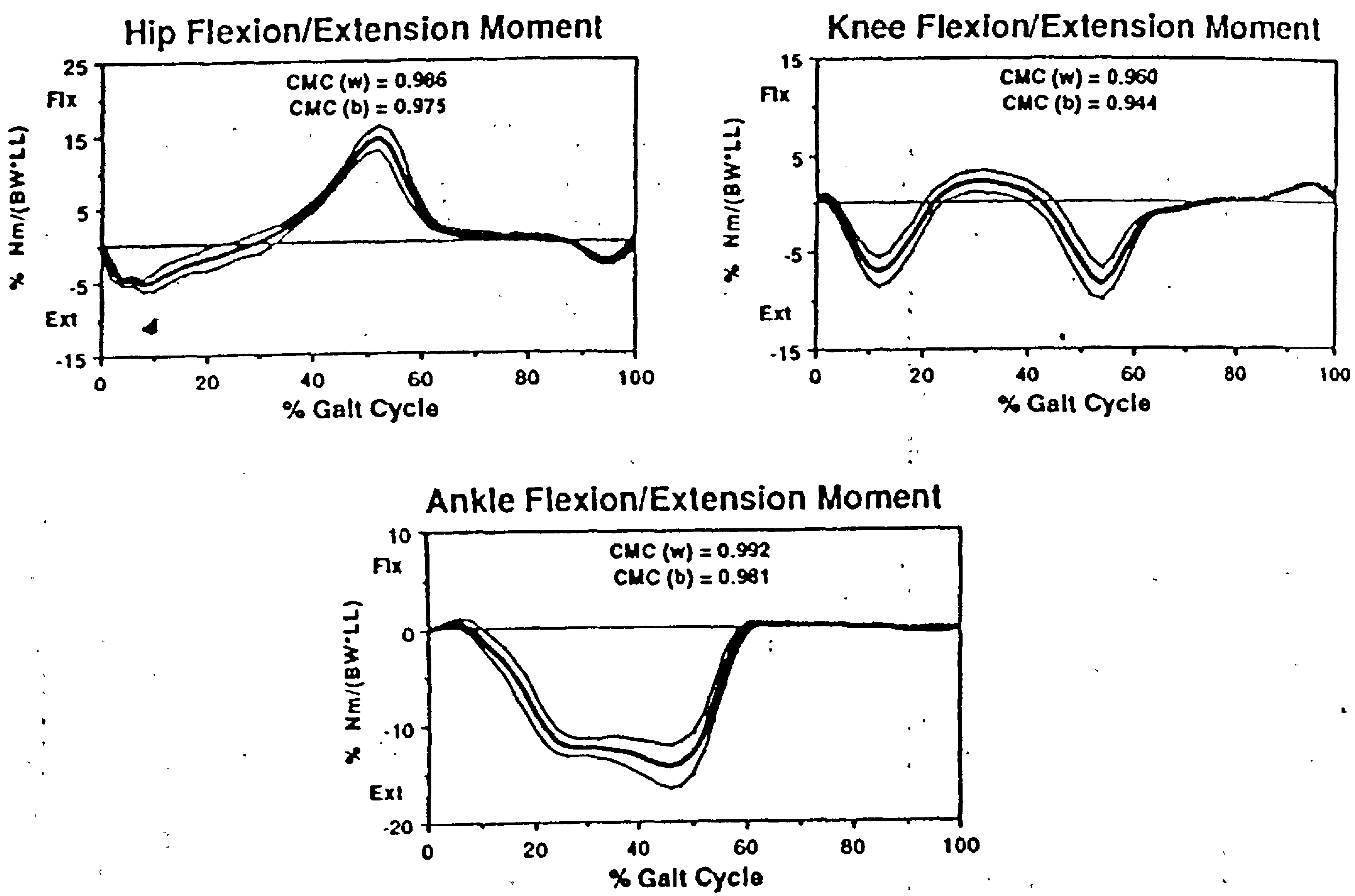


Figure 2.31 Joint moment patterns in the sagittal plane. Average of nine cycles collected on three different days for a normal subject. (CMC) is the coefficient of multiple correlations, it reflects the repeatability of the waveforms. (w) within day, and (b) between days. (from Kadaba et al 1989).

CHAPTER THREE

Lower Limb Prosthetics

3.1 Introduction

As mentioned in the previous chapter, an understanding of normal and pathological human gait has been of interest to many scientists since the end of World War II, the object being the provision of better aid for those people who lost their limbs or part of their limbs during the war. Pathological gait and the development of lower limb prostheses are discussed in this chapter, with emphasis on the gait analysis of above knee amputees, biomechanical aspects and the historical development of prostheses. As the project reported in this thesis is primarily concerned with above knee amputee's prosthetics, prosthetic aspects of below knee amputees will only be discussed briefly.

3.2 Amputation Surgery

Amputation surgery is usually carried out to serve two ends. Firstly to remove the diseased or dead tissue, and secondly to provide a stump in good condition capable of controlling the prosthesis fitted to it. Therefore, the amputation surgery should be handled with great responsibility and experience. In the ideal case, any decision for an amputation must be taken by a team of specialists consisting of the surgeon, the prosthetist, the physiotherapist and the patient himself. Each member of this team must take his/her part with great responsibility and must perform to the best of his/her experience. However, as the reason for any amputation is always clinical, the surgeon must approach the clinical need and should always have the last word. This suggests that the surgeon should also be aware of the requirements of every member in the team. Before amputation takes place, the clinic team should decide whether amputation is the correct treatment and if the answer is yes, the level of the amputation should be selected very carefully, even with the present advanced prosthetic systems. There are several factors to be taken into consideration in order to provide the best results from an amputation. These factors are pathological, anatomical, surgical, prosthetic and personal such as the

Table 3.1 Lower extremity amputations in Denmark from 1978 to 1983. (from Ebskov 1988)

Year	Total	01 Arterio-sclerosis (%)	02 Diabetes mellitus (%)	03 Trauma (%)	05 Tumour (%)	06 Congenital deformity (%)	08 Infection (%)	09 Pseud-arthrosis (%)	10 Miscellaneous (%)
1978	1838	61.0	26.6	3.4	2.2	0.1	2.9	0.1	2.9
1979	1969	59.6	27.6	4.6	2.0	0.5	2.1	0.1	3.4
1980	2168	61.1	27.4	4.0	1.5	0.3	2.2	0.1	3.6
1981	2125	61.0	27.1	4.0	2.1	0.2	2.8	0.1	4.4
1982	2165	61.9	24.8	3.7	1.5	0.5	3.3	0	4.4
1983	2200	64.1	23.2	3.5	1.8	0.2	3.1	0	4.1

occupation or sex of the patient. All these factors considered together can determine the level of amputation and they always affect each other. For example, the through knee amputation is easy to perform but it creates cosmetic problems (see the following section).

Today the performance of amputation surgery has become an easier task, particularly with the use of anaesthetics which provide good conditions for the surgeon and allow him to concentrate on effective shaping of the stump and on controlling the wound.

3.2.1 Causes and Sites of Amputations

The most common reason for amputations in the developed countries is vascular disease. Because of the high standard of living, people tend to live longer until they are less healthy and vascular disease may start to appear at their distal segments. In this case amputation will be needed to remove the dead tissue and it should leave the stump in a good condition. Murdoch (1977) reported that about 86 percent of Scottish amputees are victims of vascular disease. McCollum and Walker (1992) stated that over 90% of those patients requiring amputation, are a direct or indirect consequence of CLI (Critical Leg Ischaemia). They also reported that with the increase in the elderly population and the significant increase in the life expectancy (73 years for males, and 78 years for females in the United Kingdom) over the past 10 years, the number of CLI patients is likely to increase substantially into the 21st century. In this kind of amputation, it is usually possible to save the knee joint, especially with the recent developments in methods of the tissue viability assessment .

Injury, or "trauma" is another reason for amputation. This can be found in the developed countries as a result of new technology and industrial machinery. It is very common in third world countries as well, because they are frequently engaged in wars. In this kind of amputation, the operation should be delayed as long as possible because it is difficult to assess the viability of the distal tissue.

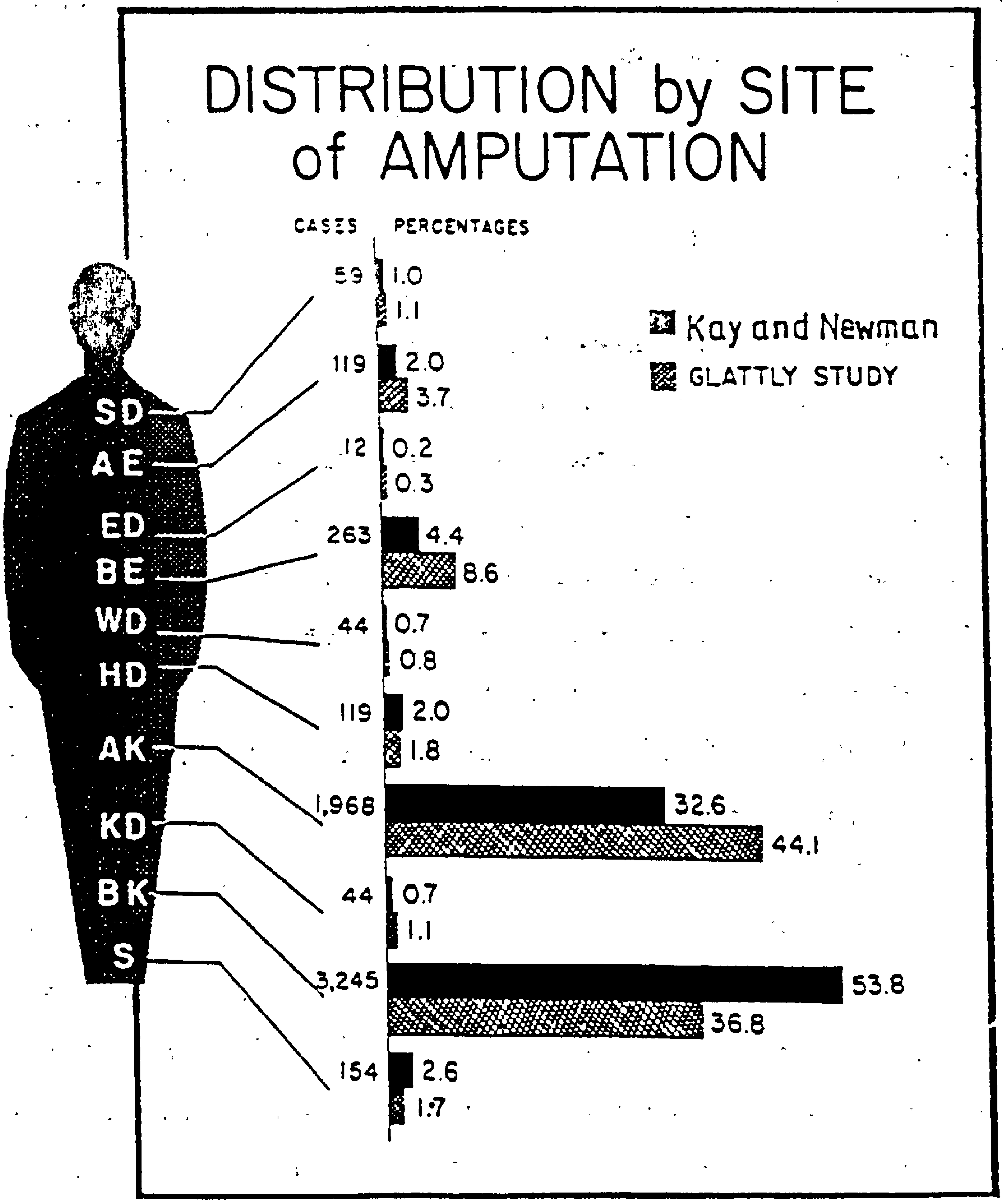


Figure 3.1 Distribution of amputations for the upper and lower limbs in the USA.(from Kay and Newman 1975)

Tumour can be a reason for amputation if radiology treatments have failed. In this case, if an amputation is to take place, it should be conducted either through or above the joint which is proximal to the tumour. This cause for amputation is not very common, and in fact Jain and Stewart (1989) stated that the experience at Dundee Limb Fitting Centre shows that over the period from 1965 to 1988, tumour amputations are only about 1.5 to 2.2 percent of all their amputations.

Chronic bone infection may sometimes be a reason for amputation. This will occur when antibiotic treatment is not successful. Amputation can be through the affected bone but proximal to the infection. It is very rare to have this as a cause of amputation especially in comparison with vascular disease amputations.

There are other reasons for the necessity of amputation such as deformity, paralysis, shortening and congenital limb deficiency. Again these causes of amputation are not very common but do exist. As the amputation in these cases is not life-saving, it is very difficult to decide whether amputation is really necessary or if the patient can manage without it. Therefore, it is always vital to consult the patient, the patient's family, colleagues and sometimes a psychiatrist before amputation takes place.

Table 3.1 shows the total number of lower extremity amputations carried out in Denmark for the period between 1978 and 1983 giving the percentage contribution of each cause of amputation. The tendency of an increase in the total number of amputations is reflected in the increase of vascular amputations, which are the result of the modern high standard of living as mentioned above. Kay and Newman (1975) carried out a statistical study of amputations in the USA, and compared the results with data obtained by Glattly in 1964. Figure 3.1 shows the distribution by site of amputation for the upper and lower limbs as presented by Kay and Newman (1975). It is clear that the number of upper limb amputations is very much less than that of lower

Table 3.2 Total lower limbs fitted with prostheses in 1987, in England, Wales and North of Ireland. (Reprinted from DHSS 1988).

	Hind Quarter	Hip Disarti-culation	Above Knee			Knee Disarti-culation	Below Knee			Ankle Disarti-culation	Partial Foot	Digital
			Upper Third	Middle Third	Lower Third		Upper Third	Middle Third	Lower Third			
England	10	35	114	1185	778	156	734	1383	57	63	52	9
Wales	- - -	2	9	23	29	5	50	13	1	- - -	- - -	- - -
N. of Ireland	1	- - -	2	30	4	3	8	48	2	2	1	- - -
Total	11	37	125	1238	811	164	792	1444	60	65	53	9
			2174				2296					

limb amputations. Table 3.2 shows the total number of lower limbs fitted with prosthesis in England, Wales and North of Ireland in 1987 as presented by the Department of Health and Social Security (DHSS). The above studies show that the above-knee amputations have the second largest percentage after those of below knee. This is one of the reasons for choosing the above knee prosthesis to be the subject for this thesis.

If a lower limb amputation is to take place, the level of amputation can be any of those seen in figure 3.2. Hip disarticulation is indicated only by disease or injury. It should be pointed out that with the present advances in the prosthetic art, there is no longer any need to leave an inch or two of femur at hip level to provide a satisfactory fitting (Murdoch 1969).

Any above knee amputation should leave the stump as long as possible but should provide adequate space for the knee unit to be fitted. As the result of an above-knee amputation, most of the hip adductors and extensors lose their insertion; this will weaken the hip adduction and extension movements, and because the hip abductors and flexors are still of normal strength, the stump will be subjected to a flexion-abduction deformity. Fulford (1969) reported that myoplasty, by re-inserting divided adductors and hamstrings, restores much of their power and reduces the muscles' imbalance. Gottschalk (1992) stated that in a transfemoral amputation, preservation of the adductor magnus is possible and helps maintain the muscle balance between the adductor magnus and the abductors. This will also allow the adductor magnus to maintain close-to-normal muscle power and be more advantageous in holding the femur in a normal anatomical position. However, it is believed that if the stump is very short, preservation of the adductor magnus may not be possible. Amputations at supracondylar level are no longer recommended because they are not truly end-bearing and they fail to provide sufficient space for the knee mechanism (Murdoch 1969).

The through-knee amputation can provide a very good end bearing area,

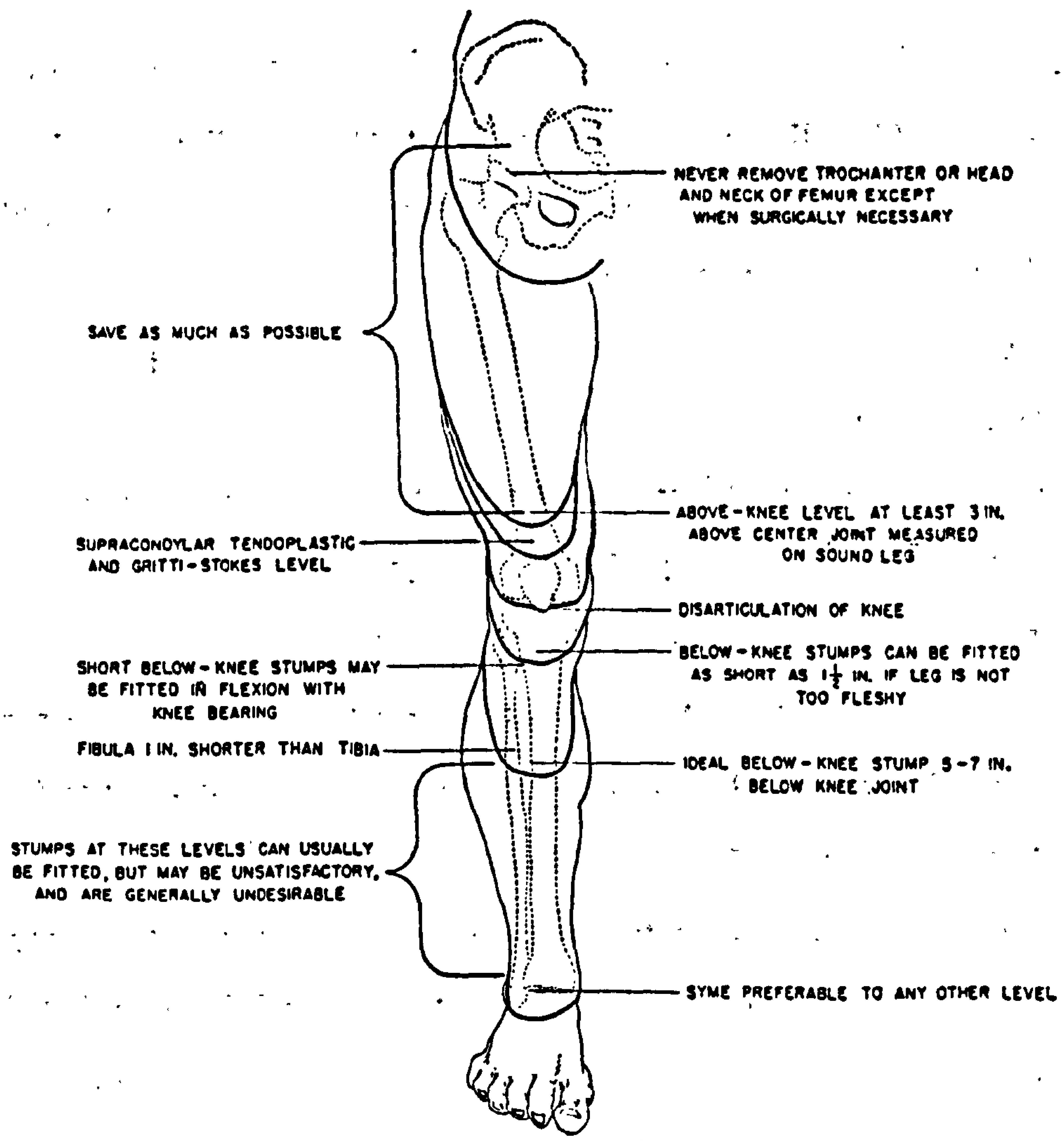


Figure 3.2 Major sites of amputation in the lower extremity. (from Alldredge and Murphy 1954)

it is relatively bloodless and is an easy operation. However, it does not provide sufficient space to install the knee unit, creating a cosmetic problem and possible knee control difficulty. Pinzur et al (1988) were in favour of the through knee amputation for two reasons. Firstly for those patients who have a low level of activity, through-knee amputation has high stability and a higher rate of wound healing. Secondly for active patients, through knee amputation is superior to above knee amputation.

For below knee amputation, the stump can usually be fitted successfully at any level, provided that the stump is at least 5 cm long; however, an ideal below knee stump should be about 15 cm long; longer stumps can also be fitted successfully. Burgess (1988) reported that for cases in which tissue viability is adequate, there is a really important place for tibiofibular synostosis. As it heals slowly, the resulting residual stump is strong, non-tender and has the ability to be end-bearing.

Syme's amputation has the same benefits and problems as the through knee amputation. Foot amputation can be as distal as possible provided that the pathological cause has been approached. However, only a few of these procedures give good functional results, therefore, this kind of amputation is not desirable.

3.3 Prosthetic Systems For The Lower Extremity

The use of lower limb prostheses is as old as traumatic amputations going back to the time of the Greeks in the fifth century B.C. Development of prostheses was very slow until the 20th century after World War II which produced injuries resulting in a great number of amputees. The requirements of these amputees have encouraged scientists to apply advanced technology to the design and manufacture of the prosthesis. This gave fast and major improvements in the function and performance of prostheses, and resulted in the present types of prosthesis which are now in satisfactorily use. It should be mentioned that, although the present prosthesis is the output of the 20th

century's work, its structure was introduced as early as 1696 by Verduin. He designed a below knee prosthesis with a wooden foot, a copper socket lined with leather, and a thigh cuff to support the device and to assist in weight bearing.

3.3.1 Above Knee Prostheses

The above knee (AK) prosthesis has three major components which are the socket, the knee unit, and the ankle/foot assembly. In addition to this is the shank tube which connects the knee unit with the ankle/foot unit. The prosthesis is produced by connecting all the components together in a particular orientation.

3.3.1.1 Sockets For Above Knee Prostheses

In practice, the socket is employed as a structural element to support the body during stance phase of the gait cycle. This should be associated with comfortable transmission of the forces to and from the stump. Several factors affect the function of the socket such as the socket type, shape, suspension system and the alignment of the prosthesis. For example, the patient may lose control of the socket if the suspension system is not effective, and the stability of the prosthesis may be lost if the alignment is not set correctly.

Eberhart (1947) used rectangular sockets to study the effectiveness of the suction method of suspension, to evaluate the gait of above knee amputees, and to improve the design of their prostheses. This study led to the understanding of the basic principles of biomechanics, design, socket fabrication, and alignment of the AK prosthesis. The present version of this socket is different in shape and is called the quadrilateral socket.

Thorndike (1949) reported on the use of the AK suction socket in the USA and especially on the work reported by Eberhart, the chairman of the Lower Extremity Committee of the Advisory Committee on Artificial Limbs. Nine failures were reported out of 211 fittings of the AK suction socket. The causes of the nine failures were listed by the author as follows: in 3 cases the patients were not cooperating and gave up during initial fitting, 2 cases had poor fit and alignment, in 2 cases cyst or abscess on adductor roll remaining from using old leg, insecurity in one case, and loss of suction in the last case. Modifications in the shape of the original design were suggested, such as reduction in the size of the ischial seat and "easing" the tightness of fit on the walls of the socket below the level of the brim. Greater emphasis on the proper alignment of the entire limb was also suggested. Thorndike (1955) confirmed the need for correct alignment in order to satisfy the patient. He also reported 15.7 percent of failures in a series of over 2000 cases fitted with the suction socket. The failures were attributed to the selection criteria of the cases for prescription and fitting and thus new selection criteria were presented. The new selection criteria consider the stump's physical condition and how convenient it is for the patient to be frequently visiting the limb shop for checking.

The " H " socket was used in Britain for above-knee amputees. It is an ischial bearing socket which used to be made of wood or metal and constructed from measurements taken from the stump.

The socket most commonly prescribed and accepted today is the quadrilateral socket. It is a socket for above knee prostheses and consists of a four sided configuration formed to prevent rotation of the prosthesis and to distribute the load on weight bearing areas. The quadrilateral socket can be one of the three following types: 1- an open ended suction socket which has an air chamber at the distal end of the stump. 2- a loosely fitted socket with an auxiliary suspension system. 3- a total contact suction socket.

It is now accepted that the quadrilateral total contact suction socket is the best quadrilateral socket for an above knee amputee. It provides good pressure distribution over the supporting areas and controls oedema which may develop at the distal end of the stump if an open ended socket is used. The load is mainly supported by the ischial tuberosity and partially by the gluteus

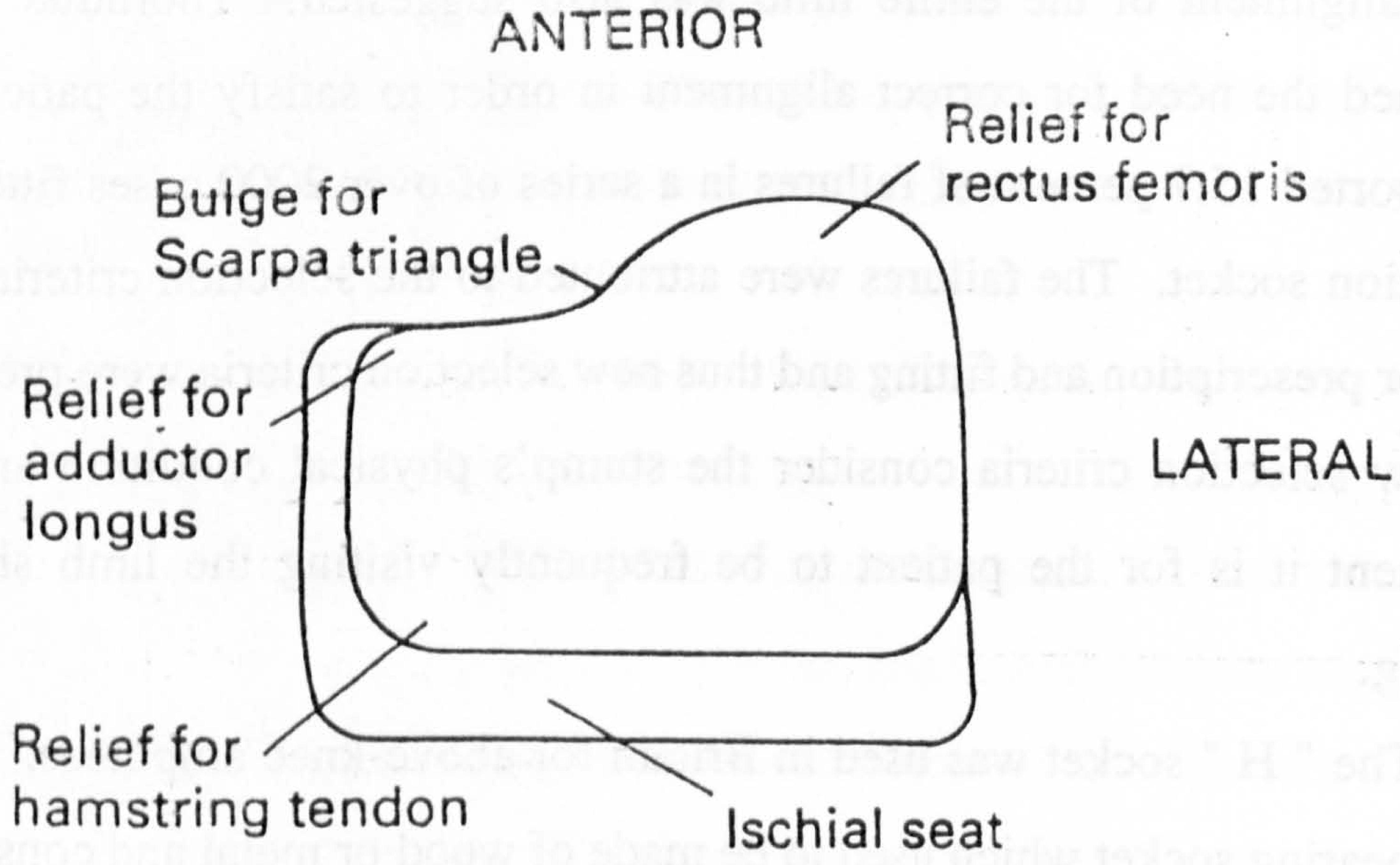


Figure 3.3 Cross section of the quadrilateral socket at brim level. (from Jacobs 1988)

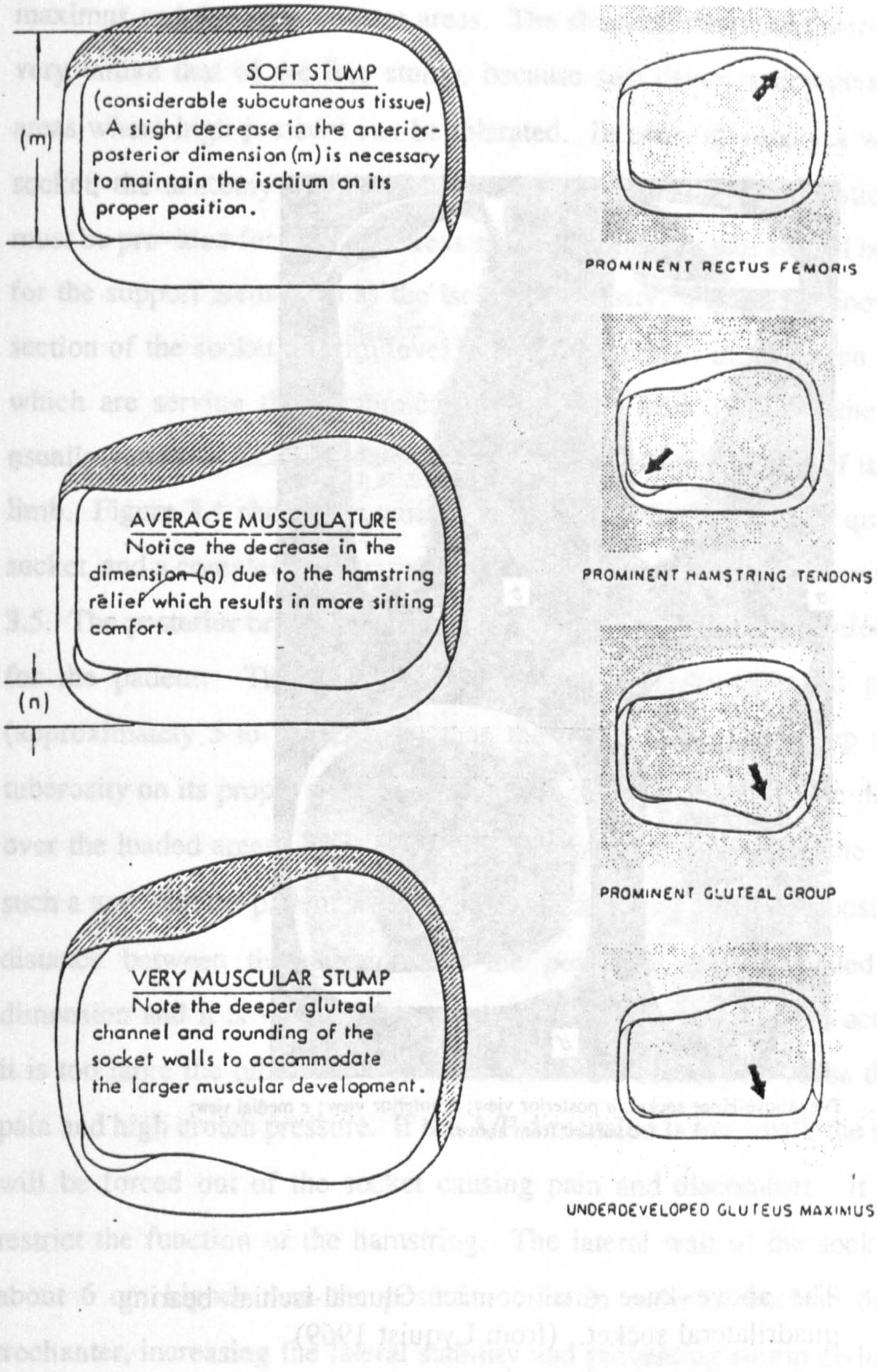
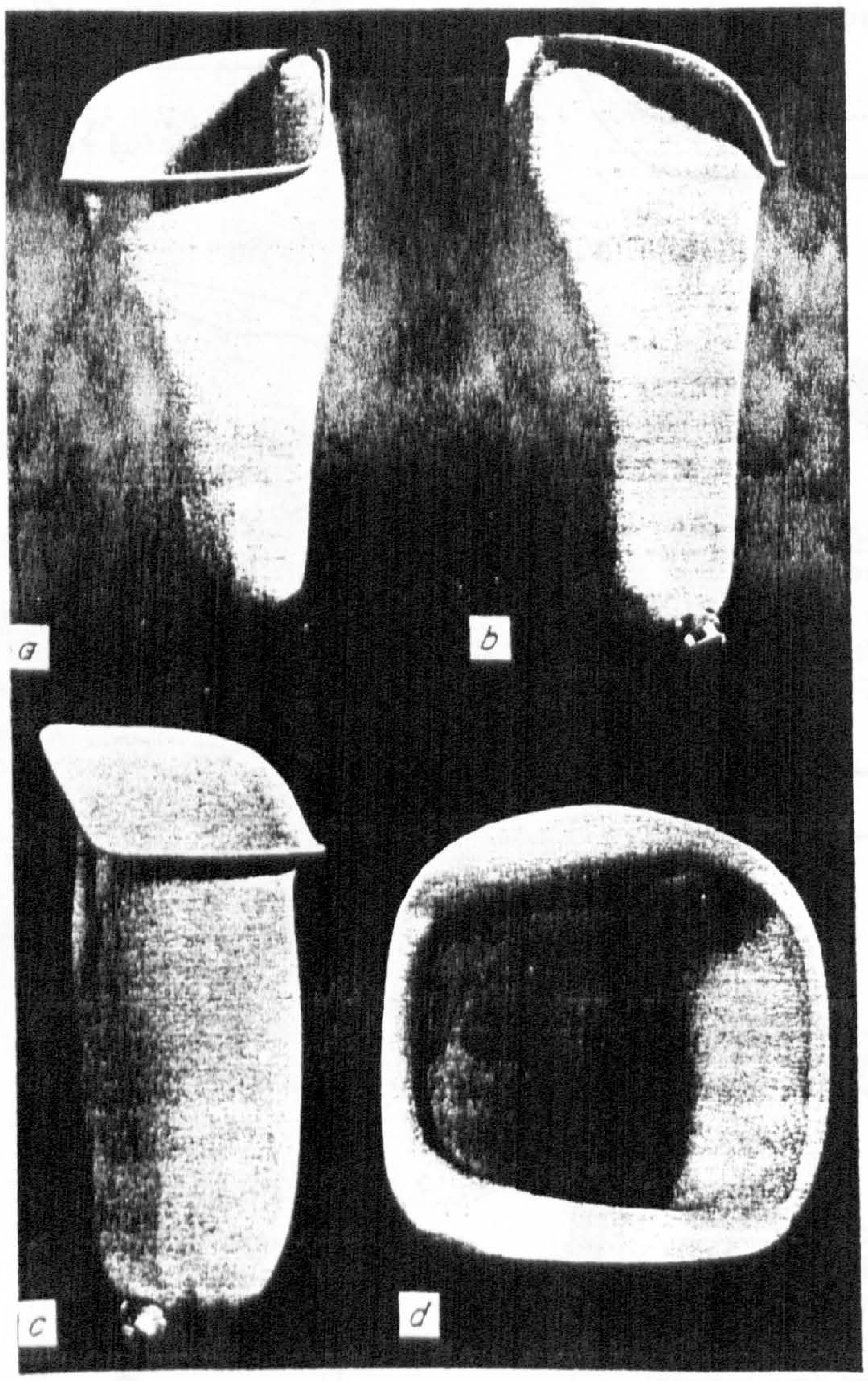


Figure 3.4 The quadrilateral suction socket shape.
(from Radcliffe 1977)



The Above-Knee socket. *a* posterior view; *b* anterior view; *c* medial view; *d* viewed from above.

Figure 3.5 The above knee total contact Gluteal-ischial bearing quadrilateral socket. (from Lyquist 1969) maximus and the other contact areas. The shape of the quadrilateral socket is very unlike that of the free stump, because soft tissue is compressed in the areas where high pressure can be tolerated. In order to supply a well shaped socket, the anatomy of the stump must be taken into consideration. Relief must be provided for sensitive areas and a comfortable seat should be provided for the support areas such as the ischial tuberosity. Figure 3.3 shows a cross section of the socket at brim level with indications of the areas on the socket which are serving the anatomical needs. The brim shape of the socket is usually varied to accommodate the amount of flesh and muscle of the residual limb. Figure 3.4 shows the various shapes of different types of quadrilateral socket, and a complete illustration of the quadrilateral socket is shown in figure 3.5. The posterior brim is left flat and horizontal to provide a comfortable seat for the patient. The anterior wall of the socket is extended proximally (approximately 5 to 7 cm higher than the posterior brim) to keep the ischial tuberosity on its proper seat and provide the best possible pressure distribution over the loaded areas. The top of the anterior brim is flared to the outside in such a way that the patient will be comfortable during the sitting position. The distance between the anterior and the posterior walls is called the A/P dimension and it is very important in socket design and must be accurate. If it is too large the tuberosity will slide inside the socket and cause the patient pain and high crotch pressure. If the A/P dimension is too small, the tuberosity will be forced out of the socket causing pain and discomfort. It may also restrict the function of the hamstring. The lateral wall of the socket is kept about 6 cm higher than the posterior brim in order to enclose the greater trochanter, increasing the lateral stability and preventing stump abduction. In order to provide medio-lateral stability and pelvis control, the lateral wall is adducted, when possible, by about 10 degrees. The medial brim of the socket is at the same level as the posterior brim and is kept horizontal with a dip to avoid contact with the pubic ramus. The medial proximal wall of the socket

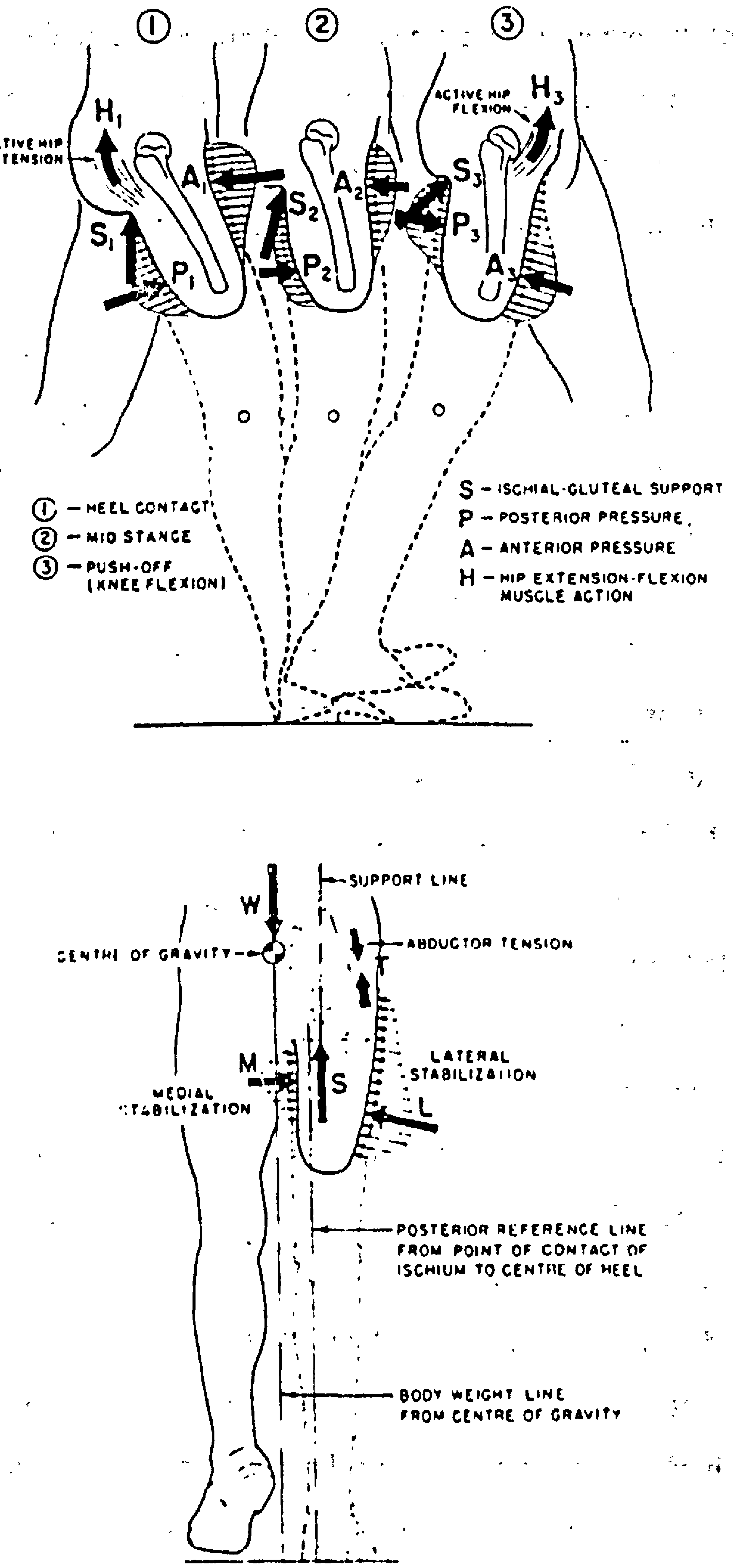


Figure 3.6 Pressure distribution between stump and socket for AK amputee, in AP (top) and ML (bottom) planes. (from Radcliffe 1977)

is kept flat to take the necessary pressure for prevention of movement of the stump in the medio-lateral direction.

Rubin (1970) has discussed some problems of AK amputees with quadrilateral socket. It was noted that if the ischial seat is too thick the patient will complain of discomfort during sitting, and if the anterior brim of the socket is too high it may contact bony prominences when the patient sits and be a source of discomfort. Rubin stated that AK amputee problems can be solved only by an understanding of the relationship between the stump and the prosthesis. Proper socket fabrication and prosthesis alignment were also suggested in order to overcome the problems of AK amputees.

Radcliffe (1955 and 1969), Radcliffe (1977), Foort (1979a and 1979b), Rolleghem and Berteele (1979), and Holmgren (1979) have reported on the design, fabrication and the biomechanics of the above knee socket. Foort has identified 13 different areas on the stump that can have a part in supporting the body weight. Radcliffe has predicted the pattern of pressure distribution between the stump and the socket for above knee amputees in the anterio-posterior and the medio-lateral planes (fig 3.6), assuming that the load is mainly taken by the ischial tuberosity and by some contribution from the gluteal musculature. It is obvious that particular attention should be paid to the areas which sustain high pressure such as the proximal and distal third portion of the stump, the pressure in these areas providing stability at heel contact and initiating knee flexion at the end of stance phase. The pressure on the medio-lateral walls which provides medio-lateral stability should be distributed in a uniform manner for better stability and in order to avoid tissue breakdown. This can be achieved by giving the socket walls proper inclination.

Redhead (1979) introduced a new type of above knee socket. It was a total surface bearing self suspending socket in which the ischial tuberosity was not the main load supporting area. It was suggested that it should be possible to treat the stump, in respect of axial compression loads, as a fixed volume

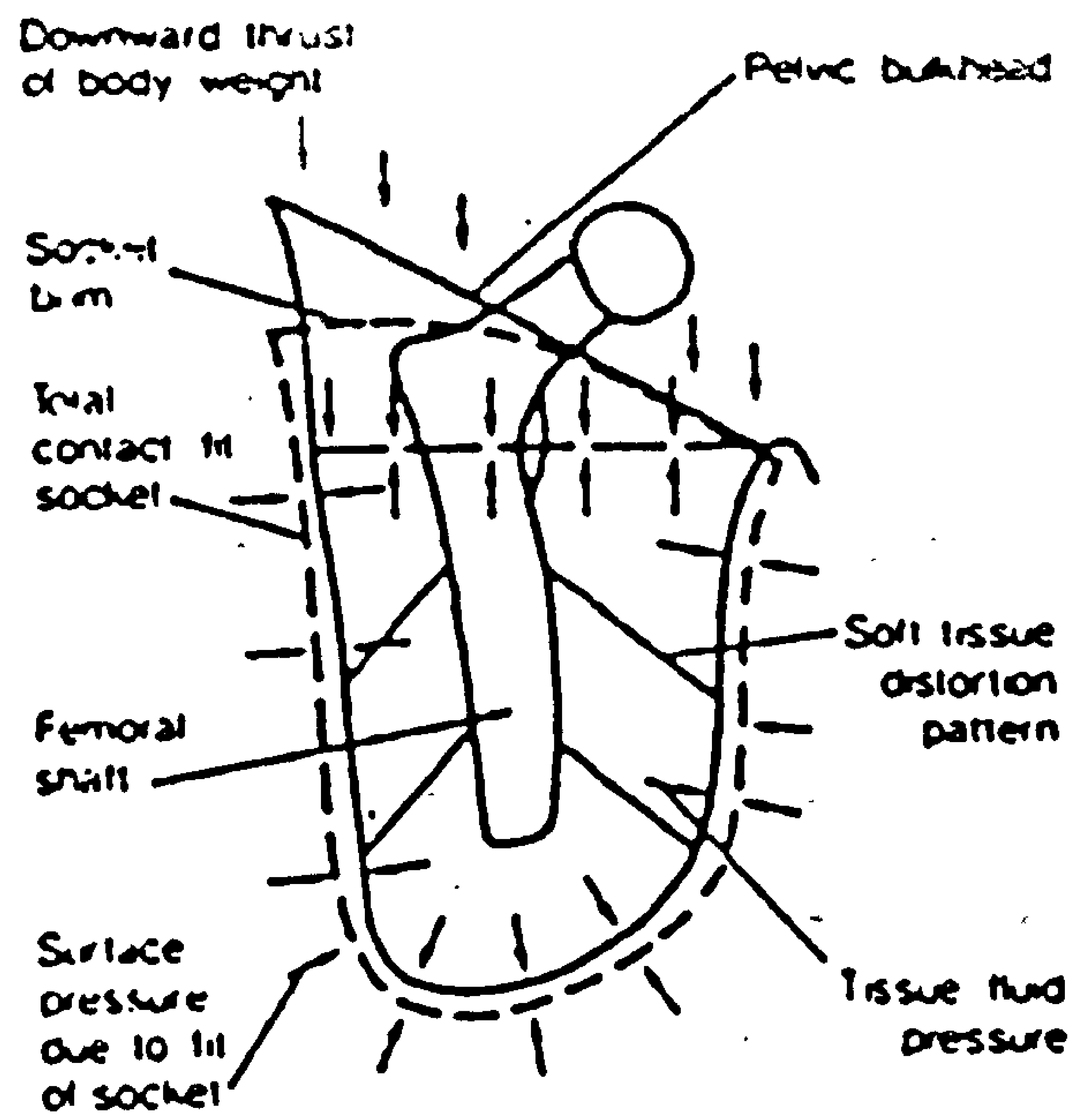


Figure 3.7 Anterior view of the total surface bearing self suspending above knee socket. (from Redhead 1979).

"bag of fluid" which behaves like a hydrostatic system. This fixed volume could be pressurised by the transference of the load of the body weight across the stump/socket interface where the pressure would be evenly distributed (fig 3.7). Radcliffe in Miami in 1987 did not agree with the hydrostatic concept in prosthetics and added that the AK residual limb is an open fluid system, and the fluid can be pushed out (Schuch 1988a and 1988b). As the lateral wall of this socket is not extended above the greater trochanter, this socket design has poor medio-lateral stability.

Kristinsson (1983) presented a flexible socket which consists of two parts: 1- a flexible socket as a tissue container which could be transparent; 2- a rigid external frame which can take loads as seen in figure 3.8. This type of socket has been used to provide excellent suspension as it can adapt itself very well to variations in stump shape. Jendrzejczyk (1985) reported on three types of supporting mechanisms for the flexible socket but recommended that the effectiveness of the flexible socket system has to be evaluated. The author did not specify how the effectiveness was to be evaluated, but it could be taken that its biomechanical aspect has to be investigated. Krebs and Tashman (1985) conducted a kinematic and kinetic comparison between the conventional and the flexible ISNY (Iceland, Sweden, and New York University) above knee socket. Only one patient was tested. The results showed no important differences in the kinetic, kinematic, or temporal data but the subject found the ISNY socket to be significantly more comfortable. Kristinsson (1988) reported again on this new system and on its advantages over the old one. From the patient's point of view, the system gives muscle activity more freedom and provides a better suspension mechanism as a result of its flexibility (fig 3.9). From the prosthetist viewpoint, as the socket can be made transparent, a visual inspection of the stump can be made and the quality of fit can be assessed. Kristinsson stated that the system is inexpensive, easy to fabricate and replace, compared to the conventional socket. This system has been well accepted by

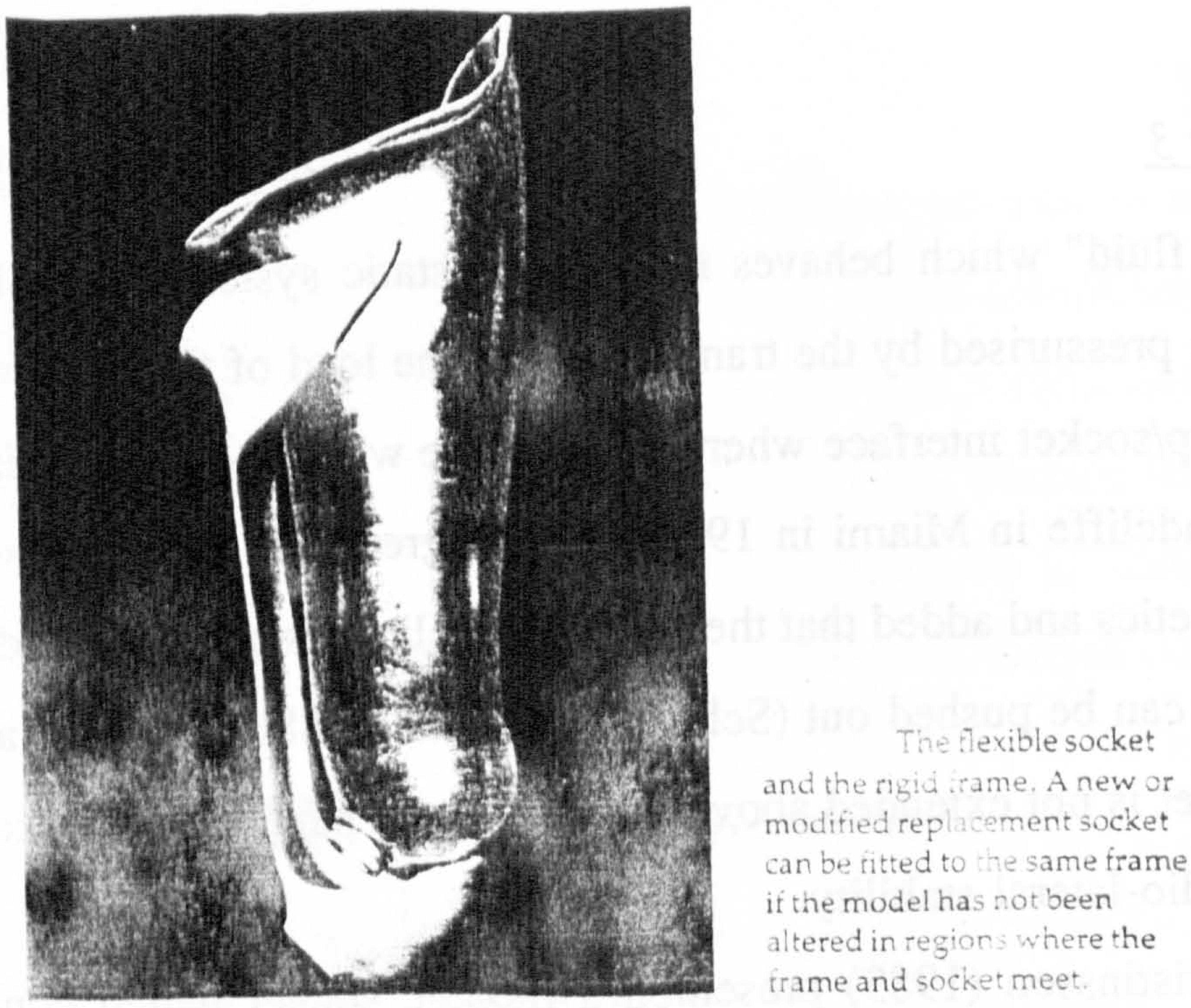


Figure 3.8 . Flexible socket in rigid frame.
(from Kristinsson 1988)

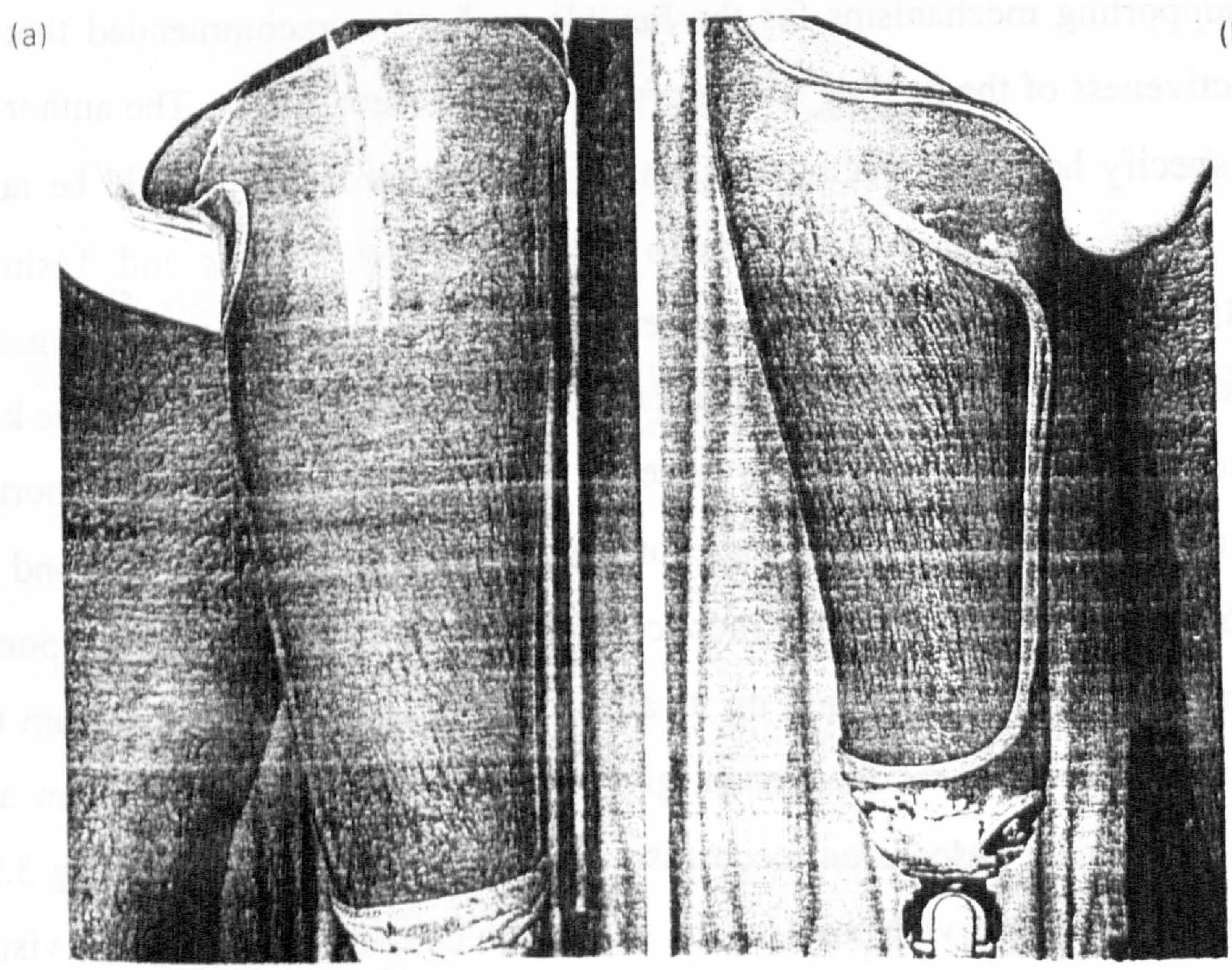
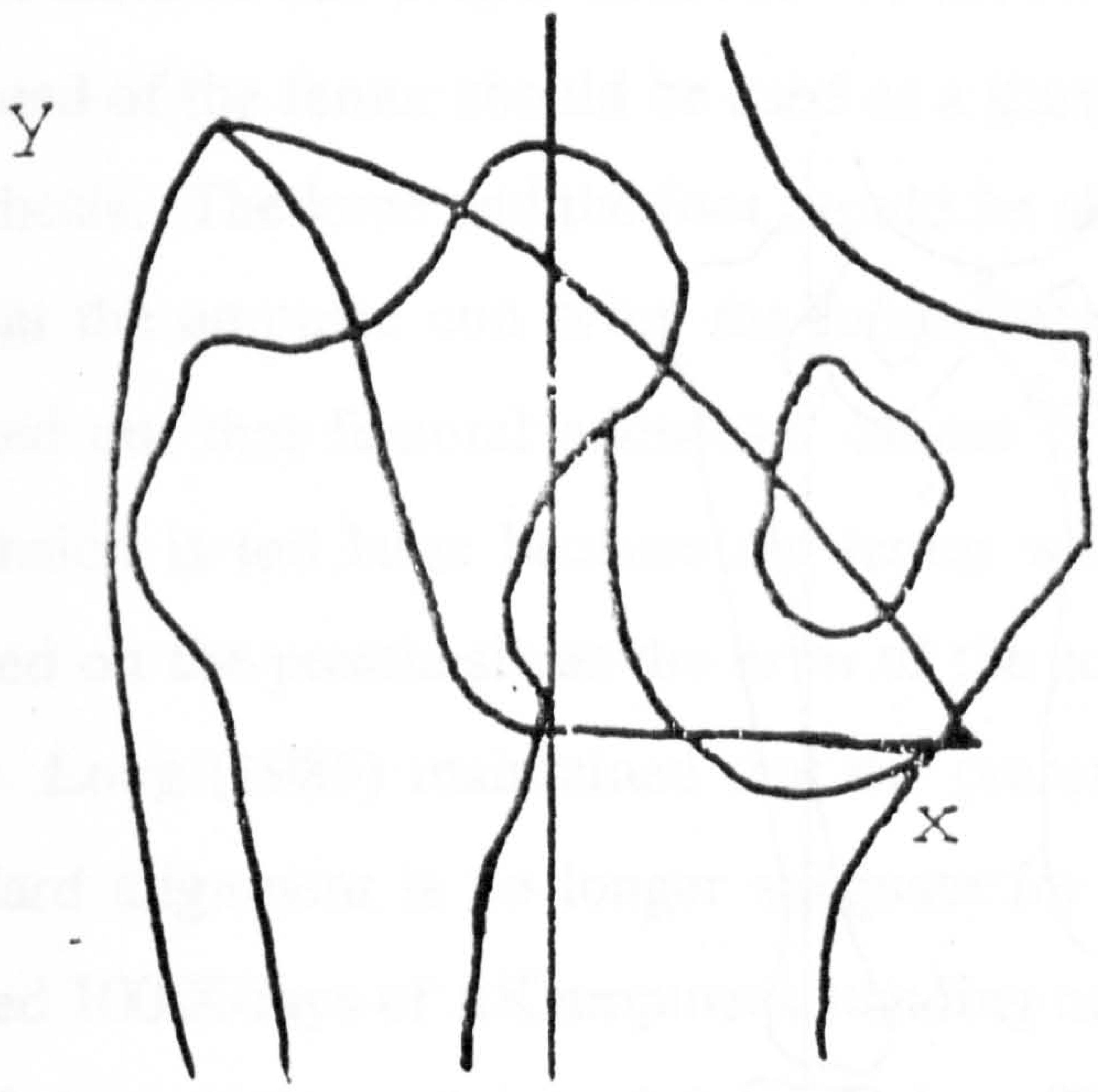


Figure 3.9 The advantages of the flexible socket. (a) Lateral view.
(b) Anterior view. (from Kristinsson 1988)



Anterior view showing relationship of medial brim (point X) to ischium and of lateral wall (point Y) to greater trochanter.

Figure 3.10 The narrow ML socket of Long 1985. Note that the ischium is inside the socket.

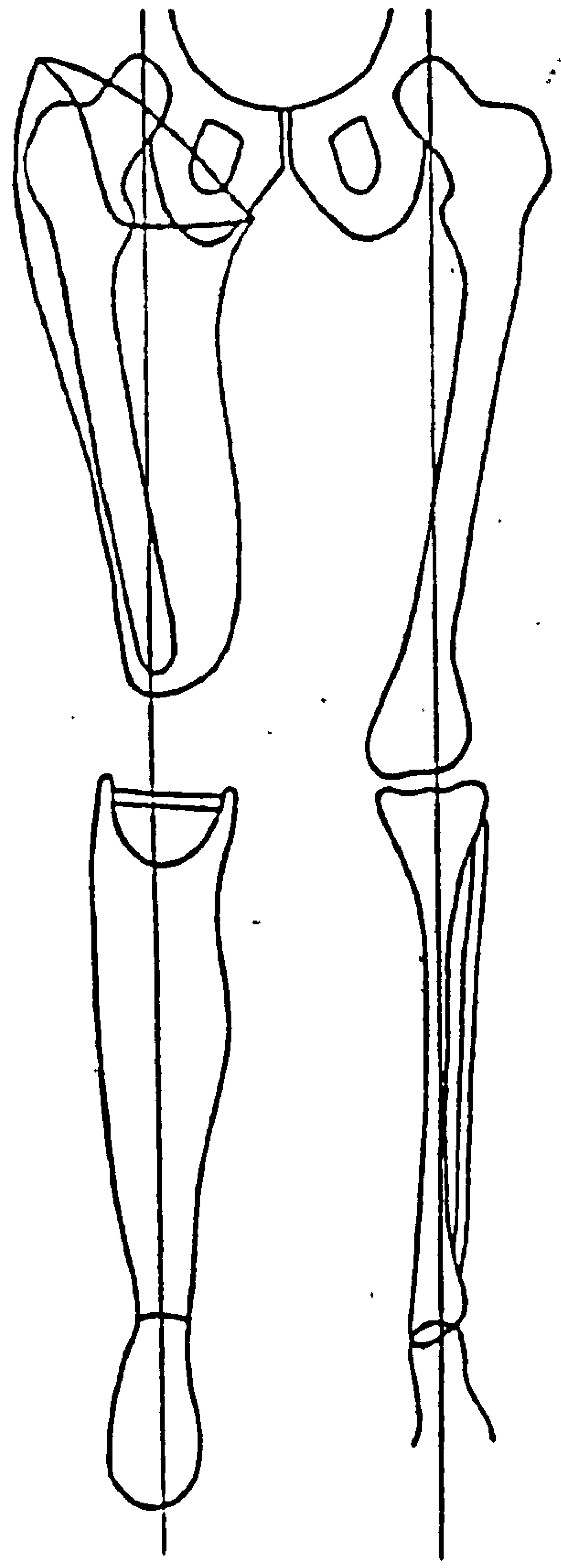


Figure 3.11 Long's line which defines the position of the foot and the distal end of the femur relative to the femoral head. Viewed from the posterior. (from Schuch 1988).

patients and by prosthetists.

Long (1975) reported that X-ray investigations showed that only a few AK prostheses had proper adduction of the femoral stump. He suggested that the head of the femur should be used as a starting point when aligning the AK prosthesis. The knee and the foot should be aligned with respect to the socket so that the amputee con bring the femur into a normal position. Long also pointed out that femoral adduction cannot be maintained if the socket M/L dimension is too large because the femur will lose its support when load is applied on the prosthesis as the brim of the socket will be shifted laterally.

Long (1985) maintained that the present quadrilateral socket with the standard alignment is no longer adequate for an above knee prosthesis. He studied 100 X-rays of AK amputees standing on their prostheses and found that 92 of them had an abducted femur. Long stated that most of above knee amputees walk with an abducted gait as a result of a too large medio-lateral (ML) socket dimension and a too small anterio-posterior (AP) dimension which also restricts the function of the active muscles. In order to overcome all these problems, Long introduced a new above knee system which he called Normal Shape Normal Alignment (NSNA) prosthesis, in which the amputee would be more comfortable and walk more like a normal subject. The socket has narrow ML and wide AP dimensions with the ischium inside the socket to prevent lateral shift during weight bearing as seen in figure 3.10. It should be noted that the lateral wall is above the trochanter. Long advised that the alignment of the prosthesis should be set according to Long's line which is defined as " a straight line from the head of the femur (located approximately at the centre of the narrow socket), through the distal femur and down to the centre of the heel" (fig 3.11). This line is not always vertical because of the shift when changing from the standing to walking position.

Lehneis (1985) questioned the biomechanics of the quadrilateral socket, suggesting that the distance from the ischial tuberosity to the ischial seat of the

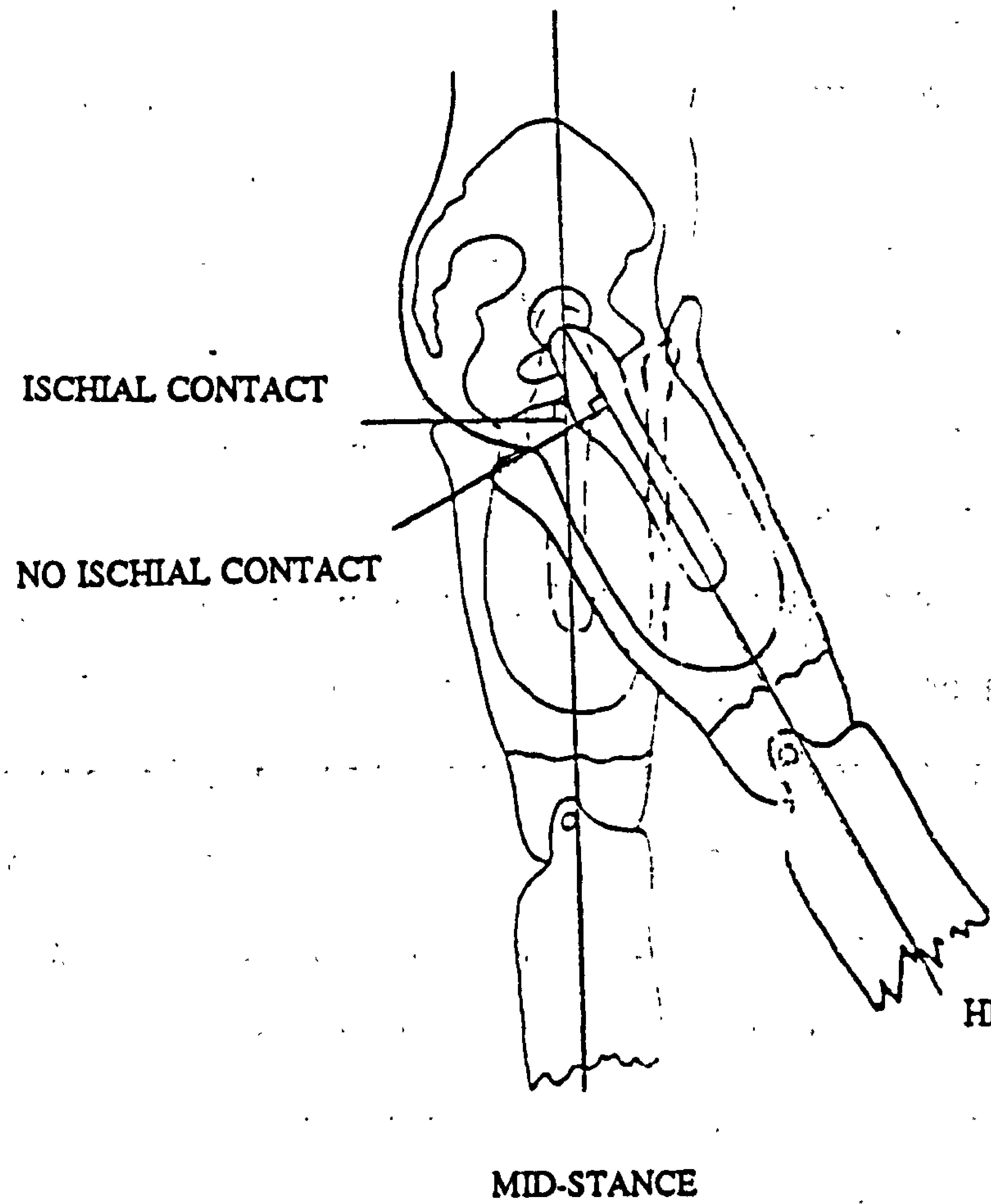
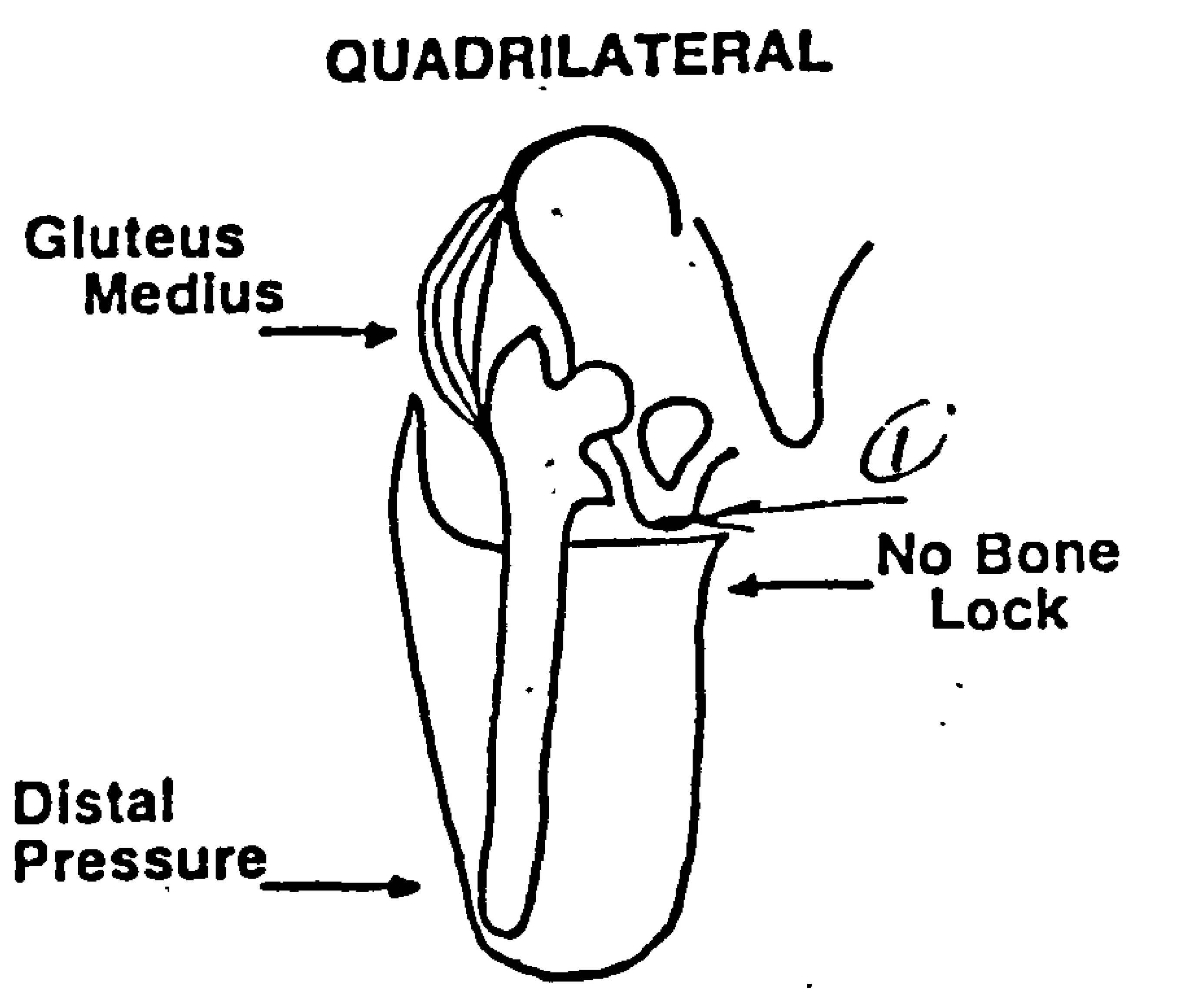


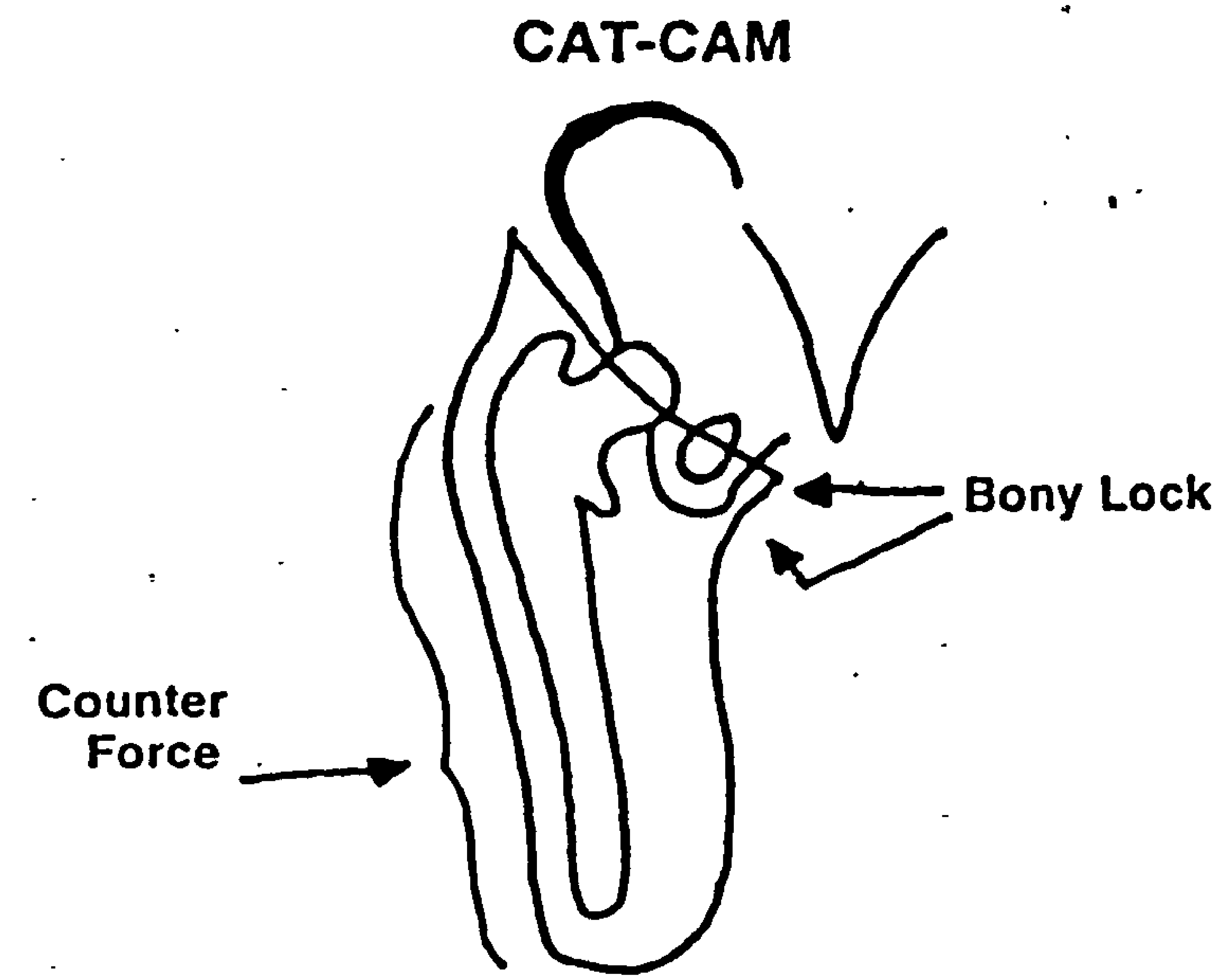
Figure 3.12 The distance between the ischial tuberosity and the socket ischial seat at heel strike. (from Lehneis 1985).

socket is increased when the hip is flexed at heel strike (fig 3.12). Lehneis stated that the socket cannot be ischial weight bearing at this point, especially considering the fact that the need for weight bearing at this point is more than at any other point in the gait cycle as body weight and " impact force " must be transmitted through the socket. Lehneis reported that discomfort arises from the quadrilateral socket, especially when the patient is provided with a manual knee lock, because the patient steps on the prosthesis (weak stump muscles, and the patient is unable to contract these muscles prior to heel strike). He therefore recommended that one should think beyond the quadrilateral socket and should consider designs such as the ischial containment sockets in which the ischial tuberosity is not the main support area.

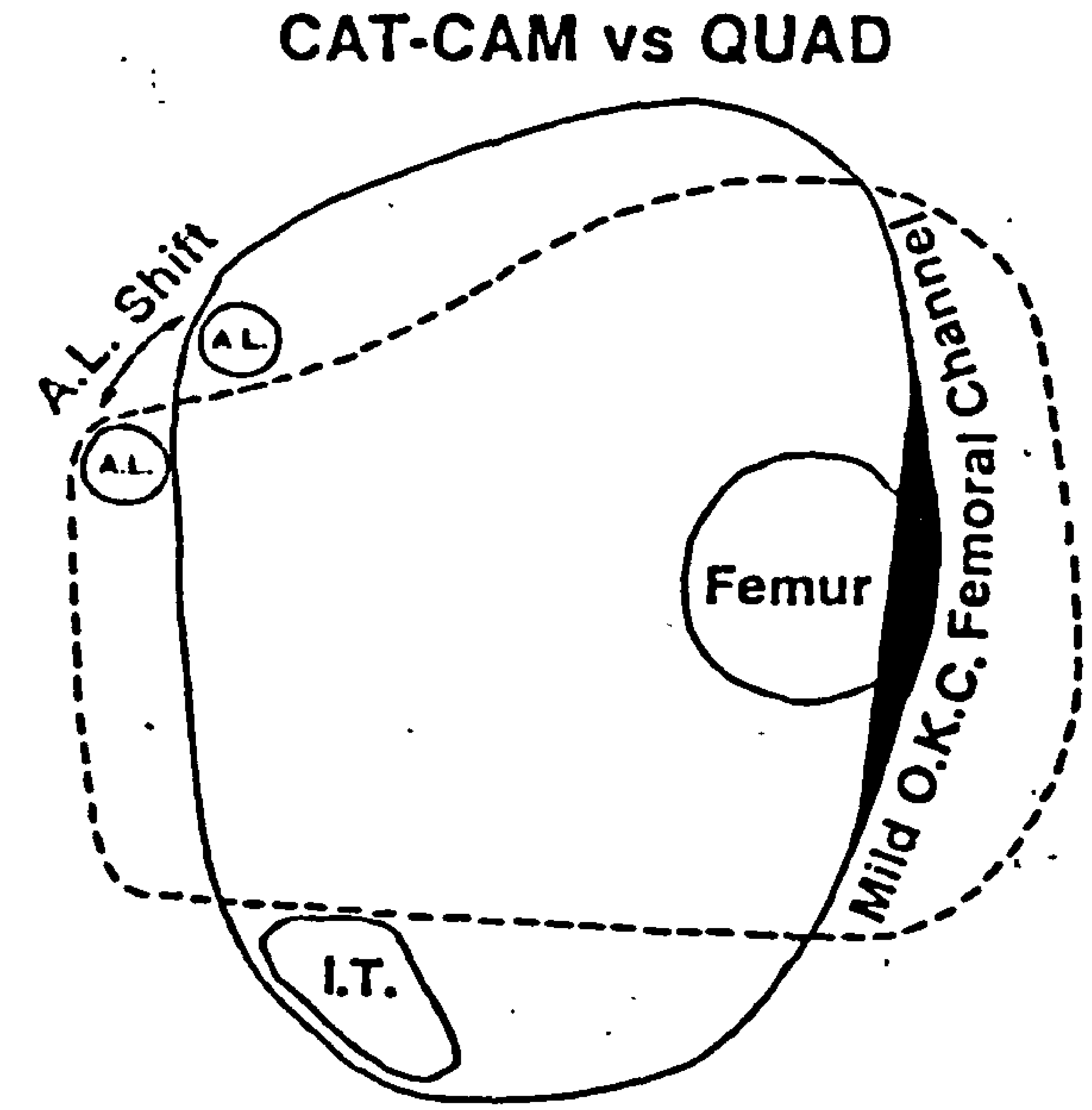
Following the recommendations of Lehneis and based on Long's work, Sabolich (1985) attempted to overcome the problems of the quadrilateral socket by introducing a new socket design which is very similar to that of Long. The socket has a narrow ML and wide AP dimension, with the ischial tuberosity and part of the inferior ramus of the ischium inside the socket. He called this socket " CAT-CAM " which stands for Contoured Adducted Trochanteric-Controlled Alignment Method. Sabolich admitted that the exact mode of weight bearing of this socket was not clear, but he assumed that the femur is able to take some weight due to the adduction angle and that hydrostatic weight bearing is playing a role with the possibility of the ischial tuberosity also bearing some weight. Figure 3.13 shows a comparison between the CAT-CAM and the quadrilateral sockets; it should be noticed that the shift of the adductor longus tendon which is marked on this figure is very unlikely to occur. Sabolich reported that he fitted about 900 sockets of this type to patients aged from 6 months to 103 years. Their subsequent X-rays have been impressive, showing the femur to be in a much improved adduction attitude. This new design of the ischial containment (IC) sockets introduced by Long and Sabolich, appears to show adequate results as reported by the developers;



No bone block and no real force system to prevent femoral or ischial drift. Ischial tuberosity acts as a fulcrum. Pelvis can rotate as well as the femur abduct.



Ischial tuberosity is locked in the socket to provide a counter force against femoral shift.



Comparison of CAT-CAM and quadrilateral sockets in a transverse view. Since the femur and ischial tuberosity are fixed in position, the adductor longus tendon has to shift a small amount. Note mild O.K.C. (Oklahoma City) channel about the femur.

Figure 3.13 Comparison between the quadrilateral and the CAT-CAM sockets. (from Sabolich 1985) the author of this thesis however believes that the IC sockets have still to be biomechanicaly evaluated. The weight support areas have also to be studied and clarified, and the hydrostatic weight bearing assumption has to be confirmed. A long term test must be carried out on the IC socket because under the weight bearing nature of this socket, the skin of the patient's stump may be subjected to high shear. This could cause serious problems for the patient after prolonged use of the prosthesis. The author in fact has been informed (Spence 1992) that a long term user of an IC socket showed evidence of skin stretching associated with increasing prominence of the distal femoral remnant. If this stretching is increased, the bone may be forced out of the skin after a short time. If that is true, and if that is the case in many patients, the whole concept of the ischial containment socket will have to be fully investigated and may in fact have to be abandoned.

Schuch (1988a and 1988b) discussed the concepts of all the new socket designs, which were presented in May 1987, at an international workshop in Miami, Florida. At the beginning of the workshop Bennett Wilson gave a brief history about socket development mentioning the CAT-CAM, NSNA and the narrow ML sockets, and ended the talk by saying " unfortunately, none of these techniques has been subjected to an evaluation programme independent of the development group". Radcliffe, who is a strong supporter of quadrilateral sockets said, that all the new techniques have been compared with a poorly fitted quadrilateral socket and not with a well fitted one. Radcliffe discussed Long's line and said that adduction and flexion of the socket for the normal use of the hip musculature has been a part of good prosthetic practice for 40 years. Referring to the ischial containment (IC) sockets, he stated that the only difference between the quadrilateral and the IC socket is in the positioning of the ischium: ie. whether it is within or on the brim of the socket and he suggested that "catchy names" should not be used to define a new design.

Flandry et al (1989) converted five above knee amputees from

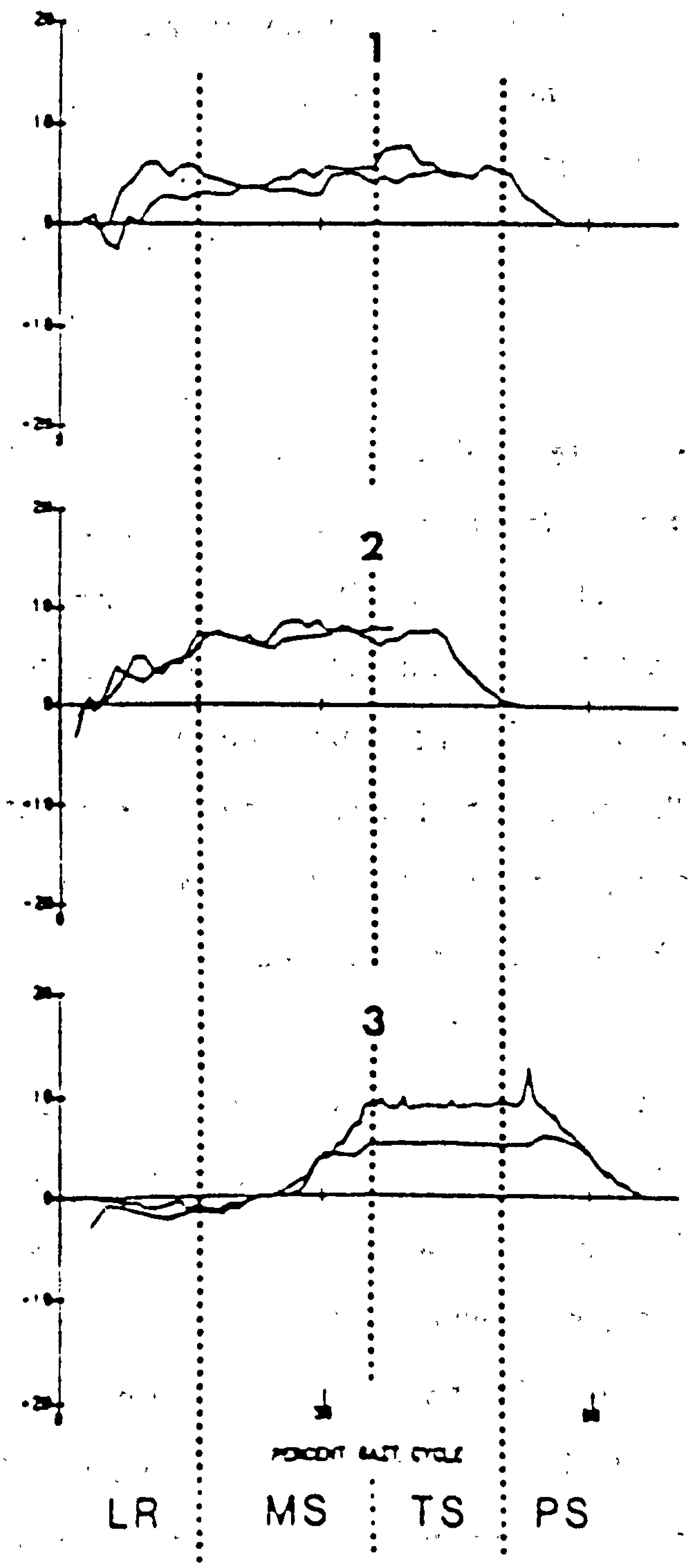


Figure 3.14 Hip moments for different patients.
The unit of the vertical axis is percent of BW/leg length. LR. loading phase, MS. mid-stance phase, TS. terminal stance phase, PS. pre-swing stance, BW. body weight as defined by the authors. (see text for more specifications). (from Flandry et al 1989).

quadrilateral to CAT-CAM sockets and studied the effect of the CAT-CAM socket on the function of the prosthesis. The CAT-CAM socket was declared superior to the quadrilateral by four patients, the fifth being incorrectly aligned. X-Ray investigations showed that the femur was adducted on average 6.5 degrees from its quadrilateral position. The "hospital's observational gait analysis system ", referred to by the authors, was used to record the movement of the patients and to calculate moments at the joints of the leg. The lateral trunk inclination which existed in the quadrilateral was found to disappear after conversion to the CAT-CAM socket. It was reported that the hip moment showed no significant difference between the quadrilateral and the CAT-CAM sockets. However, the author of this thesis sees noticeable differences in the hip moments especially of patient number 3 (fig 3.14), but unfortunately the curves in this graph have not identified with the respective sockets. There was no indication whether this moment is in the A/P or M/L plane and no definition of the axes convention which was used. Also the authors did not make it clear whether the data presented was for the amputated or the sound side, and whether it was repeatable or not. No information was presented regarding the specifications of the gait analysis system used, such as accuracies, number of cameras or method of calibration. The mean gait velocity increased from 40.4 m/minute in the quadrilateral socket to 44.5 m/minute in the CAT-CAM socket. The mean stride length was increased after conversion to CAT-CAM from 0.99 m to 1.02 m, and the average energy expenditure decreased as the mean quantity of oxygen expenditure per metre decreased from 0.364 ml to 0.286 ml. Flandry et al reported that the patients were in favour of the CAT-CAM socket particularly in the areas of comfort and stability, and only one patient whose prosthesis was "poorly aligned", requested reconversion to his quadrilateral socket. However, the author of this thesis believes that stability could be achieved with either type of socket by achieving the correct alignment, and no socket will function satisfactorily if it is poorly aligned. The above study does not provide enough information to objectively determine which type of socket is superior, and biomechanical analysis of the whole body should be carried out to evaluate the two sockets. Movements and moments at the various limb joints should be analysed, and the reaction force acting on the foot should also be studied.

Esquenazi et al (1989) reported on the problems of the quadrilateral socket and pointed out the advantages of using the new socket designs, particularly the CAT-CAM, the NSNA and the narrow ML socket. They are completely in favour of these new designs but pointed out that the need for having several casts and check sockets to obtain a satisfactory socket has made the manufacture of these sockets expensive and time consuming. They also supported the use of flexible materials in the fabrication of above knee sockets such as the ISNY socket.

Mitchell and Versluis (1990) have presented one case study of an above knee amputee, to report on the superiority of the CAT-CAM socket over the quadrilateral socket, especially for patients with complex medical problems. They stated that the quadrilateral fitting failed to provide the patient with a satisfactory function, but the CAT-CAM fitting has shown very good results on the same patient. However, this case cannot be taken into serious consideration because the quadrilateral socket was evaluated by the patient and the therapist as ill-fitting, while the CAT-CAM socket was fitted properly after a home exercise and training protocol were properly followed by the patient.

Pritham (1990) discussed and clarified the predicted biomechanics and shape of the above knee sockets which were classified in only two groups, quadrilateral and ischial containment. Pritham looked at the shape of the sockets, the predicted biomechanics, and the alignment point by point and subjectively studied the differences and the similarities in these points according to the two types of socket design. The vast majority of the points which have been discussed were found to be compatible with the analysis by

Radcliffe of the quadrilateral socket. The two socket shapes and alignment configurations were found to have nearly the same principle but, certain modifications were carried out on the quadrilateral socket by workers such as Long and Sabolich in order to overcome some of the problems which were found with it. However, Radcliffe did not agree that these modifications were necessary as he believed that in fact the prosthetists were not following the recommended principles, and had they done so the problems would not have occurred. The author concluded the paper by stating some issues and some questions which are still to be resolved such as; what are the support points in the ischial containment (IC) socket ? can claims made by the advocates of the IC style sockets be verified ? for whom is the IC socket indicated ?. Answers for these questions will clarify the concept of the IC sockets and may clarify the differences between the IC and the quadrilateral sockets.

Lawrence et al (1991) compared the energy cost of walking for AK amputees wearing quadrilateral sockets (13 subjects were tested) to those wearing CAT-CAM sockets (9 subjects were tested). Two speeds of walking were performed and the results were compared with those of 8 normal subjects. At a speed of 1.25 mph (2.01 km/h) the CAT-CAM and quad groups expended energy at rates of 16% and 24% respectively higher than the group of normals. Also at a speed of 2.5 mph (4.02 km/h) the CAT-CAM group and the quad group expended 20% and 37% more energy respectively compared to the group of normals. No more information about the data was presented as the source is an abstract.

Visser-Meily et al (1992) have converted thirteen AK amputees from quadrilateral socket users to CAT-CAM socket users. Then, after an average use of 7 months of the CAT-CAM socket, a subjective evaluation was made of their present comfort situation in comparison to that with the quadrilateral socket. The evaluation was conducted by interviewing the patients. It was stated that 80% of the subjects reported an improved walking style and 90%

reported better fitting. In addition there was a clear reduction in complaints related to the lower back and groin. Therefore, they recommended that above knee amputees be equipped with NML (Narrow Medio-Lateral) sockets if a choice of suction socket has to be made. The author of this thesis believes that such a recommendation cannot be based on the work presented in this paper, particularly since the work was subjective and depends only on patients' comments; and biomechanical analysis based on objective methods should be carried out to verify the above conclusions.

In conclusion, the above knee quadrilateral socket has been accepted and used for more than 30 years. This socket provides good support and function for the users, but it has some problems which may be solved by using the new above knee technique ie (the IC socket). If the new technique proves to be unable to overcome the problems in the old system, a new modification on the old system will be needed, or other new concepts will have to be implemented. In fact the new IC sockets have been shown to solve some of the quad problems such as comfort and the abducted gait (Flandry 1989), but it has still to be subjected to a biomechanical analysis in detail and to a long term evaluation especially with respect to the stump's conditions.

Socket Suspension Systems

The suspension system has a great effect on the function of the socket and on the performance of the whole prosthesis. A poor suspension system may alter the gait of the patient, and may cause discomfort and gait deviations such as "vaulting" which is very common in above knee amputees who have poorly suspended prostheses.

There are several methods of socket suspension for above knee amputees, but the most common methods are by suction, a mechanical hip joint with pelvic belt and the Silesian belt. Socket suspension can be provided by one of the above methods or by a combination of more than one method, usually the combination is between the suction method and one of the other

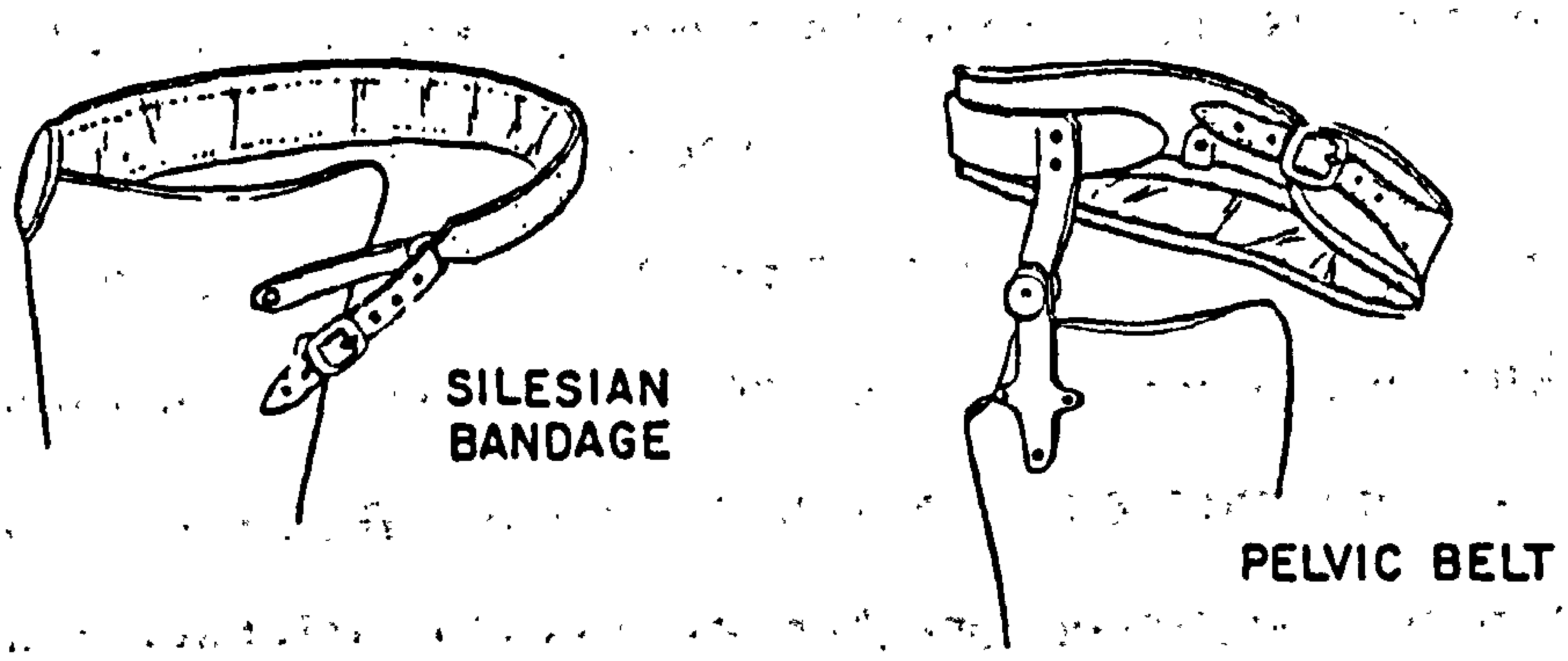


Figure 3.15 Two methods of suspension for AK socket.
(from Muilenburg and Wilson 1984)

two methods.

The suction suspending method was introduced to the USA in 1947 by Eberhart. Atmospheric pressure sustains the socket on the stump during the swing phase by creating a reduction in pressure between the socket and the stump. It is usually preferred with total contact sockets where no air chamber is needed at the distal end of the stump. This method requires a valve at the distal end of the socket and accurate socket fabrication to allow the socket walls to be in direct contact with the skin. This method is suitable for patients who are in good physical condition and have a stump with good muscles.

The Silesian belt (fig 3.15) has no mechanical joints and is connected to the socket by a flexible link. The belt is usually of leather, attached to the lateral wall of the socket and extended to the patient's back to be attached to the anterior wall of the socket. This is a comfortable, lightweight, simple method to use and it can provide a definite control of rotation and adduction of the prosthesis.

The mechanical hip joint and pelvic belt method (fig 3.15) is suitable for patients who have poor ability to control their prosthesis and need a positive suspension (Mooney and Quigley 1981). It has a rigid hip joint which is connected laterally to the socket at its top brim. This joint is placed anteriorly and proximally to the greater trochanter so that it cause no interference with the motion of the patient's hip joint. The joint is connected to a belt which is fitted around the patient's waist. Schuch (1992) stated that this suspension method provides rotational stability plus a significant degree of mediolateral pelvic stability; however, this method is generally reserved for the above type of cases since most amputees object to the weight and bulk of this suspension.

3.3.1.2. Knee Mechanisms for Stance and Swing Phase Control

The mechanism of the knee unit is one of the most important elements in the AK prosthesis. The knee mechanism can affect the performance of the amputee and the function of the prosthesis. Therefore it should be chosen carefully by the members of the clinic team taking into account the activity level and the physical fitness of the patient. All knee units are designed to give stability during stance phase, natural smoothness during swing phase, and to be comfortable when the patient is sitting. The differences between the knee units are related to the different methods which are employed to achieve the above three functions by the knee unit. There are several types of knee units and the following is a brief description for the most commonly used knee mechanisms, outlining the principles of operation and classified as stance phase control, swing phase control, or swing and stance phase control according to their basic function.

(i) Stance Phase Control

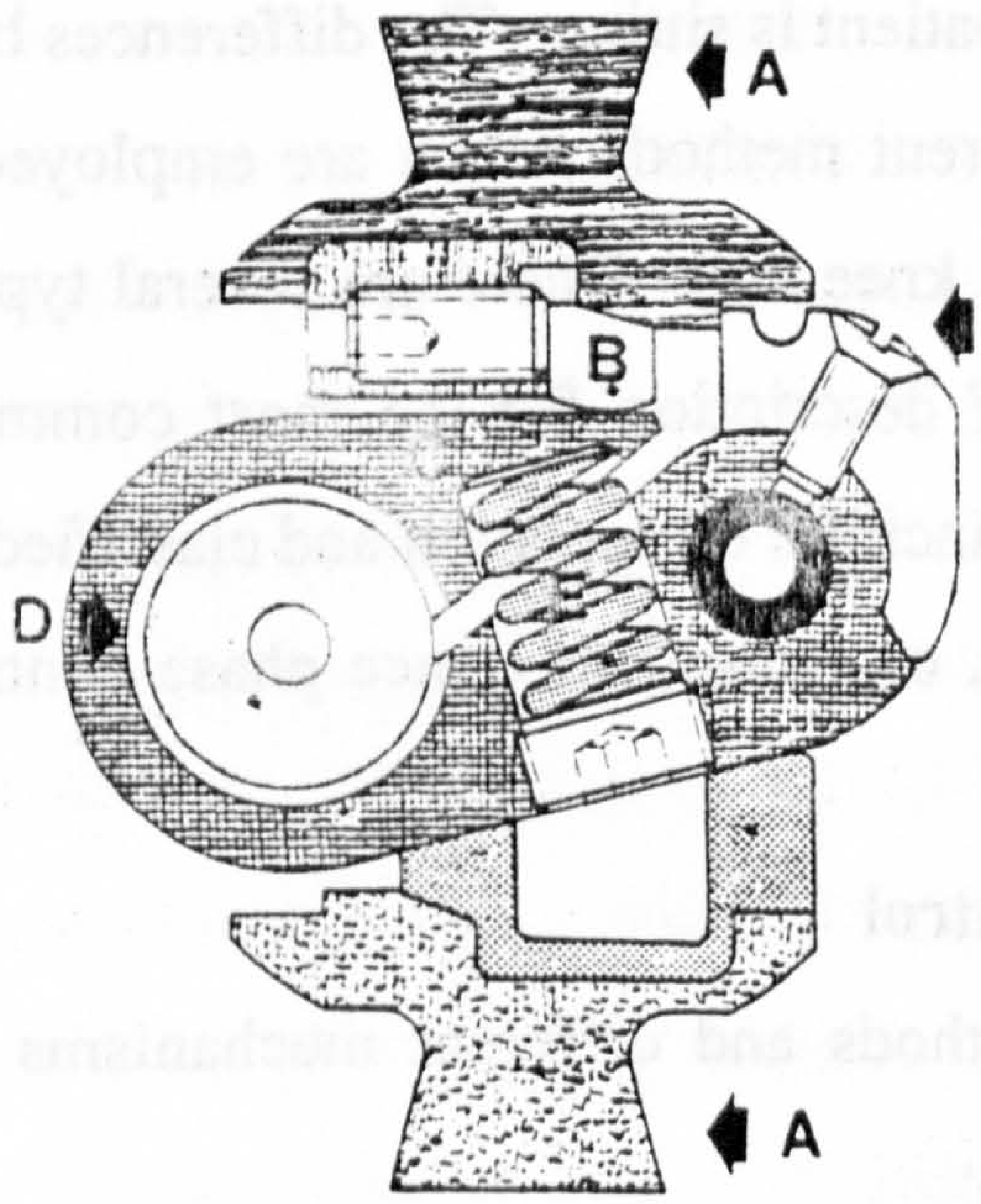
There are several methods and different mechanisms to ensure knee stability during stance phase.

(A) Alignment Method

Knee stability during stance phase is ensured in many above knee prostheses by using the alignment method. This is achieved by locating the knee axis in such a manner that the load line always passes ahead of the knee axis. Thus, the knee will be forced to be in a fully extended position and its stability will be ensured. However, such a method may not be satisfactory if the knee is too stable, because a high hip moment will need to be exerted to initiate knee flexion prior to toe off.

(B) Manual Locking Knee

The manual locking knee mechanism could be any type of knee unit provided with a positive lock which is operated manually (or semi automatic). The lock usually consists of a pin that drops through the knee to lock it in extension. When sitting, the patient manually releases the lock to obtain knee flexion, and when the patient stands, the lock will usually be automatically engaged. This knee unit can be used as a free unit if needed, therefore its stability should be ensured by the alignment method as discussed above. The



Schematic of endoskeletal weight-activated friction (safety) knee. Pylon tube is attached to upper and lower attachment extensions *(A)*. When weight is applied to top of knee during stance phase, top of joint rotates downward (counterclockwise in this case), forcing pressure block *(B)* against brake lever on lower knee block *(C)*. This narrows diameter of that portion of lower knee block surrounding knee bolt *(D)*, resulting in braking effect. Resistance to knee motion is proportional to weight applied. Amount of weight necessary to initiate braking effect is preset with adjustment screw *(E)*. When weight is released from prosthesis, spring releases friction from knee bolt, and knee functions as constant friction unit during swing phase. Constant friction setting is adjustable by tightening another screw *(F)*. (Courtesy Otto Bock Orthopedic Industries, Inc., Minneapolis, Minn.)

Figure 3.16 Weight activated friction knee unit.
(from Mooney and Quigley 1981).

manual locking knee unit is suitable for weak or unstable patients, and is used in temporary prostheses. However, due to the lack of knee flexion during swing phase, energy expenditure will increase and gait deviations often occur during ambulation with a locked knee (Schuch 1992).

(C) Weight Activated Friction Knee Unit

This knee unit is regarded as a safety knee. During swing phase it acts as a constant friction knee, but during stance phase when the load is applied to it, a special housing with high coefficient of friction generates pressure on the knee mechanism and provides a temporary lock to prevent the knee from flexing. The load required to provide the lock depends on the patient's condition and requirements. Figure 3.16 shows and explains the mechanism of this knee unit.

(D) Polycentric Knee Unit

The polycentric knee mechanism can be any device in which the position of the instantaneous centre of the knee changes as the knee flexion angle changes. In this type of knee mechanism, the stability is dictated by the position of the instantaneous centre of the knee which can be designed to ensure stability during early stance phase and to ease knee flexion at late stance phase. There are several types of polycentric knee mechanisms, but the most common one is the four-bar linkage as illustrated in figure 3.17. It should be noted that the instantaneous centre will be high and posterior to the load line at heel strike; it will also be high and anterior to the load line prior to toe off. This type of knee mechanism is recommended for patients with through knee amputation or with a short AK stump, because they are benefiting from the high instantaneous centre of rotation. Patients with weak hip extensors can also take advantage of this mechanism. However, because of its size and weight, this mechanism is not recommended for females. It is worth mentioning that Patil and Chakraborty (1991) have introduced a six-bar linkage polycentric knee mechanism with a pneumatic swing phase control. The

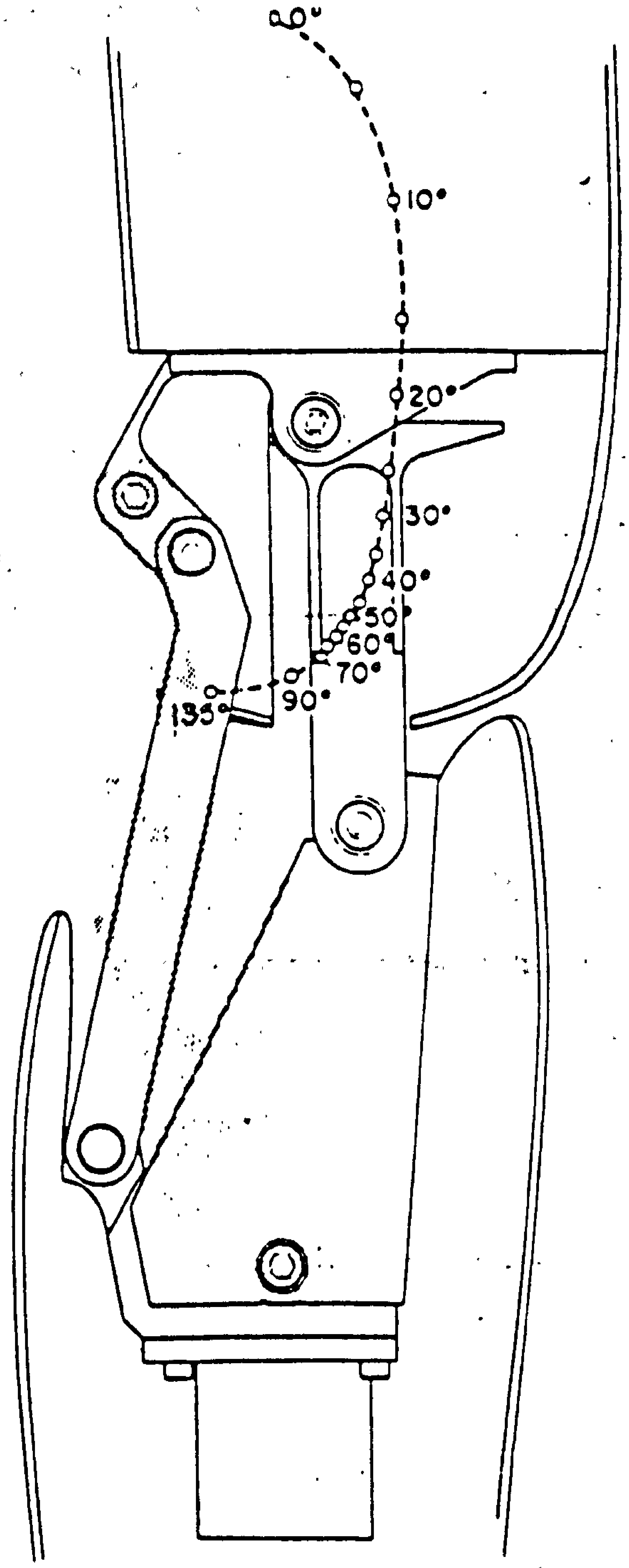


Figure 3.17 Linkage arrangement and path of the instantaneous centre, for the UCBL four-bar linkage polycentric knee. (from Radcliffe 1977)

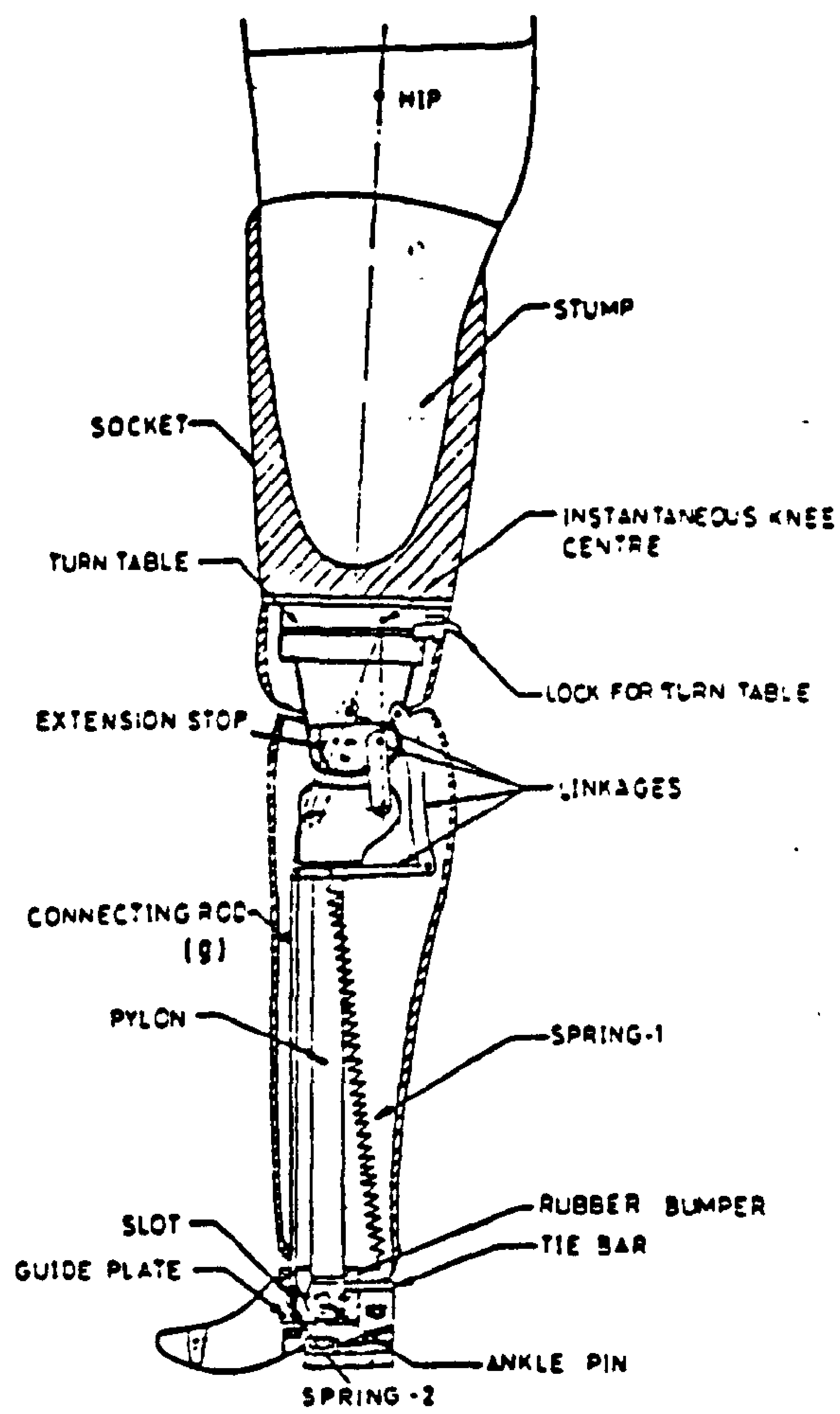
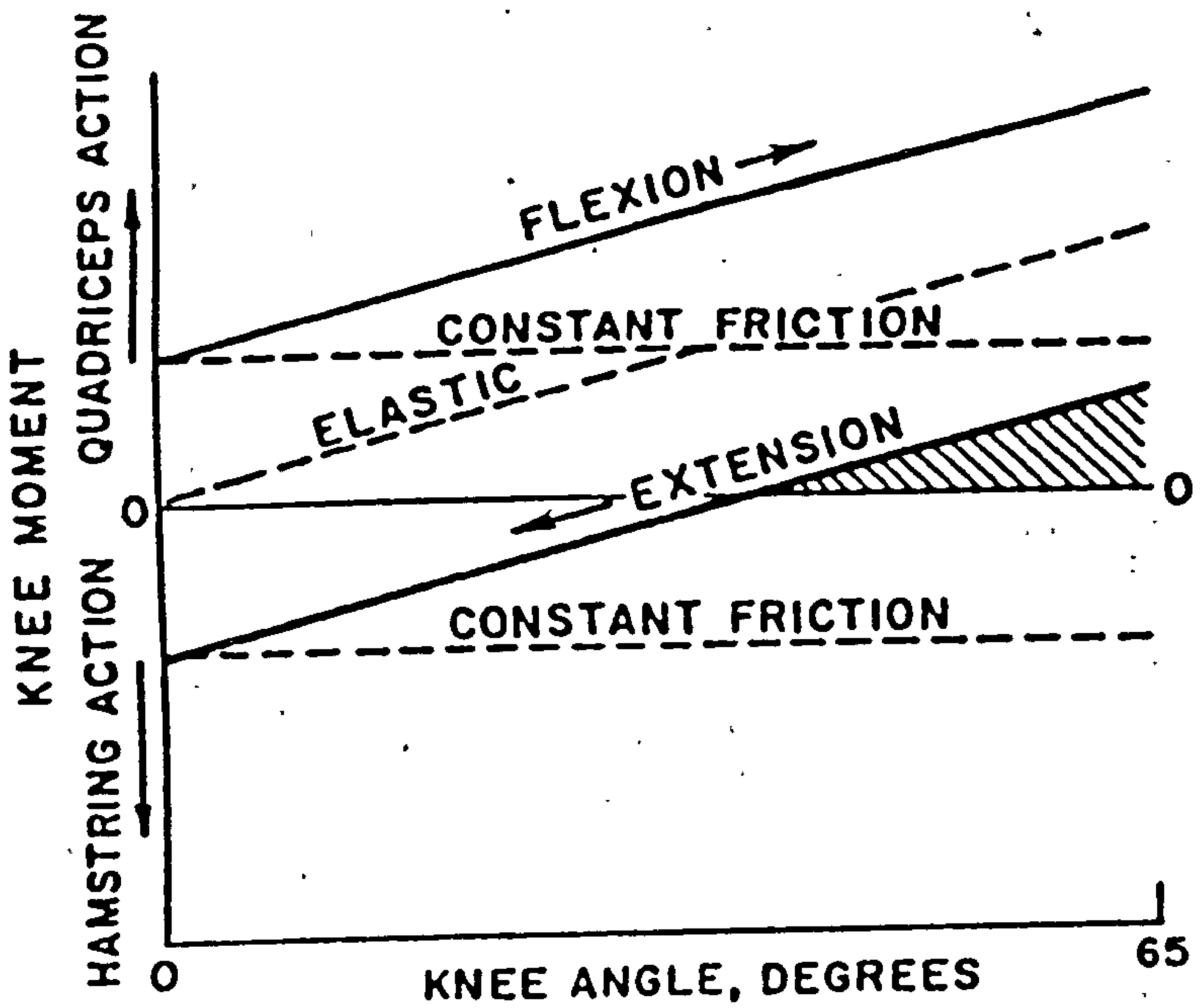


Figure 3.18 Six-bar linkage of polycentric above knee prosthesis. (from Patil and Chakraborty 1991)



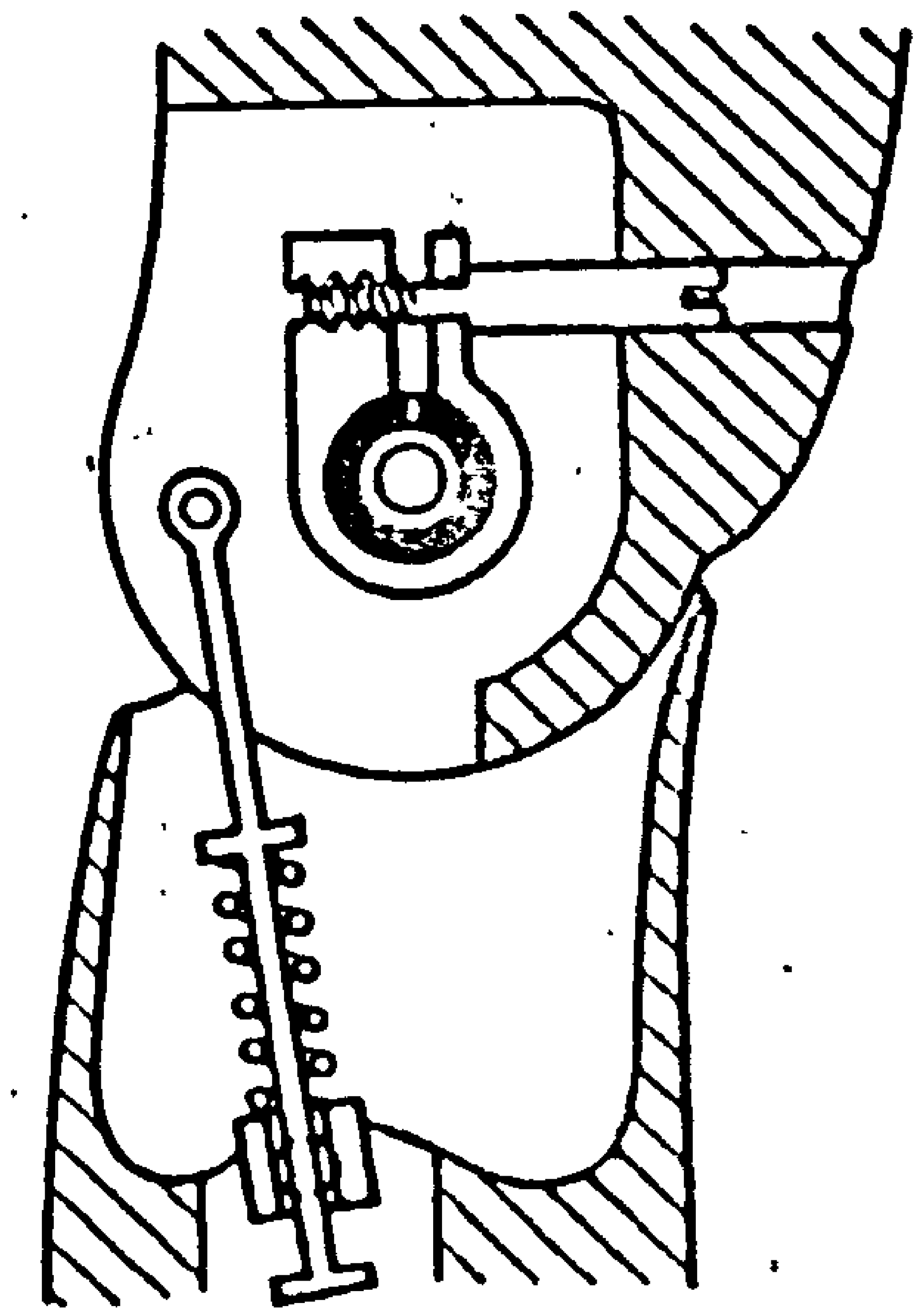


Figure 3.19 Constant friction knee unit and linear elastic extension bias. (from Radcliffe 1969)

additional two bars serve to coordinate knee flexion/extension and ankle dorsiflexion/plantarflexion to provide an above knee prosthesis with the facility for squatting for Afro-Asian amputees. Figure 3.18 shows a schematic diagram of this new prosthesis.

(ii) Swing Phase Control

(A) Constant Friction Knee Unit

The constant friction knee unit consists simply of a single axis hinge which allows the shank to swing in flexion/extension. This is controlled by a screw which applies force to a friction clamp to produce a friction moment acting on the knee bolt (fig 3.19). This friction moment can be adjusted to provide a certain angular speed of the limb during swing phase to suite the patient's cadence. This knee unit is always combined with an additional spring to provide a simultaneous elastic knee extending moment. The control screw can be adjusted to serve only one cadence, so the patient will be restricted to this cadence and will find it difficult to change the gait speed. Despite this disadvantage the unit is widely used because of its simplicity, durability, and its low maintenance cost.

(B) Hydraulic Knee Mechanism

Figure 3.20 illustrates a typical hydraulic knee mechanism which is a mechanism for swing phase control. A piston slides inside a cylinder to force fluid from one side of the cylinder to the other through a series of ports which will be closed one by one providing a variable resistance to knee flexion. This type of mechanism differs from the constant friction type in that the hydraulic system responds to cadence changes so that the resistance increases with increase in speed, thus, the patient is able to walk at different speeds. It also deffers from the constant friction type at heel strike and at toe off, as the fluid has a different amount of damping. This mechanism is rather heavy, therefore, it is used with active patients who can benefit from the cadence response function and can adapt to its mass, such as adult males. Also it is used for

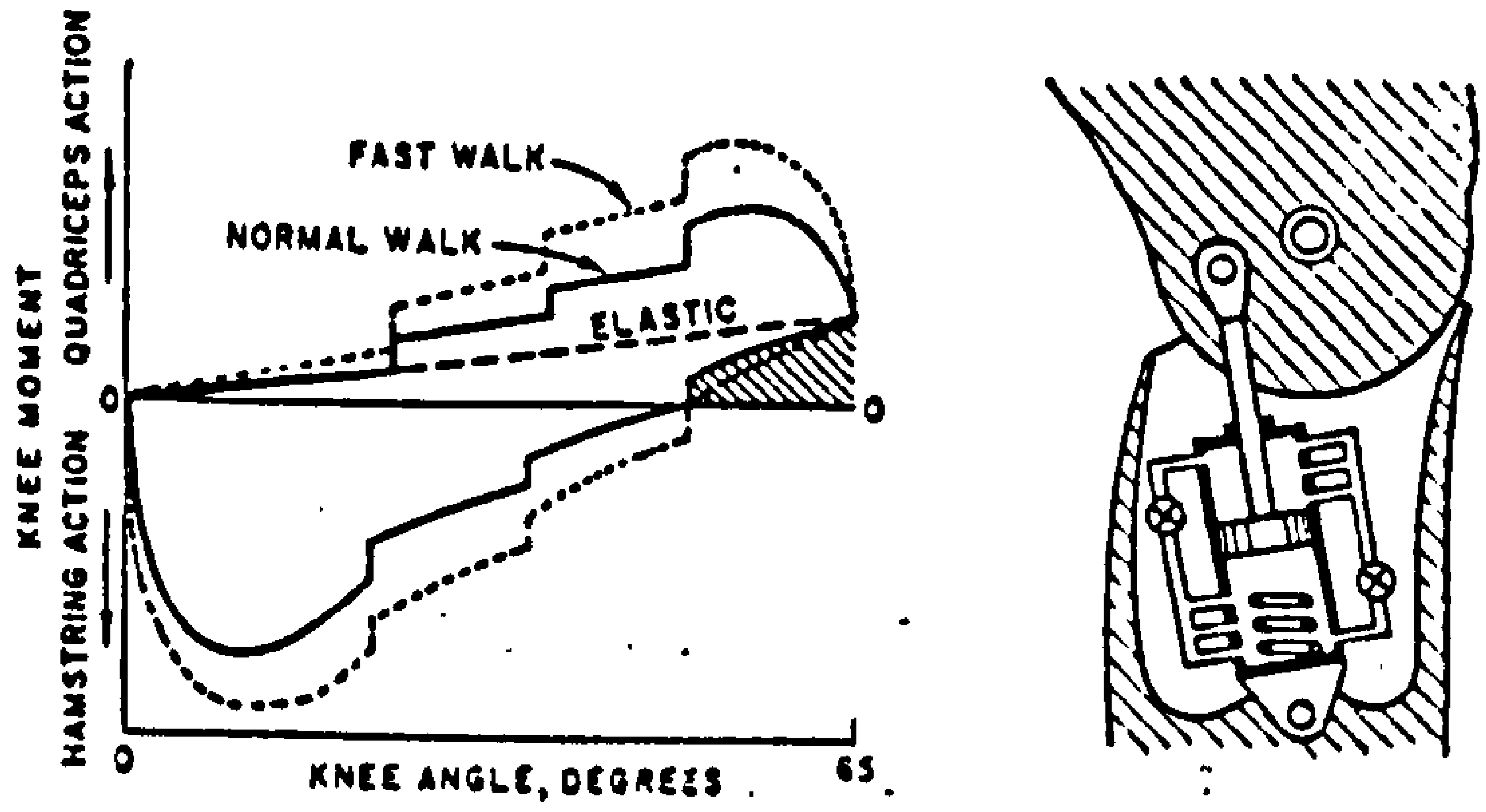


Figure 3.20 Hydraulic swing control with linear elastic extension bais. (from Radcliffe 1969)

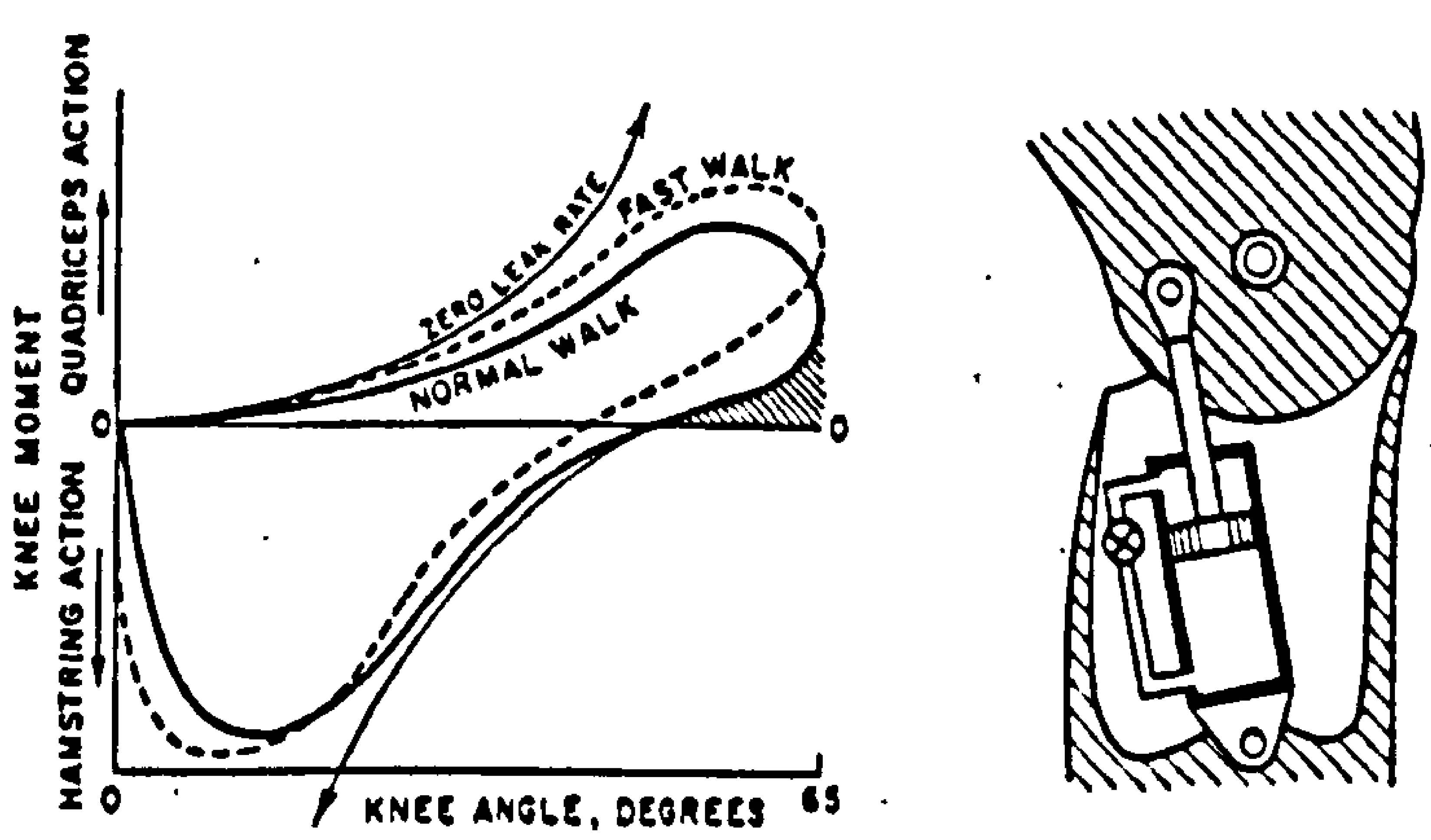


Figure 3.21 Pneumatic swing control. (from Radcliffe 1969)

adult females but not as many as males, since this mechanism has some cosmetic problems.

(C) Pneumatic Knee Unit

This type of knee mechanism is similar to the hydraulic unit in general appearance and it also provides cadence response. The pneumatic unit uses air as a compressible medium. The variable resistance of knee flexion is caused by the increase in pressure against the moving piston, and by the air transferred from one side of the piston to the other. Figure 3.21 shows the pneumatic swing control mechanism and a graphic presentation of the characteristics of its resistance. The pneumatic system has advantages over the hydraulic system in terms of weight and simplicity, but the hydraulic has smoother action, therefore, the pneumatic system is preferred only when low weight is required. The pneumatic unit is also considerably less expensive.

iii Swing and Stance Control

This type of knee mechanism provides adjustable and variable resistance to knee rotation, and it is cadence responsive during swing phase. Stance phase is also controlled by means of fluid resistance. The Mauch S-N-S Hydraulic System is an effective unit of this type. Schuch (1992) stated that the Mauch (S-N-S) system is the most advanced system of hydraulic control and the only system that includes hydraulic stance-phase control. The swing phase is controlled by a multi-orifice mechanism which enables the amputee to be in good control of his prosthesis. This system provides fluid resistance to knee flexion to ensure stability during the stance phase from heel strike to the beginning of the swing phase. For more details about this system the reader should consult Henschke and Mauch (1972).

3.3.1.3 The Foot and Ankle Assembly

A lower limb amputation usually results in losing the foot and ankle joint (except for partial foot amputation). Engineers have made extensive efforts to provide an ankle/foot unit which simulates the function of the normal

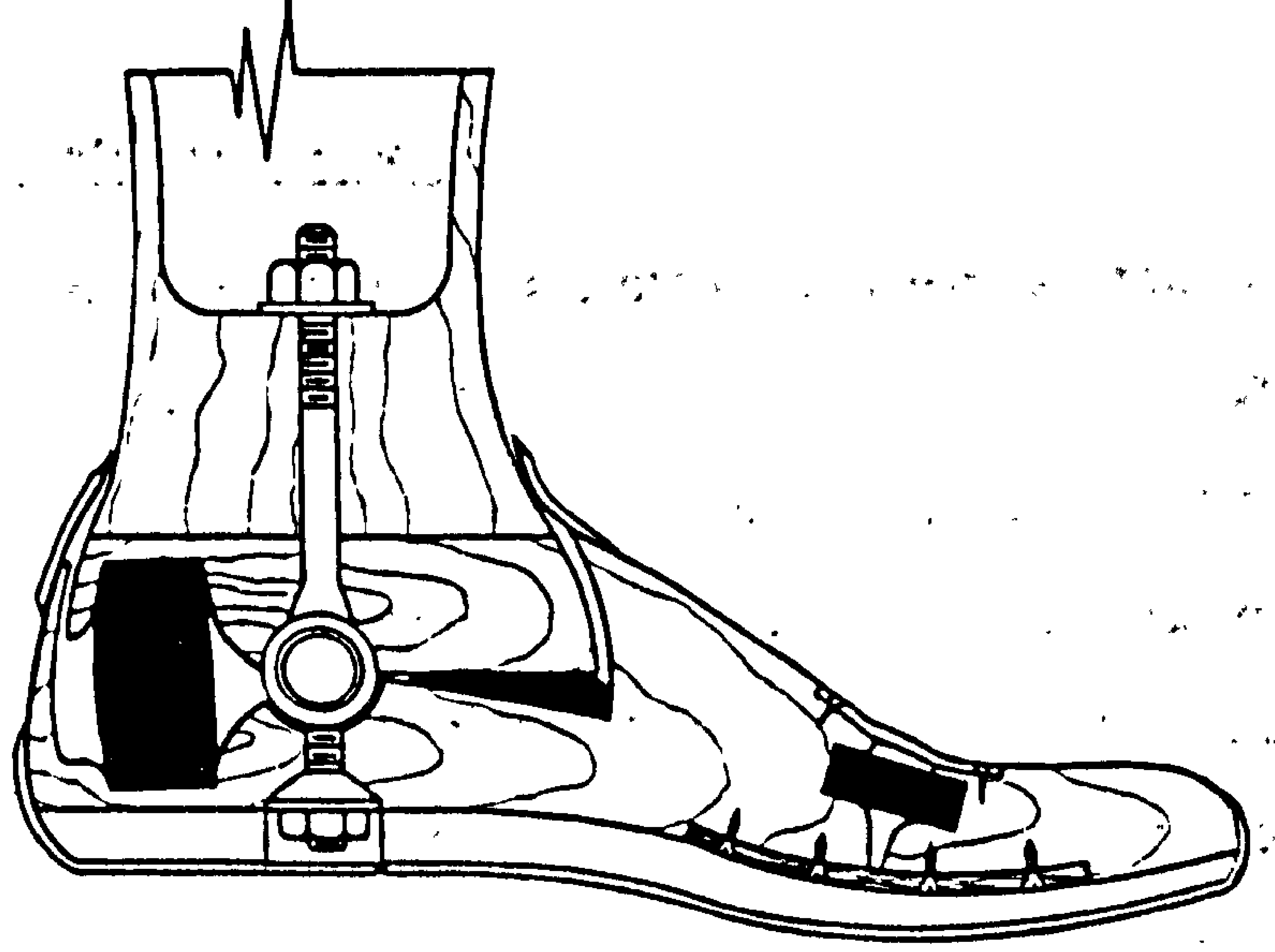


Figure 3.22 The uniaxial ankle/foot assembly. (from Condie 1969)

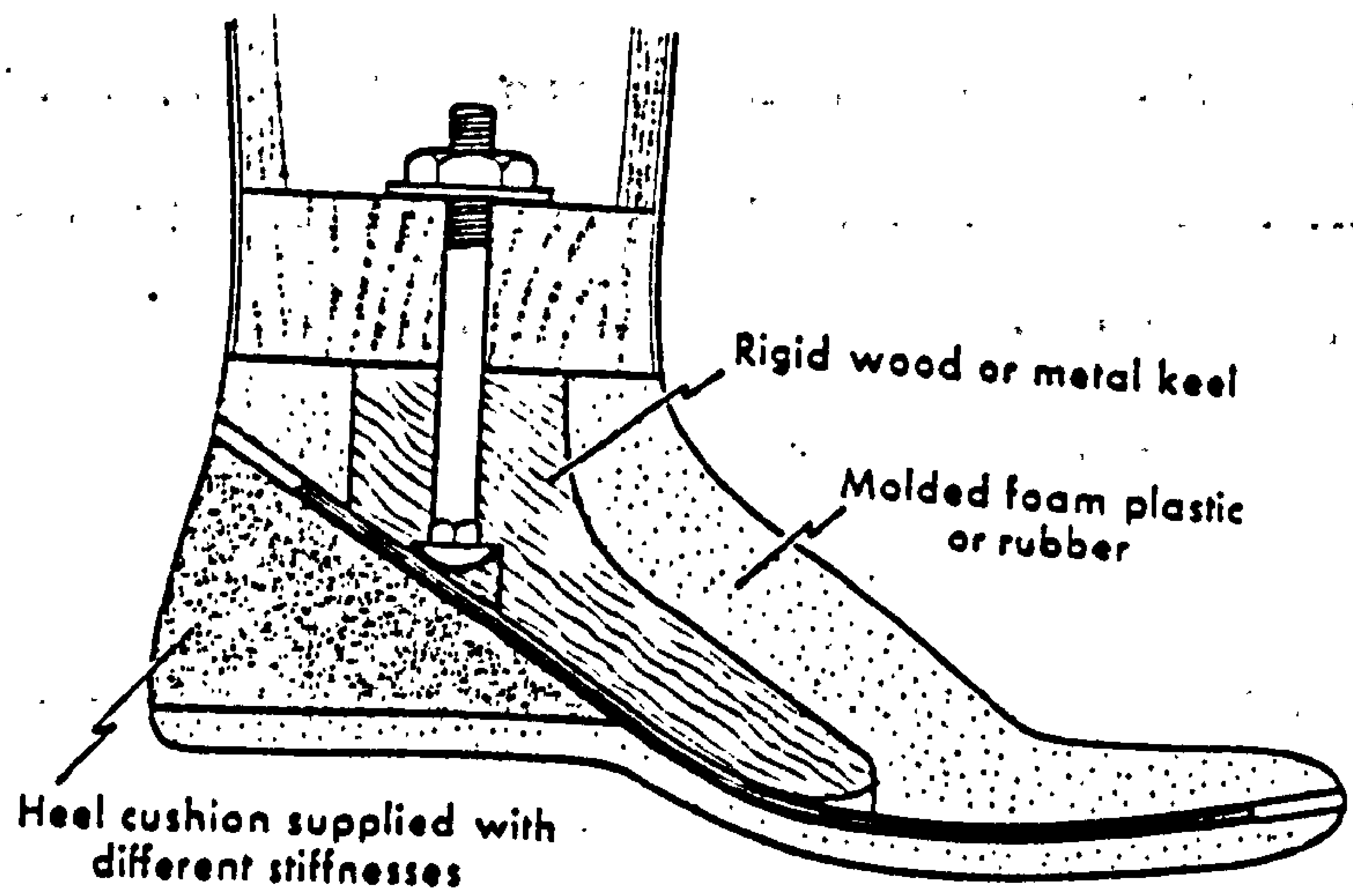


Figure 3.23 A/P cross section of SACH foot.
(from Inman et al 1981)

ankle/foot joint. Many types of feet have been introduced and several are commercially available (Michael 1987, Edelstein 1988 and Esquenazi et al 1989). Most of these feet function very well, but their complexity, high mass, and the need for excessive maintenance have prevented their wide use. Although there are many types of ankle/foot units, the following are the most common and present the principles of the vast majority of the prosthetic ankle/foot assemblies.

(A) The Uniaxial Ankle-Foot Assembly

This is the conventional ankle/foot which is widely used. It consists of a foot and a simple horizontal hinge which provides dorsiflexion and plantarflexion only. The simulation of the normal ankle joint function is obtained by employing a rubber plantarflexion bumper behind the hinge and a stiffer dorsiflexion bumper placed anterior to the hinge to provide restraining and restoring moments (fig 3.22). The stiffness of the posterior bumper is dictated by the need of the patient, i.e. whether to have quick or a slow plantarflexion after heel strike. Another rubber is placed between the foot and the toe sections to provide the toe brake. This foot unit provides fairly acceptable function but does not provide any inversion or eversion and is designed for level walking and for only one height of shoe heel. The foot is quite durable but its mechanical parts can wear quickly and the rubber bumpers need to be changed frequently depending on usage.

(B) The SACH Foot

The SACH (Solid Ankle Cushion Heel) foot which was developed by the University of California in the 1950's, was the first alternative to the uniaxial foot. It has been claimed that this design is more durable and simpler than the uniaxial design, and it provides smoother and more natural ankle/foot action. As seen in figure 3.23 the ankle of this foot has no moving element compared with the hinge in the uniaxial foot. The function of the SACH foot is provided by means of two principal elements. A wedge of cushioning material is

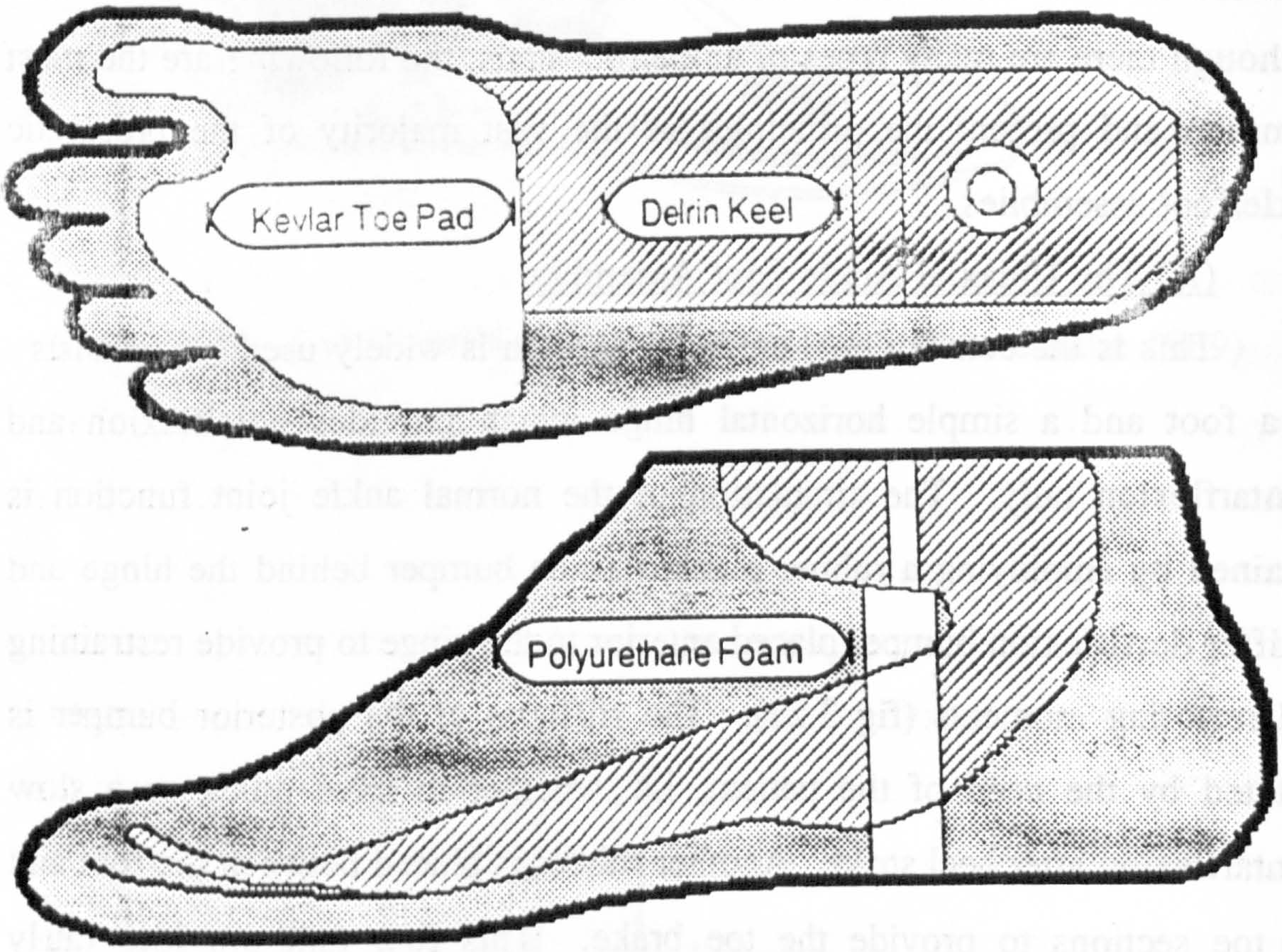


Figure 3.25 The Seattle foot. (from Michael 1987)

employed at the heel to be used as a shock absorber at heel contact and provide plantarflexion which is needed after heel contact to simulate the normal foot. The second principal element is a rigid keel which is located at the core of the foot and controls the transformation of weight from mid-stance until toe off. The keel also controls the dorsiflexion/plantarflexion of the foot by means of its shape during the period between mid-stance and toe off. In this design a toe brake is employed to strengthen the foot and allow a smooth action during toe off; it also springs the toe back to its original position when the load is removed. The cushioning material which surrounds the keel allows medio-lateral motion to some degree. This design has a wide acceptance, especially in the USA, however, some practitioners still prescribe the uniaxial foot, particularly, for above knee amputees.

Goh et al (1984) stated that in the UK about 85 percent of below and above knee amputees are fitted with uniaxial feet, but in the USA about 80 percent are fitted with SACH feet. By using subjective assessment Goh et al found no clear evidence of preference for either SACH or uniaxial feet, and the amputees showed a preference for the foot to which they were accustomed. They also reported that with proper alignment and heel stiffness both types of feet can be made to function in a manner that provides the same kinetic pattern of the whole body.

(C) Multiaxial Foot

The multiaxial foot has been developed to overcome the problems of the SACH and the uniaxial foot. The advantages of the multiaxial foot unit are the provision of foot inversion/eversion, absorbing some of the rotation and the ability of the system to be adjusted to suit different heel heights (Condie 1988). The Greissinger foot ankle assembly is a multi-axis unit. It has a wooden ankle section and a plastic foot. The shank is connected to the foot by means of a U-bolt and yoke type assembly which allows controlled plantarflexion/dorsiflexion, inversion/eversion and transverse motion. The

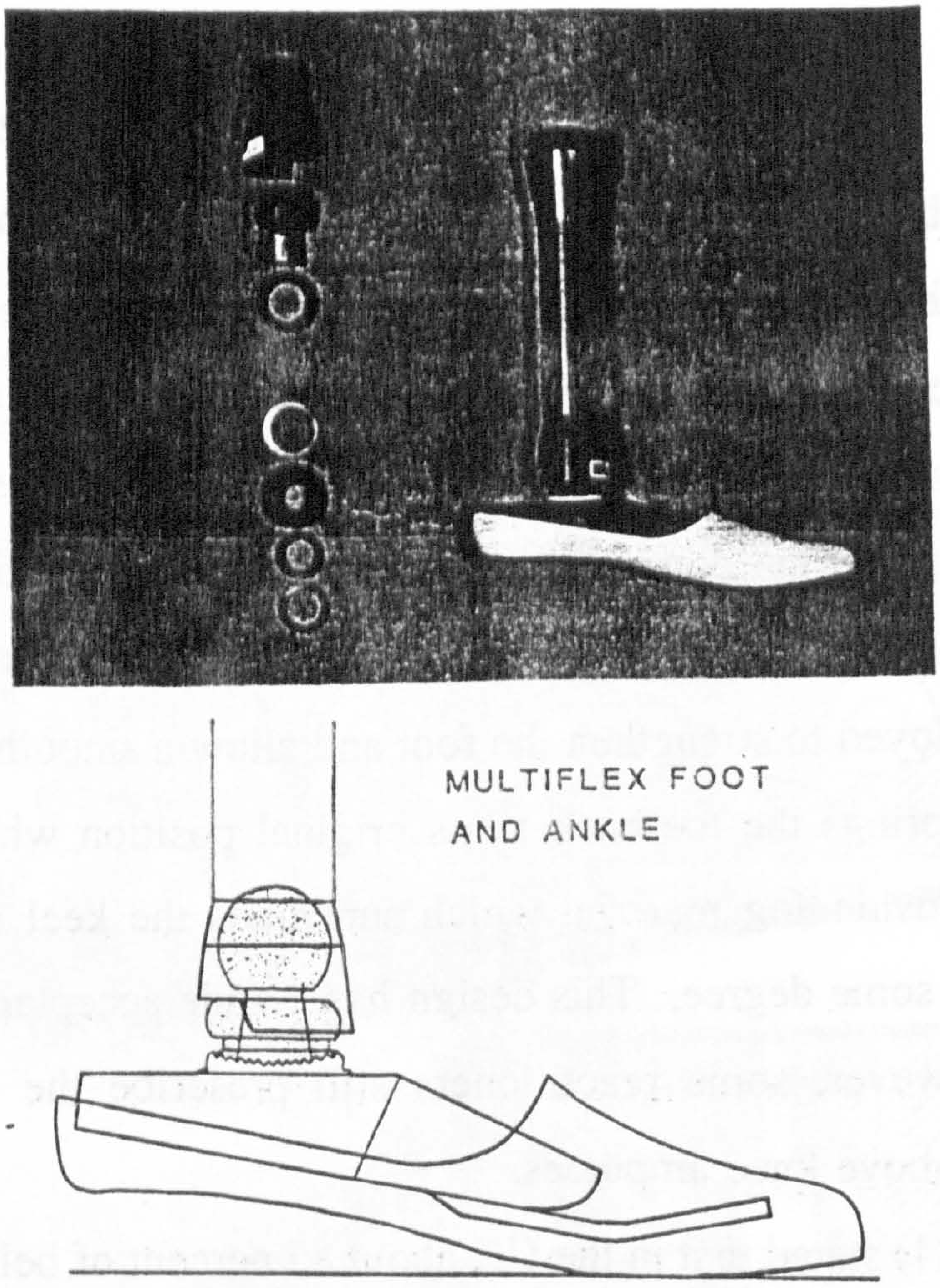


Figure 3.24 Blatchford Multiflex ankle/foot assembly. (from Condie 1988)

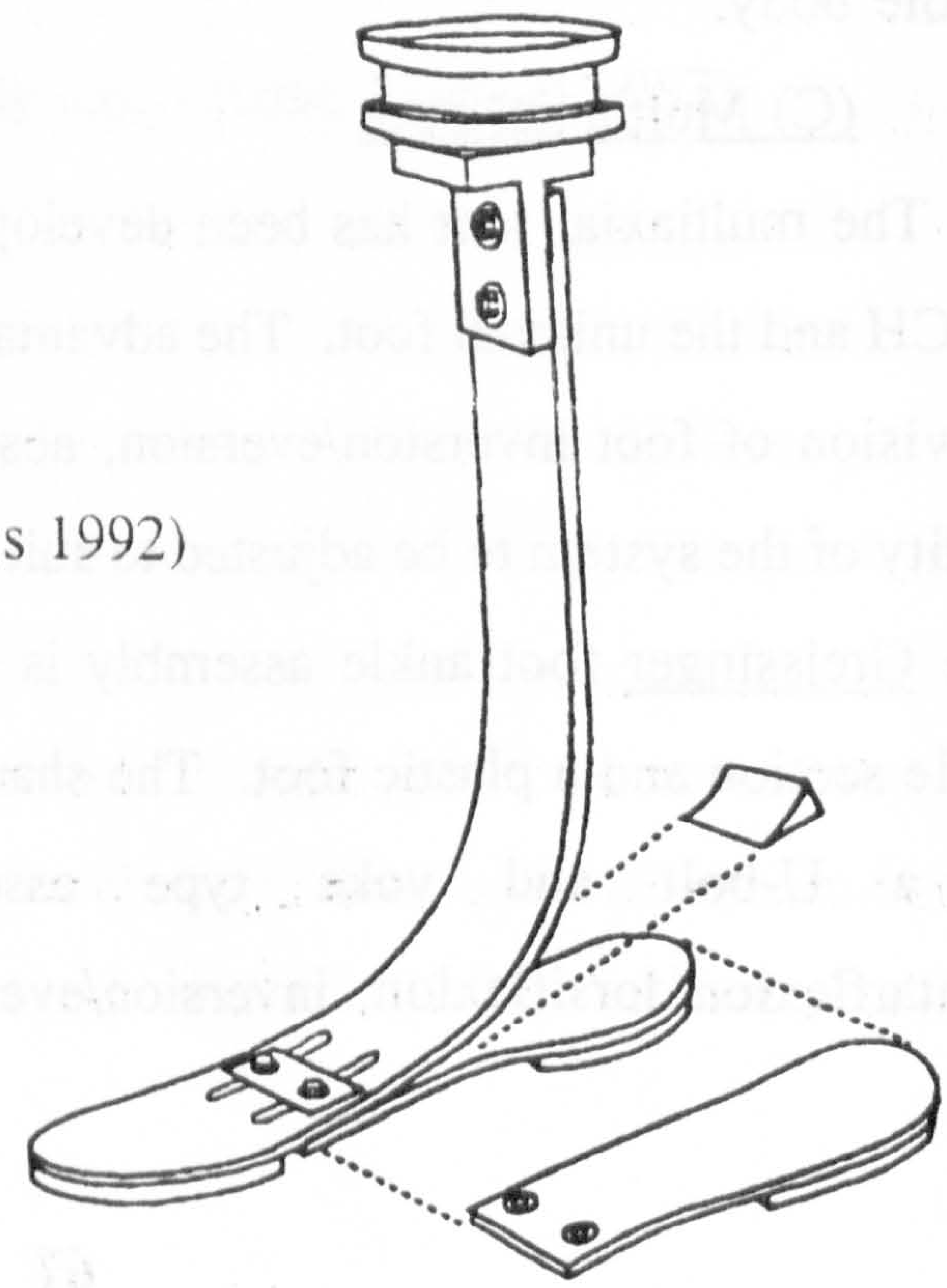


Figure 3.26
Flex-Foot (from Kapp and Cummings 1992)

Blatchford Multiflex ankle foot unit (fig 3.24) is another example of a multiaxial foot. This foot has a carbon-fibre keel and metal ankle assembly which contains the ankle motion controlling unit which has a rubber ball and stem assembly. The resistance to dorsiflexion is provided by means of a 'snubber O ring" which is located below the ball. The foot of this unit is connected to the ankle by a single bolt, and it can be adjusted to cover 40 mm variation in heel height.

In conclusion it seems to be possible to fit any patient with any of the above ankle foot units, provided that a correct selection of the unit components and alignment is achieved. The multiaxial foot is still to receive more attention from designers and its reliability and durability are yet to be evaluated.

(D) The Seattle Foot

Recently new types of prosthetic ankle/foot units have been introduced to serve active patients and their sporting activities, particularly running. These type of units are called energy storing ankle/foot units, however, it is believed that most of these feet are of the dynamic elastic response (DER) type, and are not able to store energy. One of the most common units of this type is the Seattle foot (fig 3.25) which consists of a monolithic keel made of Delrin embedded in a foam which has the shape of a normal foot. Michael (1987) stated that the keel acts as a spring as it stores energy during initial loading of the foot and releases it during push off. He also added that, although, this type of foot is relatively heavy, it is found to be favoured by patients for the "lively" step permitted.

The Seattle foot was shown to be successful in 50 active patients (Burgess 1988).

(E) The Flex-foot

The Flex Foot is another kind of dynamic elastic response foot. A flat carbon graphite shaped strip in the form of a long spring is extended into the prosthetic shank (fig 3.26). The carbon graphite at the foot and at the shank

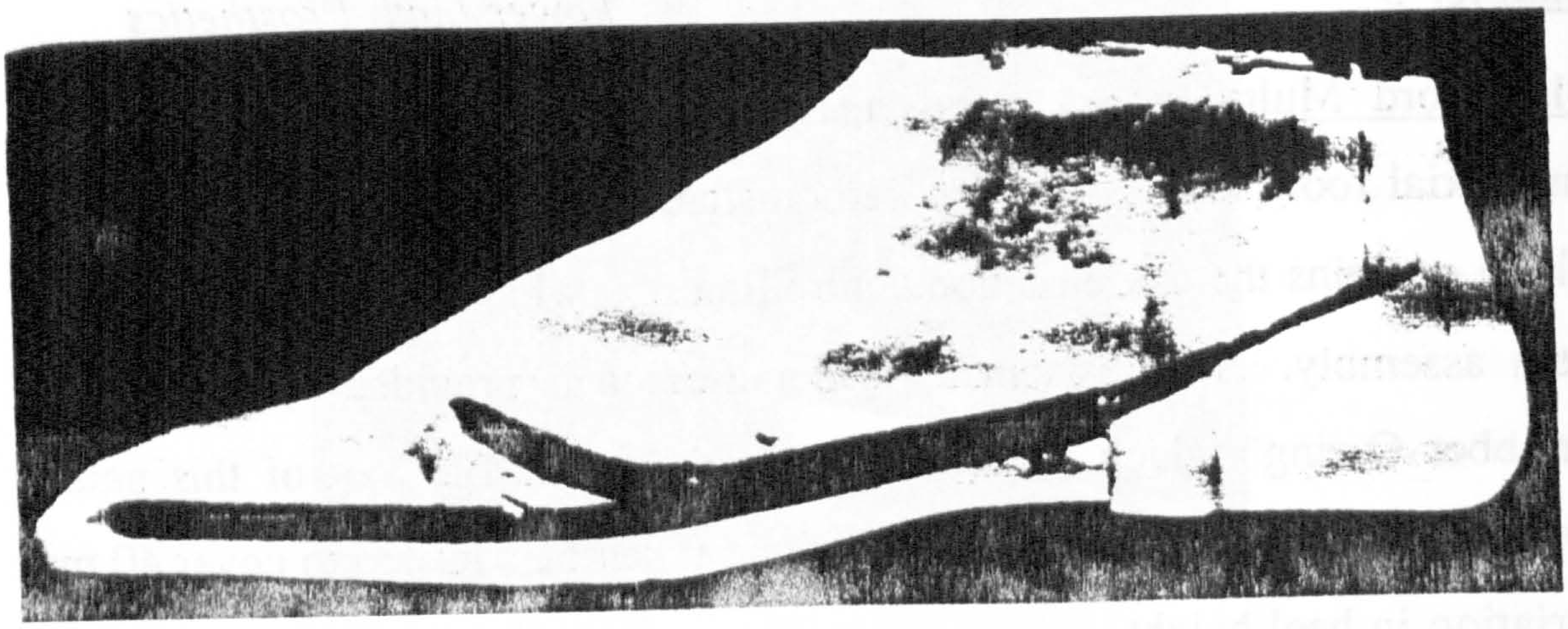


Figure 3.27 Carbon Copy II foot. (from Michael 1987)

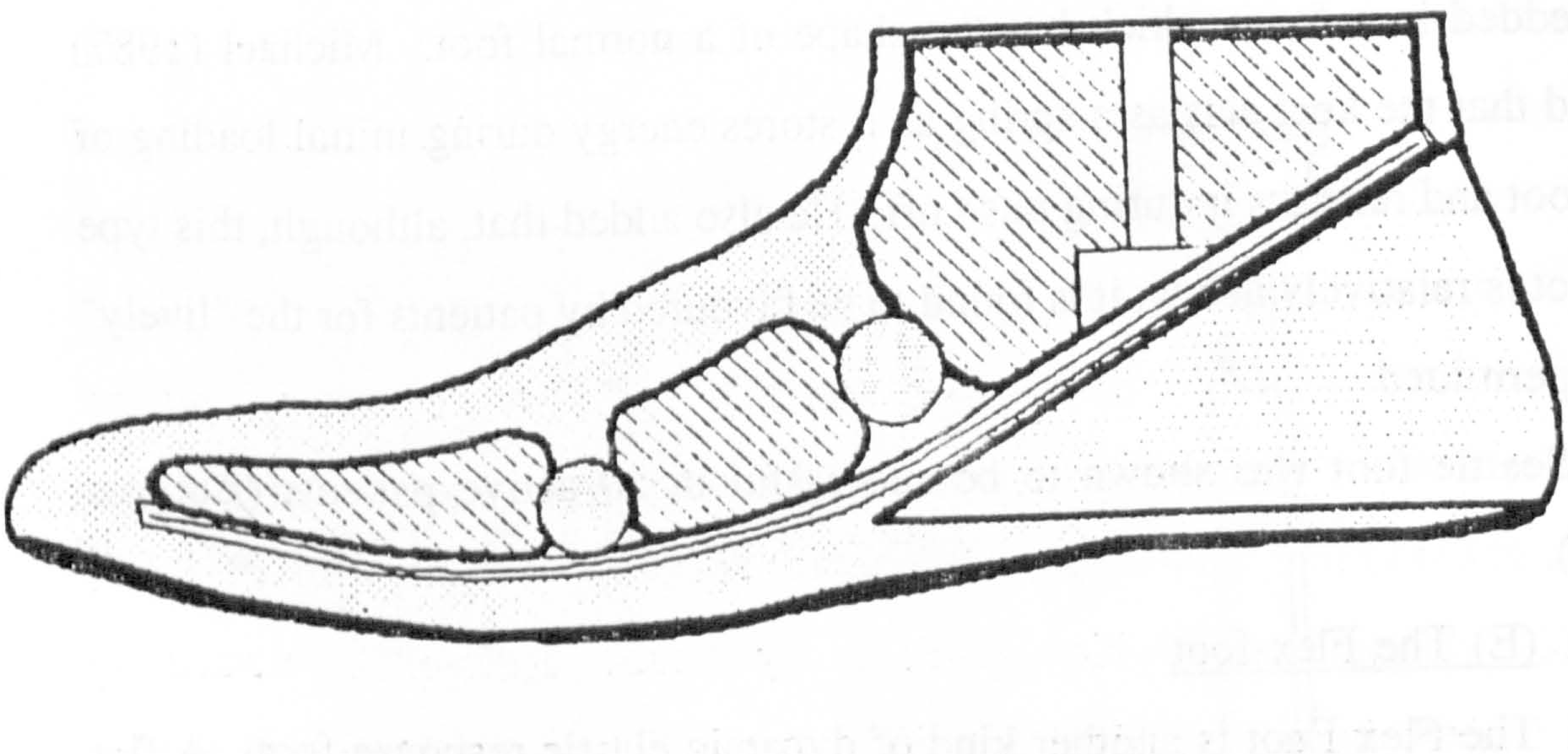


Figure 3.28 The STEN foot. (from Michael 1987)

have the ability to store energy during early stance and release it at toe-off (Esquenazi et al 1989). Michael (1987) stated that the Flex Foot stores energy throughout the entire length distal to the socket rather than just within a four inch (9.8 cm) keel. The fabrication of the Flex Foot requires several weeks. However, the Modular Flex-Foot, which was introduced in early 1987, has a reduced fabrication time. This foot is recommended to the very active patient and has mainly been used on BK amputees.

Wagner et al (1987) conducted comparative tests between the SACH foot and the Flex Foot using six moderately active BK amputees to find out whether they benefited from energy storing feet (as they called it) or if these feet should only be used on active patients. The results showed that the temporal parameters of gait did not differ between the SACH foot and the Flex-Foot, but the symmetry did improve with the Flex-Foot. The conclusion arrived at was that moderately active BK amputees are getting biomechanical benefits from the energy storage prosthesis, and the use of these prostheses should not be limited to active amputees. Edelstein (1988) reported that the Flex-Foot provides maximum assistance to the wearer because it stores energy through a long keel. However, she pointed out that the foot is expensive, needs a long time for the application of the cosmetic cover, requires special equipment and techniques for alignment, and does not provide much transverse or frontal plane motion.

(F) The Carbon Copy II

The Carbon Copy II foot is also a dynamic elastic response (reported by Michael 1987 to be energy storing) foot. It consists of a solid-ankle posterior bolt block made of reinforced nylon/kevlar and combined with two flexible anterior deflection plates which permit energy storage dependent on the activity of the patient during use (fig 3.27). This foot is lighter than the conventional SACH foot and is recommended for use by BK and AK amputees (Esquenazi et al 1989).

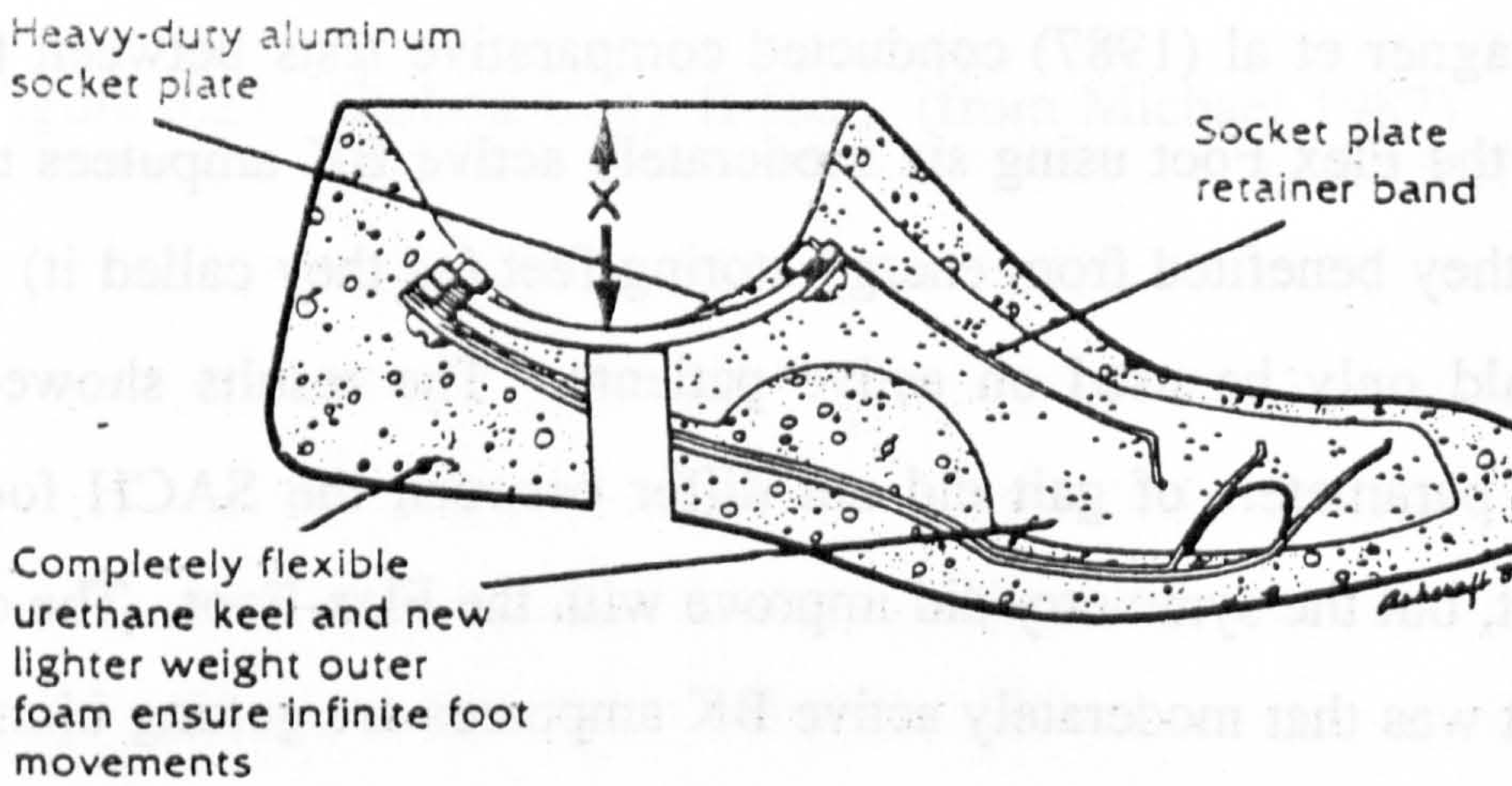


Figure 3.29 The SAFE foot. (from Pritham 1988)

(G) The STEN Foot (STored ENergy)

Figure 3.28 shows the STEN foot. It has a dual articulated keel which permits a smooth and gradual roll-over. This device has been introduced as an energy storage foot, but Michael (1987) stated that "Although the name stands for STored ENergy foot, it is our clinical impression that it does not accomplish this goal as effectively as the previous designs" (by the previous design, the author means the Seattle and the Carbon Copy II). This foot is not recommended for the AK amputees because of the soft forefoot which may cause knee instability.

(H) SAFE Foot

The SAFE foot (Stationary Ankle Flexible Endoskeleton) has been advertised as the original energy storing foot (fig 3.29). However, it is believed that its flexible keel serves primarily to dissipate energy as it accommodates to irregular surfaces, and it can be considered as a solid ankle with a foot which can deform three-dimensionally and is an alternative to the well known Greissinger foot (Michael 1987).

(I) Quantum Modular Foot

The Quantum Foot was introduced in September 1988 by The Hanger company in England, and is perhaps, the newest dynamic response foot. The foot consists of a long sole spring, a secondary spring, and an ankle base that are inserted into a hollow cosmetic foot module. It is similar to the Carbon Copy II in terms of weight and price.

Finally, a conclusion can be drawn that each type of foot has merit and can be successfully used if a correct selection is achieved. Table 3.3 shows a summary for the indications and contraindications for most of the current feet designs.

3.3.2 Modular Assembly Prosthesis (MAP)

The need for providing the patient with a comfortable, practical, and good prosthesis using minimum skills in the shortest time created the concept

Table 3.3 Clinical comparison of prosthetic feet. (from Michael 1987)

Component	Cost/Weight	Indication	Advantages	Disadvantages	Typical Sports Applications	Ankle Mechanism	Permits Forefoot Pro/Supination	Permits Hindfoot in/ Eversion/Rotation
Sach	Low/medium	General use	Reliable, inexpensive, accommodates numerous shoe styles	Fairly rigid, limited range of motion	Sprinting	Fixed	No	No
Single axis	Mod/heavy	To enhance knee stability	Adds stability to prosthetic knees	Slightly increased cost, weight, maintenance	Limited	Articulated	No	No
Greissinger	Mod/heavy	Accommodate uneven surfaces, absorb rotary torques	Multi-directional motion	Slightly increased cost, weight, maintenance less ML stability	General, to absorb stresses	Articulated	Yes	Yes
Safe	Mod/heavy	Accommodate uneven surfaces, absorb rotary torques, smooth roll-over	Multi-directional motion, moisture & grit resistant	Slightly increased cost, weight, less ML stability	General, to absorb stresses	←Fixed→	Yes	Yes
Sten-Foot	Mod/medium	Smooth roll over	Moderate cost & weight; accommodates numerous shoe styles; ML stability similar to Sach	Slightly increased cost, weight	General, for smoother roll-over	←Fixed→	Yes	No
Seattle Foot™	High/heavy	Jogging, general sports, "conserve energy"	"Energy storing" smooth roll-over	Increased cost, weight, difficult to fit in shoes	General, jogging	←Fixed→	No	No
Carbon Copy II	High/light	Jogging, general sports, "conserve energy"	"Energy storing", smooth roll-over, very stable ML, highest solid ankle foot	Incredsed cost, difficult to fit in shoes	General, jogging	←Fixed→	No	No
Flex-Foot™	Very high/ very light	Running, jumping, vigorous sports, "conserve energy"	Most "energy storing", most stable ML, lowest inertia, wide range of applications	High cost, complex fabrication & alignment, not feasible for very long residual limbs	Vigorous sports jumping; running	Flexible	No	No

of the modular system. Most of the prosthetic components can be standardised and made to be " off the shelf " products ready to be assembled when needed. The socket is the only part of the prosthesis which should not be made beforehand and should be fabricated for each individual patient. Foort (1979b) defined the modular system as " A system of elemental parts which can be combined in a variety of ways to arrive at the functional entity desired from among a variety of functional options which the system allows". Any modular system design should be efficient in operation, reliable, easy to manufacture and easily put right if it goes wrong. In addition, the system should be inexpensive, adjustable and able to be built very quickly. By 1975 a number of modular systems were introduced and used for both AK and BK amputees.

Solomonidis (1975) carried out a series of tests to evaluate the available BK modular systems. The evaluation considered four parameters: 1- The alignment of the system 2- The time needed to build the prosthesis 3- The mass properties and 4- The comments of the evaluation team. Six BK systems were evaluated; they were the Biomechanical Research And Development Unit (BRADU), Blatchford MAP, Otto Bock, Hanger, VA and Winnipeg. Three of these systems were found to be unacceptable: Winnipeg demonstrated inadequate strength in service, Hanger, had insufficient alignment adjustability; and the VA system could not be satisfactorily locked in service. Following the same procedure Solomonidis (1980) evaluated four AK modular systems; the Otto Bock, the US manufacturing company (USMC), the Blatchford MAP, and the Hosmer/Dorrance above knee modular prosthetic system. Figure 3.30 shows the Blatchford and the Otto Bock modular systems. All the systems were found to be clinically usable, but significant differences in the function and physical properties were detected, and it was found that the Blatchford MAP and Bock systems were the preferred systems for the limb fitting service in Britain.

Modular prosthetic systems are still under development and the latest

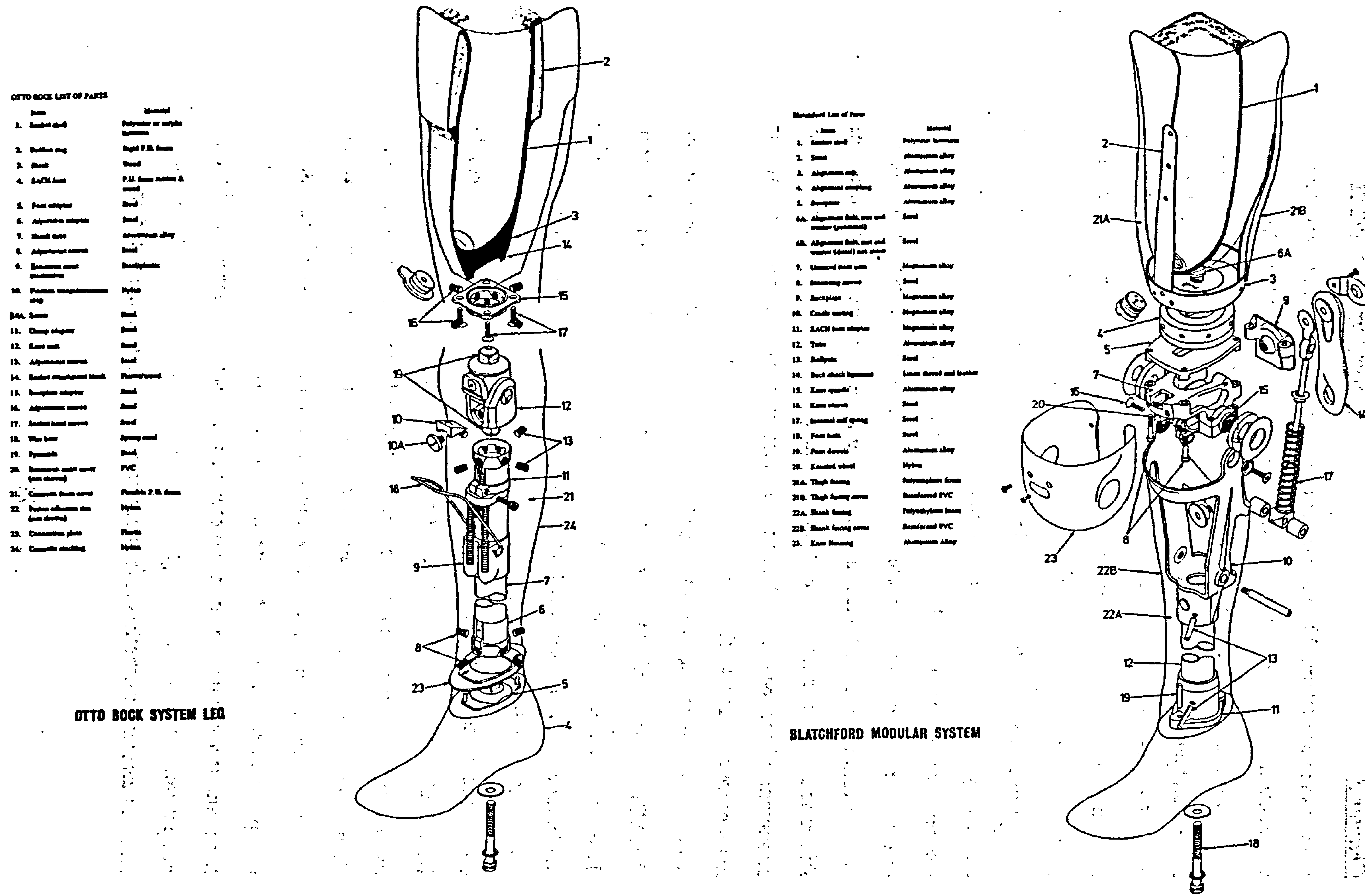


Figure 3.30 Two AK Modular systems. (from Solomonidis 1980)

versions have not been subjected to such an evaluation program. Nevertheless, the modular systems have accelerated the process of building the prosthesis and reduced the time needed to initiate the prosthetic management of the patient.

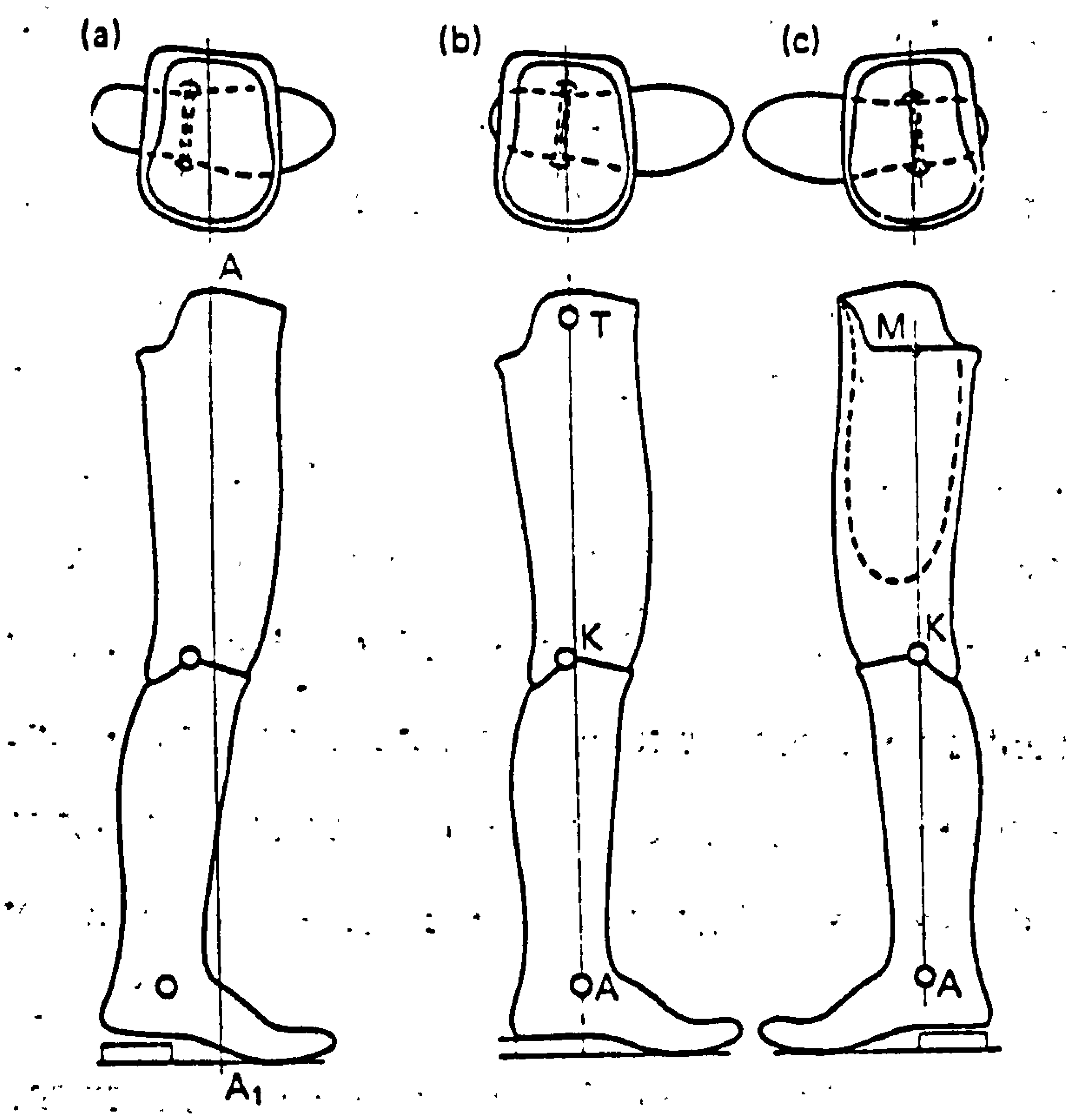
3.3.3 Alignment of Lower Limb Prostheses

This section is mainly concerned with above knee prostheses. However, since there is a dearth of data regarding the effects of alignment changes on amputee performance of AK prosthesis, BK prostheses will also be discussed.

Alignment of above knee prosthesis is defined as the relative position and orientation of the socket, knee and foot. It plays an important role on the forces generated between the stump and the prosthesis, and it has a major effect on the stability of the knee. Failure to achieve satisfactory alignment will be reflected in the patient's gait. Large forces will be generated on the stump to cause the amputee discomfort, pain, and instability during the stance and swing phases of the gait. An unsatisfactory alignment can also result in high energy expenditure.

Radcliffe (1954, 1955, 1969 and 1977) and Foort (1979c) have described the different sets of alignment to cover all the cases of above knee amputees. The alignment of an above knee prosthesis is achieved in three stages: bench alignment, static alignment and dynamic alignment.

Bench alignment is the configuration of the prosthesis when all its components are first put together. It should be set in a certain configuration which ensures knee stability and provides the prosthesis with the optimum dynamic alignment by requiring the minimum further adjustment. Static alignment is done when the patient is standing with his prosthesis, in order to adjust the height of the prosthesis and to ensure stability. The height should be approximately equal to the height of the sound side. If the prosthesis is long, the pelvis will drop on the normal side, and if the prosthesis is short, the patient will lean over to the prosthetic side. Dynamic alignment is the last stage in aligning a prosthesis; it is performed during walking trials and is the



Bench alignment systems. (a) German system, (b) TKA reference, (c) MKA reference.

Figure 3.31 Bench alignment systems. (from Radcliffe 1977) adjustment needed to the alignment to achieve a satisfactory gait.

There are several systems of bench alignment (fig 3.31) to provide knee stability. According to Radcliffe (1977) the system used in Germany considers a plumb line which passes from the centre of the socket brim to the bisector of the length of the foot, with the knee set behind this line. An air space under the heel of the foot is used to improve the knee stability at heel contact, and to ease the hip rolling action over the foot. Figure 3.31 shows the German system in that position when the amputee would begin to roll over the ball of the foot in walking. In the USA it is recommended that a similar reference line be used but it has the trochanter as the upper reference point and locates the ankle joint directly under the trochanter. The knee joint is located on or behind this reference line, TKA line, when the heel is in contact with the floor. This system necessitates a convenient way to check bench alignment prior to walking trials with adjustable devices without the need for limb assembly fixtures (Radcliffe 1977), therefore, it has been modified to a new reference line called MKA in which the knee is located on the line which is between the bisector of the medial wall of the socket and the ankle joint. In this line the knee is located on the medial aspect of the prosthesis rather than the lateral side as is the case with the TKA line. In the case of MKA, the air space was also used during bench alignment.

There are general principles for achieving a satisfactory dynamic alignment. However, these principles can only be used as a guide and they should always be modified to accommodate the individual differences depending on the physical condition of the patient and on the length of his/her stump. Figure 3.32 shows the general principles for setting the alignment according to the length of the stump.

For short stumps, the socket is slightly abducted; this is to improve the pelvis medio-lateral stability, as the hip adductors, which were weakened by the amputation, will be in an advantageous position, and the medial stabilising

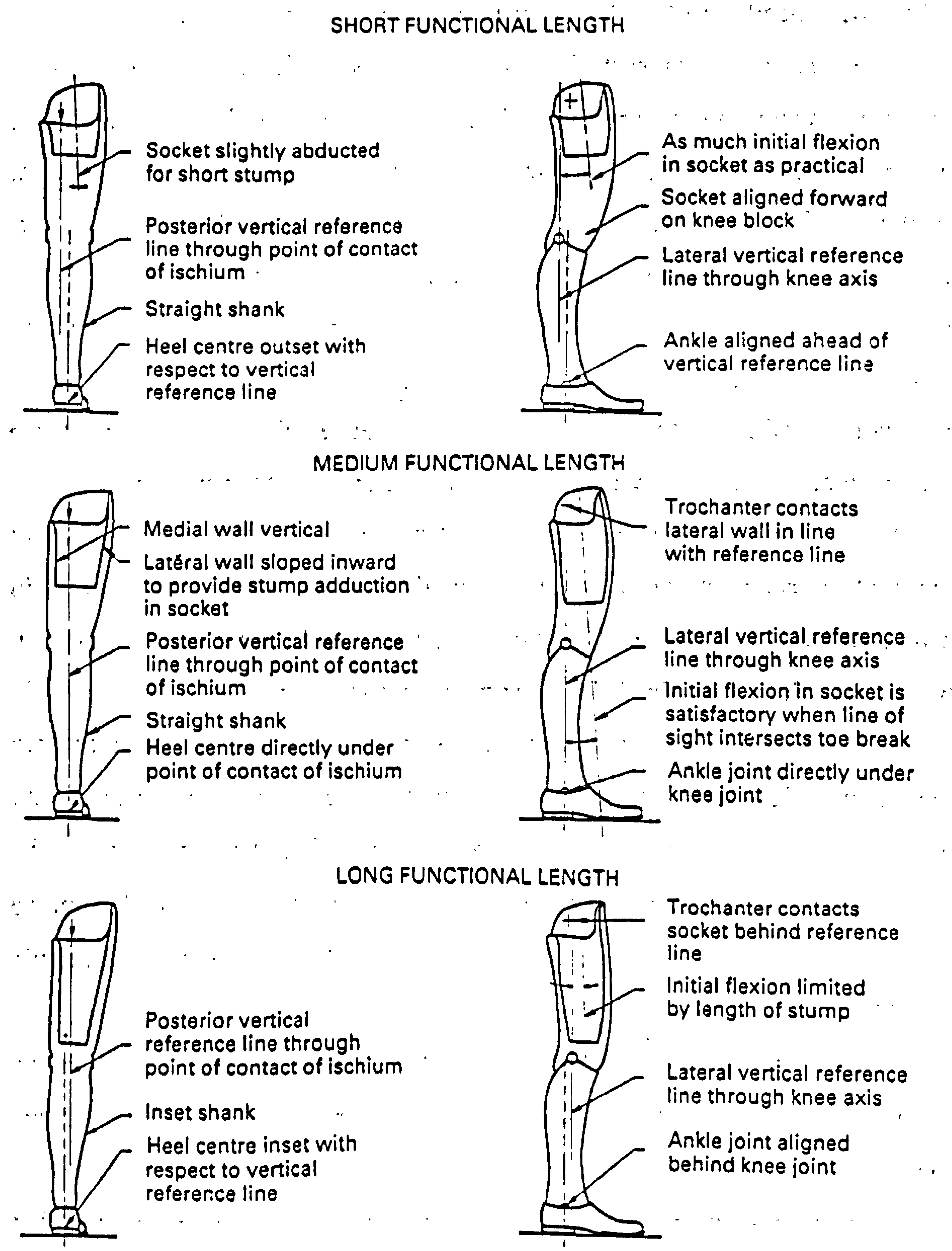


Figure 3.32 Alignment variations to accommodate stumps of various functional lengths. (from Radcliffe 1955).

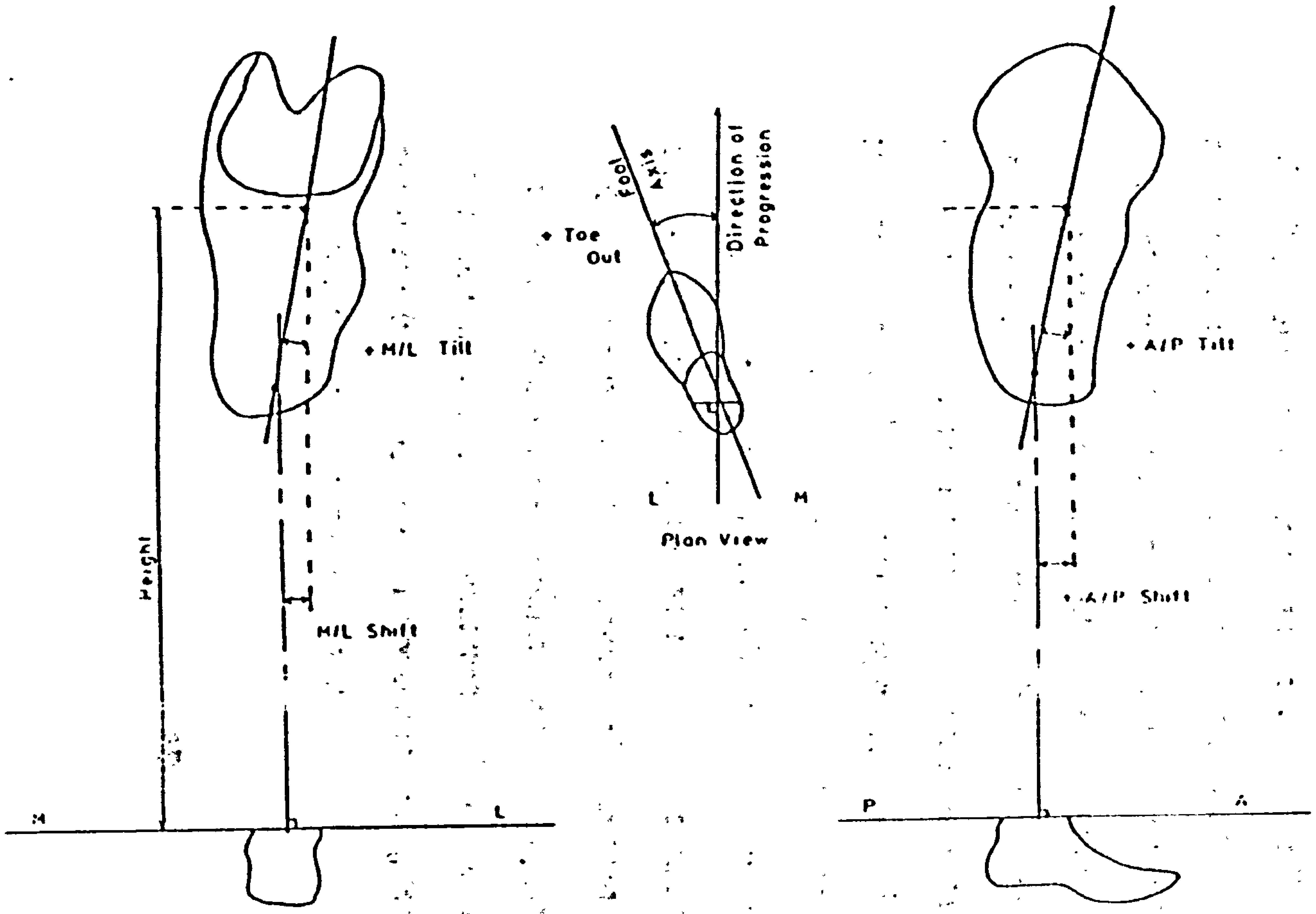
force will be increased. Also, abducting the socket will shorten the hip abductors since they are still powerful and will therefore, generate high pressure at the distal lateral end of the femur. This high pressure will be reduced as the hip abductors are shortened, however, it will be increased with abducting the socket as the medio-lateral force will be increased. Therefore, the socket should not be largely abducted. In fact, with the new method of alignment, the socket should be adducted as much as possible regardless of the length of the stump in order to provide a normal gait. For a short stump, the foot is set slightly laterally to help improve the medio-lateral stability as the medial force on the foot will be increased. In the sagittal plane the socket is set in as much initial flexion as practicable (Radcliffe 1955). This will make the hip extensors which are affected by the amputation, more effective in providing knee stability. Knee stability is also ensured during the stance phase, by locating the knee joint behind the trochanter-ankle line.

For stumps with medium length, the socket is slightly adducted to keep the stump adducted and to make the hip abductors more effective in providing pelvic stability. The foot may have to be located laterally to help medio-lateral stability especially for patients who are physically weak. In the sagittal plane the socket is flexed by approximately 5 degrees to give an advantage to the hip extensors in providing knee stability at heel strike and during early stance phase. Because of this and the medium length of the stump, the knee can be located on the trochanter-ankle line and there is no need to locate the knee behind this line.

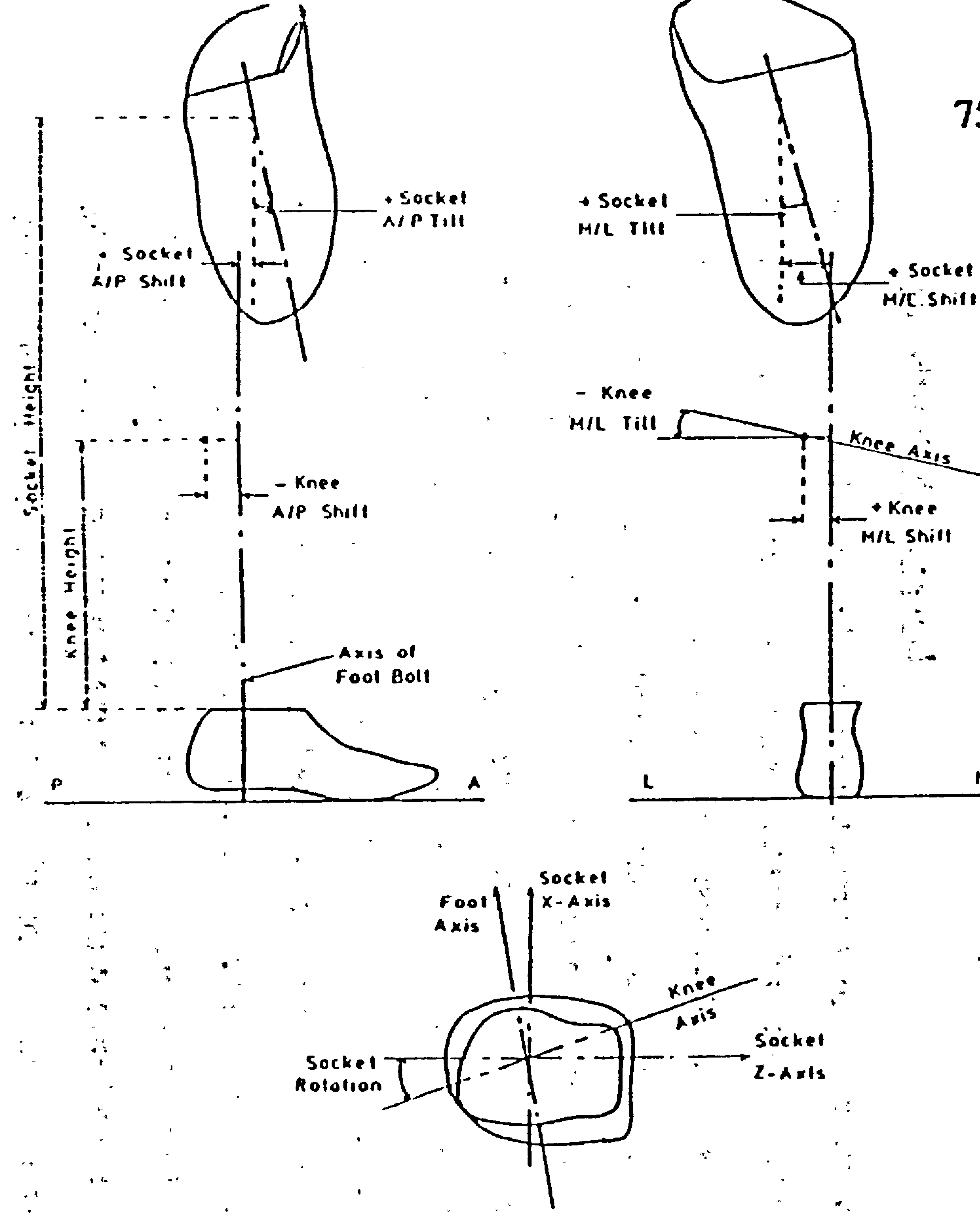
For long stumps, the socket is adducted more than for medium length and the foot can be positioned medially, so that the patient will walk with a narrow base. On the other hand the initial socket flexion can be made small because the stump can control the stability, and large initial flexion may give cosmetic problems. The knee joint can be positioned ahead of the hip-ankle line because it is easy for the patient to control the knee stability, and moving the knee forwards will help to provide a smooth knee flexion at push off. It should be mentioned that in the present practice, sockets for medium and long stumps are set as adducted as possible to provide a normal walking pattern.

Many scientists have had a special interest in the alignment configuration and in establishing a means of providing the optimum alignment of the prosthesis. The group at the University of California Berkeley (UCB, 1947) pointed out that the effect of changing the position of the elements of the prosthesis on the comfort, ability to walk and on the gait characteristics must be determined. As a result, two adjustable legs were designed, one for BK and one for AK amputees which allowed adjustment of all the alignment elements of these legs. Two AK amputees were tested and the results showed that the socket tilt, the angle between the upper and the lower leg, the position of the weight bearing line, and the position of the knee axis and the ankle axis relative to each other and to the line of progression (turned in/out) are all very critical and inter-related. The above results were based on qualitative assessment for comfort and walking; also, the type of socket which was used for the test was not reported but it was stated that the ischial bearing type was used at this centre to test the suction socket. However, it was reported that no conclusion should be made at that time, since only two subjects were tested.

Appoldt et al (1968) measured the local pressures experienced between the stump and socket wall in two AK amputees wearing quadrilateral total contact suction sockets. Also, the effect of alignment changes on the stump/socket pressure was investigated. Alignment changes of 2 degrees in the socket abduction/adduction and 5 degrees in the socket flexion showed no effect on the stump/socket pressure. This may question the accuracy of their instrumentation because a change 5 degrees in the socket flexion will result in dramatic changes in the function of the prosthesis (based on observations by the author of this thesis). However, an error of 3 percent in the pressure readings was recorded.



Alignment parameters for below-knee prostheses.



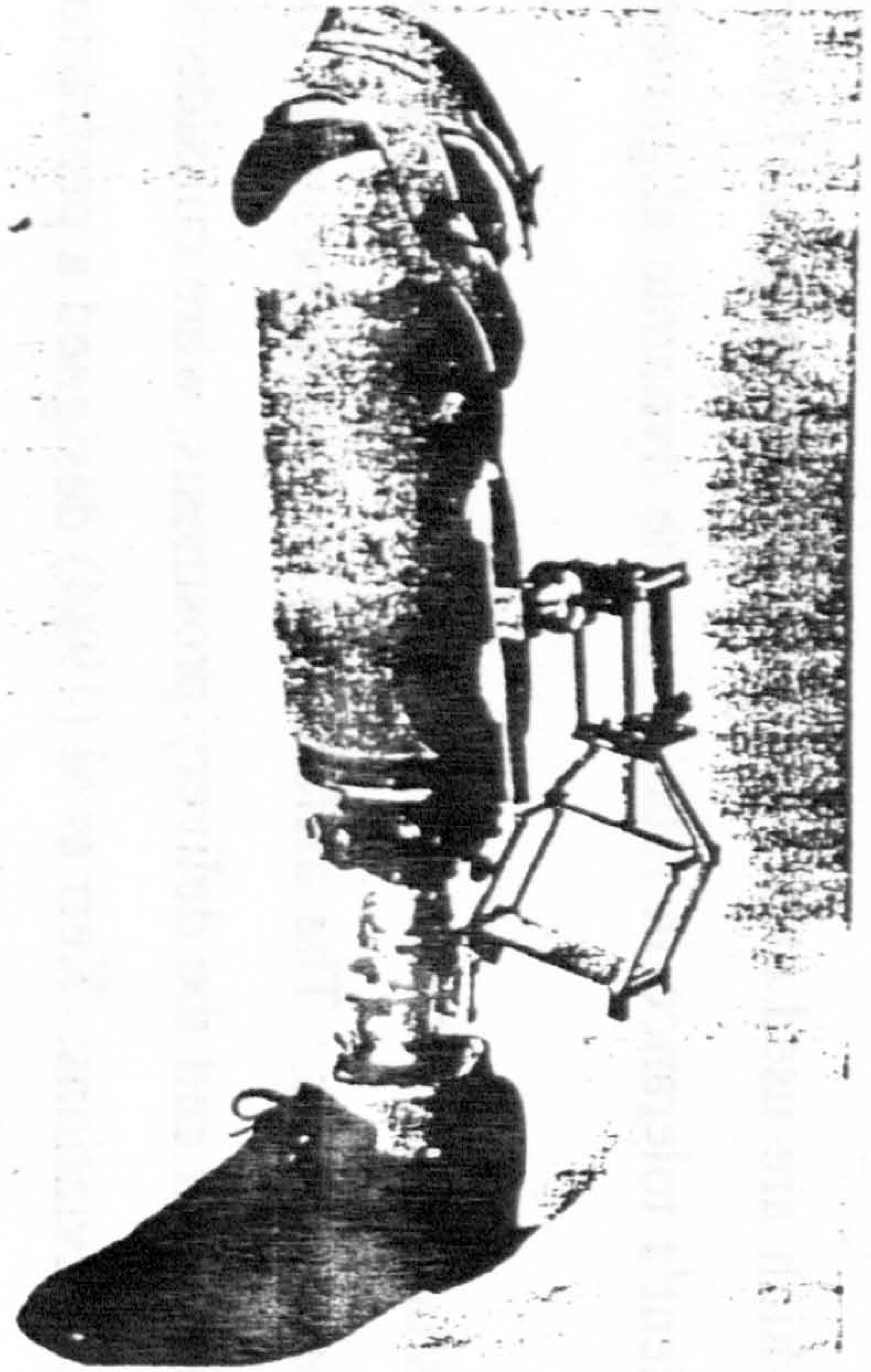
Alignment parameters for above-knee prostheses.

Figure 3.33 Alignment parameters for AK and BK prostheses. (from Zahedi et al 1989)

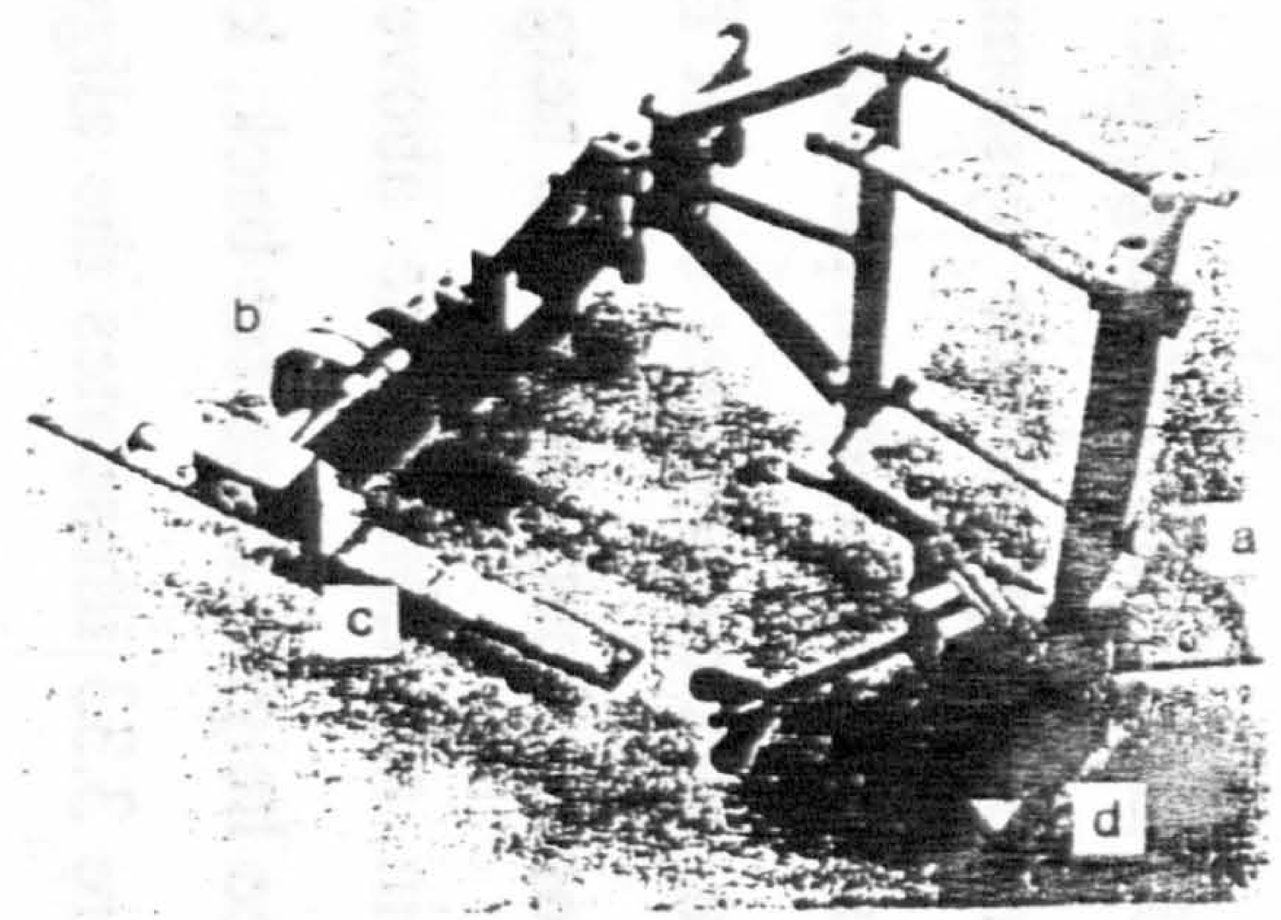
Pearson et al (1973) studied the effect of alignment changes on the pressure at critical regions of the stump of BK amputees. Ten BK amputees were tested and alignment changes were applied in the sagittal and medio-lateral planes. The investigations showed that the pressure at the distal anterior tibial region tended to be largest and most sensitive to alignment alterations. It was found that pressure variations were more sensitive to angular rather than linear alignment changes.

Solomonidis (1975 and 1980) during the course of an objective evaluation of available modular prosthetic systems, defined six alignment parameters for the BK prosthesis and eleven parameters for the AK prosthesis. The below knee alignment parameters are: socket flexion, socket lateral tilt, socket lateral set, socket forward set, prosthesis height and foot toe out angle. The AK prosthesis has in addition to the above, further five alignment parameters which are: knee height, knee set back, knee set out, knee ML tilt and socket rotation. Figure 3.33 illustrates the alignment parameters for the AK and BK prostheses as defined by Solomonidis. The results of Solomonidis's work showed that a range of dynamic alignments was acceptable for any patient, and that a relationship might exist between the various alignment parameters. He recommended that an evaluation should be done on the various procedures which are used to set the bench alignment, and more understanding of the patient's tolerance in relation to the dynamic alignment should be obtained.

Saleh et al (1983) studied the effects of mass properties of the prosthesis on the gait of below knee amputees. The differences in weight between the prosthesis used during alignment and the delivery prosthesis, were considered to be a major factor in gait deviations. Kerr et al (1984) designed a protractor to measure the angular alignment of BK prostheses. It is a parallelogram of linkages attached to fixed reference points on the socket and on the pylon tubing as seen in figure 3.34. This device has an accuracy of ± 1 degree in AP



The protractor fitted to a BK prosthesis.



The angular alignment protractor— a) antero-posterior scale; b) medio-lateral scale; c) fastening strip; d) clamp.

Figure 3.34 The angular alignment protractor of Kerr et al 1984.

and ± 2 degrees in ML angular alignment. The author of this thesis believes that this accuracy is not acceptable as a change in the socket (or any component) angle by ± 1 degree in the AP plane is believed to produce a large effect in the kinetics and kinematics of the prosthesis (Morimoto et al 1987, Zahedi et al 1988, and observations by the author of this thesis). Also the weight of the device and its complexity may alter the gait of the patient during the dynamic alignment session.

Zahedi et al (1985) proposed that using biomechanical analysis, by measuring the angular movements and calculating the moments at the major joints, the most suitable alignment can be selected from a range of acceptable alignments for any patient.

Zahedi et al (1986) examined the alignment results of 183 fittings on 10 below knee amputees, and 100 fittings on 10 above knee amputees. The results showed that an amputee can tolerate several alignments ranging in some parameters by as much as 148 mm in shifts and 17 degrees in tilts. The prosthetists could not repeat a given alignment; however, a range of alignments was acceptable to the patient and to the prosthetist. Figure 3.35 shows nineteen different acceptable alignments made by the same prosthetist for one below knee patient using one prosthesis. This fact was also found for above knee amputees, and it was also found that the acceptable range of alignment varied from one patient to another depending on how much control the patient had of his prosthesis. A linear relationship was found between socket AP tilt and knee AP shift in the case of above knee amputees. It was also found that the above knee amputee can tolerate more variability in the ML direction than in the AP direction. Zahedi et al pointed out that a biomechanical evaluation of the amputee's gait is needed, and comparative optimization procedure of the kinetic and kinematic parameters would be useful.

Hannah et al (1984) investigated the effects of alignment changes on the gait of five below knee amputees. The investigation was carried out using

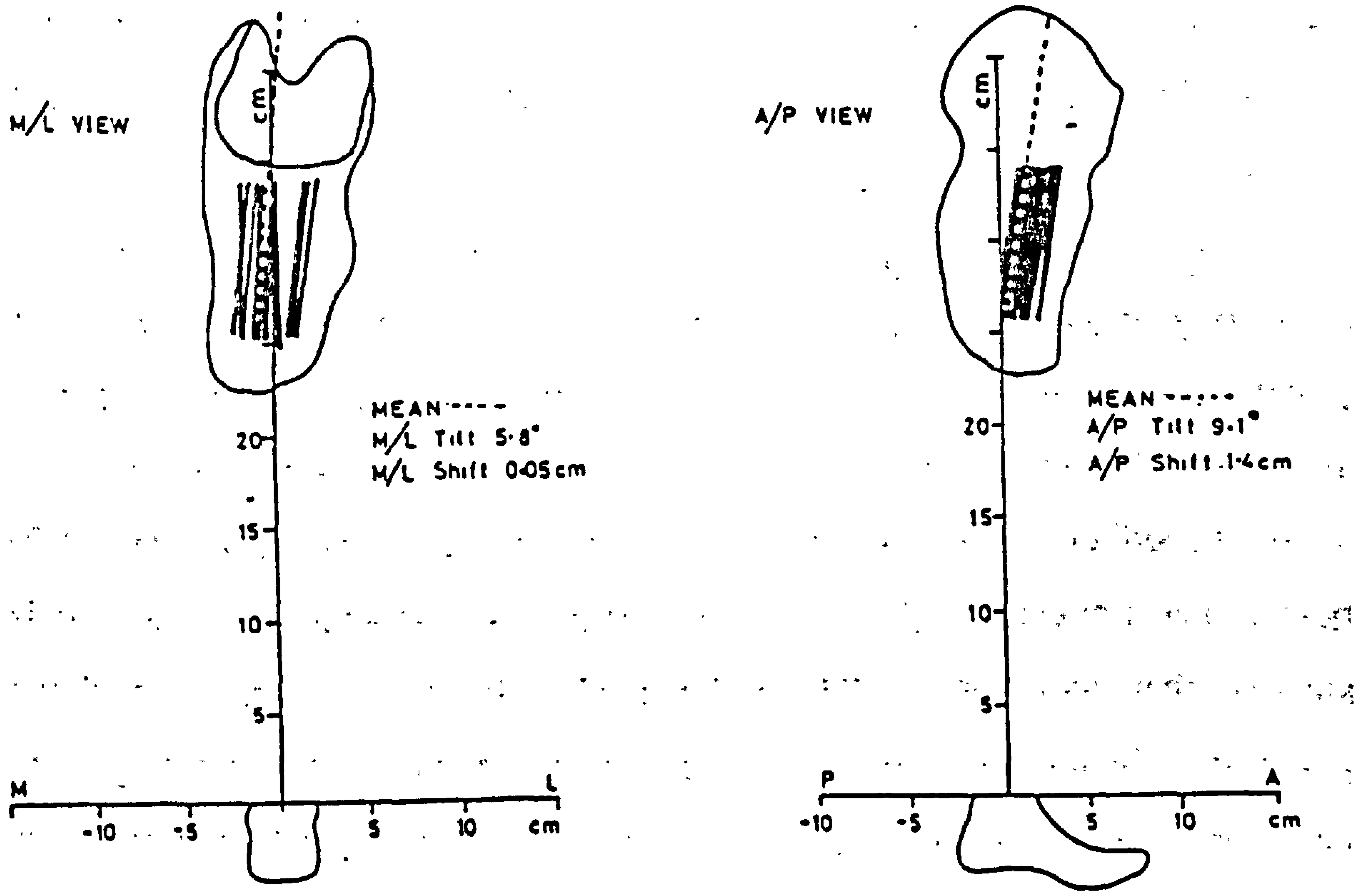


Figure 3.35 Nineteen different acceptable alignments by the same prosthetist for a below knee amputee.
(from Zahedi et al 1986)

goniometry to measure kinematic parameters at the knee and hip of the sound and the prosthetic legs. It was thought that maximum symmetry of the measured kinematic variables coincided with the optimum alignment of the prosthesis. They found that with alignment changes, 40 indices out of 180 were significantly different from the indices achieved at the optimum alignment setting. Thirty eight of the 40 indices were of a lower value than the indices at the optimum alignment setting, showing that for the detected differences in symmetry, the change in alignment resulted in an increase of lower limb asymmetry. Therefore, they concluded that "symmetry of kinematic lower limb variables of a person with below-knee amputation is maximum at the optimal alignment of the prosthesis". It was also found that foot dorsiflexion is the most important variable in the alignment changes, and that hip flexion/extension motion is the most sensitive to alignment changes. It is believed that the authors did not have adequate data to draw the above conclusion, because the optimum alignment was considered as the alignment which the patients were familiar with, and in fact it may not be the optimum alignment. Furthermore, the concept of having maximum symmetry in the kinematic variables coincident with optimum alignment, was undermined when Winter and Sienko (1988) showed by means of biomechanical analysis, which was applied on 5 below-knee amputees, that the motor patterns of the residual muscles at the hip and knee were modified after amputation, and it is not known whether the modified motor patterns are optimal.

Mizrahi et al (1986) studied the effect of alignment changes on the symmetry of the ground reaction force pattern between the sound and the prosthetic sides of an AK amputee. The results showed that by plantarflexing the foot, the push off force on the prosthetic leg was increased and the braking force of the sound leg was reduced. Mizrahi et al found that all the disorders in the patterns of the anterioposterior force at the sound and prosthetic sides, can be eliminated by achieving the optimal alignment. The author of this thesis believes that the ground reaction force pattern alone does not provide enough information to determine the effect of alignment changes on the gait and kinematics of all the body joints should be considered. Furthermore, results obtained by testing only one AK patient cannot be considered conclusive.

Winarski and Pearson (1987) established a quantitative relationship between the loads on the prosthesis and the normal pressure on the stump in BK prostheses using a matrix equation. The equation was used to relate two loads on a shank pylon (axial load and flexion/extension moment) to the normal pressure acting at the patellar tendon and at the gastrocnemius. This relationship was established over " both time and a range of changes in prosthesis alignment " as referred to by the authors. It can be understood that the time is referred to percentage of the gait cycle or the stance phase, and the range of alignment changes is referred to different angular positions of the pylon of the prosthesis. Their aim was to obtain the normal pressure between the socket and the stump at the above areas if the pylon loads were known. The equation has two similar forms, one of them presents the relationship of the axial load, extension/flexion moment and the normal pressure at the patellar tendon. The other form reflects the relationship of the axial load, extension/flexion moment and the normal pressure on the gastrocnemius.

$$\sigma_p(\Theta , t) = [W_{11} + (dW_{11} /d\theta)* \theta]* F_z(\theta ,t) + [W_{12} + (dW_{12} /d\theta)* \theta]* M_x(\theta ,t)$$

where: $\sigma_p(\Theta , t)$ is the normal pressure on the patellar tendon (Pascal)

W_{11} is the influence factor of the pylon axial force ($1/m^2$)

$F_z(\Theta , t)$ is the axial force on the pylon (N).

W_{12} is the influence factor of the flexion/extension moment ($1/m^3$)

$M_X(\Theta , t)$ is the flexion/extension moment on the pylon (Nm).

t is the time (s)

Θ is the flexion/extension angular change of the pylon about the VAPC (Veterans Administration Prosthetic Centre) alignment unit

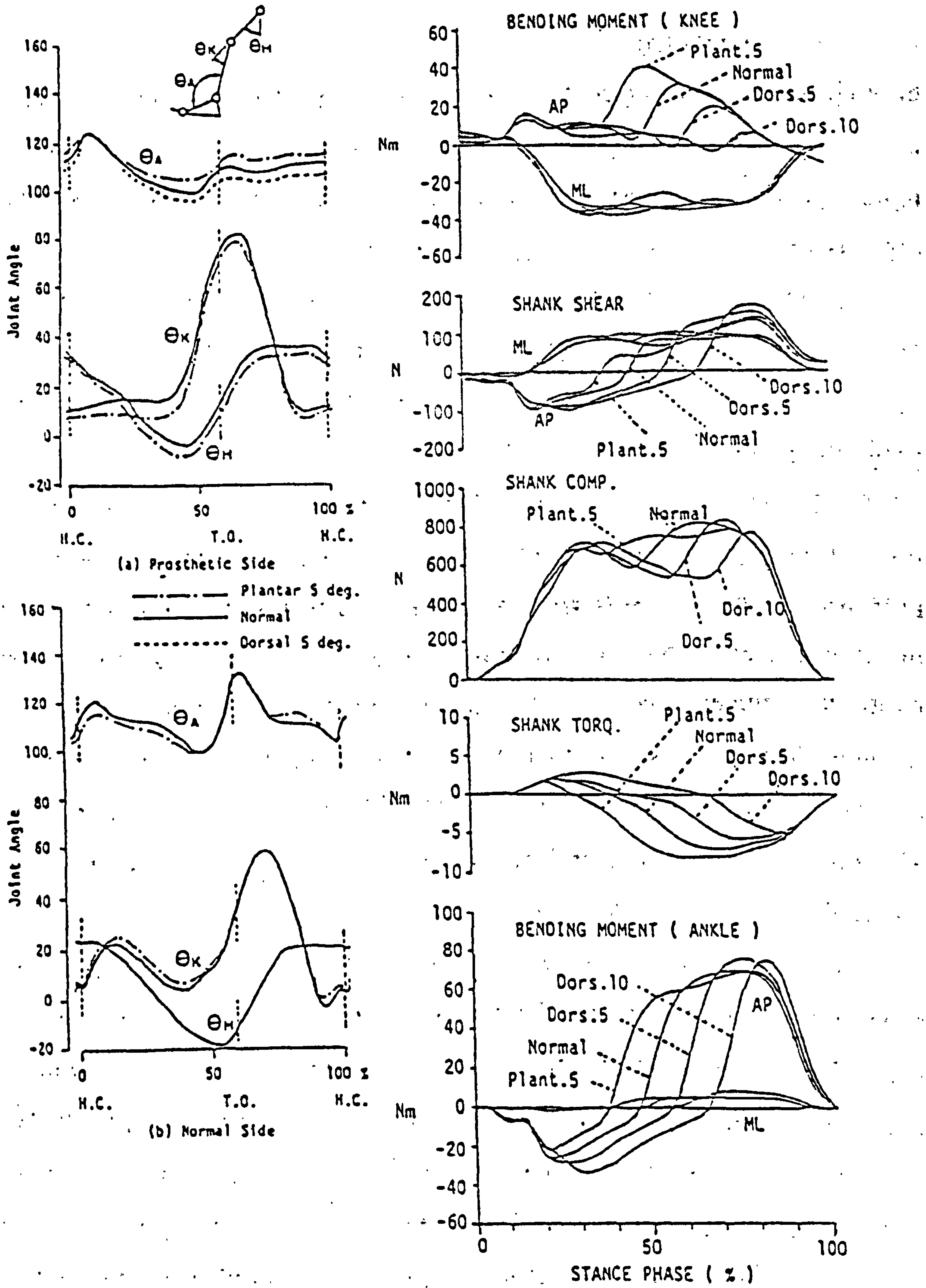


Figure 3.36 The effect of foot plantar/dorsi flexion changes, on the gait pattern of a BK amputee. (from Morimoto et al 1987)

(radians).

The influence factors and their derivatives should be calculated for every subject, and were calculated by firstly; summing the square of the error between actual (measured) and theoretical patellar-tendon pressures, secondly; differentiating the summing equation with respect to the influence factors and to their derivatives, and finally setting the derivatives to zero in order to minimize the sum of the squares of the errors between the actual and the theoretical pressures. However, the authors did not present any computer program to serve this calculation.

The above equation shows that a certain percentage of the normal pressure on the patellar tendon is produced by the axial force, and the rest is produced by the flexion/extension moment. The results showed that the patellar tendon pressure has the greatest sensitivity to the pylon flexion/extension adjustments. This study calculated only the normal pressure without considering vertical shear. Also, it was concerned with only two areas of the stump socket interface of a BK prosthesis, and it should be extended to cover the whole surface between the socket and the stump of BK and AK amputees.

Morimoto et al (1987) used a pylon load cell (PLC) to measure six quantity loads at the shank of a BK prosthesis with a PTB socket and a uniaxial foot. Single axis flexible electrogoniometers were also used at the hip, knee and ankle of the prosthetic and sound legs to measure the angular movements of the joints. The above equipment was used to study the effect of prosthetic alignment changes on the gait pattern of a BK amputee. The foot was shifted in the AP and ML directions, and then also tilted in the AP and ML directions, and the toe was rotated in and out from the normal position. The patient walked along a straight level path, a ramp, and a stair way when the data were collected. Figure 3.36 shows samples of data which reflect the effect of foot dorsi/plantar flexion changes on the gait pattern. All the gait

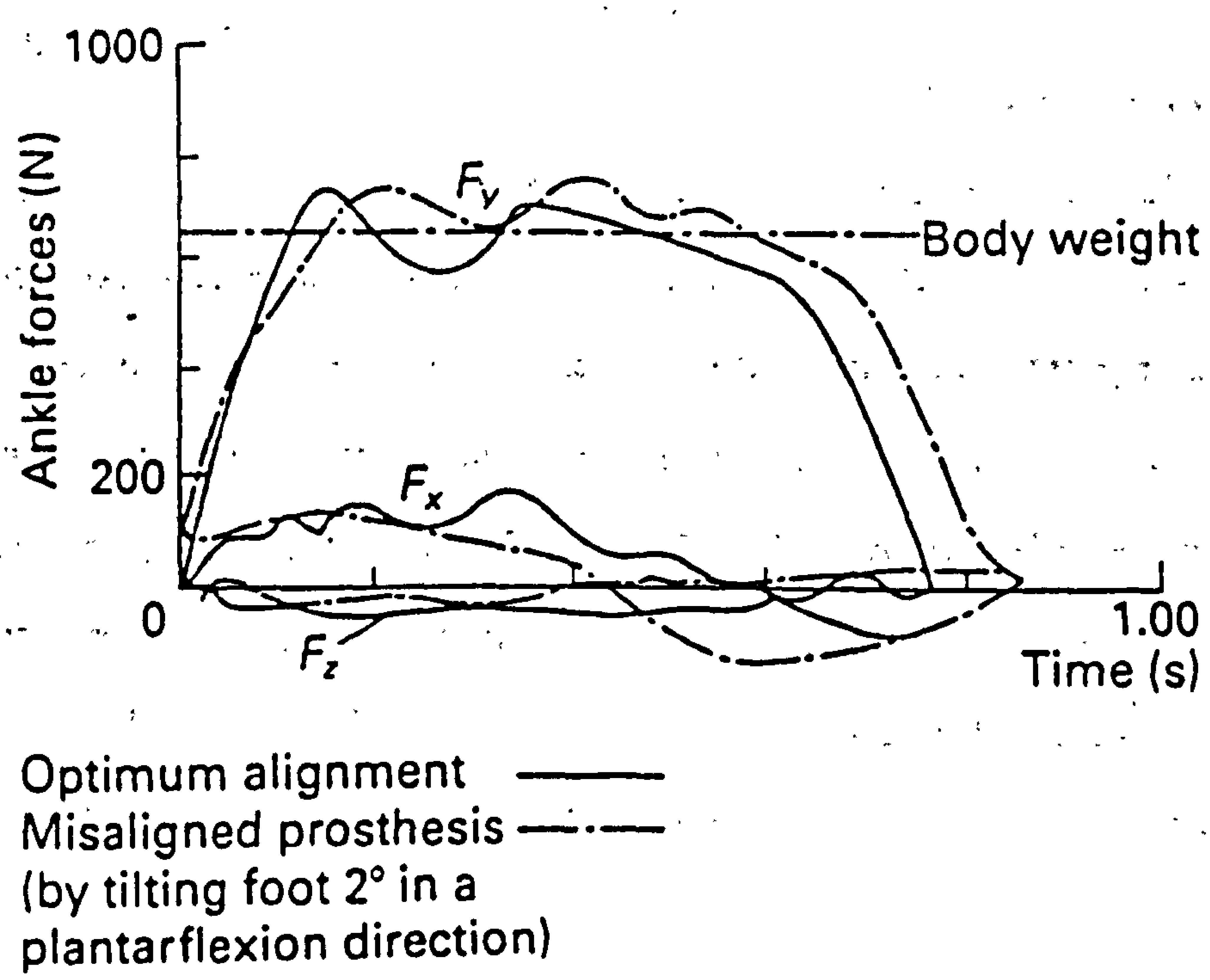


Figure 3.37 The effect of misalignment on the ankle forces of AK amputees. (from Zahedi et al 1988)

parameters were changed, but the changes were most noticeable at the ankle and knee bending moment in the AP plane, and in the torque of the shank. Dorsiflexing the foot has extended the time of the plantarflexing moment at the ankle joint (longer cross over period) and also increased the value of this moment. This study was limited to only one patient, therefore, more patients should be tested to establish the true relationship between the gait parameters and the changes in the alignment of the prosthesis. Also, this study was only concerned with the BK prosthesis and its extension to include the AK prosthesis is desirable.

Saleh (1988) discussed the optimization of the prosthesis and stated four main areas which affect it, they are: stump-socket interface, mass properties, knee and ankle joint mechanism components, and alignment. Saleh reported that, deliberate anterioposterior misalignments were made in five BK amputees, thereafter their gait was assessed by visual observation and by using an automated gait analysis system. The results proved that the subjective methods of assessment were inadequate.

Zahedi et al (1988 and 1989) performed and measured 300 alignments on ten BK and ten AK amputees. It has been found that the step to step variations are noticeable on the pattern of the amputees gait. These variations varied for each individual patient and also with the alignment of the prosthesis. A deliberate misalignment of 2 degrees plantarflexion on the foot of an AK prosthesis which was accepted by the patient and by the prosthetist, resulted in a considerable change on the loads at the ankle joint, as seen in figure 3.37. The changes in the loads at the ankle joint resulted in changes at the knee and hip joints. It was found that patient and prosthetist can accept a number of alignments and it has been recommended that by using biomechanical analysis, it is possible to select the most suitable alignment from a number of acceptable alignments. The above results were also confirmed, by Solomonidis and Spence (1988).

Yang Lang (1988) and Lang et al (1991a) studied the effect of alignment changes on the gait of above knee amputees. Four patients were tested but only two were used for the complete analysis. Two force plates and three TV cameras were employed to measure the kinetic and kinematic data of the subjects at the sound and prosthetic sides, and the effects of the alignment changes on the gait pattern were accurately assessed. As the number of patients was very small (2 patients), it was difficult to draw definite conclusions especially on the sound leg and on the trunk's variables. However, the effect of alignment changes was noticeable on the prosthetic side, and particularly on the value of the ground reaction forces and the positions of their line of action. Lang pointed out that the centre of pressure and the fore-aft shear force were very sensitive, and they could be very useful in studying the effect of alignment changes on the amputee's gait.

Marmar and Solomonidis (1989) carried out a biomechanical analysis of AK amputees in order to study the effect of alignment changes on the gait parameters. Their aim was to provide the prosthetist with a method for achieving optimum alignment regardless of the patient's comments. The alignment changes were in the sagittal plane only, and it was found that the prosthetic kinetic and kinematic data were very sensitive to the alignment changes. The results will be presented in the results chapter in this thesis, and further details regarding the prosthetic and the sound leg of more new patients will be discussed. It is evident that a number of researchers have investigated the effect of alignment on prosthetic gait, but most of these investigations were concerned with BK prostheses and very few have tested AK prostheses. This indicates that much more work should be done regarding the alignment of AK prostheses.

3.4 Gait Analysis of Lower Limbs Amputees

This section deals mainly with the gait of above knee amputees, and the gait of below knee amputees will be briefly dealt with, only when it is

Table 3.4 Means and SD of temporal parameters obtained for 20 AK amputees and 20 normal subjects.

(Reprinted from Zuniga et al 1972).

Temporal Parameters [s]	Amputees		Normal Subjects	
	Prosthetic	Anatomic	Left	Right
Duration of walking cycle	1.43 ± 0.16	1.42 ± 0.13	1.40 ± 0.11	1.39 ± 0.12
Duration of stance phase	0.83 ± 0.10	0.94 ± 0.09	0.86 ± 0.08	0.86 ± 0.10
Duration of swing phase	0.60 ± 0.06	0.48 ± 0.05	0.54 ± 0.04	0.53 ± 0.07
Duration of double stance	0.18 ± 0.04	0.14 ± 0.04	0.14 ± 0.05	0.15 ± 0.03

necessary to help in understanding the gait of AK amputees. All temporal distance, kinetic and kinematic data, and energy expenditure will be presented in this section.

3.4.1 The Temporal-Distance Parameters

The temporal-distance parameters can be used in gait evaluation and in the evaluation of the function of the prosthesis, although they are not enough for the whole evaluation.

Zuniga et al (1972) used foot-switch signals to measure the temporal parameters of 20 AK amputees which were described as "good prosthesis wearers" and had "good gait patterns". In addition, 20 male normal subjects were tested for reasons of data comparison. Table 3.4 shows the temporal parameters of the amputees and the normal subjects. It was reported that the results were reproducible when a subject was tested and retested within one day. However, the author of this thesis believes that the wiring of the foot-switches and the cables of the goniometer which was used, together with the foot switches, may have disturbed the patient and altered the gait.

James and Oberg (1973) studied the gait of 34 above knee amputees. The temporal-distance parameters were measured using two micro-switches, one at the heel and the other at the front part of the sole. All patients were fitted with total-contact suction sockets and in most cases the prosthesis was provided with a stabilising knee mechanism and swing phase control (the type was not defined). The data were collected at normal and fast speeds and were used to assess the gait asymmetry regarding the temporal-distance parameters of the sound and the prosthetic legs. The gait parameters were compared with control data which were obtained from normal subjects. The normal speed of the amputees was found to be 38% lower than the normal speed of normal subjects. The cycle duration was about 33% longer, and the stride length was about 17% shorter than those of normal subjects. At normal speed the stride width, (measured as the perpendicular distance between two parallel lines, one

Table 3.5 Temporal distance parameters of 7 AK amputees and 30 normals. (from Murray et al 1980)

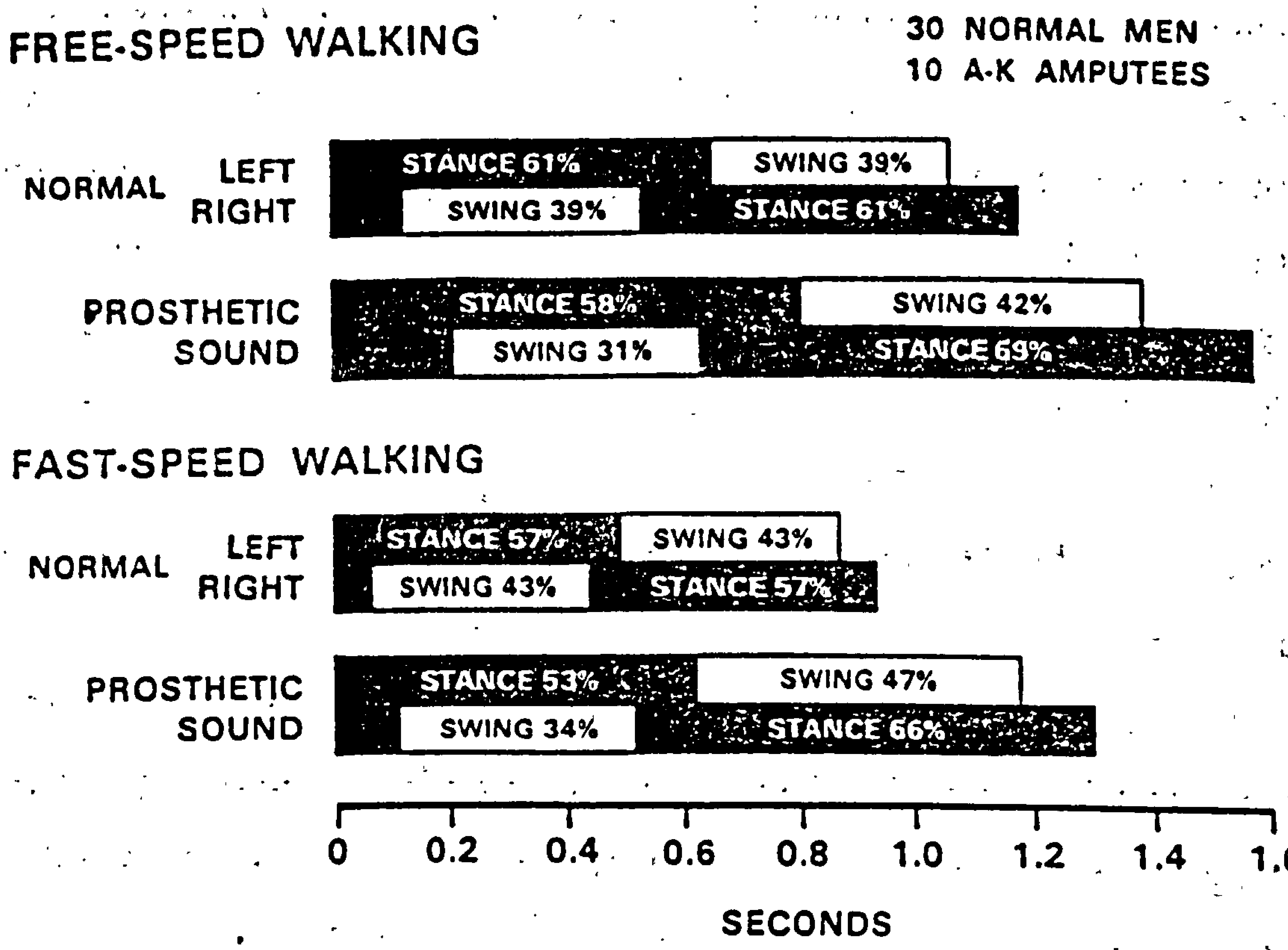
	FREE (Mean ±1 S.D.)	FAST (Mean ±1 S.D.)
Walking speed (cm/sec)		
Amputees	100 ±16*	140 ±25*
Normal men	151 ±20	218 ±25
Cycle duration (sec)		
Amputees	1.38 ±.11*	1.18 ±.11*
Normal men	1.06 ±.09	.87 ±.06
Cadence (steps/min)		
Amputees	87 ± 7*	102 ± 9*
Normal men	113 ±10	138 ±10
Stride length (cm)		
Amputees	136 ±15**	164 ±22†
Normal men	156 ±13	186 ±16
Step length (cm)		
Sound	64 ±9*	81 ±12**
Prosthetic	72 ±8	83 ±11†
Normal	78 ±7	93 ± 9
Stride width (cm)		
Amputees	17.4 ±3.6*	18.3 ±4.3*
Normal men	7.7 ±3.5	9.1 ±4.1
Foot angles (deg)		
Sound	10.9 ±7.2	10.2 ±5.8†
Prosthetic	2.3 ±5.0	2.1 ±4.6
Normal	6.3 ±5.7	5.3 ±5.5
Stance Phase (sec)		
Sound	.94 ±.12*	.78 ±.09*
Prosthetic	.80 ±.07*	.62 ±.08*
Normal	.65 ±.07	.49 ±.05
Swing Phase (sec)		
Sound	.43 ±.04	.41 ±.04†
Prosthetic	.58 ±.06*	.56 ±.04*
Normal	.41 ±.04	.38 ±.03
Double-Limb Support (sec)		
1st in cycle	.20 ±.06**	.11 ±.03**
2nd in cycle	.17 ±.04**	.10 ±.04†
Normal	.12 ±.03	.06 ±.03

*The amputees were significantly different from normal ($p<.001$); **($p<.01$); †($p<.05$)

passing through the right heel point and the other passing through the left heel point and both parallel to the line of progression) of the patients was 31% greater than that of normal subjects. The size of the stride width for the amputees was related to the possibility of stabilising the pelvis in the frontal plane during the stance phase of the prosthetic side. It was found that the patient took a longer step with the prosthetic leg (distance from sound heel strike to prosthetic heel strike) than with the sound leg (distance from prosthetic heel strike to sound heel strike). The prosthetic gait was found to have considerable asymmetry with regard to stance and swing phase duration and to step length. The stance phase duration of the prosthesis was 13% shorter than that of the sound leg, while the swing phase was 25% longer, and the step length 10% longer than that of the sound leg. The asymmetry of the gait was found to be unaffected by increasing the speed of the patient from normal to fast speed.

Godfrey et al (1975) measured the temporal-distance parameters for 7 above knee amputees in order to evaluate 6 different knee mechanisms. Each of these patients was fitted with the six knee mechanisms, and the temporal parameters were measured by means of foot switches. However, the method of measuring the distance parameters was not discussed. The results were comparable with those of other researchers (James and Oberg 1973) as presented by the authors, and it has been found that the more complicated knee designs have no great advantages over the simpler ones when walking on flat level surface.

Murray et al (1980) studied the gait pattern of 10 adults with above knee amputations, and their temporal-distance parameters were measured. Table 3.5 shows a comparison between the amputees' data for two speeds and the data of 30 normal men. The results showed that the duration of the sound stance phase was longer than that of the prosthetic stance phase, but the duration of the sound swing phase was shorter than that of the prosthetic swing phase (fig



Average stance and swing durations during free-speed and fast walking for the sound and prosthetic limbs of 10 above-knee amputees with constant-friction knee components (which are adjusted for free speed) and for the left and right limbs of 30 normal men. The numbers within the timing bars express the stance and swing durations in percent of the walking cycle durations.

Figure 3.38 Stance and swing durations during free and fast walking. (from Murray et al 1980)

3.38). The stride length was found to be 77 percent of the stature for the amputees walking at their normal speed (compared to 89 percent for the normal men). The stride widths of the amputees were found to be wider than that of the normals, and the successive step lengths tended to be uneven. The author of this thesis found the results of Murray et al to be reasonably comparable with the results of James and Oberg. For example, at the free speed, the stance phase duration was 0.77 s and 0.89 s for prosthetic and sound legs respectively in James and Oberg. The corresponding figures are 0.80 s and 0.94 s in Murray et al. This is evidence to prove the validity of their methods, and these data can be used as reference for future investigation.

Murray et al (1983) tested seven above knee amputees with two different mechanisms for controlling the prosthetic knee during swing (constant friction and hydraulic mechanisms). The temporal distance parameters were measured at slow, free, and fast speeds and the results found to be very similar to the previous report (Murray et al 1980). Table 3.6 shows the results for the two types of knee mechanisms at three different speeds. In this table the control data are for normal men in an appropriate age range.

3.4.2 Kinetic and Kinematic Analysis

The research carried out at the University of California, Berkeley (UCB, 1947) supplied the world with a basic study of human locomotion concerning normal, below knee and above knee amputee subjects. The glass walkway studies provided the measurement of the kinematic variables of the human body and the pylon studies provided the measurement of kinetic variables of the prostheses of lower limb amputees. Two AK amputees were tested using the glass walkway and the data were compared with data for normal subjects. Figures 3.39 and 3.40 show some results for an above knee left amputee (the prosthetic side). It is noted that the hip elevation curve of the prosthetic side of an above knee amputee rises only during the swing phase as the prosthesis is being lifted through its cycle, and the length of the prosthesis is controlling

Table 3.6 Temporal-distance parameters for seven AK amputees, with two knee mechanisms at three different speeds. The control data is for 11 normal men. (from Murray et al 1983)

Gait measurement		Condition of walking: Slow Mean ± 1 SD	Free-speed Mean ± 1 SD	Fast Mean ± 1 SD
Velocity (cm/sec)				
Constant friction		78± 8*	107±11*	149±22*
Hydraulic		75±11*	120±23*†	169±21*†
Controls		88±11	151±20	218±25
Cadence (steps/min)				
Constant friction		78± 5	89± 3*	104± 7*
Hydraulic		81± 6	99± 7*†	120± 9*†
Controls		81± 5	113±10	138±10
Stride length (cm)				
Constant friction		121±12	142±14	172±21
Hydraulic		111±11*	144±17	171±18
Controls		126±13	156±13	186±16
Cycle duration (sec)				
Constant friction		1.55±.11	1.34±.04*	1.15±.08*
Hydraulic		1.49±.12	1.21±.09*†	1.01±.07*†
Controls		1.48±.09	1.06±.09	.87±.06
Swing phase (sec) (% cycle)				
Constant friction	-sound	.49±.10 (31%)	.43±.05 (32%)	.40±.03 (34%)
	-prosthetic	.61±.06 (39%)*	.57±.05 (42%)*	.55±.04 (48%)*
Hydraulic	-sound	.45±.04 (30%)*	.43±.04 (35%)	.39±.04 (38%)
	-prosthetic	.53±.03 (36%)†	.49±.02 (40%)*†	.45±.03 (45%)*†
Controls		.52±.05 (35%)	.41±.04 (39%)	.38±.03 (44%)
Stance phase (sec) (% cycle)				
Constant friction	-sound	1.07±.05 (69%)*	.90±.06 (68%)*	.76±.06 (66%)*
	-prosthetic	.94±.07 (61%)	.78±.04 (58%)*	.60±.06 (52%)*
Hydraulic	-sound	1.04±.11 (70%)	.79±.10 (65%)*†	.64±.07 (62%)*†
	-prosthetic	.95±.14 (64%)	.72±.09 (60%)	.55±.06 (55%)*†
Controls		.96±.07 (65%)	.65±.07 (61%)	.49±.05 (56%)
Step length (cm)				
Constant friction	-sound	58±7	67± 8*	86±10
	-prosthetic	63±7	76± 8	86±13
Hydraulic	-sound	54±5*	69± 8*	84±10*
	-prosthetic	56±6*	75±10	87± 9
Controls		63±7	78± 7	93± 9
Foot angle, out-toeing (deg)				
Constant friction	-sound	11.3±7.3	9.3±5.8	8.4±4.8
	prosthetic	3.4±4.9	2.2±4.7	1.8±4.9
Hydraulic	-sound	11.2±5.8	7.9±5.7	6.5±4.7
	-prosthetic	1.0±4.2*	0.7±3.6*	−1.4±4.1*
Controls		6.0±5.8	6.3±5.7	5.3±5.5
Stride width (cm)				
Constant friction		17.5±6.1*	16.4±4.8*	18.1±4.4*
Hydraulic		19.8±7.1*	18.6±6.0*	18.4±5.3*
Controls		10.5±5.7	7.7±3.5	9.1±4.1

*Significantly different from normal ($p<0.05$); †Hydraulic significantly different from constant friction ($p<0.05$).

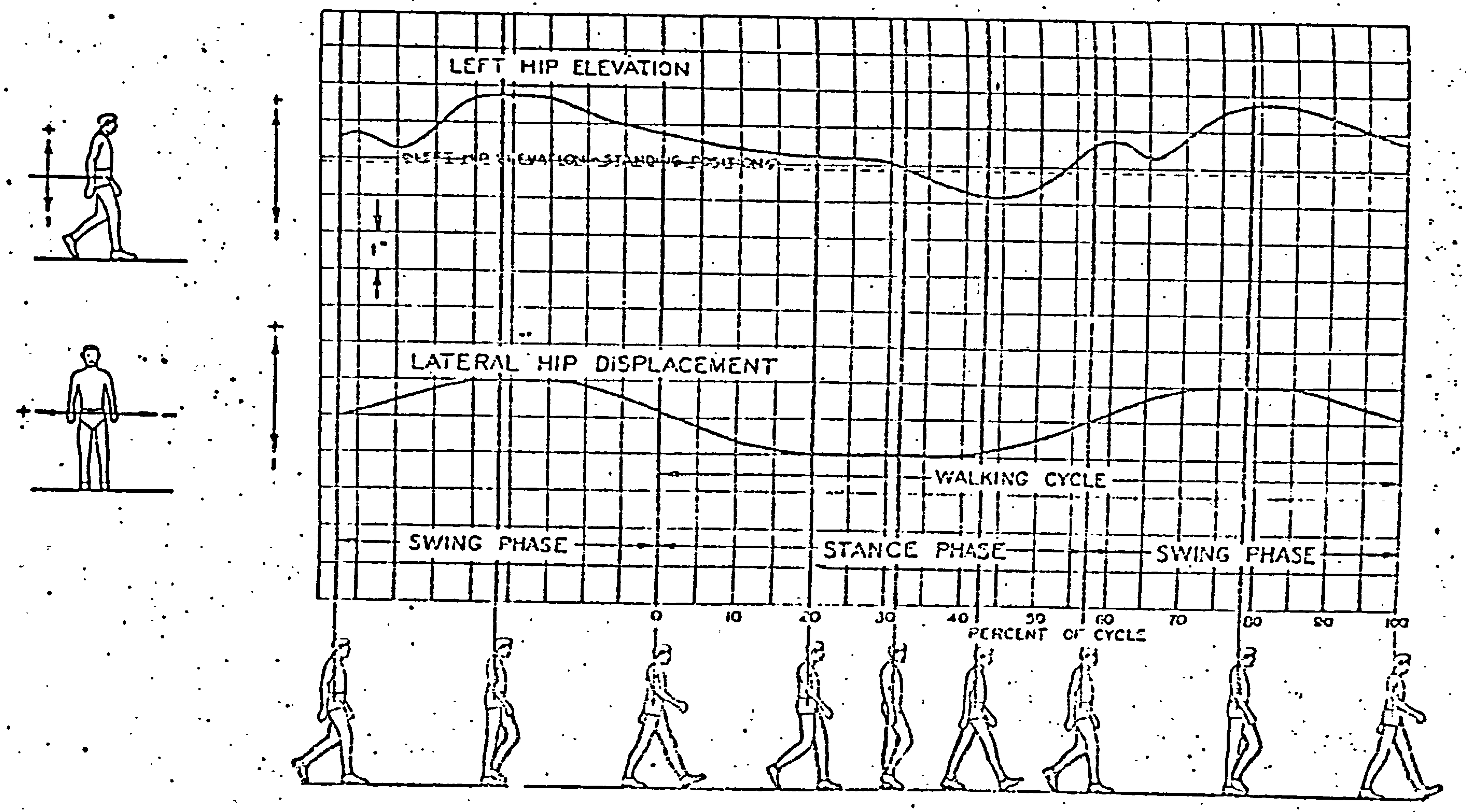


Figure 3.39 Vertical and lateral displacements of the hip joint for AK amputee. (from Eberhart 1947)

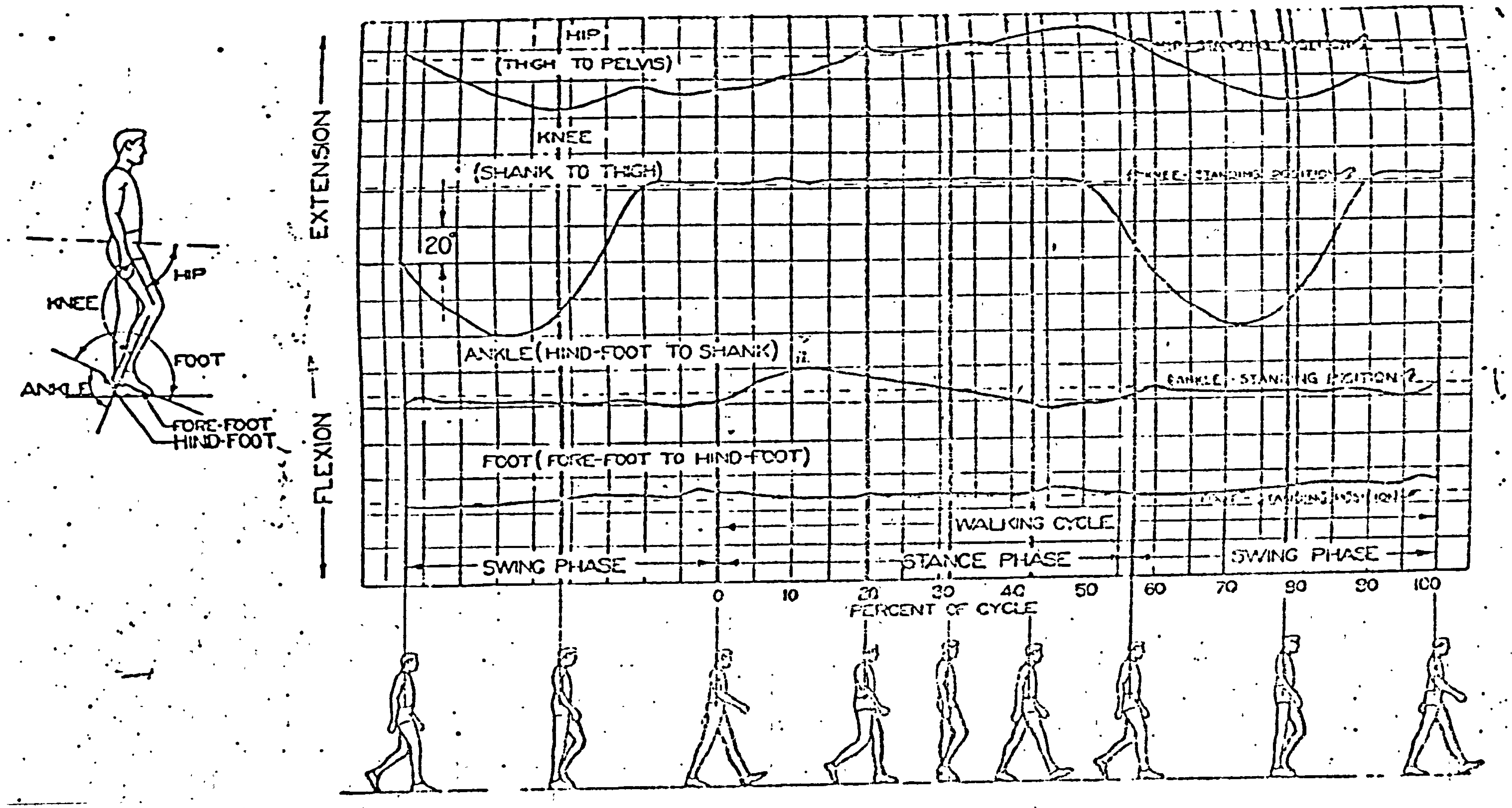


Figure 3.40 Angular displacements of the joints on the prosthetic side for AK amputee. (from Eberhart 1947)

the elevation of this curve during the stance phase. The knee of the AK amputee remains locked during the greater part of the stance phase, and it is seen (fig 3.40) that the artificial knee does not have the double peak phenomenon which is seen during the stance phase of the normal subject. Also, the ankle of the prosthetic leg was found to be unable to dorsiflex or plantarflex the artificial foot. The absence of prosthetic foot plantarflexion makes the patient unable to lengthen his prosthetic leg, therefore, the patient would take a shorter step with the sound leg (distance from prosthetic heel strike to sound heel strike). The pylon studies were performed on three BK and three AK amputees for level walking, up and down stairs, and up and down a ramp. Although the results were very useful, it would be unwise to draw any definite conclusions as the number of subjects was very small. Also, an above knee amputee was tested with a suction socket and with the usual pelvic belt leg. It was found that the patient put more load on his suction prosthesis than on his usual pelvic belt prosthesis. This was attributed to the fact that the patient had more confidence in his suction leg as it was more comfortable. Figure 3.41 shows data based on loads from a pylon test for an above knee amputee with a suction socket and with pelvic belt socket for level walking.

Cunningham (1950) presented force plate data for 10 normal subjects, 7 BK amputees, and 11 AK amputees. Figure 3.42 shows the results for four above knee amputees for level walking. The differences between the pattern of the amputees gait and the pattern of the normal subjects were related to differences in the walking speed and to the ratio of leg length to the stride length. For above knee amputees it is clear that the forces transmitted by the prosthetic leg are smaller than the forces transmitted by the sound leg. There was an indication of a third peak in the pattern of the vertical forces on the sound legs of the amputees. This is caused by the "Vaulting" action which takes place when the patient rises on his sound foot to allow clearance for his

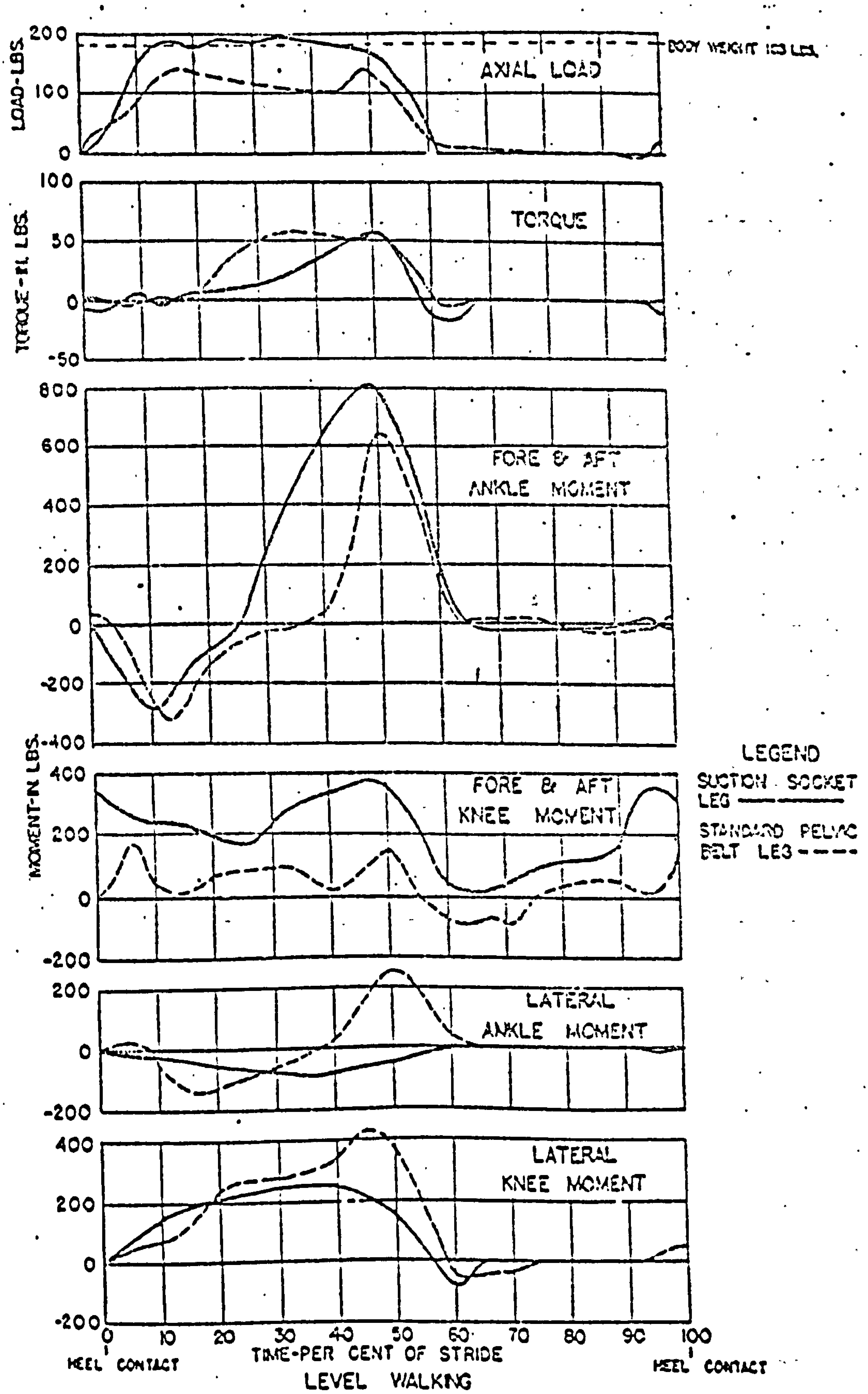


Figure 3.41 Pylon data for an AK amputee wearing suction socket and pelvic belt socket, for level walking. (from Eberhart 1947) Note: LB = 0.454 kg; IN = 0.0254 m; IN LB = 0.113 Nm.

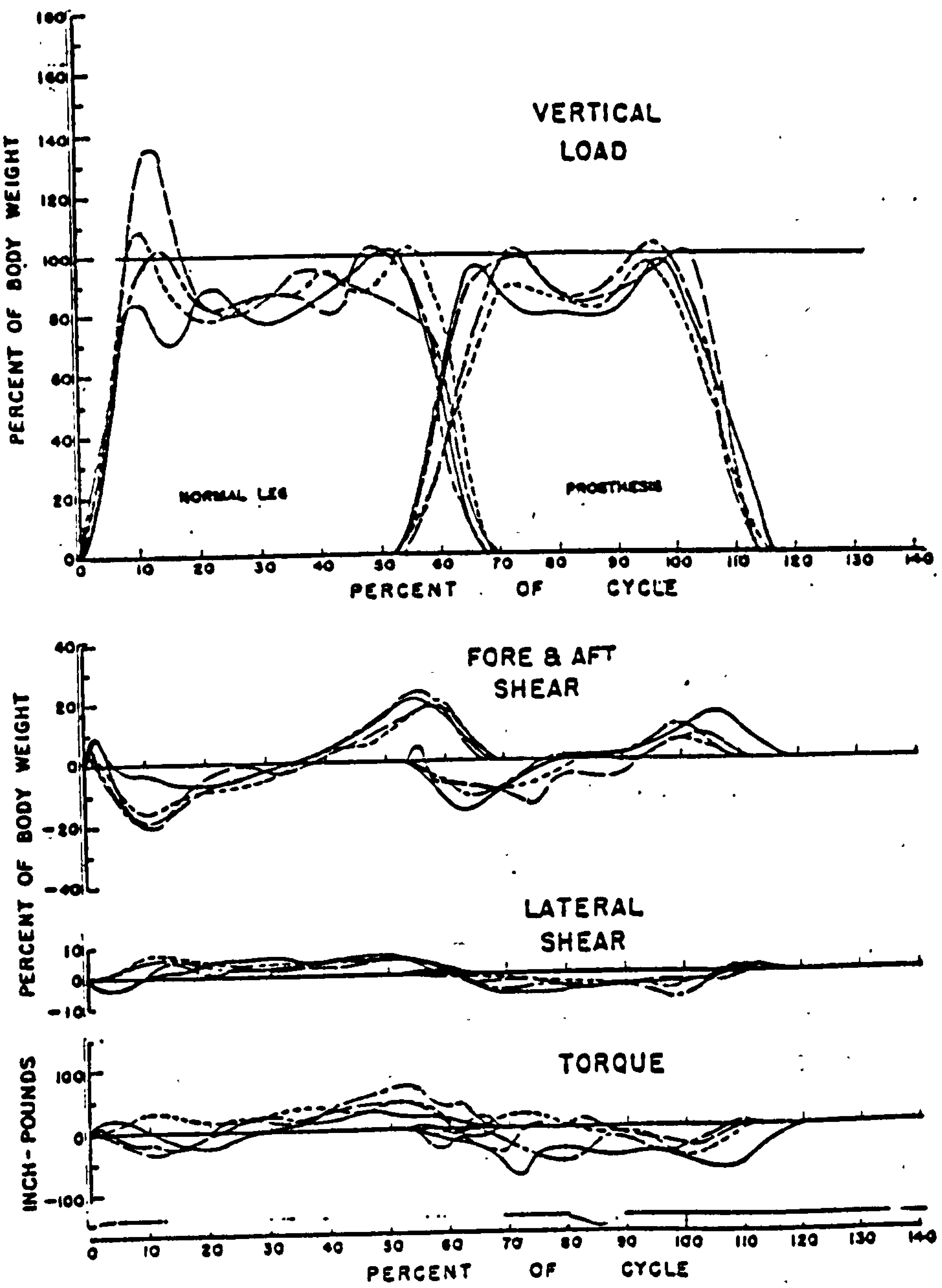


Figure 3.42 Floor reaction data for four above knee amputees for level walking. (from Cunningham 1950).
Note: INCH-POUNDS = 0.113 Nm.

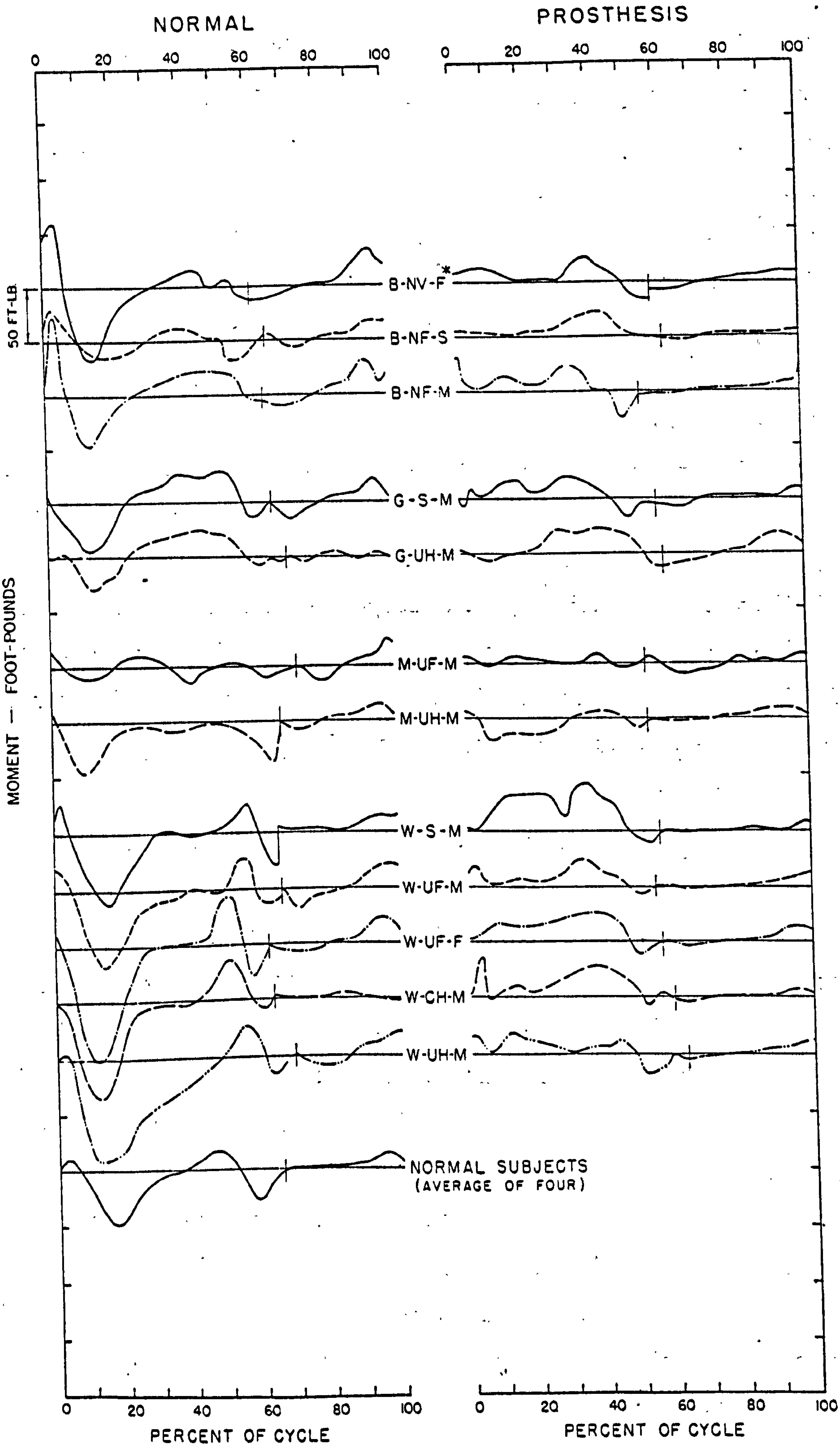


Figure 3.43 Knee flexion/extension moment, for AK amputees.
12 runs made on four patients (S, G, M and W) fitted with different knee mechanisms, and walked with fast (F), medium (M) and slow speed. Normal = sound side. (from Besler et al 1957)

Note: POUND = 0.454 kg, FOOT-POUNDS = 1.356 Nm.

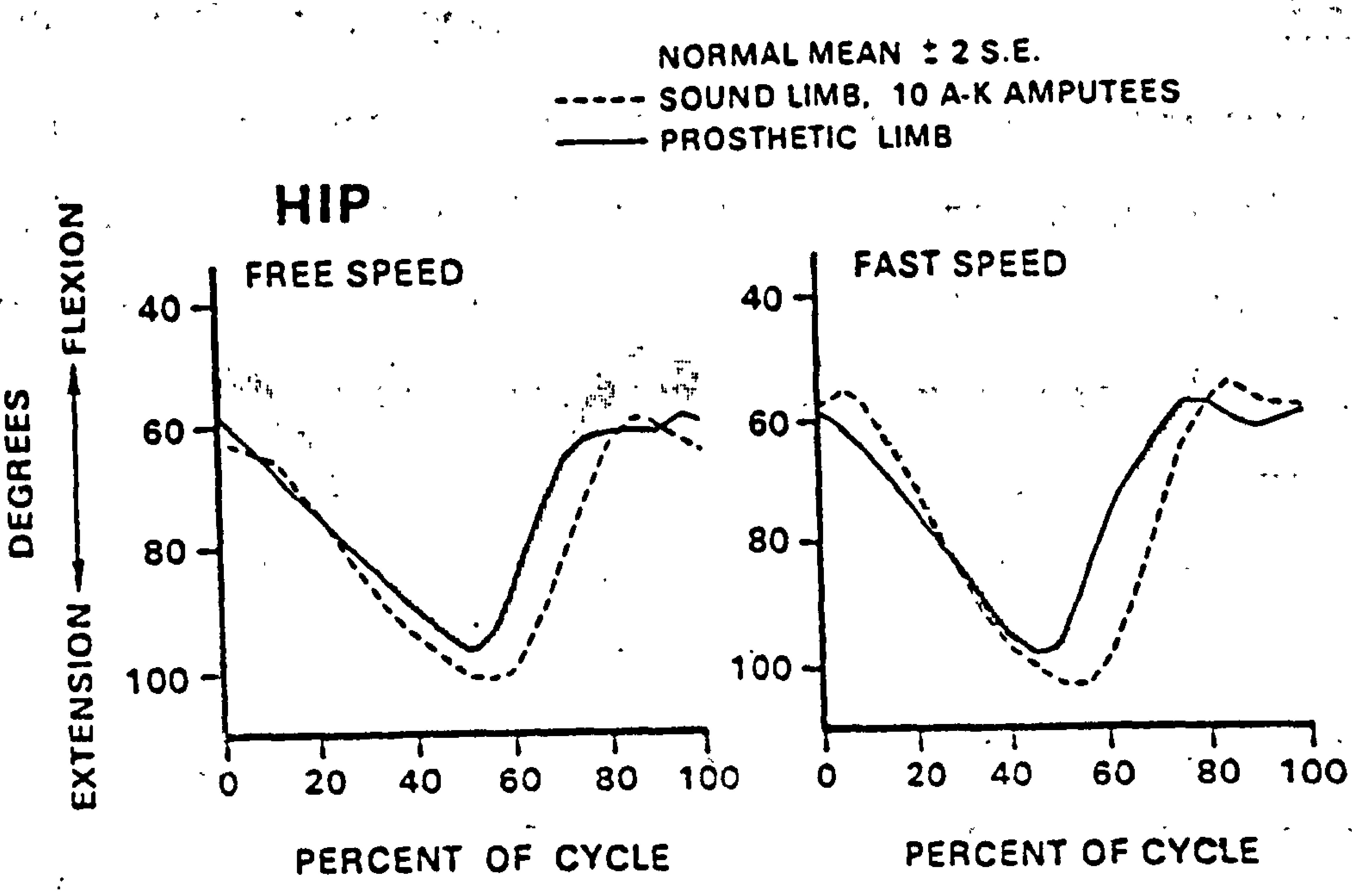
Decoding example: G-S-M = G, is the subject, S, Single axis knee mechanism, M, Medium speed.

prosthetic toe and prevent stumbling. The "Vaulting" action was also found to be caused by a bad suspension system or by a long prosthesis. A point to be mentioned about the data in figure 3.42 is that the vertical force on the prosthetic side for two subjects was below the body weight line; the reason for this phenomenon was not explained.

Bresler et al (1957) studied four above knee amputees using different types of knee mechanisms, walking at slow, medium, and fast speeds. The knee mechanisms as listed by the authors were: (1) NV = Navy Leg with "variable cadence" device. (2) NF = Navy Leg with constant friction device. (3) S = Conventional single axis knee. (4) UH = U.C.Polycentric knee with hydraulic damping. (5) UF = U.C.Polycentric knee with Navy Friction device. (6) CH = Catranis Utility Hydraulic Leg with muscle control. The angular displacement and the moment at the hip, knee and ankle joints were presented for the sound and the prosthetic legs, and compared with data for four normal subjects. The results showed that as heel-off of the prosthetic leg was delayed, the shank continued to rotate about the ankle causing an increase in the ankle angle (dorsiflexion) of the prosthetic side. On the prosthetic side the knee was found to have no flexion during stance phase. The hip flexion angle of the prosthetic side did not increase after heel contact, instead, the hip started extending just after heel strike. This was attributed to the lack of knee flexion at heel contact. The moment at the hip joint was found to have no significant differences between the sound and the prosthetic legs. Figure 3.43 shows the AP knee moments for the four subjects fitted with different knee mechanisms and walking with slow, medium and fast speeds.

Cappozzo et al (1976) studied the kinematics of two AK amputees wearing prostheses with uniaxial knee mechanisms and SACH feet. The results showed that the pronounced disadvantages of the prostheses used were the stiff knee during the stance phase and the fixed ankle during the swing phase.

Murray et al (1980) tested ten active above knee amputees using



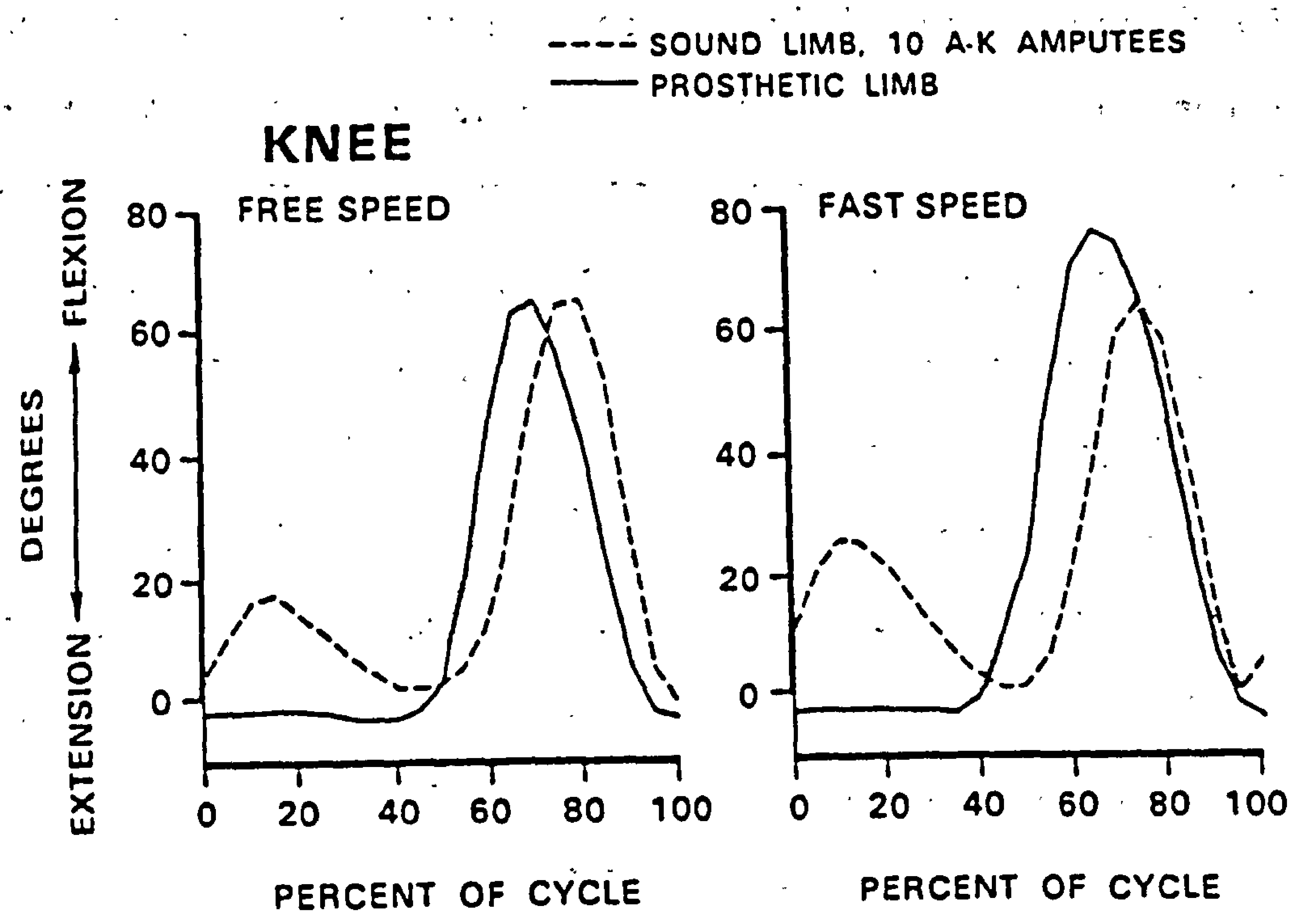


Figure 3.44 Knee and hip angles in the sagittal plane, for free and fast speeds. (from Murray et al 1980). The hip angle was measured between the line connecting HJC to KJC and the line which connects the anterior superior iliac spine with the posterior superior iliac spine.

constant friction knee components. The kinematics of the sound and prosthetic legs was measured and compared with data from 30 normal subjects (fig 3.44). The data confirmed the fact that the prosthetic hip did not flex after heel contact, and that the prosthetic knee has no flexion at stance phase.

Cappozzo (1982) measured the motion of the upper body (head, shoulder, pelvis) for three above knee amputees walking at two different speeds. Differences with respect to relevant normal patterns are obvious. Gait asymmetry and deviations in the displacement of the head, shoulder and pelvis in comparison with the normal pattern, are depicted.

Goh (1982) conducted an evaluation of SACH and Uniaxial feet, by measuring the kinetic and kinematic data in six BK and five AK amputees. The kinematic data showed that the uniaxial foot movement resembles a normal foot more closely in providing plantarflexion in early stance. Figure 3.45 shows the angle-time diagram for the sound and prosthetic sides at the joints of an above knee amputee in the sagittal plane when the amputee was fitted with SACH and uniaxial feet. The kinetic data of the AK amputees showed no differences between the two types of feet (fig 3.46), and it has been concluded that with proper selection of heel stiffness and alignment, both types of feet can be made to produce similar kinetic patterns. The results of this work for the hip angular motion are comparable to the results of other researchers (Murray et al 1980), therefore, the results of this work can be considered as a reference for further work, and the method of analysis can also be used.

Krebs and Tashman (1985) made a comparison between the conventional rigid socket and the flexible ISNY socket by measuring the kinematic and kinetic parameters for an above knee amputee. Data were collected for the two sockets at the preferred, slow and fast speeds. No significant differences were found between the two sockets or between the speeds, but it has been found that with the ISNY socket, the patient can load

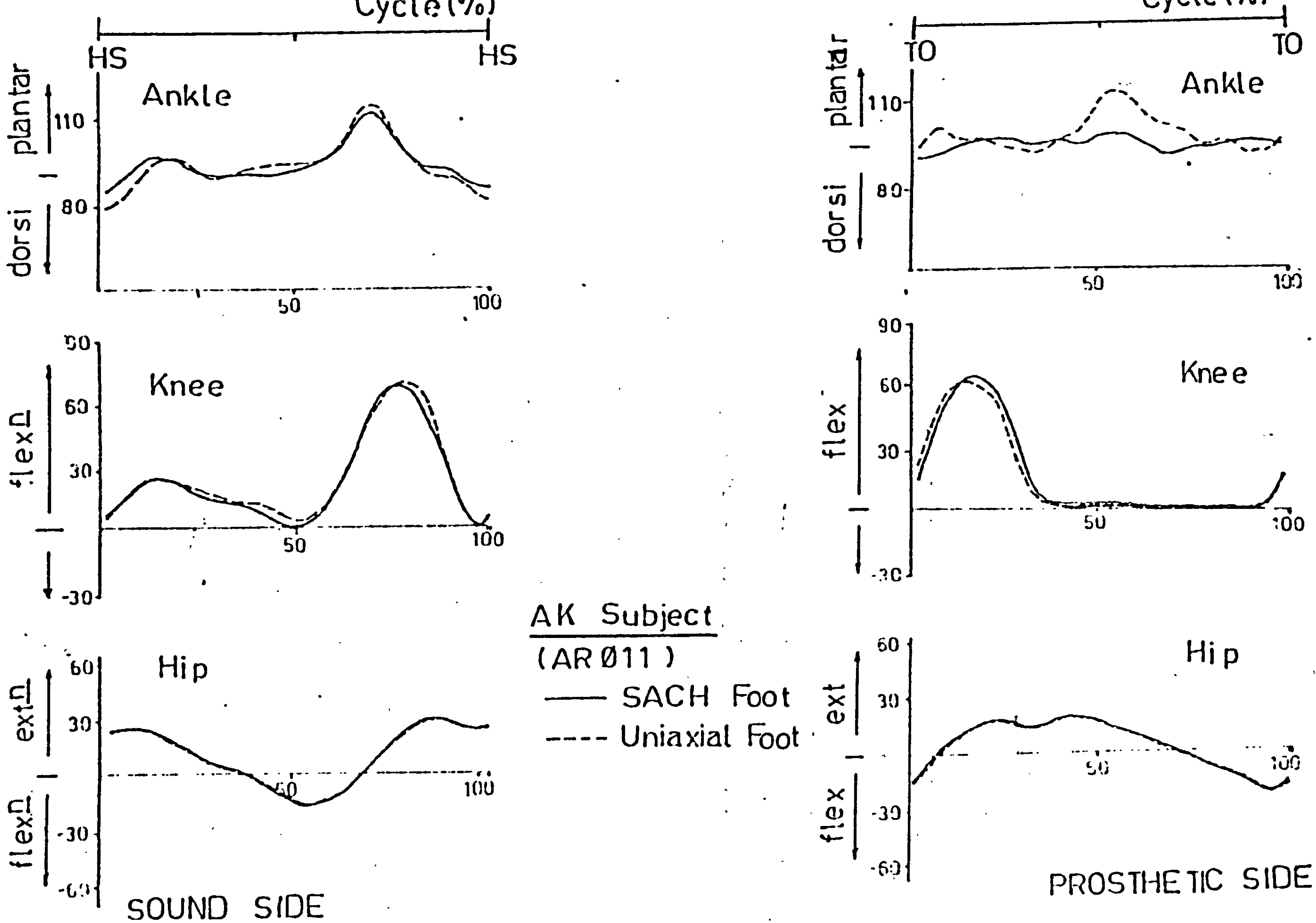


Figure 3.45 Angular displacements in degrees at lower limb joints of an AK amputee. (from Goh 1982)

Figure 3.46 Lower limb joint moments of an AK amputee. (from Goh 1982)

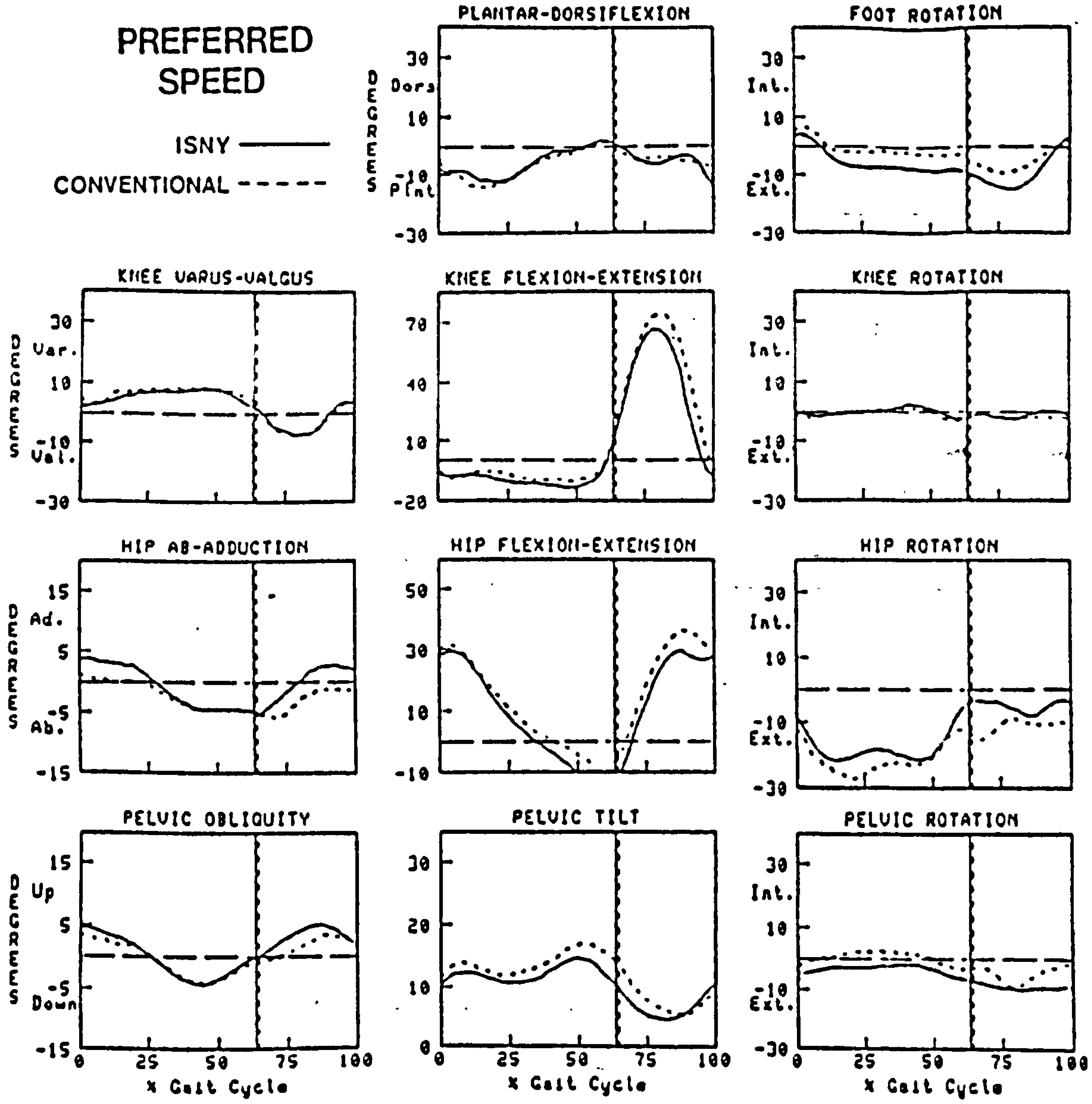


Figure 3.47 Kinematics of an AK amputee. (from Krebs & Tashman 1985)

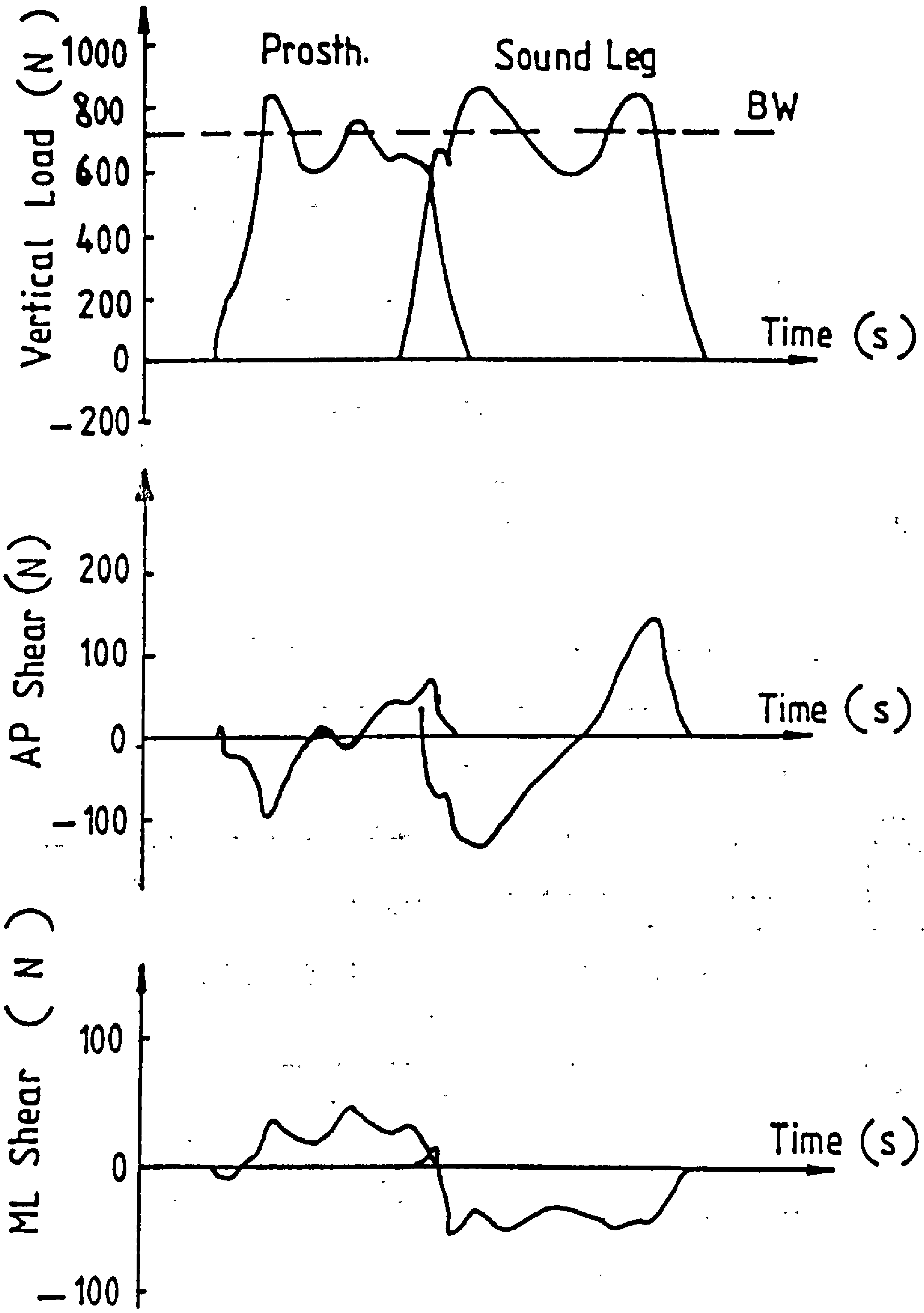


Figure 3.48 Ground reaction forces acting on the sound and prosthetic feet of an above-knee amputee. Data were averaged for five runs. (Redrawn from Akidele 1987)

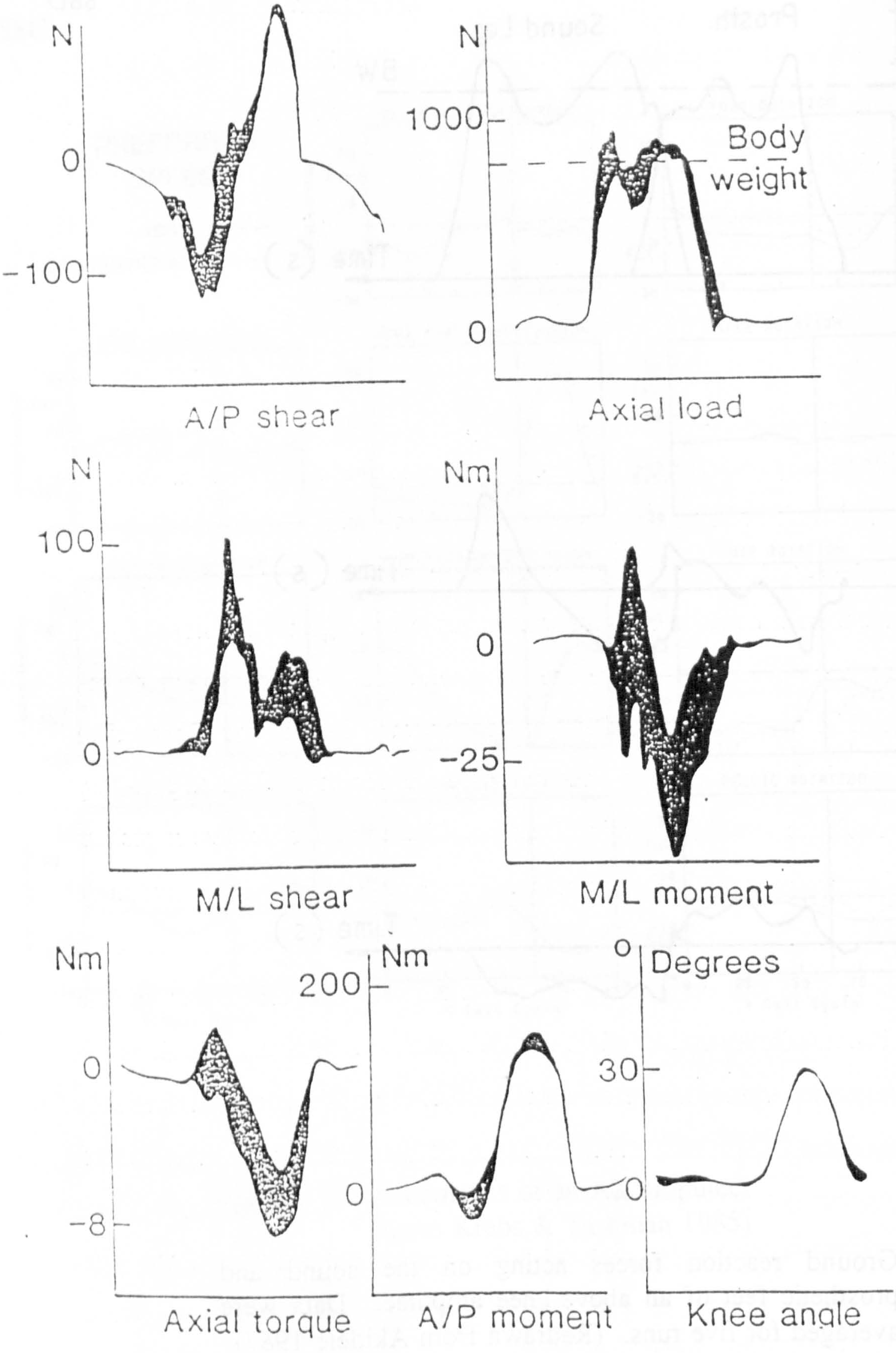


Figure 3.49 Forces and moments for 60 steps by an AK amputee at the ankle level. (from Zahedi et at 1989)

on his prosthesis more quickly. This was considered as an indication of a comfortable socket. Figure 3.47 shows the kinematic results which were obtained at the preferred speed. Because only one patient was tested, the results cannot be generalised, but it can be considered as a pilot study, and the methodology can be used on an extended study.

Akidele (1987) measured the forces acting on the feet of the above-knee amputees using the two Kistler force plates installed in the biomechanical laboratory of Strathclyde University. Figure 3.48 showed the forces obtained for the sound and prosthetic legs of one above-knee amputee. The data shown was averaged for five runs obtained during level walking.

Zahedi et al (1988 and 1989) studied the gait of lower limb amputees, and the effect of prosthetic alignment changes on the gait variables. Pylon transducers were used to measure the prosthetic loads on the shank of the prosthesis. The results showed that step-to-step variations in the kinetic data of the prosthetic leg of an above knee amputee are depicted. The step-to-step variations could be contained within a "repeatability envelope" and were greater for the AK amputee than the BK amputee. Figure 3.49 shows the repeatability envelopes for the forces and moments for 60 steps by an above knee amputee at the ankle level of the prosthetic leg. It is clear that the variation is greater in the medio-lateral plane. It is believed that the step to step variation depends on the patient and on the degree of control which he has on the prosthesis.

Yang Lang (1988) and Lang et al (1991a) investigated the effect of alignment changes on the gait pattern of above knee amputees. The kinetic and kinematic data were measured for four above knee amputees and the gait pattern found to be in agreement with many of the previous researchers work (Zahedi et al 1989, Goh 1982). The effect of alignment changes has been discussed above (section 3.3.3), and the gait pattern which was obtained can always be used for the analysis and understanding of the gait of AK amputees.

3.4.3 Energy Expenditure

The analysis of energy expenditure in lower limb amputees can play an important role in determining the degree of leg pathology and in evaluating the effectiveness of certain prostheses and their components. By determining the energy required for movement and by understanding the mechanism of energy transfer within the body segments, improvements may be achieved in the design, function, and alignment of the prosthesis, and only the minimum energy expenditure would then be required for an amputee's movement. Bresler et al (1957) studied methods of calculating the mechanical energy of the body segments during locomotion, and discussed the use of energy methods in evaluating the prosthesis. Complete energy calculations on the sound and prosthetic legs were carried out on four AK amputees using five different prostheses and walking at slow, medium, and fast speeds. The results were not conclusive because the data were affected by a large number of variables, and four subjects only are not sufficient to provide statistically valid results. However, the data can be used for understanding the mechanism of energy expenditure during walking. It has been found that the net energy generated by the prosthetic leg is not sufficient to provide the required movement at its joints. The average energy deficiency is about 15 ft lb (20.33 joules) , which is usually compensated for by the sound leg. The energy required for an amputee during walking is 25 to 50 percent greater than that required for the normal subject. This was attributed to the inability of the prosthesis to provide the same function as the normal leg. It has also been pointed out that by improving the function of the prosthesis and by proper training of the amputee, a substantial reduction in the total energy requirement may be achieved. It was recommended that the function of the prosthesis can be improved by improving the initial knee stability, minimising vaulting, push-off compensation for the prosthetic ankle, preparation for swing and swing phase control, because these are the parameters which were found to have an effect on the energy cost of

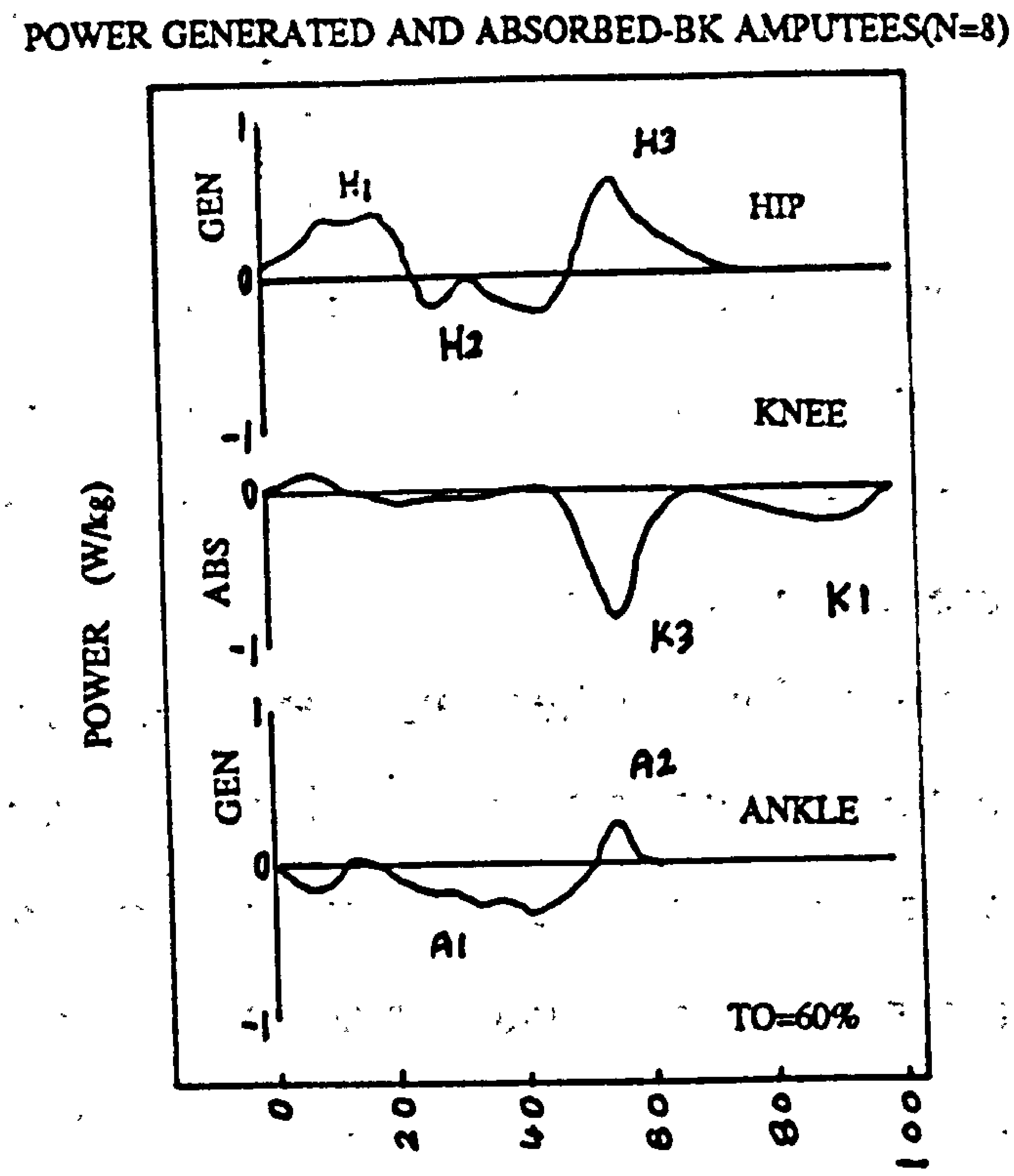


Figure 3.50 Mechanical power for the prosthetic leg of BK amputees. (redrawn from Winter and Sienko 1988).

walking. Cappozzo et al (1976) calculated the mechanical work at the hip, knee and ankle of the sound and prosthetic legs for two AK amputees. It was found that the normal ankle of the amputee had to perform extra work in order to keep the upper body in a suitable trajectory and to gain clearance for the swinging prosthesis. It was also found that the muscles of the stump have to generate more work to maintain stability during the stance phase and to flex the knee prior to the swing phase as the prosthetic ankle does not have an active push off.

Winter and Sienko (1988) have studied the mechanical power at the hip, knee and ankle of five BK amputees. Again it was found that the lack of push off at the prosthetic ankle is compensated for by more power generated at the hip joint of the same leg (fig 3.50).

Lang et al (1991b) have studied the mechanical energy output, and its relation to prosthetic alignment, during the gait of four AK amputees. The results showed that the mechanical energy at the joints of the prosthetic and the sound legs are in agreement with those reported by Bresler et al (1957). Although the effect of the alignment changes was not very conclusive, several common trends were identified in some patients such as: plantar/dorsiflexing the foot increased/decreased the mechanical energy absorbed by the prosthetic foot, and flexing/extending the socket increased/decreased the energy absorbed by the prosthetic foot during the single support (SS) phase and decreased/increased the energy output of the prosthetic knee.

CHAPTER FOUR

THE EXPERIMENTAL WORK

This chapter deals with all experimental aspects of the work presented in this thesis. A description of the above knee prostheses used for the test together, with the coordinates of the reference system for each component is given. The method used for measuring the physical properties and the alignment of the prosthesis is also presented in this chapter. A description of the biomechanical laboratory, and the Strathclyde TV Computer System is also included. In addition the body model, the marker system and the test procedure are described.

4.1 The Above Knee Prosthesis

All subjects were fitted and tested with an Otto Bock modular system (fig. 3.30). This system was chosen because all the components can be readily tilted in the A/P and M/L planes and rotated in the transverse plane in relation to each other. This feature facilitates the alignment procedure of the prosthesis, and the alterations to a given alignment. The ability of this system to permit changes in the alignment in a controlled manner, has made it very suitable for serving the aims of this project. For example, the screws item number 8 and 16 (diagram in fig. 3.30) allow A/P and M/L rotations at the foot and the socket respectively, and the toe in/out angle can be adjusted using the screws item number 13.

Two types of sockets were fitted to the subjects, the quadrilateral (quad) and the ischial containment (IC) type socket. A constant friction uniaxial knee unit and SACH feet were supplied to all subjects except one who was supplied with a uniaxial foot and safety uniaxial knee unit.

4.1.1 Coordinate Systems

Figure 4.1 shows the cartesian coordinate system which is used for the analysis of the work presented in this thesis. The X axis is along the line of progression and positive forward, the Y axis is vertical and positive upward, and the Z axis is horizontal and positive to the right regardless of whether it

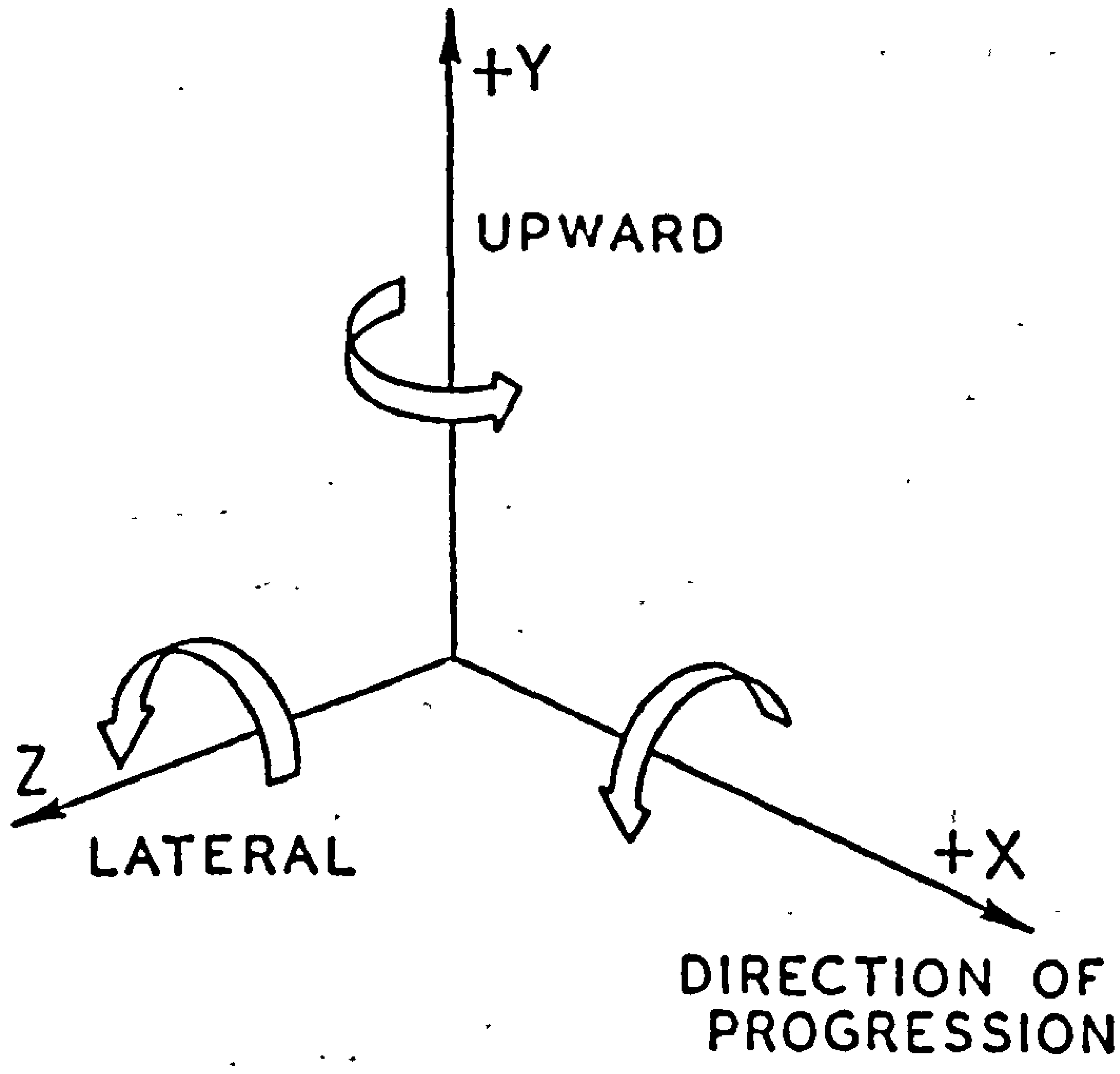


Figure 4.1 Cartesian coordinates system used for the ground frame of reference. (from CPRD 1975).

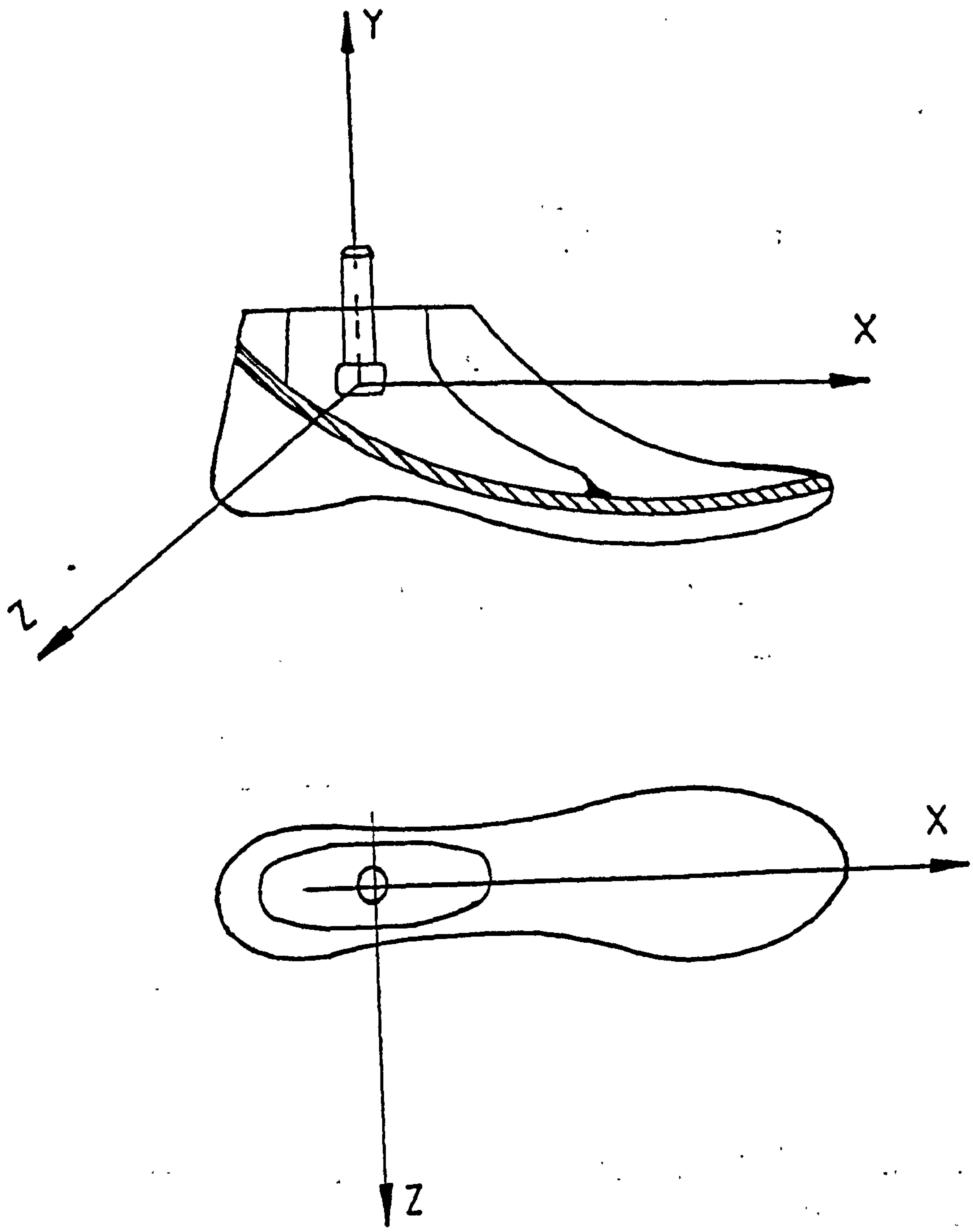


Figure 4.2 SACH foot reference system.

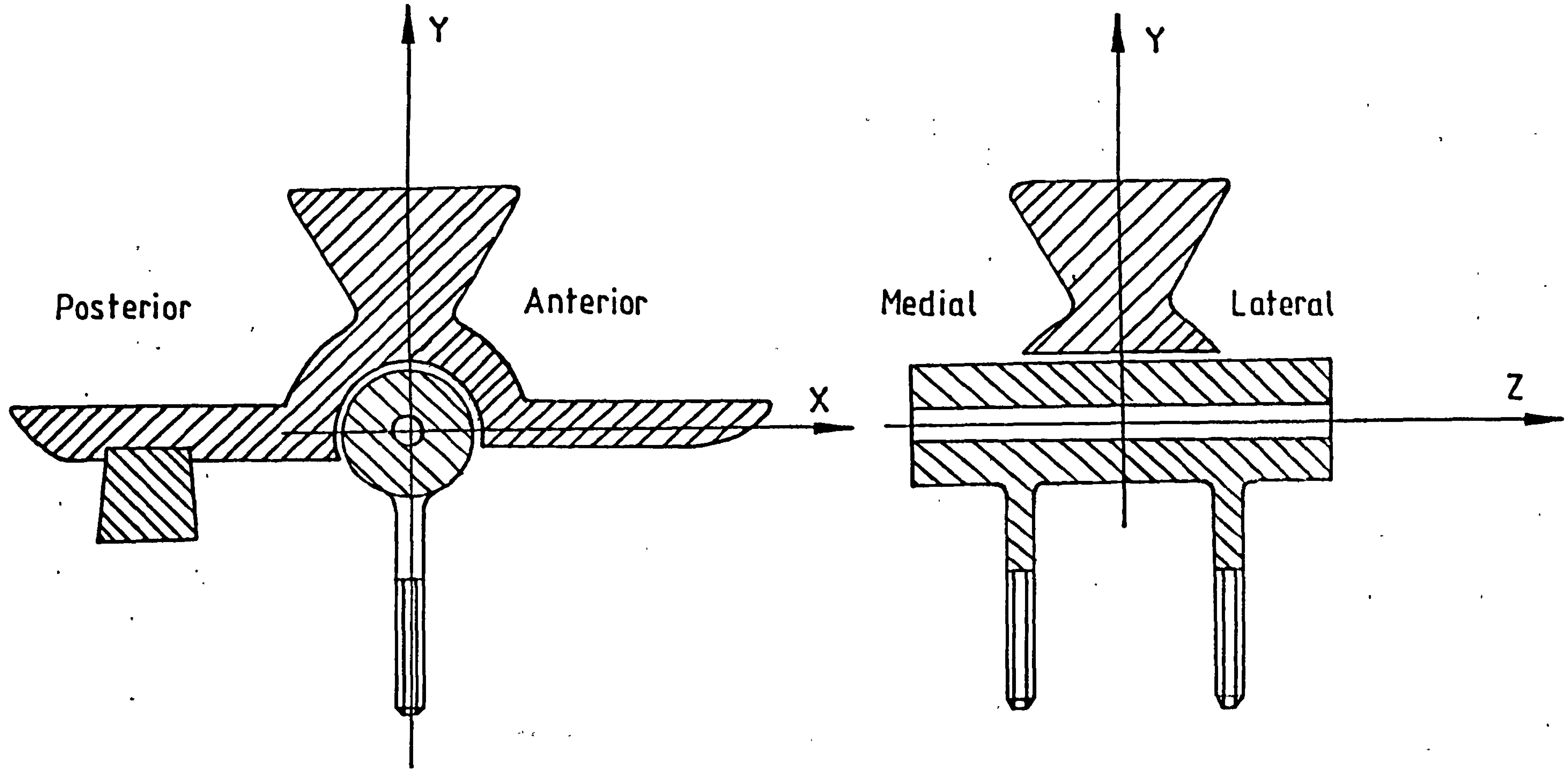


Figure 4.3 Uniaxial foot reference system.

is a right or left amputee. The positive rotation about these axes is anticlockwise looking against the positive direction of the axis (clockwise looking along the positive direction). This system was recommended in 1975 by a task force set-up by the Committee on Prosthetics Research and Development USA (CPRD75), and it is now widely used for biomechanical analysis. In this work, the above system was considered to be the laboratory system with its origin in the mid-point between the two force plates (fig. 4.13), and it is called the ground frame of reference.

4.1.1.1 The Prosthetic Foot

For the SACH foot, a coordinate system of reference was defined so that the origin of the system is at the centre of the bolt hole. The X and Z axes are located in the plane which contains the origin and is parallel to the top surface of the foot as seen in figure 4.2. The X axis is passing through the marked point which is equivalent to the tip of the second toe, Z axis is normal to the X axis and directed to the right. The Y axis is normal to X and Z axes, and is positive upward. For the uniaxial foot, a plane was defined to be parallel to the flat surface of the ankle adopter which is located at the top of the foot and contains the foot's pivot. The origin of the system is at a centre defined between the two bolt holes (U-bolt) and is contained in the defined plane. The Z axis is parallel to the ankle axis and is directed to the right (fig. 4.3), the X axis is normal to the Z axis and is contained in the defined plane. The Y axis is normal to X and Z axes and it is directed upwards.

4.1.1.2 The Prosthetic Shank, Knee and Thigh

Two frames of reference are defined, one for the shank and the other is for the thigh of the prosthetic leg. Thus, the data of the prosthetic knee can be expressed in either of the two frames of reference. As the knee joint is connecting the shank to the thigh, it is convenient to have the knee pivot as a common axis between the shank and thigh frames of references. Only the uniaxial type of knee unit was used through out this project and its pivot axis

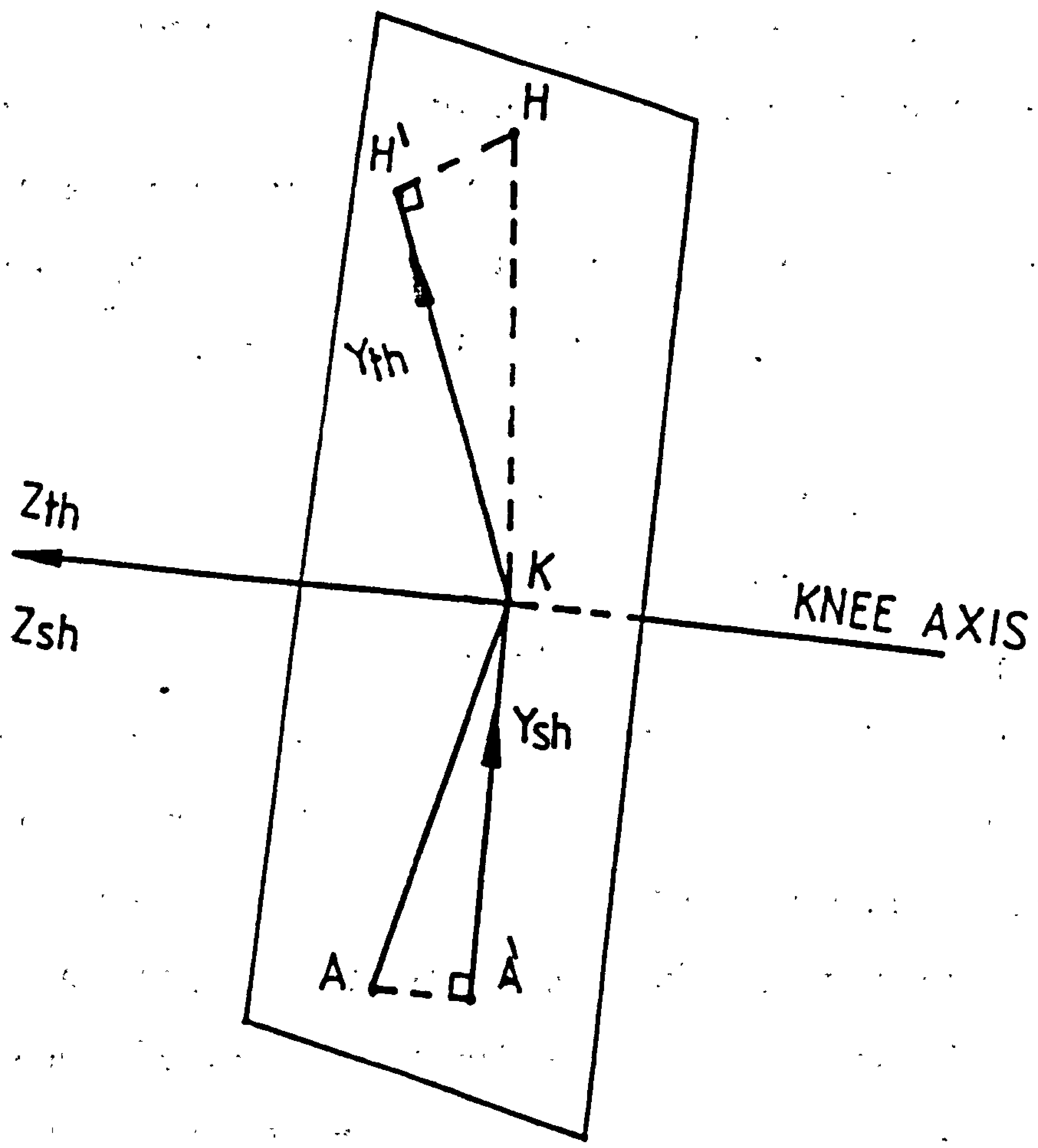


Figure 4.4 Relative orientation of the prosthetic shank and thigh axis system.

was considered as the Z axis of the shank and thigh frames of reference. The origin of these two reference systems is located on the pivot and is equidistant from the medial and lateral ends of the pivot (fig. 4.4). To define the X and Y axes of the shank and thigh, a plane is defined to be normal to the Z axis and contains the origin K. The line which connects the projection point A' of the ankle joint centre A on this plane to the origin K is considered to be the shank Y axis. The line which connects the origin K to the projection point H' of the hip joint centre H on that plane is considered to be the thigh Y axis. Simply, the X axes of the shank and thigh frames of reference were considered to be normal to Y and Z axes in each system. The knee joint can be related to the shank or thigh frame of reference, but throughout this work the knee data is related to the shank frame of reference unless another system is mentioned.

4.1.1.3 The Socket Frame of Reference

To define a frame of reference for the quadrilateral socket, two planes were first defined (fig. 4.5). One was approximately 25 mm above the inner distal end of the socket and the other was about 25 mm below the ischial seat of the socket. The planes were perpendicular to the long axis of the socket which was considered to be the Y axis of the socket frame of reference. This axis (Y) is defined by the line which connects the geometrical centres of the lower and upper sections formed between the socket and the planes and it is positive upwards. The X axis is defined to be parallel to the medial wall of the socket and perpendicular to the Y axis. Its positive direction is pointed forwards. The Z axis is perpendicular to X and Y axes, and its positive direction is to the right regardless of the side of amputation. The same procedure was considered with the ischial containment (IC) socket, but difficulties were faced in positioning the X axis because the medial wall of the IC socket takes the shape of the stump and it cannot be considered a flat plane. The method of determining the socket axes of the quadrilateral and the IC

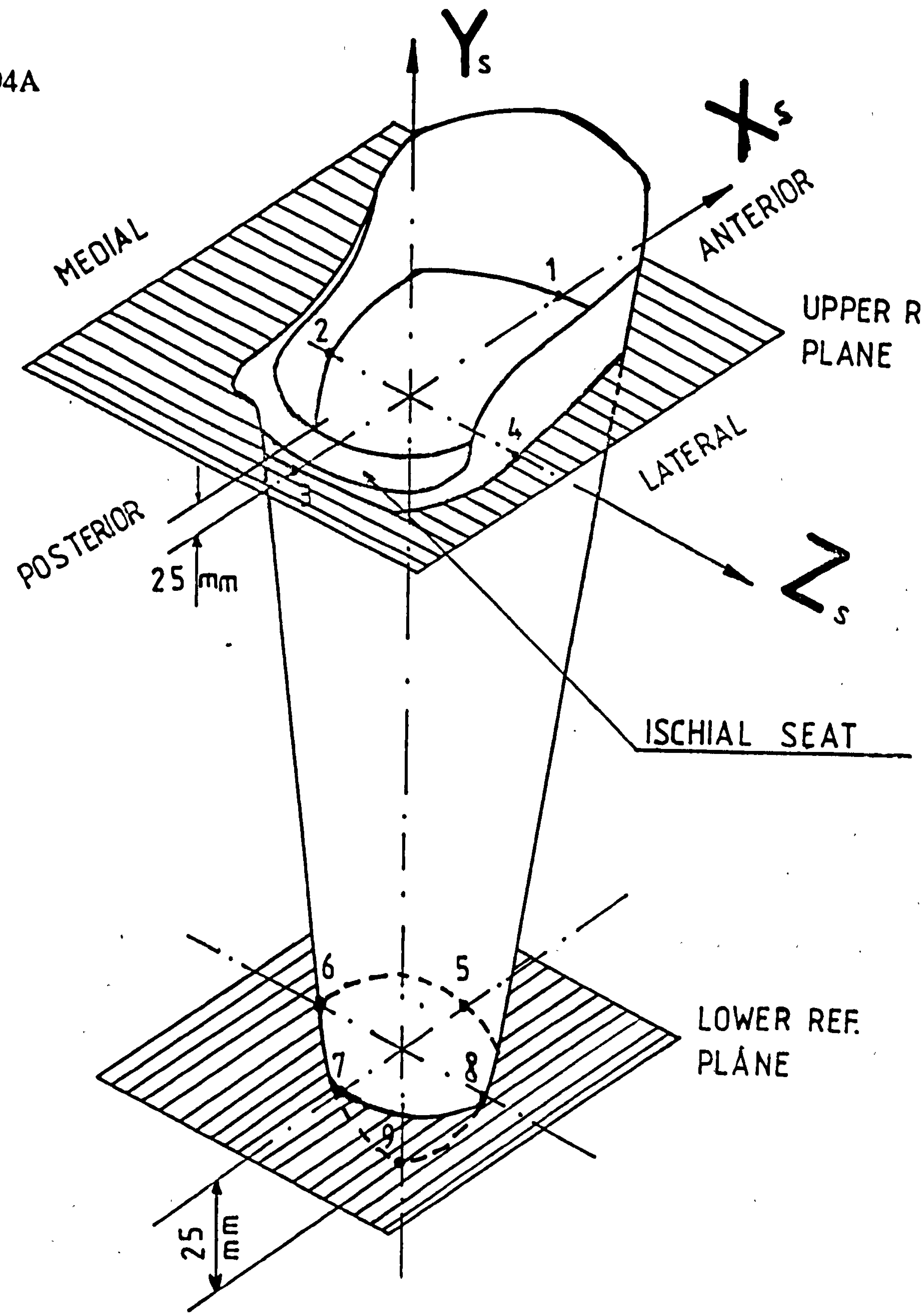


Figure 4.5 Socket frame of reference.

1 to 9 are the contact points between "SAL" and the inner walls of the socket.
2 is the socket left top (SLT).
4 is the socket right top (SRT).
6 is the socket left bottom (SLB).
8 is the socket right bottom (SRB).

sockets is discussed in section 4.2.1.

4.1.2 The Measurement of the Prosthetic Mass Properties

The mass, centre of gravity and mass moment of inertia of each component of the prosthesis was determined. The prosthesis was disconnected at the knee joint into the two separated segments, the socket with the part which is above the knee joint of the prosthesis and the shank-and-foot including the shoe. The mass was measured by weighing the segment using an accurate spring balance scale with a resolution of 20 gram (0.2 N) and an accuracy of 1:227. The centre of gravity was determined to an accuracy of ± 1 mm by balancing the segment against a sharp edge and the distance from the centre of gravity to the knee joint centre was recorded for the two segments. To calculate the moment of inertia I of the prosthetic segments, the pendulum method was used. If a pendulum swinging with small amplitude oscillations Θ and subjected to gravity and inertia effects, the equation of motion can be written as (see fig. 4.6a):

$$I\ddot{\theta} = -mgL\theta \quad \Rightarrow \quad I = -\frac{mgL\theta}{\ddot{\theta}}$$

Where:

I is the moment of inertia about the pivot O.

$$\ddot{\theta} = -\omega^2\theta \quad : \quad \omega = 2\pi f = \frac{2\pi}{T}$$

T is the time of an oscillation [s]

Thus

$$I = \frac{mgLT^2}{4\pi^2} \qquad [kgm^2] \tag{4.1}$$

To use equation 4.1 for calculating the moment of inertia of the prosthesis

Mass Properties of prosthesis

Name:

Affected side:

Type of socket:

Date:

Shank Pendulum Test [time in seconds for ten oscillations]						
	1	2	3	4	5	Mean
Time						
Socket pendulum test [time in seconds for ten oscillations]						
	1	2	3	4	5	Mean
Time						

Mass Parameters		
Parameter	Shank	Socket
Mass [kg]	m_{Psh} =	m_{So} =
Distance from CG to KJC [m]	G_{Psh} =	S_{So} =
I_{XX} about KJC [kg m^2]	I_{PK}^{PSH} =	I_{PK}^{SO} =

Figure 4.6 Form for determination of mass properties of the prosthesis.

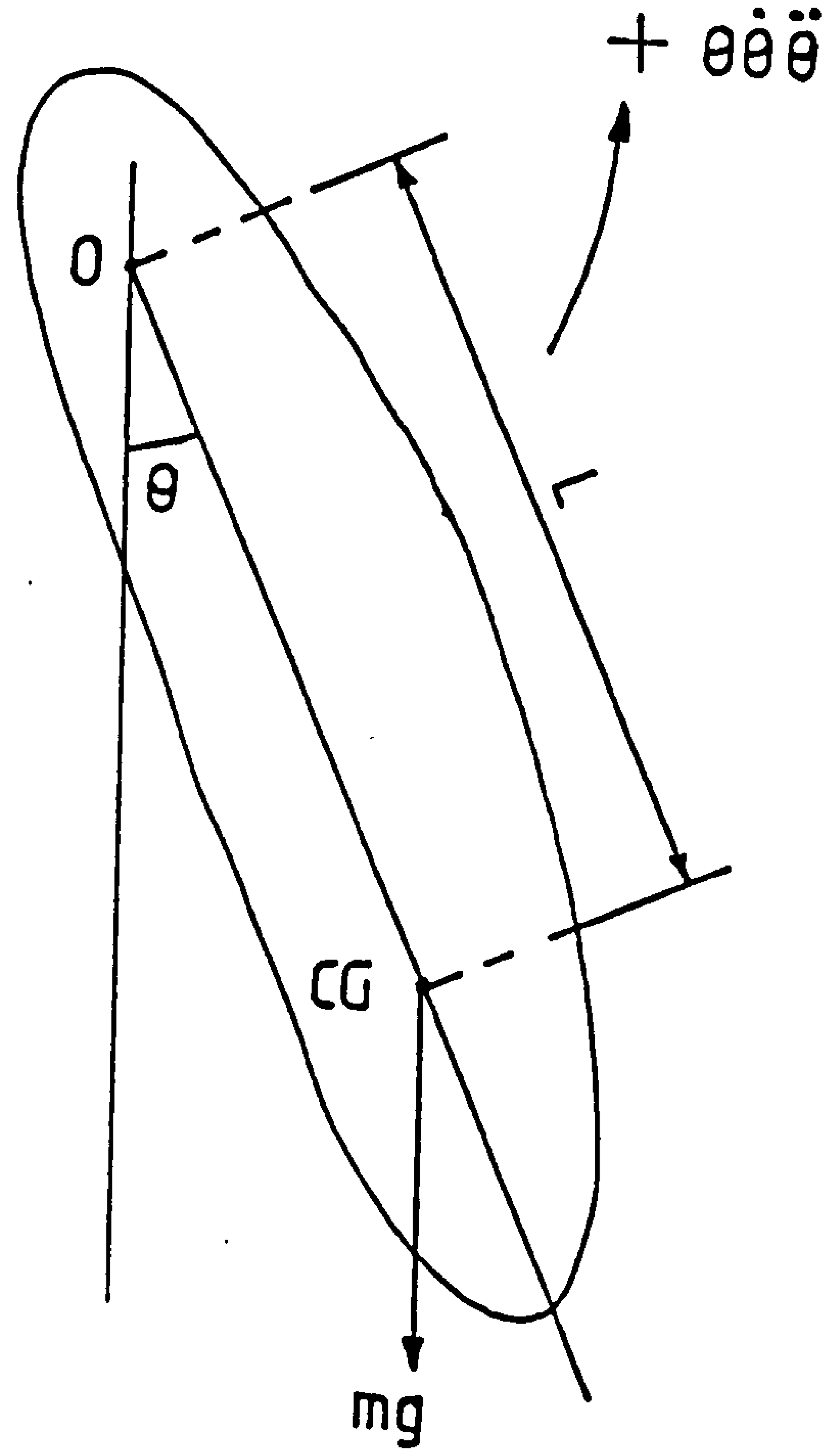


Figure (4.6a) The motion of the pendulum.

components the socket and shank/foot segments were freely oscillated about the pivot of the knee joint, and the time taken for ten small oscillations was measured using a stop watch. The moments of inertia I_{xx} and I_{zz} were considered to be equal and I_{yy} was considered to be negligible (I_{yy} is less than 20% of I_{xx} or I_{zz}, Chandler et al 1975). All the measurements were recorded in the special form shown in figure 4.6.

4.2 Alignment Measurement

A special apparatus has been used for quick and accurate measurement of the prosthetic alignment parameters. It mainly consists of a socket axis locator which allows the determination of the socket axes, and a coordinate measuring system which allows accurate measurement of all the required dimensions in order to calculate the alignment parameters (see section 4.2.2 about the accuracy of the coordinate measuring system). This apparatus is not convenient for use during the biomechanical test as the patient has to take off his/her prosthesis. Therefore, a special method was used to measure the angular changes created by alignment adjustments during the dynamic test procedure.

4.2.1 The Socket Axis Locator (SAL)

The socket axes locator is a device designed by Szulc (1983) to define the socket axes. It consists of eight pointed bars which can move in pairs up and down or in and out in relation to the central axis which has the bars connected to it, arranged like an umbrella shape, as seen in figure 4.7. The eight bars define two planes perpendicular to each other, one of them forms the "SAL" anterior-posterior plane and the other forms the "SAL" medio-lateral plane. Two main conditions should be satisfied when locating the "SAL" in the socket. a- the "SAL" AP plane should be parallel to the medial wall of the socket. b- the pointed ends of the eight bars and the lower end of the "SAL" axis should touch the inner surface of the socket. These conditions were achieved accurately on the quadrilateral socket. This was proved by the ability

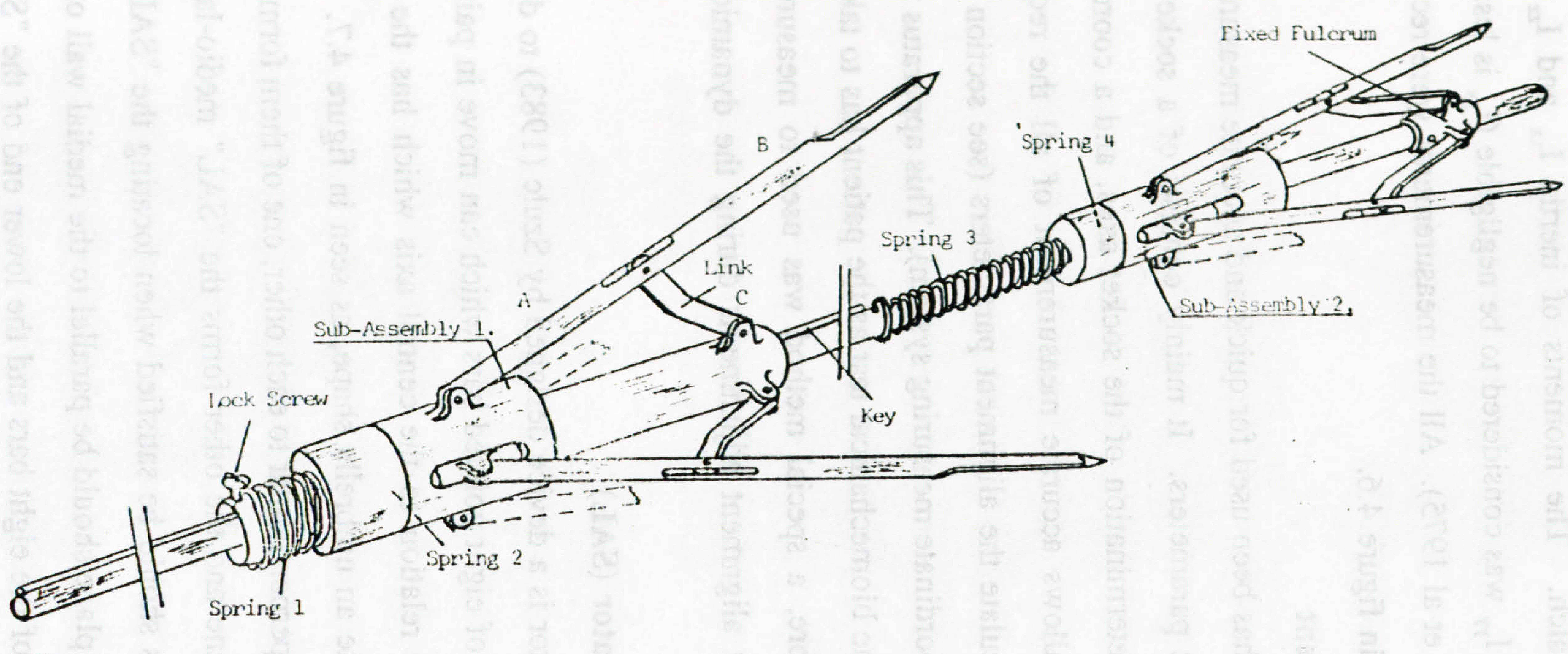


Figure 4.7 The socket axis locator (SAL). (from Szulc 1983)

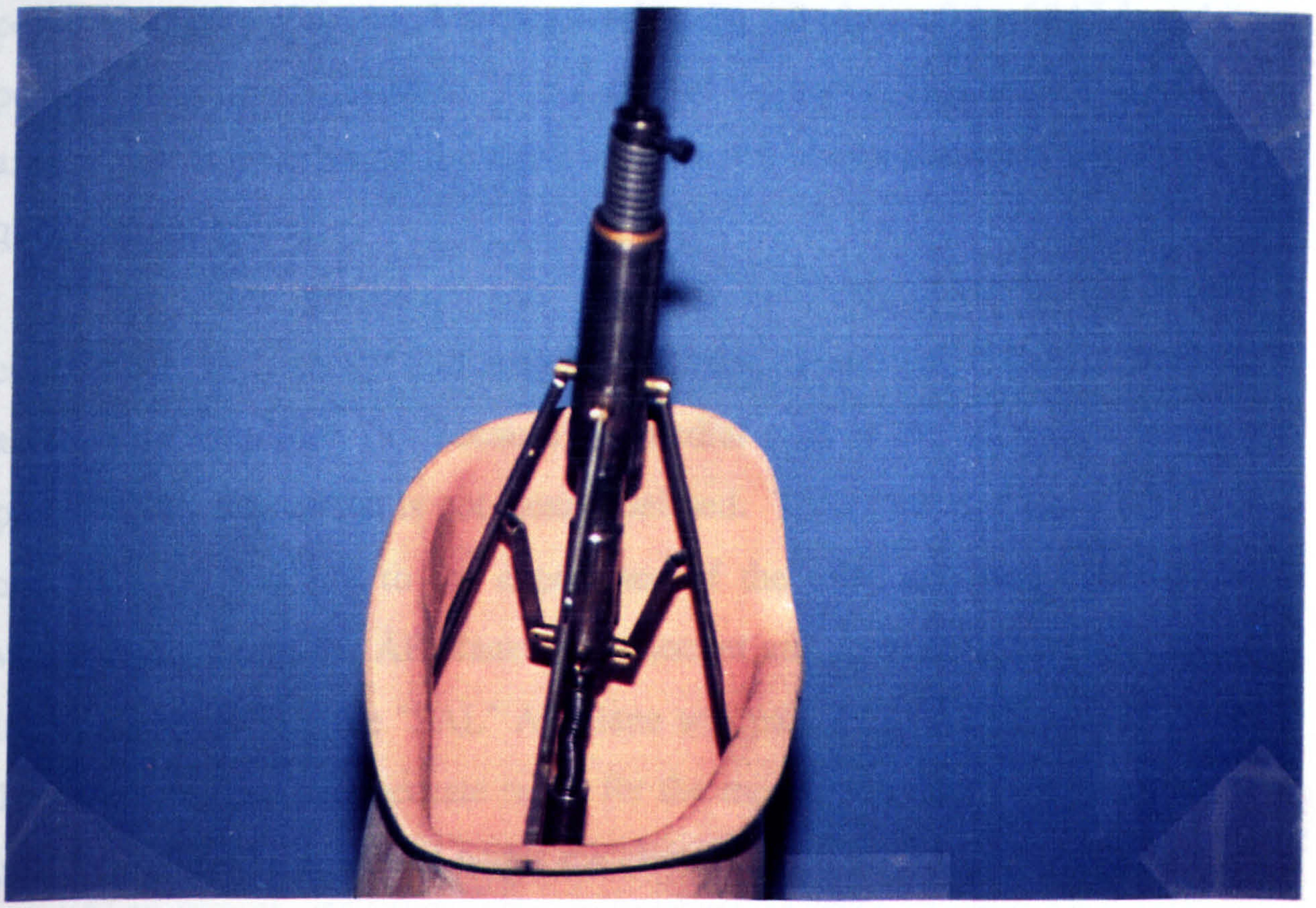


Figure 4.8 "SAL" located in a quadrilateral socket.

of reproducing the same contact points between the socket walls and the pointed bars when the "SAL" was repeatedly located. After mounting the "SAL" in the socket, four points were marked on the inner walls of the socket. They are the contact points between the socket and the four bars which form the medio-lateral plane of the "SAL". The "SAL" central axis defines the socket Y axis and it is positive upwards. The X axis of the socket is perpendicular to Y axis and is contained in the AP plane of the "SAL", and its positive direction is forward. The Z axis of the socket is normal to X and Y axes and it is positive to the right. Figure 4.8 shows the SAL located in a quadrilateral socket.

The same procedure was followed to locate SAL in the ischial containment (IC) socket and defining its axes. However, it was difficult to achieve the condition (a), because the medial wall of the IC socket is not a plane and its orientation is not clearly defined. Therefore, the achievement of condition (a) was left to the experience of the user, and was achieved by visually predicting the AP plane of the socket and thereafter, locating "SAL" in the socket so that the "SAL" AP plane was parallel to the socket AP plane. This may not give an accurate result for the measured parameters but it is the best that could be achieved under the circumstances.

4.2.2 The Coordinate Measuring System

The coordinate measuring system is a surface table with an axis reference system (fig. 4.9). The X axis is vertical and directed downward, Y axis is horizontal and parallel to the long axis of the table and Z axis is horizontal and parallel to the short dimension of the table and it is positive to the right. A sheet of perspex is mounted horizontally on the top surface of the table and it is marked with a grid of 1 cm squares. This enables the user to take any measurement in the Y and Z directions to an accuracy of more than 0.3 mm. A V shaped clamp is used to hold the pylon tube of the prosthesis parallel to the measuring system. The prosthesis is mounted so that the foot

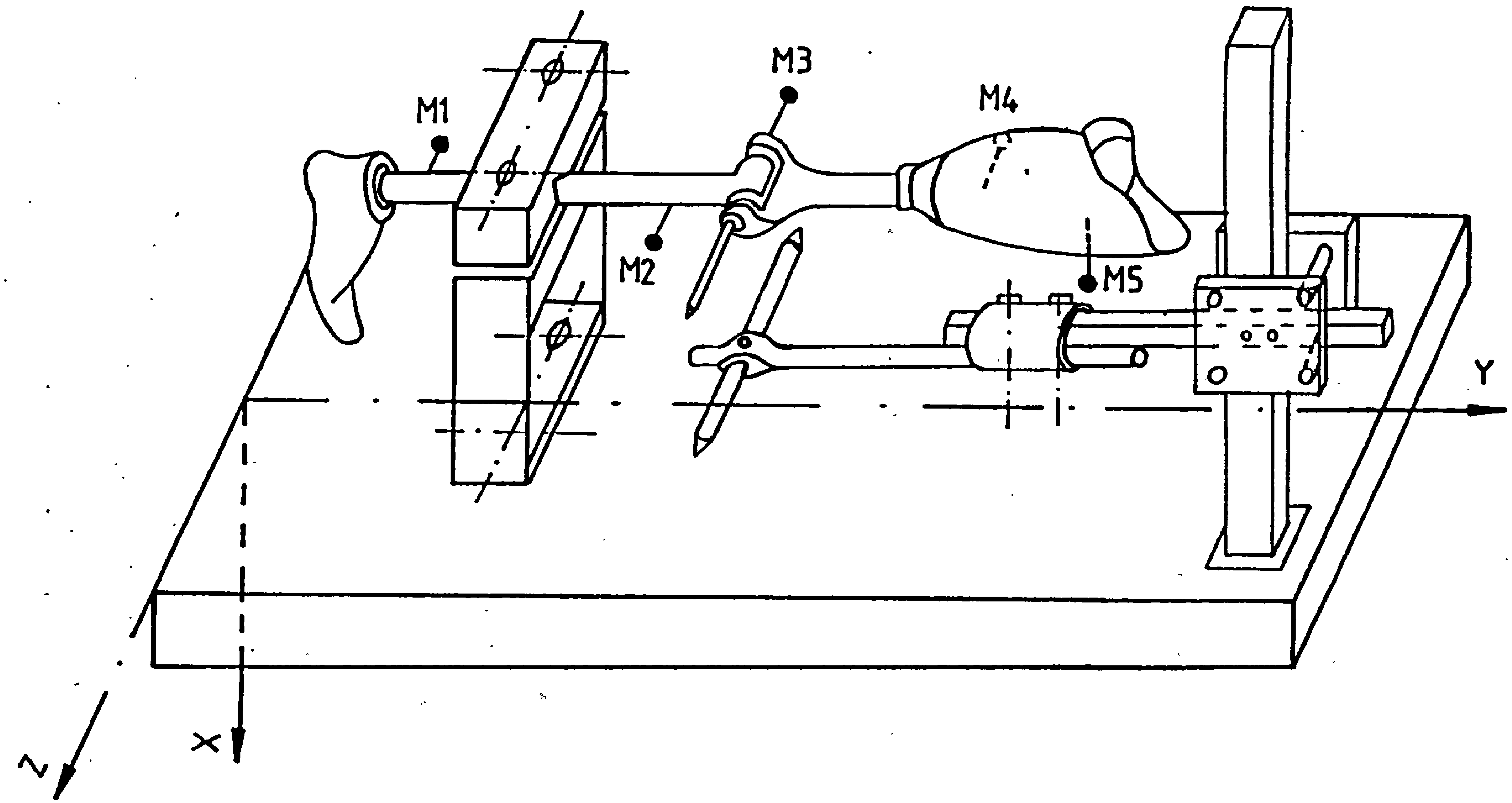


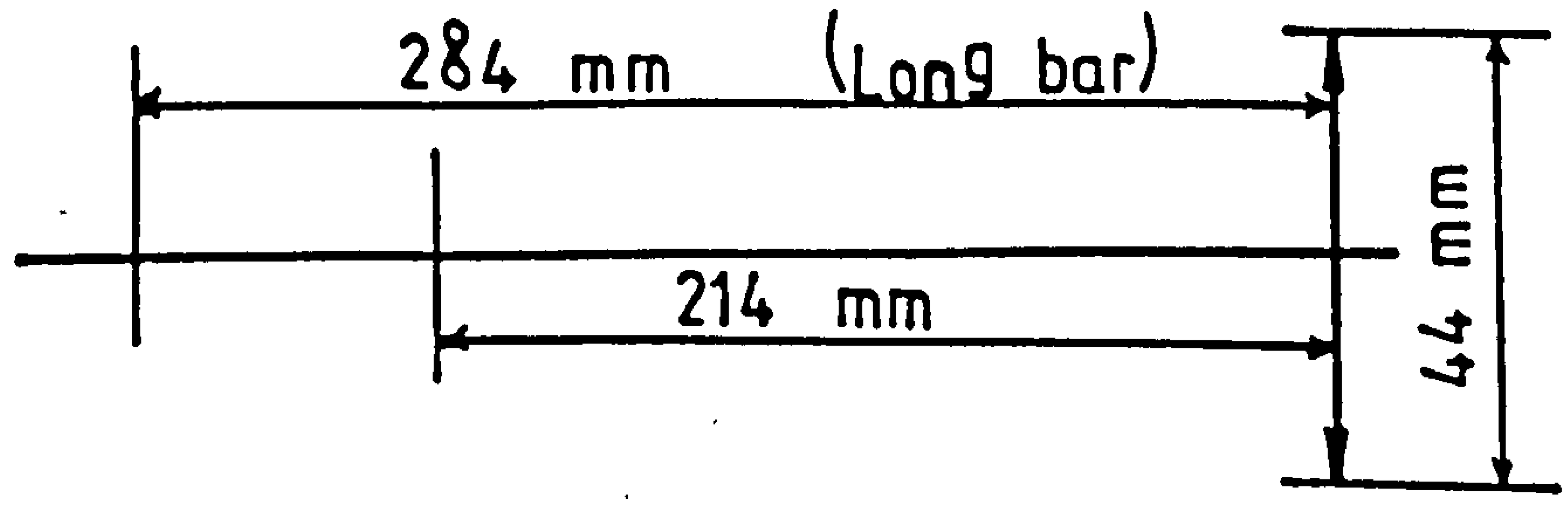
Figure 4.9 The coordinate measuring system with the prosthesis mounted on it.

is directed downward and the knee is fully extended and stabilised under the weight of the prosthesis. The shank tube of the prosthesis should be located horizontally and be parallel to the Y axis of the system. This is done by means of the clamp which is adjustable. The knee axis must be brought to be horizontal. Two pointed metallic bars each 8 cm long, were used at the right and left ends of the knee axis to improve the accuracy of the measurement of the knee axis. A specially constructed device was used to measure the coordinates of the points marked inside the socket (fig. 4.9 and 4.10). The coordinates of the ankle joint centre, toe tip, knee right and left ends, four points inside the socket and the socket centre of rotation were all measured and recorded in a form as seen in figure 4.10. The coordinates of the five markers which were fixed on the shank and socket of the prosthesis (see fig. 4.16) were also measured. This is to relate the prosthetic ankle and knee joint centres to the marker frame of reference during the dynamic test as these joints will not be defined in the static test.

The accuracy of the coordinate measuring system cannot be checked in terms of accuracy in the alignment parameters because there is no other way of measuring the alignment parameters and comparing the result. However, the repeatability of the measured parameters was checked by measuring the alignment of a prosthesis four times, three times by the author and once by a different operator, and the prosthesis was taken off after measurement number two. Table 4.1 presents the alignment parameters and the standard deviation (SD) of the repeatability test. Measurement number 3 was done by the different operator. The largest SD was found to be in the ML socket adduction parameter (0.868 deg.) and it was due to the difference in measurement by the other operator. This error can be related to the user (personal) and his way of approaching the centre of the marked points inside the socket.

4.2.3 Alignment Measurement During the Biomechanical Test

The coordinate measuring system cannot be used during the



Name:
Socket type:
Affected side:
Date:

Measurements [mm]	X	Y	Z
Ankle Joint Centre			
Toe Tip			
Right Knee RK.			
Left Knee LK.			
SRT			
SRB			
SLT			
SLB			
M1			
M2			
M3			
M4			
M5			
SRC			

Figure 4.10 The alignment measurement form.

SRT Socket Right Top (see fig. 4.5)
SRB Socket Right Bottom   SLT Socket Left Top
SLB Socket Left Bottom   SRC Socket Rotation Centre

M1 to M5 are the markers of the socket and the prosthetic shank, numbered from the lowest on the shank to the highest on the socket(see fig. 4.9).

Socket top and socket bottom measured by the short and long bar respectively. The long bar should be also used to measure markers M1 and M4 with the right tip located at the centre of the marker (see fig. 4.9).

Table 4.1 The repeatability of the coordinate measuring system.

Measurement No.	1	2	3	4	SD
K Height* cm	44.15	44.4	44.3	44.4	0.102
K Set Back cm	-1.36	-1.32	-1.365	-1.23	0.054
K Tilt (LSH) deg.	3.526	2.967	3.197	2.573	0.347
K Set Out cm	-0.975	-1.125	-1.05	-0.7	0.16
S Flexion deg.	-6.020	-6.2	-6.096	-5.137	0.424
Socket OR deg.	-5.399	-5.034	-4.875	-5.794	0.355
S Adduction deg	6.539	5.254	7.56	5.77	0.868
S Forward Set cm	4.565	4.75	4.76	4.02	0.301
S Set Out cm	-0.675	-0.7	-0.525	-0.3	0.159
S Height cm	71.22	71.2	71.17	71.45	0.109
Toe Out deg.	2.603	2.356	2.017	2.210	0.213

K = Knee S = Socket

LSH = Lateral Side Higher

OR Outwards Rotation

* See section 4.2.4 for the definition and calculation of the alignment parameters.

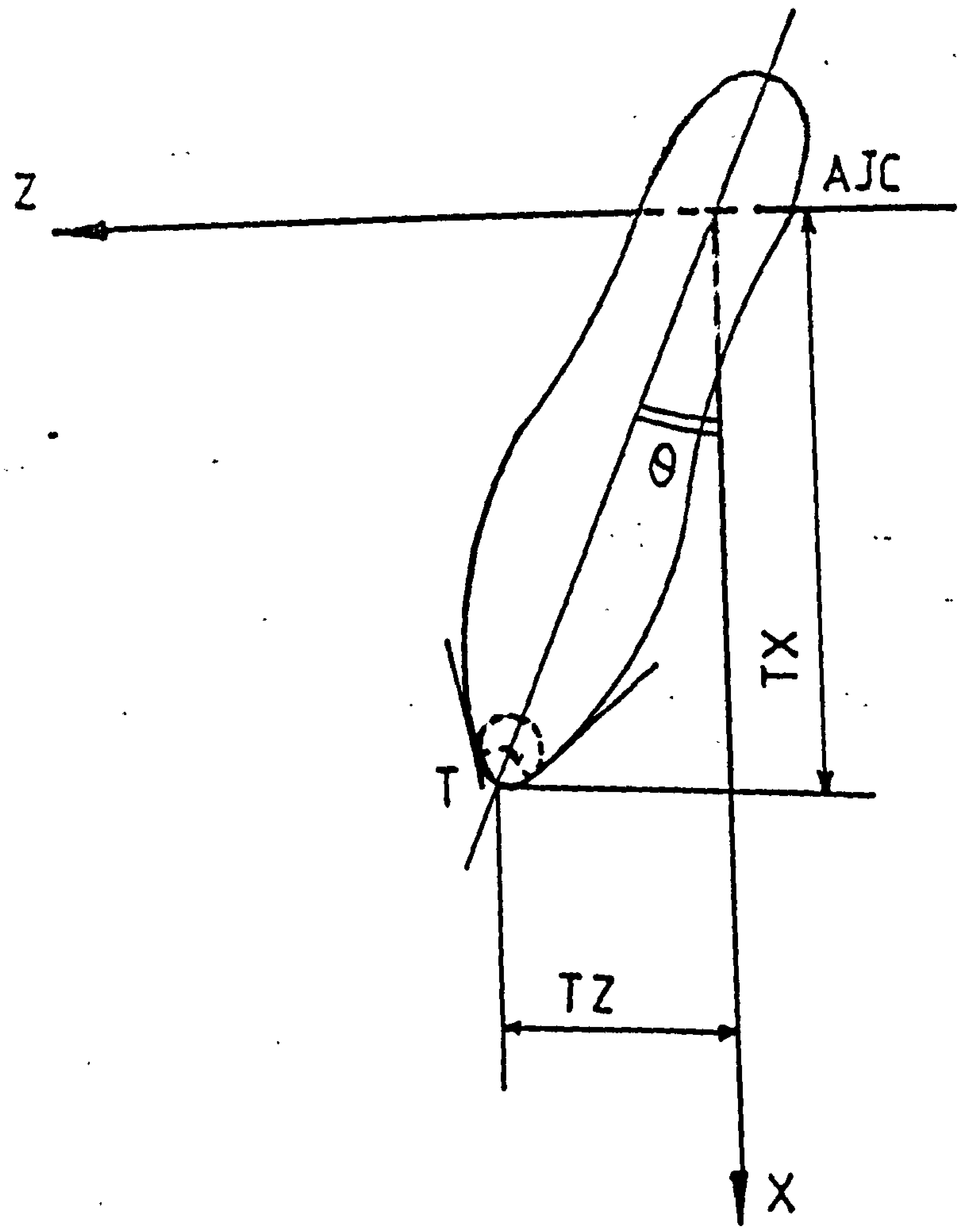


Figure 4.11 Calculation of the foot toe out angle (TOUT).
Θ = TOUT angle

biomechanical test because the measurement may take about 40 minutes and the patient should not be left waiting during this time. The prosthesis should not be taken off and put on again during the test because this will change the marker positions on the patient as the prosthesis can never go back to exactly the same place. Therefore, this method was used to measure only the normal alignment of the prosthesis which was set by the prosthetist. To measure the alignment changes during the gait test, Yang Lang (1988) used a protractor to measure the socket flexion/extension and two metallic bars with a perspex plate to measure the ankle dorsi/plantar flexion. This method was found to be inadequate, inaccurate and time consuming. It is very difficult to align the protractor to the socket centre of rotation (centre of the alignment adjustment) and the angular changes at the socket and at the ankle are always combined with shifts at the centres of rotation. Thus, a new procedure was used to measure the alignment changes during the biomechanical test. The alignment changes were carried out by adjusting the screws at the socket and ankle units. Before calling the patient for a test the prosthesis was calibrated using the coordinate measuring system, and the angular changes resulting from one turn of the adjustment screws was determined. During the test, the position of the screws was changed by a certain number of turns to obtain the desired angular changes.

4.2.4 Calculation of the Alignment Parameters of the AK Prosthesis

After measuring the prosthesis on the coordinate measuring system, the measured coordinates were related to the ankle joint centre (AJC) by subtracting the coordinate of the AJC from all other measures. This was done in order to allow calculation of the alignment parameters as defined in section 3.3.3.

The foot toe out angle was calculated using the coordinates of the tip of the foot TX, TY and TZ (see sketch in fig. 4.11):

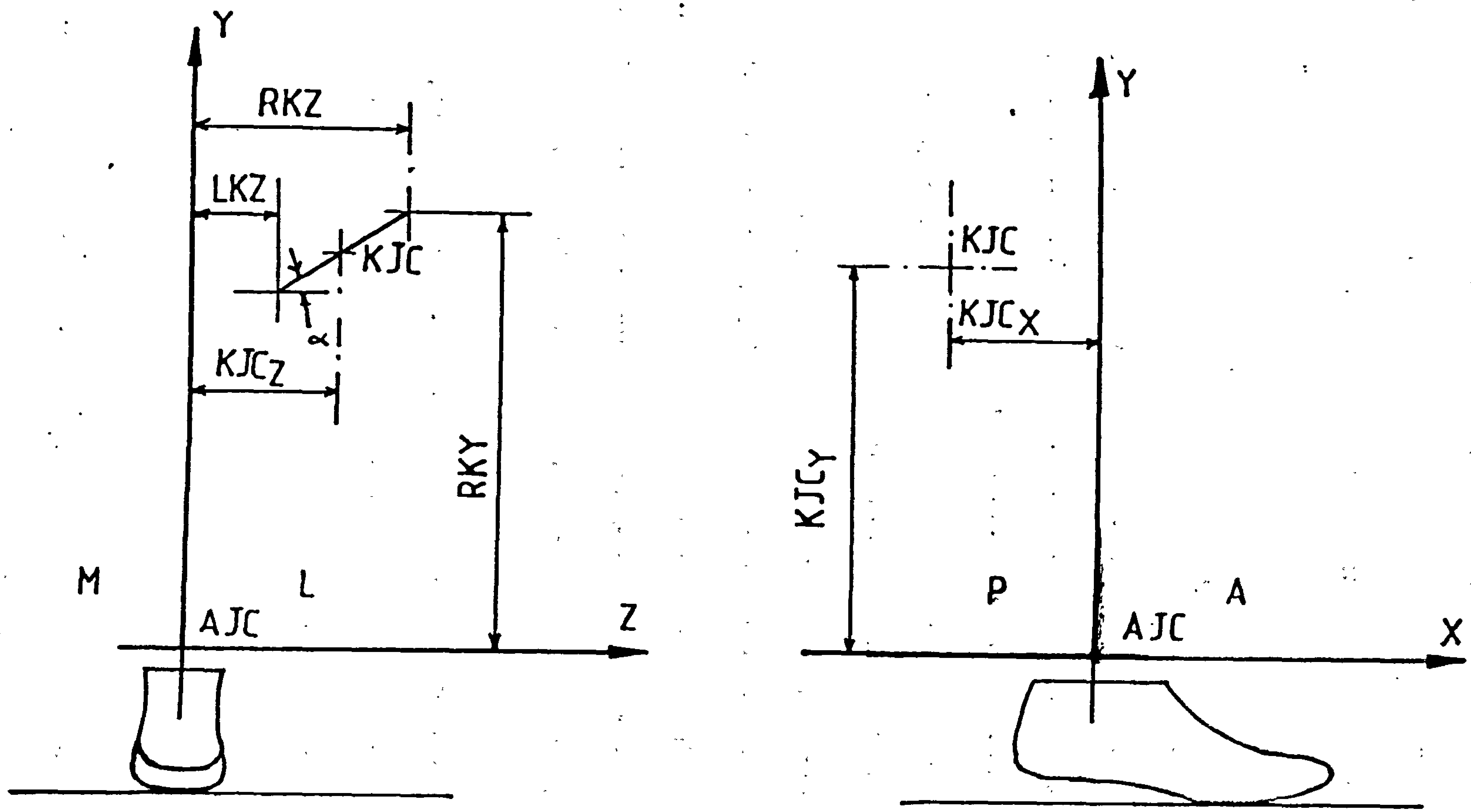


Figure 4.12 Diagram showing the knee set back (KJC_X) and the knee set out (KJC_Z). The knee tilt (KMTLSH) is also shown.
α = KMTLSH

$$TOUT = \arctan(\frac{TZ}{TX}) \,.\, ILEG \qquad (4.2)$$

Where

ILEG = 1 for the right amputee and -1 for the left amputee

To calculate the knee parameters, the coordinates of the knee joint centre (KJC) were calculated as (see fig. 4.12 for the abbreviations):

$$KJC_X = \frac{RKX + LKX}{2}, \quad KJC_Y = \frac{RKY + LKY}{2}, \quad KJC_Z = \frac{RKZ + LKZ}{2}$$

Where: RK(X, Y, Z) and LK(X, Y, Z) are the coordinates of the right and left ends of the knee axis respectively, relative to the ankle joint centre.

Thus, the knee parameters are:

Knee set back = - KJC_X

Knee set out = KJC_Z . ILEG

Knee height = KJC_Y

For the knee set back parameter, the (-) sign was used so that knee set back will be positive and knee set forward will be negative. Similarly, the negative knee set out means that the knee is set inward.

The knee medial tilt, lateral side is higher (KMTLSH):

$$KMTLSH = \arctan\left(\frac{RKY - LKY}{RKZ - LKZ}\right) \,.\, ILEG \qquad (4.3)$$

The socket parameters are calculated from the four points which were marked inside the socket. The points are the right top (RT), the left top (LT), the right bottom (RB) and the left bottom (LB) points. At first two points present the socket upper (SU) and socket lower (SL) points were found. These points define the Y axis of the socket and calculated as:

$$SU = [SU_X, SU_Y, SU_Z] = \frac{[RT] + [LT]}{2}$$

$$SL = [SL_X, SL_Y, SL_Z] = \frac{[RB] + [LB]}{2}$$

Then the alignment parameters of the socket can be calculated:

$$\textit{Socket External Rotation} = \arctan\left(\frac{RTX - LTX}{RTZ - LTZ}\right) . ILEG \tag{4.4}$$

$$\textit{Socket Adduction} = \arctan\left(\frac{SU_Z - SL_Z}{SU_Y - SL_Y}\right) . ILEG \tag{4.5}$$

$$\textit{Socket Flexion} = \arctan\left(\frac{SL_X - SU_X}{SU_Y - SL_Y}\right) \tag{4.6}$$

Socket Forwards Set = SU_X

Socket Height = SU_Y

Socket Set Out = SU_Z . ILEG

All the above calculations are carried out within less than ten minutes, this time is consumed when feeding the raw data which are measured on the coordinate system to computer subroutine called "ALIGN" which has been developed on the main frame computer of the University to calculate all the alignment parameters. The subroutine is ran by a command file called "RUNALIGND".

4.3 The Biomechanics Laboratory

In the biomechanics laboratory at Strathclyde University, two Kistler force plates and three TV cameras were installed as seen in figure 4.13. The force plates and the TV cameras were connected to a PDP11 minicomputer

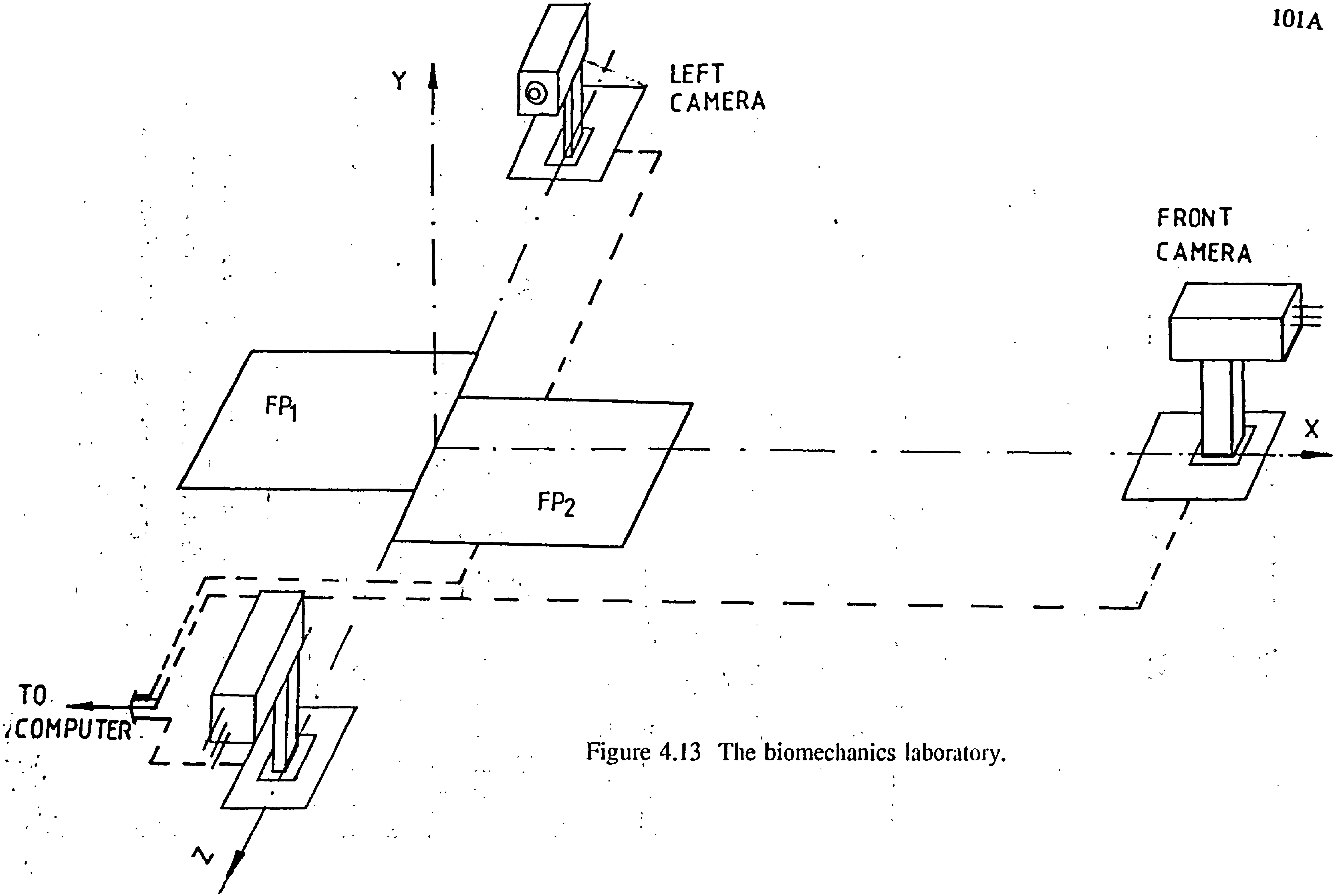


Figure 4.13 The biomechanics laboratory.

which controlled the timing of data collection and stored the collected data. The data could be collected from the force plates or from the TV system or from both force plates and TV system simultaneously.

The two force plates (FP1) and (FP2) were situated in the middle of the walk path which is about 20 metres long. The force plates may be sampled at a rate of 50 Hz, and their signals are taken through 8 charge amplifiers, two summing amplifiers (see section 2.5.2). After that the signal is taken to an A/D convertor and then to the computer to be stored on a disk. Now the data can be displayed on screen, checked and prepared to be transferred to university's mainframe computers (VAXs) for further analysis. The force plates are precisely calibrated and the validity of the calibration factors was checked through out the period of this project. The description and the accuracy of Kistler force plate were covered in section 2.5.2.

The TV cameras are mounted so that the front and the two sides of the subject can be monitored, so that a biomechanical analysis for the left and right sides of the whole body can be carried out at the same time. The three TV cameras are aligned so that their optical axes meet approximately above the origin of the ground frame of reference and run at a rate of 50 Hz. Each camera has a ring of LEDs around its lens. The LEDs emit infra-red light which is reflected off the markers which are covered by reflective material. As the right and left cameras are looking at each other, they may blind each other by the light emitted from their LEDs. Therefore, the opening instants of their rotary shutters were arranged to be out of phase. The opening instant of one of the side cameras was delayed by 10 ms from the opening instant of the other side camera. When reflected light strikes the camera's sensor, an individual unit will be excited on this sensor. A coordinate generator will generate the coordinates of the excited unit and transfer these coordinates for storing in the computer in special code which identifies the coordinates of the excited unit and the camera to which the excited unit belong to. Specially

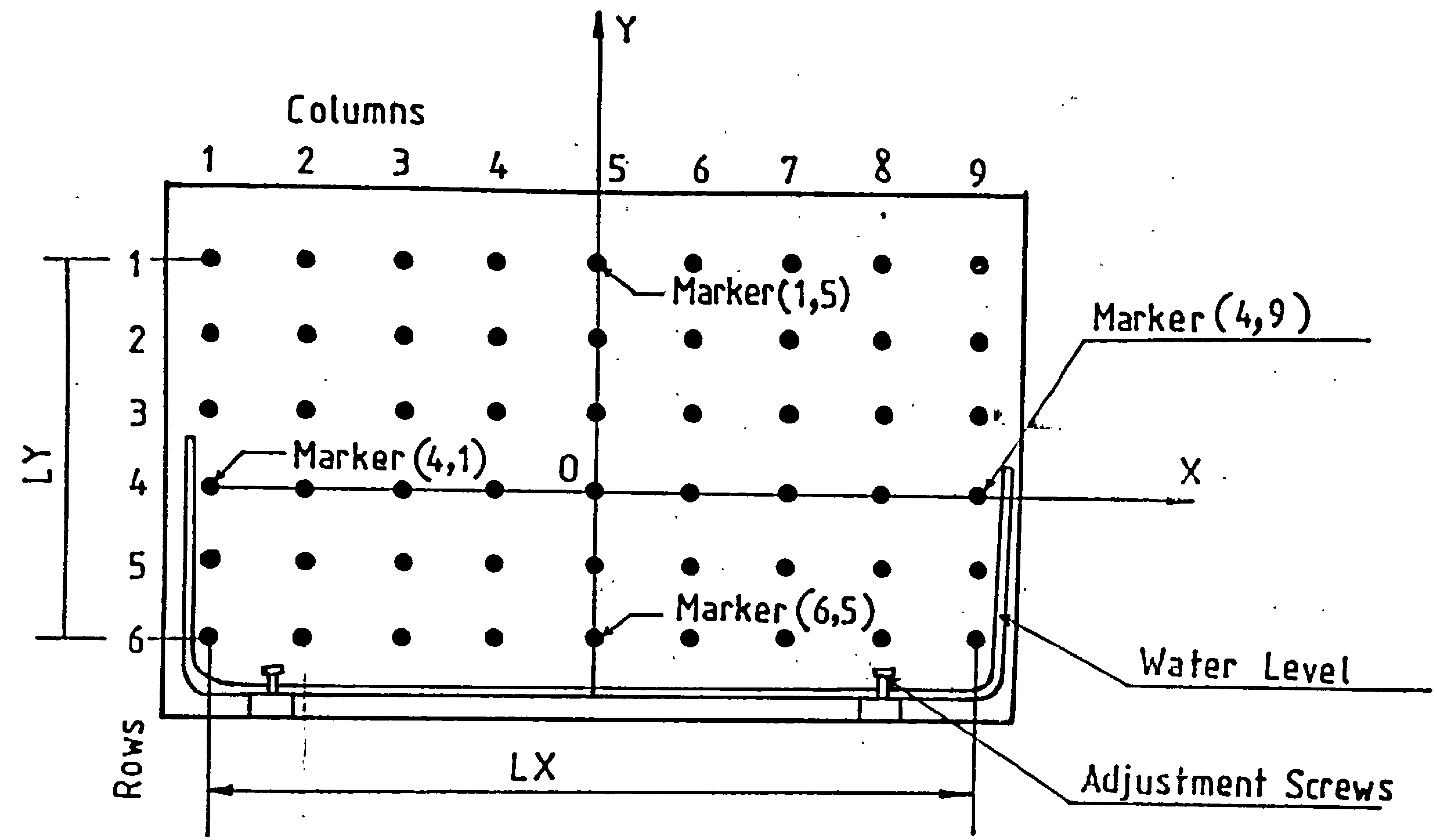


Figure 4.14 Calibration board used for the assessment of the four points and ten points method of calibration.
(from Yang 1988)

developed software installed in the PDP11 computer operates at each stage of the above procedure.

4.3.1 Calibration of the TV System

It is of special importance to use an accurate calibration method in transforming the 2D data of the cameras into 3D data. The accuracy of the calibration method affects the accuracy of the TV system and plays an important role in the validity of the kinematic data. Several calibration methods have been used in gait analysis, but it has been reported, as discussed below, that the most practical method is the direct linear transformation (DLT) which was introduced by Marzan and Karara (1975). This method is widely accepted because it is accurate and reasonably simple. Yang Lang (1988) has compared this method with two other calibration methods using the TV system which was used in the present project. The two calibration methods were the "four points" and the "ten points" which were simple and conventional. The four points method used four control markers located at the ends of a cross, formed by the fourth row and the fifth column of the board seen in figure 4.14. The magnification factors in the horizontal (DX) and vertical (DY) directions for each camera were then simply determined as follows:

$$DX = \frac{LX}{HC(4,9) - HC(4,1)} \quad , \quad DY = \frac{LY}{VC(1,5) - VC(6,5)}$$

Where: LX = 1.600 m is the horizontal distance between marker (4,1) which is on the fourth row and first column and marker (4,9) which is on the forth row and ninth column. LY = 1.000 m is the vertical distance between markers (1,5) and (6,5) which were on the fifth column. HC and VC are the horizontal and vertical TV readings of the control markers respectively ,obtained by the camera under consideration.

It was reported that the parallax errors were corrected when the "four points" method was used. In this method, the three cameras must be aligned in the same horizontal plane and the optical axis of each camera should be parallel to the relevant axis of the ground frame of reference. It was difficult to achieve such alignment for the cameras, therefore, Yang Lang also discussed the "ten points" calibration method in which ten markers on the fourth row and fifth column of the board seen in figure 4.14 were used, to determine constants c and d of the modified magnification factor (MAG) defined below. The least-square statistical method was employed.

The modified magnification factor was: $MAG = c + d\,X$ where X is known coordinates of the ten markers [m], and similarly for the Y and Z co-ordinates.

When the constants c and d are determined the coordinate x of any marker can be found as:

$x = X' \, MAG$ where X' is the TV reading of the considered camera.

Yang Lang reported that the DLT method is superior to the other two and an improvement of 50% was achieved in the accuracy of the TV system when the DLT method was used. Therefore, the DLT method was used for the calibration of the TV system throughout this work. The theoretical concept of this method is presented in section 5.1.

4.3.2 Accuracy of the TV System

Accuracy is the closeness of the measurements to the actual value of a physical quantity and it should be evaluated for any measuring device before use. Assuming an actual value M which can only be measured as an observed value m, the difference E between the observed value and the actual value is called the error of observation. The ratio E/M represents the accuracy of the measurement and it shows the closeness of m to M. Precision is another parameter which must be evaluated before using any measuring device. Considering that a physical quantity measured N times by the same device; precision ot this device is the closeness with which the measurements agree

Table 4.2 Frame to frame variations for each individual camera (2D assessment).

Marker	Mean [machine units]		Range [machine units]		SD [machine units]	
	Vertical	Horizon.	Vertical	Horizon.	Vertical	Horizon.
	Front Camera					
1	230.91	927.42	1.17	3.67	0.21	1.18
2	232.48	593.08	0.5	5.00	0.19	1.03
3	232.92	240.92	0.17	2.33	0.08	0.31
4	121.05	891.55	1.00	3.00	0.33	0.69
5	122.25	608.44	0.5	1.00	0.25	0.5
6	122.25	308.55	0.5	1.9	0.25	0.5
Overall			0.64	2.817	0.218	0.702
	Left Camera					
1	240.25	527.01	1.00	1.00	0.08	0.31
2	169.75	529.63	1.00	1.50	0.08	0.40
3	92.256	534.16	0.5	1.5	0.056	0.33
4	255.25	858.71	1.00	3.5	0.08	0.45
5	174.84	861.63	1.00	3.00	0.21	0.48
6	85.744	866.11	0.5	1.5	0.056	0.39
Overall			0.833	2.00	0.094	0.393
	Right Camera					
1	261.17	917.6	0.67	4.00	0.25	1.06
2	167.54	922.61	0.5	2.4	0.25	0.47
3	67.71	926.25	0.533	3.07	0.207	0.56
4	249.75	522.75	0.5	2.00	0.25	0.52
5	167.08	526.1	0.5	1.00	0.25	0.27
6	79.75	528.2	0.5	2.00	0.252	0.46
Overall			0.534	2.412	0.243	0.556

Range = Max. value - Min. value

with one another independently of any systematic error involved. The precision shows the closeness of any measurement to the mathematical mean of the measurements and it is presented by the standard deviation.

In addition to the calibration method, the accuracy of the TV system can be affected by the system's basic resolution which is the minimum scale of measurement obtained by the system. The resolution of this TV system was assessed by Yang Lang (1988). The assessment was carried out by measuring the position of a marker moved in increments of 2 mm on a vernier calliper with a resolution of 0.02 mm. It was found that the TV system has a resolution of 4 mm in the horizontal direction and 6 mm in the vertical direction.

The accuracy of the TV system was evaluated by the author of this thesis in terms of frame to frame variations, static accuracy, dynamic accuracy and the biomechanical (i.e. with markers on a human segment) accuracy using the DLT calibration method which was used throughout this project.

4.3.2.1 Frame to Frame Variations

Frame to frame variations of the TV system were evaluated for data in 2D machine units and in 3D metric units. The 2D machine units test evaluates all errors resulting from the system instability caused by the electronics of the detector, camera resolution and coordinates generator, and all systematic errors such as the quality of data averaging model which define the centre of marker. The 3D metric test also assesses all the above errors including the error caused by the calibration method. This test reflects the overall accuracy of the TV system in the X, Y and Z directions and assesses the accuracy of the calibration method.

To assess frame to frame variations in 2D computer units, six markers of the same size as those used on the subjects were distributed in the test measurement volume. With the TV system warmed up for one hour, 80 frames were collected for these markers by the three TV cameras. Average, range and

Table 4.3 Measured coordinates of the markers used in the camera accuracy test.

Marker	1	2	3	4	5	6
X [mm]	750	750	750	-750	-750	-750
Y [mm]	96	792	1556	96	792	1556
Z [mm]	300	300	300	-300	-300	-300

Table 4.4 Frame to frame accuracy of the TV system in 3D metric units. (Front and left camera).

Marker	X mm				Y mm				Z mm			
	Mean	Range	Error	SD	Mean	Range	Error	SD	Mean	Range	Error	SD
1	753.057	4.501	-3.057	0.928	92.627	3.572	3.373	1.014	296.559	2.782	3.441	0.636
2	754.539	5.058	-4.539	1.006	795.500	4.111	-3.5	0.954	303.415	1.037	-3.415	0.297
3	746.826	3.400	3.174	0.927	1554.530	2.729	1.47	1.102	304.562	0.438	-4.562	0.1125
4	-755.290	7.619	5.29	1.211	90.83	2.529	5.17	0.59	-296.343	3.494	-3.657	0.99
5	-753.393	7.65	3.393	1.169	795.102	1.088	-3.102	0.318	-296.335	1.388	-3.665	0.000
6	-755.371	4.338	5.371	0.927	1557.154	1.684	-1.154	0.779	-304.284	1.288	4.284	0.238
Aver.		5.427	1.6 sd=3.93	1.028		2.452	0.376 sd=3.23	0.792		1.738	-1.26 sd=3.65	0.378
overall	Max overall error = 5.53 mm				Max overall error = 3.606 mm				Max overall error = -4.91 mm			

Error = Measured value - Calculated value
Range = Max. value - Min. value

standard deviation of the vertical and horizontal readings of each of the three cameras are presented in table 4.2. It is clear that all cameras have better precision in the vertical direction than the horizontal direction, and the largest standard deviation is 0.702 machine units at the horizontal direction of front camera.

To assess frame to frame variations in 3D metric units another configuration of six markers were positioned to cover the test measuring volume. The six markers were located on two chains similar to those used for the camera calibration (fig. 4.20) but at different coordinates from those of the camera calibration. The coordinates of the six markers in relation to the ground frame of reference are shown in table 4.3, these were physically measured by a vernier calliper with a resolution of 0.02 mm. However, when the coordinates were repeatedly measured an error of ± 0.5 mm was found. The coordinates of these markers were reconstructed for 80 frames of each marker using groups of data collected by front/left and front/right cameras. The mean, range and standard deviation of the coordinates which were reconstructed by data from front/left and front/right cameras are presented in tables 4.4 and 4.5 respectively. It should be noted that the reconstructed coordinates were not subjected to any kind of data filtering. It was found, as seen from tables 4.4 and 4.5 (the SD columns) that the front/left camera combination has a precision of 1.028 mm, 0.792 mm and 0.378 mm in the X, Y and Z directions respectively. The front/right camera combination was found to have a precision of 0.802 mm, 0.768 mm and 0.417 mm in the X, Y and Z directions. The precision figure in Z direction is comparable to that reported by Morris (1991) for the "VICON" system (0.303 mm), in the Y direction, the precision of this system is about 2 times larger than that of the "VICON" system (0.345), in X direction, the precision of this system is 3 to 3.9 times larger than that of the "VICON" system. The errors resulting from frame to frame instability are included in the errors calculated in the following section.

Table 4.5 Frame to frame accuracy of the TV system in 3D metric units. (Front and Right Cameras).

Marker	X mm				Y mm				Z mm			
	Mean	Range	Error	SD	Mean	Range	Error	SD	Mean	Range	Error	SD
1	747.185	4.463	2.815	1.771	92.151	3.830	3.849	1.025	296.62	3.408	3.38	0.733
2	752.96	2.682	-2.96	0.551	795.564	3.95	-3.564	0.871	303.48	1.172	-3.48	0.251
3	754.316	2.918	-4.316	0.389	1552.88	2.392	3.12	0.9	304.459	0.272	-4.459	0.112
4	-747.039	4.265	-2.961	0.841	90.652	2.473	5.348	0.584	-295.959	3.602	-4.041	1.021
5	-749.505	1.585	-0.495	0.45	795.472	1.13	-3.472	0.45	-296.244	1.537	-3.756	0.275
6	-759.407	3.666	9.407	0.811	1556.935	1.588	-0.935	0.779	-304.279	1.596	4.279	0.112
Aver.			0.248 sd=4.69	0.802			0.724 sd=3.55	0.768			-1.346 sd=3.68	0.417
overall	Max. overall error = 4.94 mm				Max. overall error = 4.27 mm				Max. overall error = -5.03 mm			

Range = Max. value - Min. value.
Error = Measured value - Mean of Calculated values.

4.3.2.2 Static Accuracy

The absolute static accuracy is also presented in tables 4.4 and 4.5. The error was calculated by subtracting the calculated mean values from the absolute measured values which are presented in table 4.3. The front/left camera set has a maximum error of 5.53 mm in X, 3.61 mm in Y and -4.91 mm in Z directions, and a minimum error of -2.33 mm in X, -2.85 mm in Y and 2.39 mm in Z directions respectively. These error values were averaged for errors of six markers, which explains the negative sign of some of the error values. The front/right camera set found to have a maximum error of 4.94 mm, 4.27 mm, and -5.03 mm in X, Y and Z directions respectively, and a minimum error of -4.4 mm, -2.83 mm and 2.33 mm in the X, Y and Z directions respectively. The absolute static accuracy obtained for the system used is higher than that obtained by Yang Lang (1988) for the same system (overall accuracy of 5.2 mm). However, the accuracy obtained in this test is lower than that reported for the "VICON" system (1.67 mm) by Morris (1991).

The relative static accuracy was also calculated using the above six markers. Distances between the highest and lowest markers of the two chains were calculated and compared with the measured values (1460 mm) of these distances. A relative accuracy of 4.1 mm with standard deviation of ± 1.1 mm was obtained for front/left camera set and an accuracy of 3.6 mm with standard deviation of ± 0.8 mm for the front/right camera set. This produces an overall relative accuracy of 3.85 mm with a standard deviation of 0.95 mm for the static test.

4.3.2.3 Dynamic Accuracy

The assessment of the dynamic accuracy of the TV system is of special importance as the system may show better static accuracy than dynamic accuracy. In the measurement of the kinematic data in gait analysis, the accuracy of relative displacements is more important than the absolute accuracy because gait analysis usually deals with distances between the joint centres.

Table 4.6 Relative dynamic accuracy of the TV system by determination of the length of a stick in various positions.

Stick Position	Mean Length mm	Range mm	Error mm	SD mm
Front and Left Camera				
Horizontal at chest level	791.73	4.21	1.27	1.12
Horizontal at knee level	788.05	5.1	4.95	1.25
Vertical with top marker at shoulder level	787.82	3.49	5.18	0.91
Rotated in field of view of the L.camera	789.1	14.02	3.9	4.59
Rotated while moving towards F.camera	793.68	-5.03	-0.68	1.18
Overall	790.08		2.924	1.81
Front and Right Camera				
Horizontal at chest level	792.00	4.07	1	1.08
Horizontal at knee level	787.00	5.69	6	1.02
Vertical with top marker at shoulder level	789.56	3.57	3.44	0.73
Rotated in field of view of the R.camera	791.65	4.89	1.35	1.12
Rotated while moving towards F.camera	794.47	4.58	-1.47	0.90
Overall	790.94		2.064	0.97

Range = Max. value - Min. value
Error = Measured value - Mean Calculated value
Measured Length = 793 mm

Therefore, the overall relative accuracy of the TV system was investigated by fixing two markers of the same size as those used in the biomechanical test on the two ends of a wooden stick. The length (L) between the centres of the two markers was accurately measured by vernier with a resolution of 0.02 mm and found to be 793 mm. The stick was carried by a person who walked over the force plates covering the test measurement volume and undertaking his normal speed and gait. The stick was held vertically and horizontally in different levels as the person walked facing the front camera. The stick was also rotated in the field of view of the left and right cameras as the person stood at the origin of the ground frame of reference. For each of the above cases three runs were collected and 80 frames were analysed for each run. The X, Y and Z coordinates of each marker were obtained using the DLT calibration method, and the length (L) was calculated for all the above cases. The results and the total error are presented in table 4.6. The data in table 4.6 were not subjected to any filtering technique. The investigations show that the front/left camera set has a relative dynamic error of 2.92 mm and standard deviation of 1.81 mm. Also the front/right camera set has a relative dynamic error of 2.06 mm and standard deviation of 0.97 mm. These errors are slightly larger than those obtained by Morris (1991) for the "VICON" system. Morris reported an average error of 0.91 ± 0.63 mm in the calculating the length of a 978 mm long rod.

Comparing relative static with the relative dynamic errors obtained for the used TV system it is found that the overall relative static error is 3.85 mm and the overall relative dynamic error is 2.49 mm. Relating these figures to the measured lengths (1460 mm for static test and 793 mm for the dynamic test) it was found that the TV system has an overall relative static accuracy of 1:379 and an overall relative dynamic accuracy of 1:318. The effect of static and dynamic accuracy of the TV system on the results of this study is discussed in appendix D.

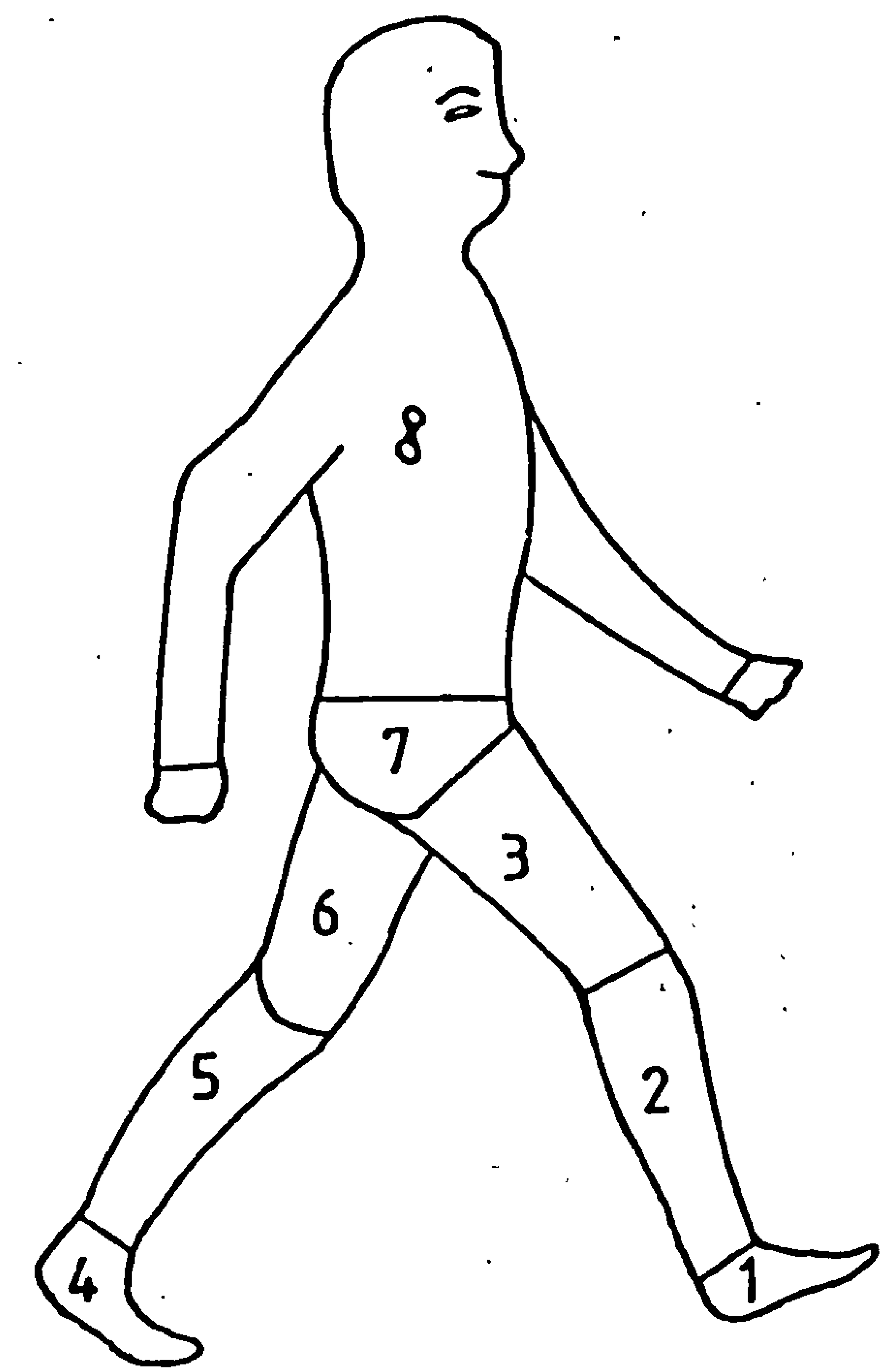


Figure 4.15 Eight segments model for the AK amputee.

4.3.2.4 Biomechanical Accuracy (Clinical Test)

It is useful to test the measuring system under the circumstances of the biomechanical test. This was conducted by fixing two markers on the left and right ASIS of an AK amputee. The distance between these two markers was calculated for 90 frames covering the distance between the left heel strike on FP1 and the right toe off at FP2 using data from the three cameras as the subject walked normally over the force plates. The inter-ASIS distance was found to vary by 12 to 17.4 mm. This variation is due to skin movement which was expected. It should be noted that the data was not filtered. This variation is also comparable to results presented by Morris (1991) of Oxford Metrics Ltd who obtained skin movement variation of 10 to 20 mm using the VICON-VX system.

4.4 The Biomechanical Test

A human body model using 8 segments was shown to give adequate results in gait analysis (Goh 1982 and Yang 1988). A similar model was adopted in this project as shown in figure 4.15. Each of the eight segments is determined by the centre of its connecting joints and assumed to act as a solid body. The segments are defined as follows:

1- The two foot segments are defined by sections through the ankle joint centres. 2- The two shank segments are defined by sections through the knee joint centres and the ankle joint centres. 3- The two thigh segments are defined by sections though the knee joint centres and the hip joint centres. 4- The pelvis is defined by sections through the hip joint centres and the fourth lumbar vertebra. 5- The upper torso segment is defined by a section though the fourth lumbar vertebra and includes the head and the upper limbs. Consideration was given to including the kinetics and kinematics of the upper limbs in the present study as it was felt that these body parts may be used by the patients to compensate for prosthetic malalignment. The limitation (3 cameras only) of the TV system used however, meant that the markers of the arms would interfere with the body markers. This would make the data analysis time consuming as marker interference may confuse the computer and manual interpretations will be needed. Because of the limitation of the TV system the subject had to put his arms on the chest or behind the back so that the body markers were not hidden by the moving arms. Since any change in the behaviour of the upper limbs is affected by the function and the movement of the lower limbs which were fully analysed, no error will be introduced to the collected data from the fact that the analysis of the upper limbs was not included. The effect of the lower limbs on the upper limbs will be transferred through the trunk which was also analysed.

The anatomical hip and ankle joints were considered as ball and socket joints allowing three dimensional rotations. The anatomical knee joint was considered to be a single hinge joint allowing only rotation in the AP plane. Finally, the joint at the fourth lumbar vertebra was considered to be a cylindrical joint allowing only transverse rotation.

Three light weight wooden markers were fixed on each segment to study its motion by recording the motion of these markers. The marker was spherical in shape and about 16 mm in diameter, it was fixed to a stick 1 to 4 cm long and the stick was fixed to a circular wooden base of diameter 18 to 38 mm according to its location. The stick length was chosen to be as short as possible to avoid marker vibrations, but it was long enough to increase the distance between the adjacent markers and to ensure that each marker was seen by at least two cameras. The markers were covered by retro-reflective material which provided effective reflecting properties. This provided good images of the markers and the cameras were focused to pick up only the images. As mentioned above, three markers were fixed on each segment, this being the least number of markers that must be used to establish the marker frame of reference and to carry out three dimensional analysis. These three markers were the "dynamic markers" and they were left on the body segments

Table 4.7 Position of the dynamic and static body markers.

Segment	Marker Position.
DYNAMIC MARKERS	
Upper Torso	Sternum, Left and Right Acromial Process
Pelvis	Sacral Flat, Left and Right Anterior Superior Iliac Spine
Socket	Anterior Wall 0.3(SL) Proximal to the Socket Lower End. Lateral Wall 0.75(SL) Proximal to the Socket Lower End.
Prosthetic Shank	Laterally on the Knee Axis, Laterally 0.25(ShL) proximal to AJC, 0.7(ShL) Proximal to AJC on the Med-Anterior Wall
Prosth.Foot	Equivalent to 2nd Toe, At Mid-heel (rear of foot, see fig. 4.16)
Sound Shank	8 cm above Lateral Malleolus, 5 cm below tibial plateau, Mid-shank on tibial flare.
Sound Foot	2nd Toe, 5th Metatarsal base on the Lateral Side, Mid-calcaneus (at rear of foot).
STATIC MARKERS ONLY ON THE ANATOMICAL JOINTS	
Hip Joint	Laterally at the Greater Trochanter
Knee Joint	Lower Edge of the Patella, Lateral Epicondyle of the Femur.
Ankle Joint	Lateral Malleolus, Anteriorly Mid-Way between the medial and Lateral Malleoli.

SL Socket Length ShL Shank Length

throughout the whole biomechanical test. In addition, other types of markers were used. These were the "static markers" which were fixed on the subject to define the joint centres and they were attached to the subject during the static test (static run) only which lasted about 10 minutes. The position of the joint centres (JC) and the anatomical frames of reference (AFR) are related to the marker frames of reference. The position of the JC and AFR is fixed in relation to the marker frames of reference during the static and dynamic tests. All marker positions were chosen to be on bony landmarks so that errors resulting from the movement of the marker relative to the skeletal system will be minimum. A total of 24 markers was fixed on the AK subject as defined in table 4.7. The position and the numbering system of these markers for right and left amputees are shown in figure 4.16. The numbers were labelled to each marker during the sorting procedure of the data. Figure 4.17 shows the marker and the numbering systems for the normal subject.

4.4.2 The Procedure of the Biomechanical Test

Before the day of the test, the biomechanics laboratory was arranged to be free at the arrival of the patient. In the day of the test, the TV system, force plates and the computer were all switched on 40 to 60 minutes before the test to allow worming up. All the switches, the wiring, the computer and disk spaces and the monitors were checked and made ready before the arrival of the patient. The TV cameras were tuned to obtain a very good image as this may take long time and if it is not good enough (the contrast and the image shape) the quality of the collected data will be affected and sorting this data will be time consuming as additional problems may result. The function of the system was examined before the test by collecting data from two or three trials on a normal subject and displaying the results for checking and to ensure that all the equipment were functioning satisfactorily. The markers were also prepared before the patient arrival by fixing them on double adhesive sellotape and putting them in order so that minimum time will be spent to mount them on

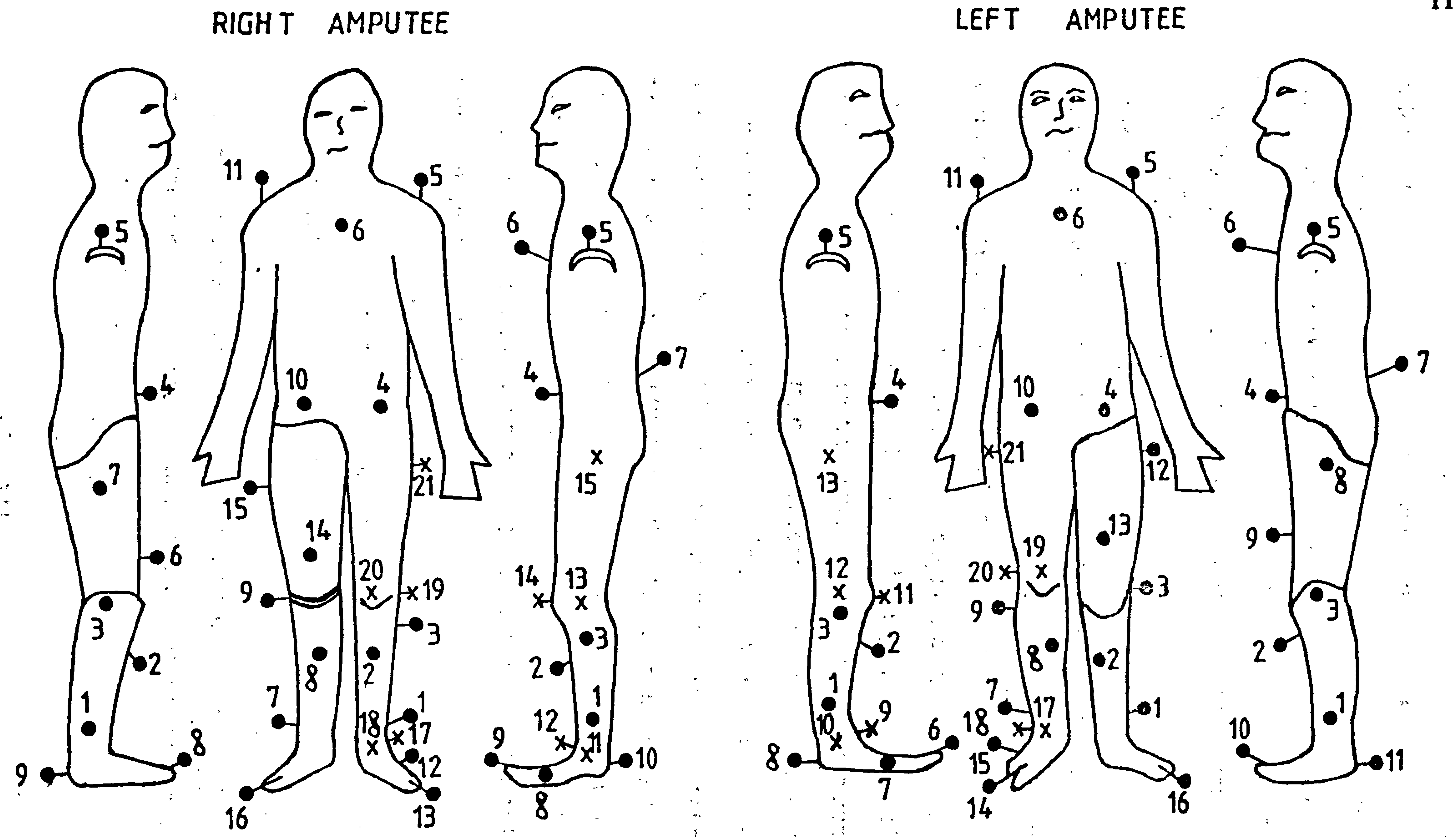


Figure 4.16 Body markers, and the numbering system used for the AK amputees.
• Markers used in dynamic and static tests. X Markers used in static test only.

the patient.

On patient arrival, if it is his/her first visit the patient was introduced to the laboratory in particular the TV cameras, the monitors and the computer but not to the force plates. This was so that the subject would use a "natural gait pattern without targeting the force platform". An explanation of the project and its aims was also given. The patient was then fitted with the experimental prosthesis which was fabricated in the workshop under the supervision of the prosthetist who also carried out the alignment (only on the first visit of the patient). After obtaining a satisfactory dynamic alignment, all the static and dynamic markers were mounted on the appropriate positions. The dynamic alignment which was achieved by the prosthetist is called "normal alignment" and this term will be used throughout this work.

4.4.2.1 The Static and Dynamic Tests

After attaching the static and dynamic markers on to the landmarks, the patient was asked to stand at around the origin of the ground frame of reference (between the two force plates) in his normal way of standing facing the front camera so that each marker was seen by the front camera and at least by one of the side cameras. Data were then collected for about 30 consecutive frames and the validity of these data was examined on the screen of the computer to ensure that all the markers were being picked up. After that, distances between some of the body markers were measured and recorded in a special form (fig. 4.18) which will also have a record for the static and dynamic tests, calibration runs and the comments of the patient. The anthropometric measurements were also taken and recorded in the form shown in figure 4.19. All the static markers were then removed and the subject was now ready for the dynamic test.

The dynamic test was conducted after asking the patient to undertake a few walking trials over the force plates and walking towards the front camera. As mentioned in the previous section, the patient was not briefed on the

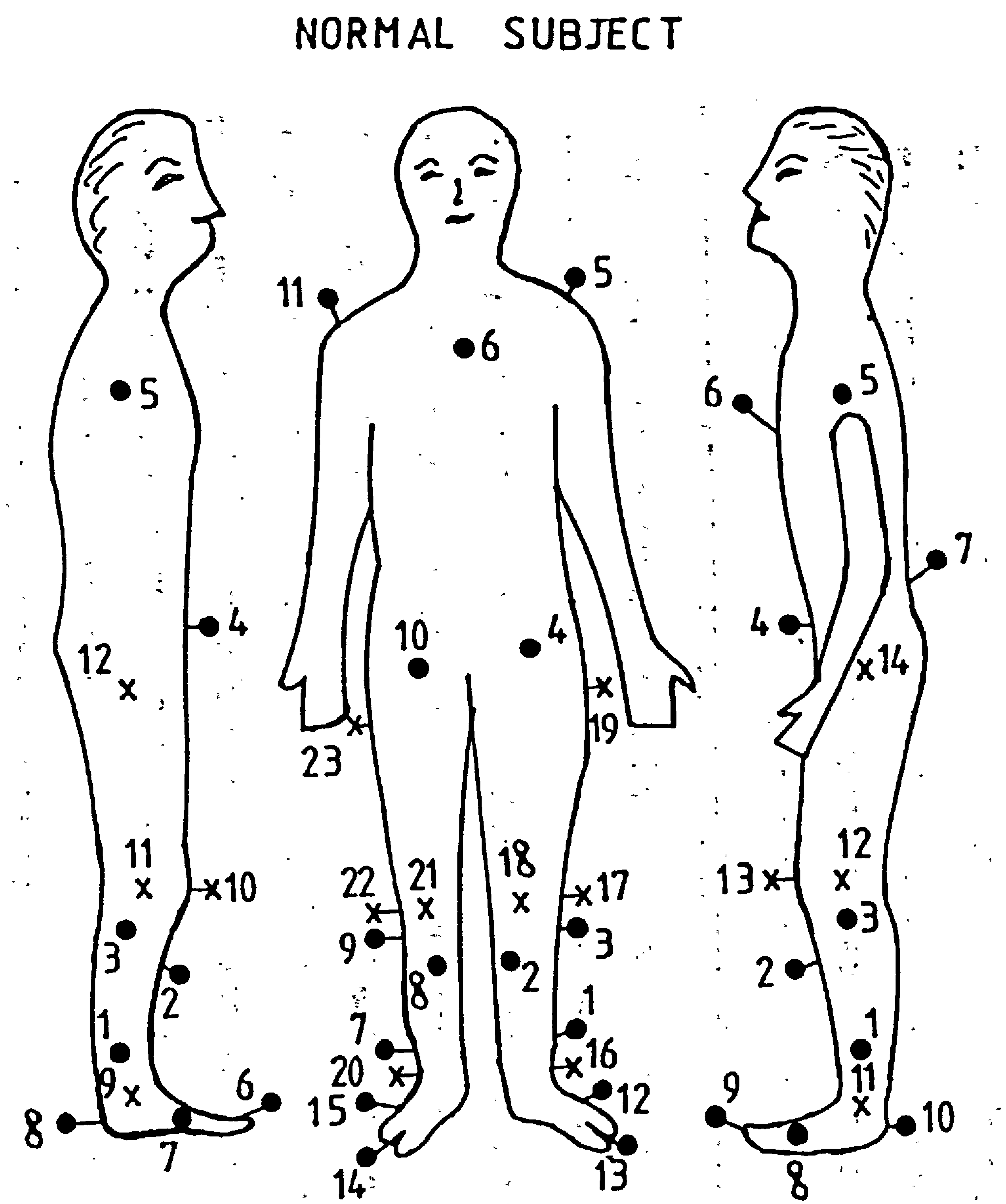


Figure 4.17 Body markers and numbering system used for the normal subjects.
• Markers used in dynamic and static tests. X Markers used in static test only.

existence of the force plates. The patient was instructed to start walking four to five steps before striking FP1; this was done in order to guarantee a natural gait when the patient stepped on the force plates. When the patient hit the force plates with his natural gait the starting point was marked on the floor and the patient was asked to consider it as the starting point for every walk. Thereafter, three successful runs were collected for each set of alignment changes starting with the normal alignment (successful run is when the TV and force plate data are appropriate, i. e. no marker was missing and the subject hit the force plates as close to their centres as possible). Four deliberate changes were made to the prosthetic alignment at the foot or at the socket or simultaneously at both sites. Thus, each patient had to visit the laboratory on three different occasions, one for the set of foot alignment changes, one for the socket alignment changes and the other for the foot/socket alignment changes.

The foot alignment changes were conducted by rotating the foot by an angle of 6 degrees towards dorsiflexion and then by 6 degrees towards plantarflexion from the normal position in increments of 3 degrees. Similarly the socket angular position was changed 6 degrees towards extension and 6 degrees towards flexion from the normal position in increments of 3 degree. Finally, the foot and socket angular positions were changed simultaneously by dorsiflexing the foot and flexing the socket or by plantarflexing the foot and extending the socket, this was also done in steps of 3 and 6 degrees in both direction from the normal position. In fact, when the changes were done at the foot and socket simultaneously no changes in the orientation of the prosthesis occurred, but the changes caused pure shift forwards or backwards in the position of the knee joint centre with respect to the hip-ankle line. All the above changes were done by the prosthetist and after each change the patient was given about 15 minutes to accustom him/her self to the new alignment. The dynamic test was completed when three successful runs were collected for each alignment. For each visit of the patient, 15 dynamic runs were collected

Record For Patient Test

Name:
Height [m] :
Mass [kg] :
Affected Side:

Date:
Camera with phase delay:
Phase delay [ms]:
Disk used:

<u>Test</u>
Static test

<u>Comments on the patient's gait</u>

Dynamic test 1:
Dynamic test 2:
Dynamic test 3:

Dynamic test 4:
Dynamic test 5:
Dynamic test 6:

Dynamic test 7:
Dynamic test 8:
Dynamic test 9:

Dynamic test 10:
Dynamic test 11:
Dynamic test 12:

Dynamic test 13:
Dynamic test 14:
Dynamic test 15:

TV Calibration Test:

<u>Distance between markers [m] :</u>
Right ASIS to Tail:
Left ASIS to Tail :
Right to Left ASIS:
Foot Marker 1 to 3:
Foot Marker 2 to 3:

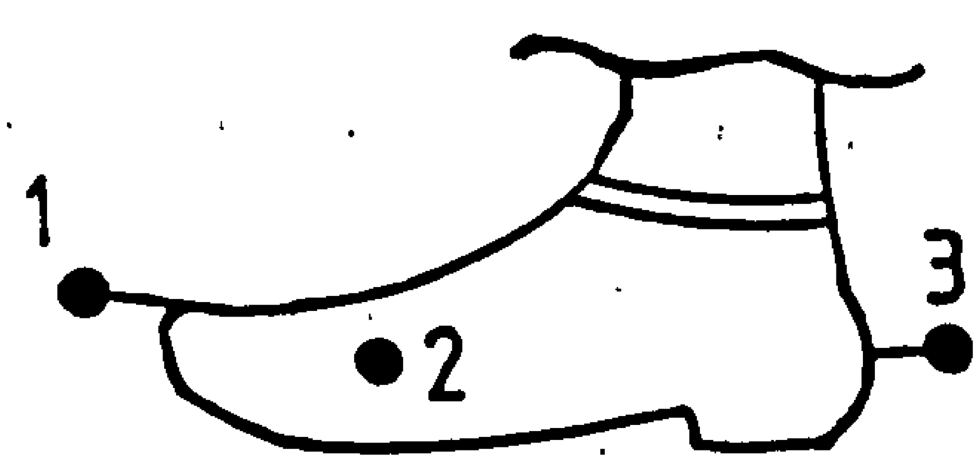


Figure 4.18 The amputee biomechanical test form.

(form in fig. 4.18). For example, during the visit for the foot alignment changes there were 3 runs for the normal alignment, 6 runs for the dorsiflexion changes and 6 runs for the plantar flexion changes. For each visit a maximum of three hours was needed to complete the test.

4.4.2.2 Calibration of the TV Cameras

At the end of each test the cameras were calibrated using a set of three flexible chains located at three different positions as seen in figure 4.20. Each chain had three markers of the same size as those used on the body segments and were located at three different heights from the floor on the chain. Data were collected for three runs during the calibration procedure. For the first run chains I, II and III were located at positions 1, 5, and 9 respectively, for the second run the chains were located at positions 7, 2, and 6, and for the third run the chains were in positions 4, 8 and 3. This resulted in the acquisition of TV coordinates for 27 points located at known positions and covering a volume of 1.8 X 1.56 X 0.8 m. These data were used to calculate the cameras constants which are needed for the DLT method to produce the 3D coordinates of any marker in the calibrated volume. The calibration method does not need 27 markers but some of these markers were used to check the accuracy of the calibration method.

4.4.3 Data Processing

At the end of test session all the data obtained were stored in a PDP11 computer in machine coded form, and, the first stage of data processing is averaging. Software in the PDP11 computer allowed reading-in of the coded data and matched the TV channels with the TV cameras. It then matched the vertical and horizontal values and set a window around each marker and averaged mathematically all the points inside this window to provide the vertical and horizontal coordinates of each marker frame by frame in machine units. The size of the window affects the accuracy of the data and it is influenced by the distance between the adjacent markers. Sorting is the second

Patient's Anthropometric Measurements

Name: Date:

Height [m]:

Mass [kg]:

Affected side:

<u>Measurements [mm]</u>

Foot Length Lf:

Ankle Circumference Ca:

Shank Circumference Cs:

Tibial Height Hti:

Thigh Circumference[1] Cth:

Iliac Crest Fat P:

Figure 4.19 The patient anthropometric measurement form.

1 The thigh circumference was taken for the sound leg at the upper level.

stage of data processing. During a data collection run, data collection was started one or two steps before the subject hit the first force plate (FP1) and stopped at least one step after the subject left the second force plate (FP2). However, useful data is only found in the range between left heel strike which is on FP1 and right toe off which is on FP2. In the sorting procedure, the markers in each channel were labelled by taking the frame number which coincided with the left heel strike instant and giving a certain number (fig 4.16 and 4.17) to each marker in this frame. Software in the computer sorted and stored automatically all the frame numbers which were between left heel strike on FP1 and right toe off on the FP2. During automatic sorting, if a marker was missing for five frames or less the program provided a linear extrapolation, but if the missing frames are over five a manual interpolation was needed. This was hardly needed because the data quality was checked during the data collection session and the trial with missing markers was ignored. The sorting procedure had some difficulties especially when sorting the data during swing phase. If the trajectories of two markers intersected within one frame, the program was unable to reidentify the markers and manual identification was needed. This was time consuming. Another difficulty was faced during data sorting: if a camera was not perfectly tuned or the markers were not giving a good contrast during data collection, the camera could pick up the marker but the sorting program failed to track the marker's position. No explanation was found for this effect but it was avoided by setting the cameras properly tuned and examining the data during the test session to ensure that this problem did not exist.

The third stage in the data processing was correcting the phase delay for the data collected by the camera which had a phase delay to avoid blinding the opposite camera. This was done by another program which uses a linear interpolation procedure to correct that phase delay. The final stage of data processing was a visual examination of the trajectory of the markers channel

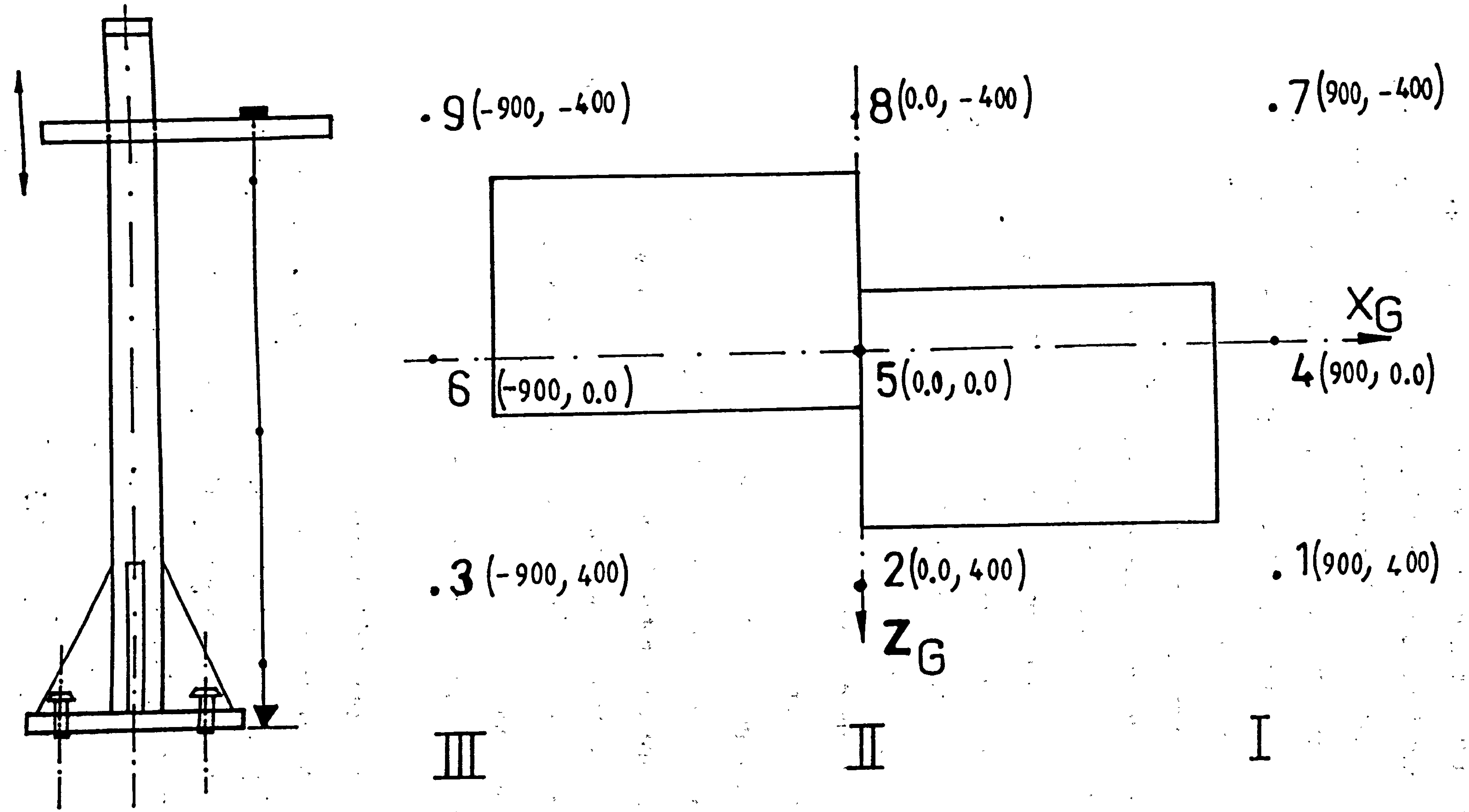


Figure 4.20 Locations of the calibration chains.

by channel to ensure that no data were missing and if the markers had any missing frames, manual modification was implemented. Then the static, dynamic, force plate and the calibration data were all merged together and transferred to the University main frame (VAXs) computers for further biomechanical analysis. The sorting procedure required about 45 to 60 minutes in order to complete the processing of the data resulted from the three cameras, for one trial only.

CHAPTER FIVE

THE THEORETICAL WORK

This chapter deals with all the theoretical aspects of the work presented in this thesis. The theoretical design of the body model, body markers and their location on the body landmarks were discussed in section 4.4.1. In this chapter the generation of the 3D coordinates of the body markers, data smoothing and normalising, definition of the joint centres, and frames of reference will be discussed. The method of obtaining the kinematic and kinetic variables is also discussed in this chapter.

5.1 Calculating the 3D Space Coordinates

Two methods were used to generate the 3D space coordinates of the body markers. The first method is the Direct Linear Transformation (DLT) and it was used for the markers which were seen by two cameras. The second method depends on the geometry of the rigid body and it was used for the heel and tail markers as these were seen by only one camera.

<u>The DLT Method</u>

The conventional method of transferring the comparator coordinates into object space coordinates is by transferring the comparator coordinates into photo coordinates and then, transferring the photo coordinates into object space coordinates. The relationship between the photo coordinates (u, v) and the observed comparator coordinates (q, P) of a point is represented by the following two equations:

$$\begin{aligned} u - u_p &= a_1 + a_2 q + a_3 P \\ v - v_p &= a_4 + a_5 q + a_6 P \end{aligned} \qquad (5.1a)$$

Where:

u_p and v_p are the photo coordinates of the principal point (see fig. 5.1).

a_1.....a_6 are transformation and correction constants. The corrections are for the linear film deformation, linear lens deformation and comparator errors.

If multiple cameras are used to film a point A in the space, the

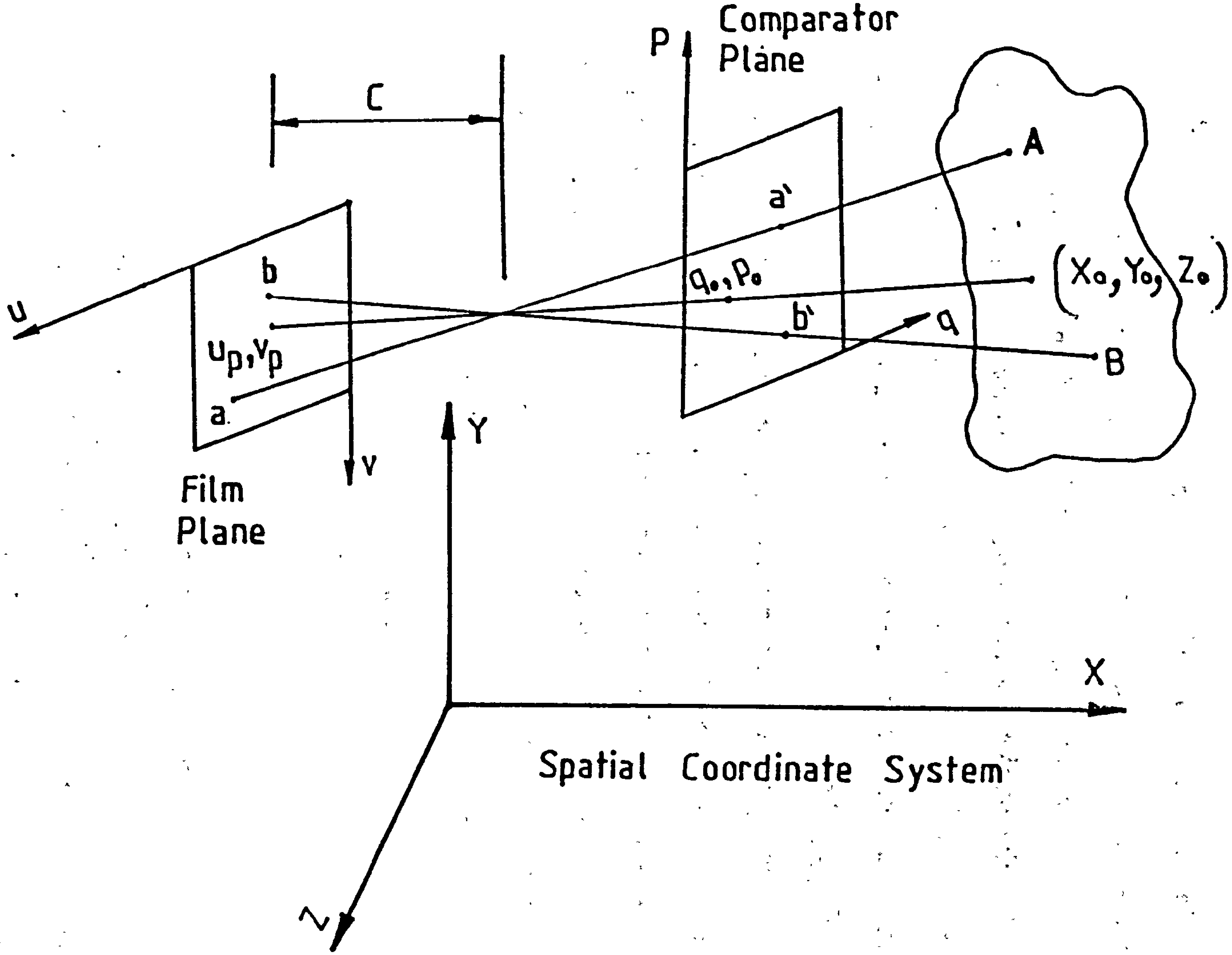


Figure 5.1 Relationship between spatial, photo and comparator coordinate system. (adopted from Miller et al 1980).

relationship between the space coordinates (X, Y, Z) and the photo coordinates (u, v) of that point is given for each camera by the equation (Appendix A in Hallert 1960, and Abdel-Aziz and Karara 1971):

$$\begin{bmatrix} u - u_p \\ v - v_p \\ - C \end{bmatrix} = \lambda \, [\, M \,] \begin{bmatrix} X - X_0 \\ Y - Y_0 \\ Z - Z_0 \end{bmatrix} \tag{5.1b}$$

Where:

C is the camera principal distance.

λ is a scale factor.

[M] is 3 x 3 rotation matrix

X_0 , Y_0 , Z_0 are the three dimensional space coordinates of the camera perspective centre.

Thus, the position of point A can be determined in space by solving equations 5.1a and 5.1b for the X, Y and Z coordinates and for more than one camera.

Abdel-Aziz and Karara (1971) proposed the DLT method which deals with direct linear transformations from the 2D comparator coordinates into 3D object space coordinates. Assuming that the photo coordinate system is parallel to the comparator coordinate system, and by regrouping the parameters of equations 5.1 they derived the basic formulae of the DLT method which are:

$$P + \Delta P = \frac{L_1X+L_2Y+L_3Z+L_4}{L_9X+L_{10}Y+L_{11}Z+1} \tag{5.2a}$$

$$q + \Delta q = \frac{L_5X+L_6Y+L_7Z+L_8}{L_9X+L_{10}Y+L_{11}Z+1} \tag{5.2b}$$

Where:

ΔP and Δq are systematic errors in the comparator coordinates.

L_1 , L_2 , ...L_{11} are the DLT parameters.

P, q and X, Y, Z are explained above.

Equations 5.2 can be solved for the eleven unknowns which are the eleven DLT parameters in which the linear systematic errors are implicit (Marzan and Karara 1975). The DLT parameters are constant for a fixed camera with fixed lens, and they can be determined for any camera by using control points (at least 6 points) to solve equations 5.2. This can be explained as follows: Suppose we have N control points, (in this work N = 27), for one camera equations 5.2 can be written in the form seen in equation 5.3a:

$$\begin{bmatrix} X_1 & Y_1 & Z_1 & 1 & 0 & 0 & 0 & 0 & -X_1P_1 & -Y_1P_1 & -Z_1P_1 \\ 0 & 0 & 0 & 0 & X_1 & Y_1 & Z_1 & 1 & -X_1q_1 & -Y_1q_1 & -Z_1q_1 \\ X_2 & Y_2 & Z_2 & 1 & 0 & 0 & 0 & 0 & -X_2P_2 & -Y_2P_2 & -Z_2P_2 \\ 0 & 0 & 0 & 0 & X_2 & Y_2 & Z_2 & 1 & -X_2q_2 & -Y_2q_2 & -Z_2q_2 \\ \cdot & \cdot & \cdot & \cdot & \cdot & \cdot & \cdot & \cdot & \cdot & \cdot & \cdot \\ \cdot & \cdot & \cdot & \cdot & \cdot & \cdot & \cdot & \cdot & \cdot & \cdot & \cdot \\ \cdot & \cdot & \cdot & \cdot & \cdot & \cdot & \cdot & \cdot & \cdot & \cdot & \cdot \\ X_N & Y_N & Z_N & 1 & 0 & 0 & 0 & 0 & -X_NP_N & -Y_NP_N & -Z_NP_N \\ 0 & 0 & 0 & 0 & X_N & Y_N & Z_N & 1 & -X_Nq_N & -Y_Nq_N & -Z_Nq_N \end{bmatrix} \begin{bmatrix} L_1 \\ L_2 \\ L_3 \\ L_4 \\ \cdot \\ \cdot \\ \cdot \\ L_{10} \\ L_{11} \end{bmatrix} = \begin{bmatrix} P_1 \\ q_1 \\ P_2 \\ q_2 \\ \cdot \\ \cdot \\ \cdot \\ P_N \\ q_N \end{bmatrix} \quad (5.3a)$$

Where:

X_i, Y_i, Z_i, are the known object space coordinates of the control points.

P_i and q_i are the known comparator coordinates of the control markers regarding the camera under consideration.

The DLT parameters can be found for each camera by solving equations 5.3a for L_1, L_2 ...L_{11} , providing that the control points are $N \geq 6$. Thus, the calibration of the TV cameras is achieved by determining the values of the DLT parameters (L_1, L_2, ...L_N).

Now after determining the DLT parameters, equations 5.2 can be solved for the X, Y and Z coordinates of any marker seen by at least two cameras. Equations 5.2 can be written as :

$$\begin{bmatrix} P_1L1_9-L1_1 & P_1L1_{10}-L1_2 & P_1L1_{11}-L1_3 \\ q_1L1_9-L1_5 & q_1L1_{10}-L1_6 & q_1L1_{11}-L1_7 \\ P_2L2_9-L2_1 & P_2L2_{10}-L2_2 & P_2L2_{11}-L2_3 \\ q_2L2_9-L2_5 & q_2L2_{10}-L2_6 & q_2L2_{11}-L2_7 \end{bmatrix} \begin{bmatrix} X \\ Y \\ Z \end{bmatrix} = \begin{bmatrix} L1_4-P_1 \\ L1_8-q_1 \\ L2_4-P_2 \\ L2_8-q_2 \end{bmatrix} \qquad (5.3b)$$

Where:

X, Y, and Z are the marker 3D object space coordinates for which equations 5.3b are solved.

P_1, q_1 are the comparator coordinates of the marker in camera number 1 (the front camera in this study).

P_2, q_2 are the comparator coordinates of the marker in camera number 2 (one of the side cameras in this study)

$L1_i$, $L2_i$ are the DLT parameters for cameras 1 and 2 respectively.

The DLT method was used in this study by employing a computer program based on that which was developed by Marzan and Karara (1975) and the accuracy of this method was assessed in chapter four.

Markers Seen By Only One Camera

If a marker is seen by only one camera, only two equations will be obtained by applying the DLT method on the data of this camera. They can be written as:

$$\begin{aligned} (PL_9-L_1)X+(PL_{10}-L_2)Y+(PL_{11}-L_3)Z &= L_4-P \\ (qL_9-L_5)X+(qL_{10}-L_6)Y+(qL_{11}-L7)Z &= L_8-q \end{aligned} \qquad (5.4a)$$

The X, Y, and Z coordinates cannot be obtained by using only two equations, therefore the theory of properties of the rigid body was also employed. Considering the tail marker as an example: the tail marker and the two ASIS markers belong to one body (The pelvis) which is assumed to be a rigid body. The distance D1 between the tail marker and the right ASIS marker is constant and so is D2 the distance between the tail and left ASIS markers. The mathematical form of the relationship between the tail and ASIS markers can be written as follows:

$$(X-X_1)^2 + (Y-Y_1)^2 + (Z-Z_1)^2 = D_1^2$$
$$(X-X_2)^2 + (Y-Y_2)^2 + (Z-Z_2)^2 = D_2^2 \qquad (5.4b)$$

Where the subscripts 1 and 2 belong to the right and left ASIS respectively.

Equations 5.4a and 5.4b can be solved for X, Y and Z which are the coordinates of the tail marker. They were solved by a computer subroutine called "E04GEF" which was implemented in the University main computers "VAXs"

5.2 Filtering, Differentiating and Normalising the Data

A - Filtering and Differentiating

Any measurement system used in the study of human motion introduces noise into the measured signal. In the kinematic study of human motion, this noise may not be evident in the displacement data, but it will be amplified during the differentiation procedure to cause marked inaccuracies in the velocity and acceleration data. Therefore, data filtering must be applied on the raw data to remove or reduce its noise. To design a suitable filter, information about the frequency spectrum of the signal and noise should be obtained. Winter et al (1974) have conducted spectral analysis of digital gait kinematic data and found that the spatial data is of low frequency, and a low-pass filter will cut off the high frequency noise contained in the data to such an extent that linear and angular velocities can be calculated by direct digital differentiation. Andrews (1982) stated that the upper frequency limit of the distinguishable gait signals is approximately 10 Hz. Pezzack et al (1977) have compared the accelerations obtained by three different techniques of data filtering and differentiation with a measured acceleration signal to determine their efficiency. The techniques were: (a) the second order finite difference equation cited by Miller and Nelson (1973), (b) Chebyshev least squares polynomial fitting followed by polynomial differentiation, and (c) second order recursive Butterworth digital filter used by Winter el al (1974) followed by first order finite difference differentiation used by Miller and Nelson (1973). The acceleration obtained by the third technique was found to be more similar to the measured signal than the other two. Therefore, the idea of digital filtering followed by a differentiator was adopted in this work. A fourth order Butterworth low-pass digital filter given by the equation shown below was used in this work. It was designed by Andrews in 1975 (published in Andrews 1982) and is now widely in use (Tooth 1976 and Yang Lang 1988):

$$y_k = \frac{1}{C_6}\,[C_1(x_k+4x_{k-1}+6x_{k-2}+4x_{k-3}+x_{k-4}) - (C_2y_{k-1}+C_3y_{k-2}+C_4y_{k-3}+C_5y_{k-4})] \quad (5.5)$$

Where:

x_k is coordinate of the point to be filtered

y_k is coordinate of the filtered point

C_1, C_2, C_3, C_4, C_5, C_6, are constants expressed as functions of the cutoff frequency (5 Hz in this work) and the sampling time interval T (0.02 s in this work).

As the filtering produces a phase lag, the data were filtered twice, one in the forward and one in the backward directions. This produces output data with no phase lag, and with a cut off frequency of 10 Hz.

After filtering, the data were numerically differentiated to obtain the velocities and the accelerations. The quality of the calculated derivatives is affected by the quality of the measured data and by the quality of the differentiation technique. If there is too much noise in the measured data or the sampling rate is inadequately low, no differentiation technique will produce good estimates of the derivatives. On the other hand, if the noise level of the measured data is low and the sampling rate is sufficient, any differentiation technique will provide adequate derivatives. Having the data filtered as seen above and sampled at a rate of 50 Hz, the following differentiator which was developed by Tooth (1976) and used by Yang Lang (1988) under the same circumstances, was used in this work:

$$\dot{x}_n = \frac{1}{12T}\left[(x_{n-2}) - 8(x_{n-1}) + 8(x_{n+1}) - (x_{n+2})\right] \qquad (5.6)$$

where x_n is the point to be differentiated, and T is the sampling time interval (0.02 s).

The accuracy of the above filtering and differentiating technique was assessed by fixing a marker on a rotating disc, which was rotated with a constant angular velocity (ω) of 5.236 1/s. The marker was of the same size as those used on the subjects during the biomechanical test, and was fixed on the disc so that the distance from its centre to the disc centre of rotation (r) was equal to 0.21 m. Thus, the linear velocity (V) of the marker can be calculated as:

$V = \omega\, r = 5.236 * o.21 = 1.10$ [m/s] (similar to the average velocity of the normal subjects).

As the disc was rotated with a constant angular velocity, the linear tangential acceleration is equal to zero, thus, the resultant linear acceleration (a) can be calculated as:

$a = \omega^2 * r = (5.236)^2 * 0.21 = 5.757$ [$1/s^2$]

The cameras were operated under conditions similar to those of the biomechanical test and 70 frames were collected by each pairs of camera (front/left and front/right). Then, the data were firstly filtered using the above filter, and secondly, differentiated using the above differentiator to calculate the linear velocity of the marker, and finally the velocity was differentiated to calculate the linear acceleration of the marker.

A mean velocity of 1.09 ± 0.168 m/s and mean acceleration of 5.63 ± 1.04 m/s^2 were found for the front/left camera set. Also a mean velocity of 1.09 ± 0.086 m/s and mean acceleration of 5.72 ± 0.935 m/s^2 were found for the front/right camera set. This reflects the quality of the used filtering and differentiating technique, and it should be noted that the source of the obtained errors in the velocities and accelerations is not from the filtering and differentiating technique only, and it can be from the TV system.

B - Time Normalization

Because a single run can never accurately represent the gait of the subject, three successful (all the required data are obtained) runs were collected for each set of alignments and the output results of these three runs were then averaged as mentioned in chapter four. As the subject has speed variations from run to run and from step to step, the numbers of points obtained at the different runs were not equal. Thus, the use of a simple mathematical method of averaging may distort the data and produce misleading results. Fourier analysis was used to represent the data of each trial in the form of sine and cosine terms and a certain number of Fourier coefficients. The data were then reconstructed using Fourier coefficients, and each run was time normalised to have 100 points. Thus, the three runs of each alignment set have the same number of points (100) which can be averaged using simple methods without any data distortion. Using Fourier analysis provides several advantages. The data can be stored, compared, and also averaged using only the Fourier coefficients.

In Fourier analysis, the number of harmonics which is essential to reconstruct the data is based on the data sampling rate and on the accuracy required in reconstructing the original patterns. Winter (1974) found that for the trajectories of markers on the thigh and shank, 99.7 percent of the signal power lies below the 8^{th} harmonic. Schneider and Chao (1983) found that for the ground reaction force patterns only the first two to four harmonics plus the constant term dominate. In this work the first ten Fourier coefficients were used to reconstruct the patterns of the data.

Normalising the data q_i(i=1,2,3,.......N) where N is the number of actual points, to a new number of points M (M=100) was carried out in three stages: firstly, to avoid introducing bias in the reconstructed pattern the data was detrended (see Andrews 1982) using the equation:

$$\grave{q}_i = q_i - q_1 - \frac{q_N - q_1}{N-1}(i-1) \qquad : i=1,2,3......N \tag{5.7}$$

Where:

q'_i are the detrended data and q_i is the point under consideration.

Secondly, the Fourier coefficients of the first n harmonics were determined by using the relations:

$$A_0 = \frac{2}{N-1}\sum_{i=1}^{i=N}\grave{q}_i$$

$$A_j = \frac{2}{N-1}\sum_{i=1}^{i=N}\grave{q}_i \cos\frac{2\pi j(i-1)}{(N-1)}$$

$$Bj = \frac{2}{N-1}\sum_{i=1}^{i=N}\grave{q}_i \sin\frac{2\pi j(i-1)}{(N-1)} \tag{5.8}$$

where j = (1,2,3,....n) : n = 10 in this work.

Finally, the normalised data were reconstructed as:

$$q_k = q_1 + \frac{q_N - q_1}{M-1}(k-1) + \frac{A_0}{2} + \sum_{j=1}^{j=n} A_j \cos\frac{2\pi j(k-1)}{M-1} + \sum_{j=1}^{j=n} B_j \sin\frac{2\pi j(k-1)}{M-1} \tag{5.9}$$

Where k = 1,2,3,.......M : M=100 in this work.

The data patterns which were reconstructed in this method have 100 points in each run and it is possible to apply simple averaging methods on this normalised pattern.

5.3 Determination of the Joint Centres of the Lower Limbs

The positions of the ankle and knee joint centres of the prosthetic leg were defined and measured relative to the shank marker system (see section

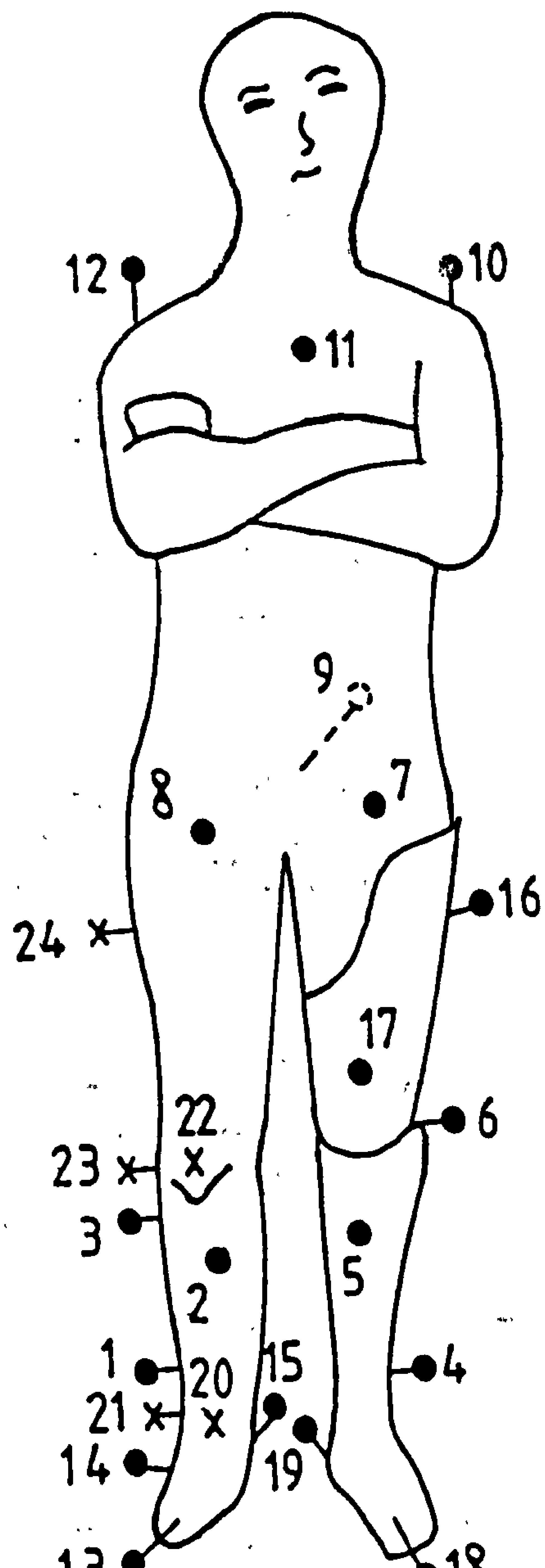


Figure 5.2 Left amputee with the 24 markers located at their landmarks.
• Dynamic markers, stay on the subject during the whole test.
x Static markers, stay on the subject during the static test only.

4.2.4). On the sound leg, the positions of the centre of ankle and knee joints were defined by the static markers used during the static test. Referring to figure 5.2 the position of the unaffected ankle joint centre can be defined by the markers 21 and 20 as:

$$\vec{R}_{UA} = [X_{UA}\ Y_{UA}\ Z_{UA}] = [X_{21}\ Y_{21}\ Z_{20}] \tag{5.10}$$

Similarly, the position of the unaffected knee joint centre is:

$$\vec{R}_{UK} = [X_{UK}\ Y_{UK}\ Z_{UK}] = [X_{23}\ Y_{23}\ Z_{22}] \tag{5.11}$$

The location method of the knee and ankle joint centres was used by Andriacchi and Strickland (1985). They recommended that an average error of 0.79 cm was estimated when this method was compared with x-rays method.

Accurate definition of the hip joint centre is important for gait analysis purposes. Unfortunately, however location of this joint on a subject is very difficult. There are several approaches to define the hip joint centre but none of them is particularly accurate. Bell et al (1990) carried out a comparison test to asses the accuracy of three different methods for locating the hip joint centre. The first method was developed by Andriacchi et al (1980, 1982) and Andriacchi and Strickland (1985) who predicted that the hip joint centre (HJC) lies 1.5 to 2 cm directly distal to the midpoint of a line joining the pubic symphysis and the anterior superior iliac spine (ASIS) in the frontal plane projection, and directly medial to the greater trochanter in the sagittal plane. The second method was introduced by Tylkowski et al (1982): they predicted that the HJC lies medial, distal and posterior to the ASIS by 11%, 12% and 21% of the inter-ASIS distance respectively. The third method was described by Cappozzo (1984) based on the premises that (a)- the thigh is a rigid body, (b)- the HJC is the centre of a sphere described by the three dimensional rotation of a point on that body. The precise location of the HJC was found by means of the radiographic method (X-Ray) in a way based on that used by Brand et al (1982), the femoral head and the skin markers which define the pelvic frame of reference were digitized from the radiographs, thus, the positions of the HJC obtained from the various methods were expressed in the pelvic frame of reference. The above three methods were compared with the X-Ray method. None of these methods was found to be particularly accurate, but Bell et al have recommended an accurate method of locating the HJC by combining the approach of Tylkowski et al with new modified percentage figures at the frontal plane, and the approach of Andriacchi et al for the anterior posterior (AP) plane. The HJC was defined in the AP plane by a marker located on the head of the greater trochanter. In the frontal plane the HJC lies medial and distal to the ASIS by 14% and 30% of the inter-ASIS distance respectively. They suggested that this method predicts the location of the HJC to an accuracy of 1.07 cm.

This method was used to determine the location of the HJC of the sound and amputated legs throughout the work of this thesis, and the effect of the location error on the results is discussed in appendix D. The following is the method of locating the HJC vector of a left amputee at the static test:

A- The Unaffected Side:

$$\begin{aligned} \vec{R}_{UH} &= [X_{UH} \; Y_{UH} \; Z_{UH}] \\ X_{UH} &= X_{24} \\ Y_{UH} &= Y_8 - 0.30D \\ Z_{UH} &= Z_8 - 0.14D \end{aligned} \tag{5.12}$$

Where D is the distance between the ASIS markers. R_{UH} is the position vector of the unaffected hip relative to the ground frame of reference. X_{24}, Y_8 and Z_8 are for markers shown in figure 5.2.

B- The Prosthetic Side

No marker was located on the greater trochanter of the prosthetic side during the biomechanical test. Thus, the prosthetic HJC was defined using the ASIS markers and the anatomical HJC. This was mathematically calculated as follows:

$$
\begin{aligned}
\vec{R}_{PH} &= [X_{PH}\ Y_{PH}\ Z_{PH}] \\
X_{PH} &= X_7 + (X_{UH} - X_8) \\
Y_{PH} &= Y_7 - 0.30D \\
Z_{PH} &= Z_7 + 0.14D
\end{aligned}
\qquad (5.13)
$$

(Left)

Where: R_{PH} is the position vector of the prosthetic hip relative to the ground frame of reference.

should be

Equations 5.12 and 5.13 are rectified for right side amputees.

5.4 Segment orientations and Inertia Properties

In order to define the position of the body segments in space, three reference systems of coordinates were defined. At any instant the body segments can be related to any of these systems, they are the ground frame of reference, the principal axes of inertia of the segment and the marker frame of reference. The principal axes and marker frame of reference were related to the ground frame of reference by direction cosine matrices which were used to transfer the positions of the body segments within the three frames of reference.

5.4.1 Principal Axes of Inertia

The principal axes of inertia were defined for each segment separately, and their direction cosine matrices in relation to the ground frame of reference were calculated from the static test as follows:

<u>The Prosthetic Foot</u>

The principal axes of inertia of the prosthetic foot (X_P, Y_P, Z_P) were defined to be parallel to those defined in section 4.1.1.1. X_P and Z_P are contained in the plane parallel to the top surface of the SACH foot, X_P connects the mid-toe point to the foot bolt and is positive forwards. Z_P is normal to X_P and directed to the right regardless of the side of amputation. Y_P was taken to be normal to the X_P and the Z_P plane and directed upwards. As

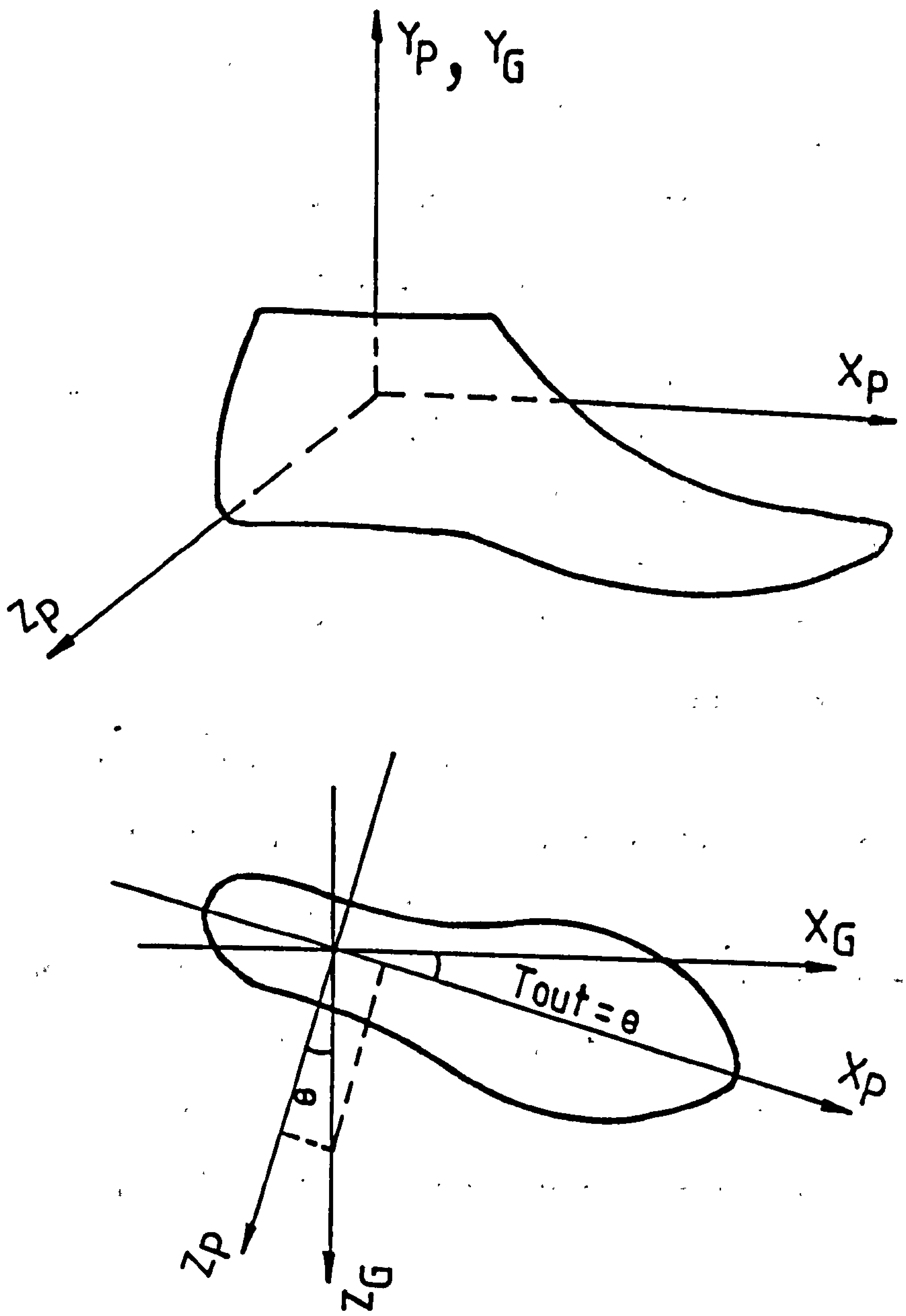


Figure 5.3 The foot principal axes related to the ground frame of reference.

Y_P is supposed to be parallel to the vertical axis of the ground frame of reference (Y_G) during standing, the foot has only toe out angle (θ) between the X_P and X_G, and it was measured on the coordinate measuring system. The relationship between the ground and principal axes can be derived (fig. 5.3):

$$\begin{aligned} X_P &= X_G COS\theta + 0 + Z_G SIN\theta \\ Y_P &= 0 + Y_G + 0 \\ Z_P &= -X_G SIN\theta + 0 + Z_G COS\theta \end{aligned}$$

Thus, the direction cosine matrix which defines the principal axes of inertia of the prosthetic foot (PF) relative to the ground frame of reference is:

$$[DCM_S^{PF}]_{P\leftarrow G} = \begin{bmatrix} COS\theta & 0 & SIN\theta \\ 0 & 1 & 0 \\ -SIN\theta & 0 & COS\theta \end{bmatrix} = \begin{bmatrix} D_X^{PF} \\ D_Y^{PF} \\ D_Z^{PF} \end{bmatrix} \qquad (5.14)$$

Where θ = toe out angle for the right amputee.

θ = - toe out angle for the left amputee.

The left hand side of equation 5.14 presents the direction cosine matrix (DCM) of the principal axes of inertia relative to the ground frame of reference (P$\leftarrow$G), and it is for the prosthetic foot (PF) calculated from the static (S) test. This notation system is used in all relations presented in this chapter.

The Unaffected Foot

The principal axes of the unaffected foot were defined using the same method as that of the prosthetic foot. During the static test, the subject was asked to stand with his feet flat on the floor, thus the Y_P of the unaffected foot was parallel to Y_G and the plane of the X_P Z_P axes was horizontal. X_P was defined by the line connecting the heel marker to the marker on the mid-toe. Therefore, equation 5.14 can be used to obtain the direction cosine matrix of the principal axes of the unaffected foot. It should be noticed that the angle

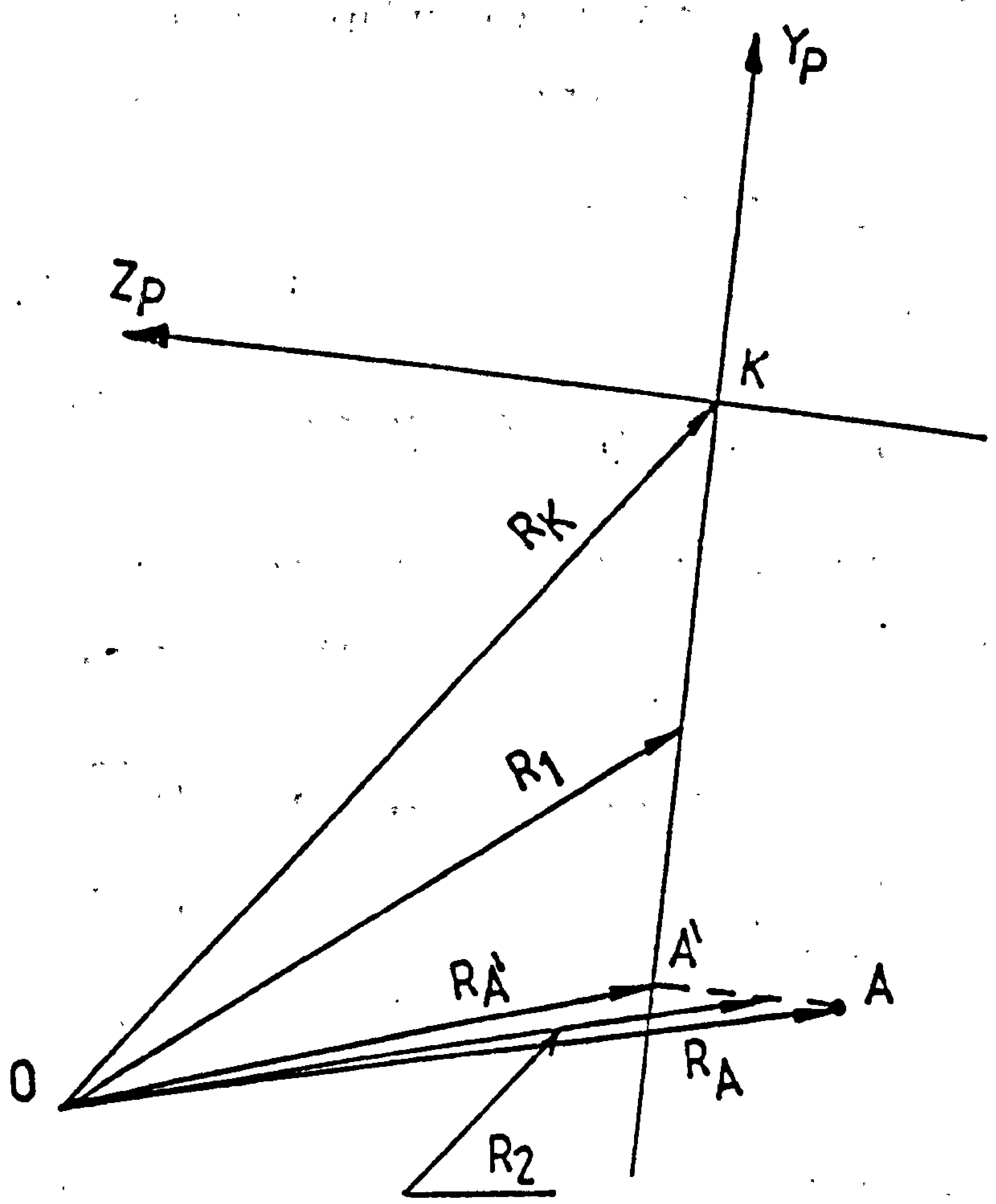


Figure 5.4 Determination of the shank principal axis Y_p for the prosthetic leg.

θ' which is the equivalent of the toe out angle, can be calculated from the toe and heel markers at the static test as follows for the right foot:

$$\theta' = \arctan\frac{Z_{toe}-Z_{heel}}{X_{toe}-X_{heel}} \tag{5.15}$$

The Prosthetic Shank

The principal axes of inertia of the prosthetic shank were defined to be parallel to those defined in section 4.1.1.2. The Z_P axis coincides with the knee axis and it is assumed to be parallel to the coronal plane. Z_P has a tilt angle α with Z_G and this angle was calculated in section 4.2.4 and considered to be positive when the lateral side is higher than the medial. Thus the direction cosine of Z_P can be calculated for a right amputee as:

$$\vec{D}_Z^{Psh} = [0 \quad \sin\alpha \quad \cos\alpha] \tag{5.16}$$

where: Psh = prosthetic shank, and α = - tilt angle for the left amputee.

To determine Y_P of the prosthetic shank, a plane was defined to be normal to Z_P and contain the knee joint centre K. The Y_P axis was then defined to be a line connecting the projection point A′ of the ankle joint centre A on that plane (see figure 5.4) to the knee joint centre K. The vector $R_{A'}$ which determines the position of A′ can be determined as follows:

The plane normal to Z_P should satisfy the condition of normality which is :

$$\vec{D}_Z^{Psh} \cdot (\vec{R}_1-\vec{R}_K) = 0 \tag{5.17}$$

Where $\vec{R}_1$ is the position vector of any point on that plane.

The line passing through the ankle joint centre and is perpendicular to the plane specified above is parallel to Z_P and should satisfy the relation:

$$\vec{R}_2-\vec{R}_A = c\vec{D}_Z^{Psh} \qquad (5.18)$$

Where $\vec{R}_2$ is the position vector of any point on the above (AA′) line and c is a constant.

Now the position vector of point A′ can be found by solving the equations 5.17 and 5.18 for the constant c when $\vec{R}_1$ equal to $\vec{R}_2$ because the point A′ is shared between the plane and the line under consideration. That is:

$$c = \vec{D}_Z^{Psh}.\ (\vec{R}_K-\vec{R}_A) \qquad (5.19)$$

Thus:

$$\vec{R}_{A'} = \vec{R}_A + c\vec{D}_Z^{Psh} \qquad (5.20)$$

Therefore, the direction cosine of the shank principal axis of inertias in the Y_P direction can be obtained as:

$$\vec{D}_Y^{Psh} = \frac{\vec{R}_K-\vec{R}_{A'}}{\left|\vec{R}_K-\vec{R}_{A'}\right|} \qquad (5.21)$$

The X_P of the prosthetic shank is normal to Y_P and Z_P and its direction cosine was calculated as:

$$\vec{D}_X^{Psh} = \vec{D}_Y^{Psh} \times \vec{D}_Z^{Psh} \qquad (5.22)$$

The Unaffected Shank

For the shank of the sound leg, the Y_P axis was considered to be connecting the ankle joint centre to the knee joint centre. Its direction cosine can be obtained as:

$$\vec{D}_Y^{Ush} = \frac{\vec{R}_k-\vec{R}_A}{|\vec{R}_K-\vec{R}_A|} \qquad (5.23)$$

The direction cosine of Z_P of the unaffected shank was calculated as the cross product of the ground X_G axis and the Y_P of the unaffected shank:

$$\vec{D}_Z^{Ush} = \vec{D}_X^{G} \times \vec{D}_Y^{Ush} \qquad (5.24)$$

Where: $\vec{D}_X^{G} = [1\ 0\ 0]$ *represents the X ground axis.*

The X_P of the unaffected shank was constructed as the cross product of Y_P and Z_P :

$$\vec{D}_X^{Ush} = \vec{D}_Y^{Ush} \times \vec{D}_Z^{Ush} \qquad (5.25)$$

The Thigh

The same method was used to define the thigh principal axes of inertia for the prosthetic and unaffected legs. The thigh Y_P axis was defined as the line connecting the knee joint centre (KJC) to the hip joint centre (HJC) and it is positive upwards. As the KJC and HJC are obtained from the static test, the direction cosine of Y_P can be calculated as:

$$\vec{D}_Y^{TH} = \frac{\vec{R}_H-\vec{R}_K}{|\vec{R}_H-\vec{R}_K|} \qquad (5.26)$$

Then, Z_P was defined to be normal to the plane which is defined by Y_P of the thigh and X_P of the shank, thus $\vec{D}_Z^{TH}$ can be obtained as the cross product of $\vec{D}_Y^{TH}$ and $\vec{D}_X^{SH}$:

$$\vec{D}_Z^{TH} = \vec{D}_X^{SH} \times \vec{D}_Y^{TH}$$

And naturally, the direction cosine of the X_P is:

$$\vec{D}_X^{TH} = \vec{D}_Y^{TH} \times \vec{D}_Z^{TH}$$

The direction cosine matrix of the principal axes of inertia for the prosthetic and unaffected thigh sections is:

$$[DCM_s^{TH}]_{P \leftarrow G} = \begin{bmatrix} D_X^{TH} \\ D_Y^{TH} \\ D_Z^{TH} \end{bmatrix} \qquad (5.27)$$

The left hand side of equation 5.27 represents the direction cosine matrix (DCM) of the principal axes of inertia relative to the ground frame of reference (P←G), and it is for the prosthetic and sound side thigh (TH) sections calculated from the static test.

The Trunk

The trunk was considered to consist of two segments, the upper torso and the pelvis. For both segments, the principal axes of inertia were defined to be parallel to the ground frame of reference when the subject is standing for the static test. Thus, the direction cosine matrix of the upper torso and pelvis relative to the ground frame of reference will take the following form :

$$[DCM_S^{Tor.Pel.}]_{P \leftarrow G} = \begin{bmatrix} 1 & 0 & 0 \\ 0 & 1 & 0 \\ 0 & 0 & 1 \end{bmatrix} \qquad (5.28)$$

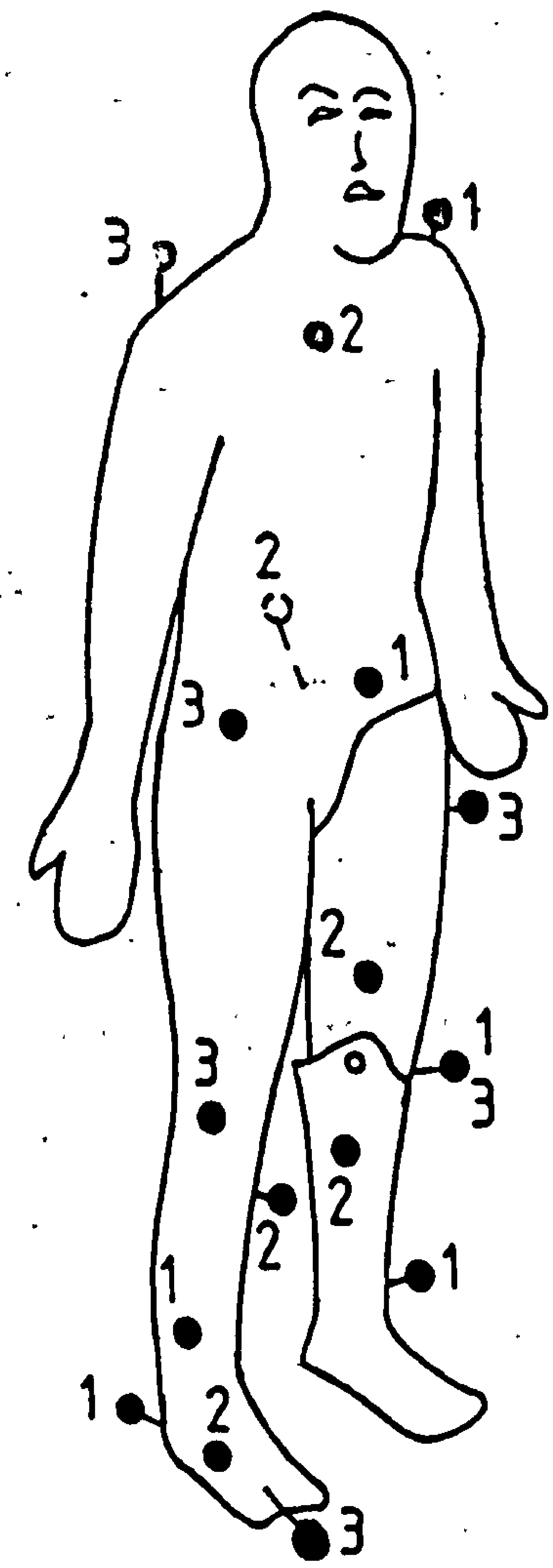


Figure 5.5 The body markers numbered in the sequence used to define the marker frame of reference.

5.4.2 The Marker Frame Of Reference

The marker frame of reference plays an important part in defining the orientation of the body segments. It provides the relationship between the principal axes of inertia and the ground frame of reference during walking. Three markers were attached to each body segment and numbered from 1 to 3 as seen in figure 5.5. The same method was used to define the marker frame of reference of all segments in relation to the ground frame of reference. This method was used during both the static and dynamic tests.

The origin of the marker frame of reference was considered to be at marker number 1. The Y_M axis of the marker frame of reference was directed from marker number 1 to marker number 3 as demonstrated in figure 5.6. The direction cosine " D_Y^M " of Y_M can be calculated as:

$$\vec{D}_Y^M = \frac{\vec{R}_3 - \vec{R}_1}{|\vec{R}_3 - \vec{R}_1|} \tag{5.29}$$

The X_M axis was defined to be normal to Y_M and to the line connecting marker number 1 to marker number 2. Thus, its direction cosine was calculated from the formula:

$$\vec{D}_X^M = \vec{D}_Y^M \times \frac{\vec{R}_2 - \vec{R}_1}{|\vec{R}_2 - \vec{R}_1|} \tag{5.30}$$

Simply, the $\vec{D}_Z^M$ was obtained from the cross product of $\vec{D}_X^M$ and $\vec{D}_Y^M$:

$$\vec{D}_Z^M = \vec{D}_X^M \times \vec{D}_Y^M \tag{5.31}$$

Thus, the direction cosine matrix which defines the marker frame of reference in relation to the ground frame of reference can be written as:

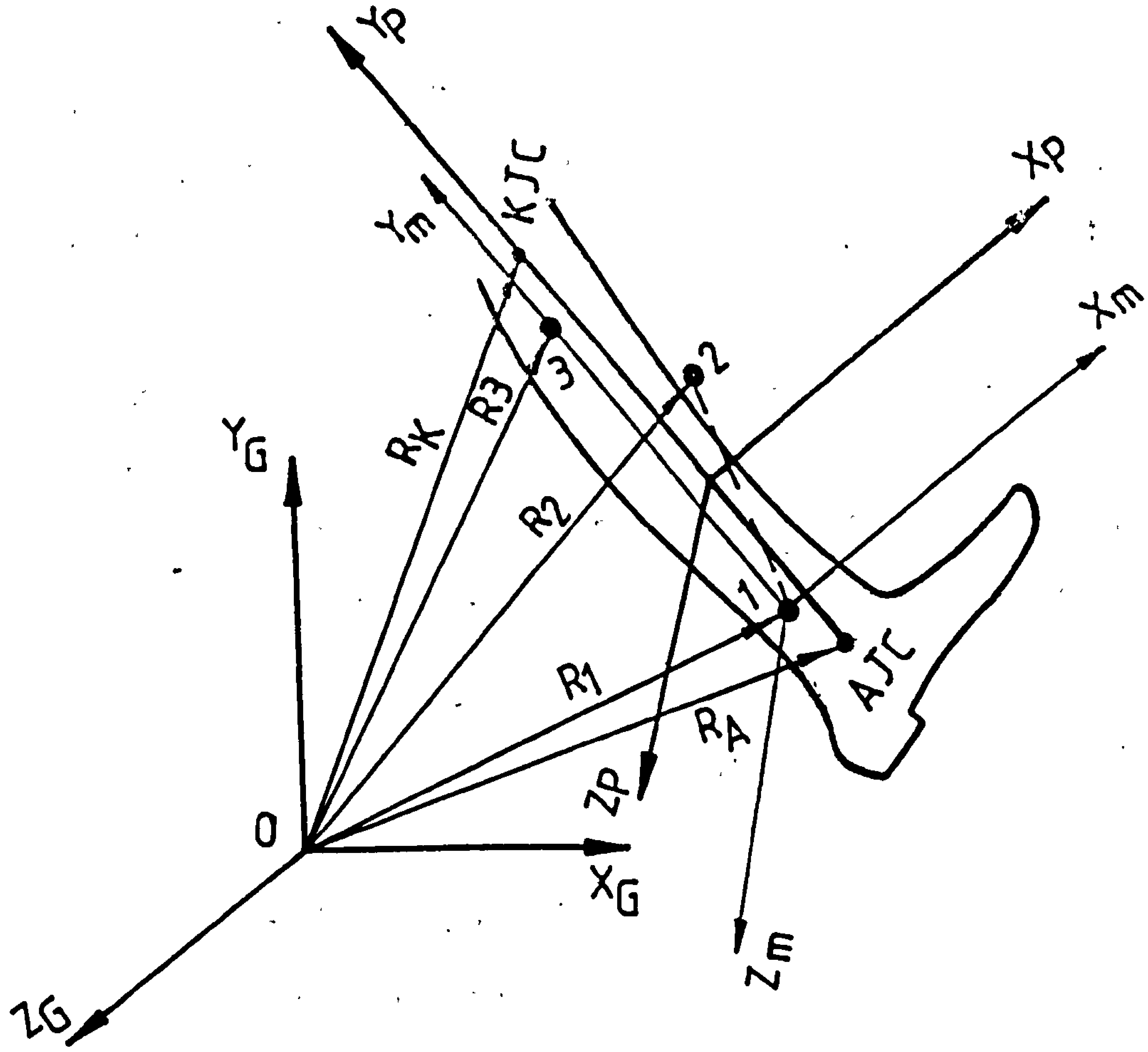


Figure 5.6 The shank marker frame of reference and its relationship with the ground and principal frames of reference.

$$[\,DCM\,]_{M\text{-}G} = \begin{bmatrix} D_X^M \\ D_Y^M \\ D_Z^M \end{bmatrix} \tag{5.32}$$

5.4.3 Segment Orientations

The static orientation of the body segments was determined in section 5.4.1 by relating the principal axes of inertia to the ground frame of reference. Similarly, the dynamic orientation of the segments can be determined by obtaining the relationship between the principal axes of inertia and the ground frame of reference during walking. This can be obtained by employing the marker frame of reference and was calculated as follows:

Firstly, the relationship between the principal axes of inertia and the marker frame of reference was found from the static test. It has the form:

$$\left[DCM_S\right]_{P\text{-}M} = \left[DCM_S\right]_{P\text{-}G}\left[DCM_S\right]_{M\text{-}G}^{T} \tag{5.33}$$

Secondly, as the relation between the principal axes and the marker frame of reference does not change during walking, the $[DCM_S]_{P\leftarrow M}$ remains the same during the static and dynamic tests. And thus, the dynamic relationship between the principal axes and the ground frame of reference can be obtained as:

$$\left[DCM_D\right]_{P\text{-}G} = \left[DCM_S\right]_{P\text{-}M}\left[DCM_D\right]_{M\text{-}G} \tag{5.34}$$

5.4.4 Determining the Inertia Properties of the Body Segments

The mass, centre of gravity and moment of inertia of the body segments were determined using the following methods:

(i) The Sound Leg

The mass of the sound leg segments was calculated using the method presented by Clauser et al (1969). More than one anthropometric measure was involved in calculating the mass of a segment. The mass of the shank and foot (m_{sh}) was calculated as one segment from:

$$m_{sh} = 0.111C_S + 0.047H_T + 0.122C_A + 0.003BM + 0.027L_F \qquad (5.35)$$

The mass of the anatomical thigh was calculated from:

$$m_{th} = 0.074BM + 0.123C_{th} + 0.027(0.78p - 0.27) \qquad (5.36)$$

In the above equations (5.35 and 5.36), p is the thickness of the folded iliac fat and it is measured in mm, while all the other dimensions are measured in cm. C_S is the shank circumference at the calf. H_T is the tibial height. C_A is the ankle circumference. L_F is the foot length. C_{th} is the upper thigh circle. BM is the mass of the body [kg].

The position of the centre of gravity (CG) and the mass moment of inertia (I) were calculated using the coefficients presented in table 2.1 which were derived by Goh (1982) by averaging available data as given below. The distance L_{GSH} from the KJC to CG of the shank/foot is:

$$L_{GSH} = C_2 L_{SH} = 0.4673 L_{SH} \qquad (5.37)$$

Where L_{SH} = the static ankle height + the shank length.

The length L_{GTH} from the HJC to the CG of the anatomical thigh was calculated as:

$$L_{GTH} = C_2 L_{TH} = 0.4169 L_{TH} \qquad : L_{TH} \textit{ is the thigh length} \qquad (5.38)$$

The moment of inertia of the shank/foot segment about the X and Z axes is:

$$I_{XX}^{SH} = I_{ZZ}^{SH} = m_{sh}(C_3\ L_{SH})^2 = m_{sh}(0.3467L_{SH})^2 \qquad (5.39)$$

The moment of inertia of the thigh is:

$$I_{XX}^{TH} = I_{ZZ}^{TH} = m_{th}(C_3 L_{TH})^2 = m_{th}(0.2912L_{TH})^2 \qquad (5.40)$$

I_{YY} of the shank and I_{YY} of the thigh were assumed to be negligible.

(ii) The Prosthetic Leg

A- The Prosthetic Shank

The mass (m_{psh}), the position of the centre of mass (CG_{PSH}) and the moment of inertia of the prosthetic shank/foot were determined and presented in section 4.1.2. However, the moment of inertia was calculated about the knee joint, and further calculation is needed to transfer that to axes through the shank centre of gravity. This was calculated by employing the parallel axis theorem. The moment of inertia of the prosthetic shank (shank/foot) about its centre of gravity was calculated as:

$$I_{XX}^{PSH} = I_{ZZ}^{PSH} = I_{PK}^{PSH} - m_{psh}(CG_{PSH})^2 \qquad (5.41)$$

I_{PK} is the shank moment of inertia about the prosthetic knee

B- The Prosthetic Thigh

The prosthetic thigh actually consists of two elements, the socket section of the prosthesis and the stump. Its inertia properties should be calculated by combining the two elements. The mass of the prosthetic thigh (m_{pth}) can be simply found as the sum of the stump mass and the mass of the socket section of the prosthesis:

$$m_{pth} = m_{st} + m_{so} \qquad (5.42)$$

The mass of the socket section of prosthesis (m_{SO}) was measured in section 4.1.2. To calculate the stump mass (m_{st}), centre of gravity (CG_{ST}) and stump moment of inertia (I_{ST}); the stump was modelled as a truncated circular

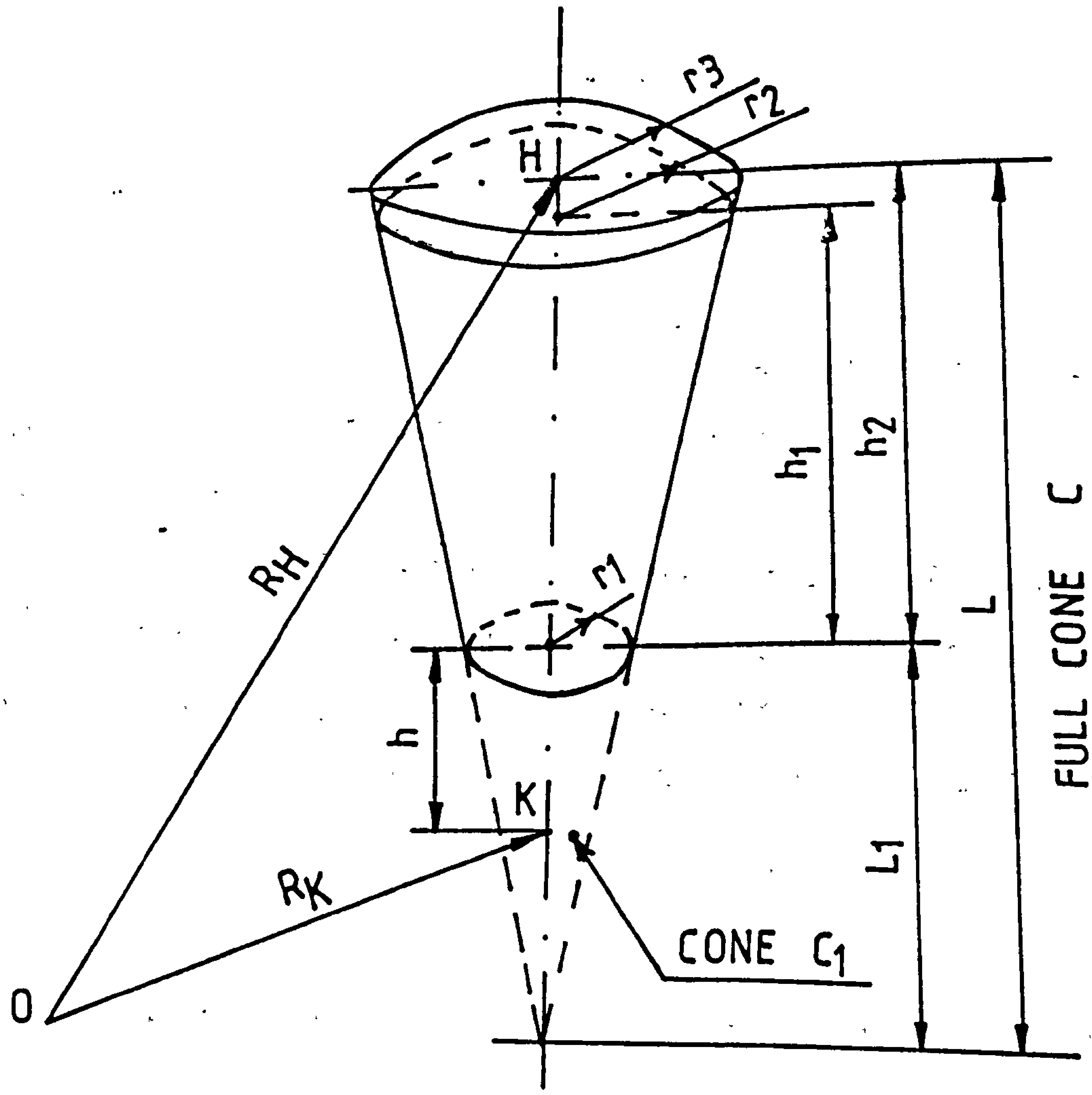


Figure 5.7 Stump model for the calculation of its mass properties.

cone (fig. 5.7). The top circle of that truncated cone is contained in the plane passing from the HJC and it is horizontal during the static test. The lower circle of the truncated cone is contained within the socket distal plane which was defined in section 4.1.1.3.

Now, considering figure 5.7 the mass of the stump can be obtained as:

$$m_{st} = \rho v_{st} \qquad : \rho = 1065.2 \; [kg/m^3] \; \textit{the mean density of the body}$$

v_{st} is the stump volume in m^3

The body density was calculated by Contini (1972) from data based on the work of Harless (1860) and Dempster (1955).

$$v_{st} = \frac{\pi}{3} h_2 \, (r_3^2 + r_3 r_1 + r_1^2) \qquad [m^3]$$

$$\textit{Thus,} \quad m_{st} = \frac{\pi}{3} \rho h_2 \, (r_3^2 + r_3 r_1 + r_1^2) \qquad [kg] \tag{5.43}$$

$$\textit{where} \; : \; h_2 = |\vec{R}_K - \vec{R}_H| - h$$

and h = socket lower point - knee joint centre (SL - K) measured at the coordinate measuring system.

r_3 is calculated as:

$$\frac{h_2}{r_3 - r_1} = \frac{h_1}{r_2 - r_1} \quad \Rightarrow \quad r_3 = r_1 + \frac{(r_2 - r_1) h_2}{h_1}$$

r_1 and r_2 are the radii of the socket lower and upper circles respectively, and were measured at the coordinate measuring system. h_1 is the distance between the socket lower and upper circles.

The CG_{st} can be obtained as a distance from the HJC (Shames 1980):

$$CG_{st}=\frac{h_2(r_3^2+2r_3r_1+3r_1^2)}{4(r_3^2+r_3r_1+r_1^2)} \qquad [m] \tag{5.44}$$

Thus, the centre of gravity of the prosthetic thigh from the hip joint centre can be calculated as:

$$CG_{PTH} = \frac{m_{st}CG_{st} + m_{so}CG_{so}}{m_{pth}} \qquad [m] \tag{5.45}$$

where CG_{so} is the distance of the socket centre of gravity measured from the hip joint centre:

$$CG_{so} = |\vec{R}_K - \vec{R}_H| - S_{so}$$

S_{so} is the distance from KJC to the socket centre of gravity measured in section 4.1.2.

The stump moment of inertia was calculated by considering the stump part from the complete cone which has the length L (fig. 5.7). The moment of inertia of this cone about the HJC is: $I + m\,(L/4)^2$ where I is the moment of inertia of the cone about its CG and m is the mass of that cone. It is possible to write the following:

$$I+m(\frac{L}{4})^2 = I_{st}+m_{st}CG_{st}^2+I_1+m_1(h_2+\frac{L_1}{4})^2 \tag{5.46}$$

where I_1, m_1, and L_1 are related to the cone C_1; I_{st} and I_1 are the moments of inertia of the respective components about their own centre of gravity.
The stump moment of inertia about its centre of gravity is:

$$I_{st} = I + m\left(\frac{L}{4}\right)^2 - \left[I_1 + m_1\left(h_2 + \frac{L_1}{4}\right)^2\right] - m_{st}CG_{st}^2 \tag{5.47}$$

Where:

$$L = \frac{r_3 h_2}{r_3 - r_1} \quad , \qquad L_1 = L - h_2$$

$$I = m\frac{3}{80}\left(L^2 + 4r_3^2\right) \qquad \textit{(Text book, Shames } 1980)$$

Finally, the moment of inertia of the prosthetic thigh about its principal axes is:

$$I_{XX}^{PTH} = I_{ZZ}^{PTH} = \left[I_{SOCG} + m_{SO}\left(\left|\vec{R}_K - \vec{R}_H\right| - CG_{PTH} - S_{so}\right)^2\right] + \left[I_{st} + m_{st}\left(CG_{PTH} - CG_{st}\right)^2\right] \tag{5.48}$$

where I_{SOCG} is the socket moment of inertia about its centre of gravity.

$I_{SOCG} = I_{so} - m_{so}\, S_{so}^2$: I_{so} is the socket moment of inertia about the KJC measured in section 4.1.2.

5.5 Kinematics

The kinematic analysis methods which were used in the work of this thesis are presented in this section. The method of determining the joint centres during walking, velocity, acceleration and the angular displacement of joint is discussed.

5.5.1 Determining the Dynamic Position of Joint Centres and Segments CG

The dynamic positions of the joint centres relative to the ground frame of reference were determined by employing the marker frame of reference and the features of its static and dynamic relationship with the ground frame of reference. The joint centres have fixed positions in relation to the marker frame of reference during the static and dynamic test. Thus, during the static test, each joint centre was related to the marker frame of reference of the adjacent segment. Later, during the dynamic test, the positions of the joint centres were first found relative to the marker frame of reference, they were then related to the ground frame of reference. The ankle and knee joint centres were related to the marker frame of reference of the shank. The prosthetic hip joint centre was related to the marker frame of reference of the socket, while the anatomical hip joint centre was related to the pelvis's marker frame of reference. The same procedure was applied on all joints and the following shows the application of the above method on the knee joint centre:

From the static test, the coordinates of the knee joint centre with respect to the marker frame of reference of the shank are X_K^M, Y_K^M, Z_K^M and can be obtained as:

$$\begin{bmatrix} X_K^M \\ Y_K^M \\ Z_K^M \end{bmatrix} = \left[DCM_S^{SH}\right]_{M-G} \begin{bmatrix} X_K^G - X_1^G \\ Y_K^G - Y_1^G \\ Z_K^G - Z_1^G \end{bmatrix}_{st.} \tag{5.49}$$

The coordinates of the joint centre with respect to the marker frame of reference do not change during walking. Therefore, the dynamic position of the knee joint centre with respect to the ground frame of reference can be obtained as:

$$\vec{R}_K^D = \begin{bmatrix} X_K^G \\ Y_K^G \\ Z_K^G \end{bmatrix}_{dy.} = \begin{bmatrix} X_1^G \\ Y_1^G \\ Z_1^G \end{bmatrix}_{dy.} + \left[DCM_D^{SH}\right]_{G-M} \begin{bmatrix} X_K^M \\ Y_K^M \\ Z_K^M \end{bmatrix} \tag{5.50}$$

Then, the dynamic position of the segments CG can be determined using the coefficients of table 2.1. Taking the thigh CG as an example, it can be written:

$$\vec{R}_{CG}^{TH} = \vec{R}_H + C_2(\vec{R}_K - \vec{R}_H) \qquad : C_2=0.4169 \qquad (5.51)$$

The above procedure was developed to determine the position of all joints and segments.

5.5.2 Calculating The Velocities and Accelerations

The linear velocities and accelerations of the joint centres were calculated by applying the differentiating technique, discussed in section 5.2, on the displacement data. The same procedure was applied on all joints and the knee joint is taken here as an example. The knee velocity V_K and acceleration a_K at any instant can be found by differentiating the position vector of the knee joint centre R_K :

$$V_K = \frac{dR_K}{dt} \quad , \quad a_K = \frac{dV_K}{dt} \qquad (5.52)$$

After calculating the linear velocities and accelerations of all joints, table 2.1 was used to calculate the linear velocities and accelerations of the CG of the anatomical shank and thigh:

$$\begin{aligned} V_{CG}^{SH} &= V_K + C_2^{SH}(V_A - V_K) \\ a_{CG}^{SH} &= a_K + C_2^{SH}(a_A - a_K) \end{aligned} \qquad (5.53)$$

$$\begin{aligned} V_{CG}^{TH} &= V_H + C_2^{TH}(V_K - V_H) \\ a_{CG}^{TH} &= a_H + C_2^{TH}(a_K - a_H) \end{aligned} \qquad (5.54)$$

The above method was also used for the prosthetic leg. It should be noted that the value of C_2 for the prosthetic shank and thigh was determined from the position of their CGs which were calculated for the shank in section 4.1.2 and for the thigh in equation 5.45.

The angular velocity of any segment was obtained from the movement of its principal axes in relation to the ground frame of reference. The position vector of any point of a segment in relation to the principal axes of inertia and ground frame of reference is:

$$\begin{bmatrix} X^P \\ Y^P \\ Z^P \end{bmatrix} = [DCM]_{P-G} \begin{bmatrix} X^G \\ Y^G \\ Z^G \end{bmatrix}, \quad \begin{bmatrix} X^G \\ Y^G \\ Z^G \end{bmatrix} = [DCM]^T_{P-G} \begin{bmatrix} X^P \\ Y^P \\ Z^P \end{bmatrix}$$

By differentiating the position vector in relation to the ground frame of reference it was found (text books, Shames 1980 P. 412):

$$\begin{bmatrix} V^G_X \\ V^G_Y \\ V^G_Z \end{bmatrix} = \frac{d}{dt}[DCM]^T_{P-G} \begin{bmatrix} X^P \\ Y^P \\ Z^P \end{bmatrix} = \left(\frac{d}{dt}[DCM]^T_{P-G}\right) [DCM]_{P-G} \begin{bmatrix} X_G \\ Y_G \\ Z_G \end{bmatrix}$$

$$= \begin{bmatrix} 0 & -\omega_Z & \omega_Y \\ \omega_Z & 0 & -\omega_X \\ -\omega_Y & \omega_X & 0 \end{bmatrix} \begin{bmatrix} X_G \\ Y_G \\ Z_G \end{bmatrix}$$

Thus,

$$\begin{bmatrix} 0 & -\omega_Z & \omega_Y \\ \omega_Z & 0 & -\omega_X \\ -\omega_Y & \omega_X & 0 \end{bmatrix} \cong \begin{bmatrix} \dot{m}_{11} & \dot{m}_{21} & \dot{m}_{31} \\ \dot{m}_{12} & \dot{m}_{22} & \dot{m}_{32} \\ \dot{m}_{13} & \dot{m}_{23} & \dot{m}_{33} \end{bmatrix} \begin{bmatrix} m_{11} & m_{12} & m_{13} \\ m_{21} & m_{22} & m_{23} \\ m_{31} & m_{32} & m_{33} \end{bmatrix} \qquad (5.55)$$

Therefore, the angular velocity of any segment in relation to the ground frame of reference is:

$$
\begin{aligned}
\omega_X &= \dot{m}_{13}m_{12}+\dot{m}_{23}m_{22}+\dot{m}_{33}m_{32} \\
\omega_Y &= \dot{m}_{11}m_{13}+\dot{m}_{21}m_{23}+\dot{m}_{31}m_{33} \\
\omega_Z &= \dot{m}_{12}m_{11}+\dot{m}_{22}m_{21}+\dot{m}_{32}m_{31}
\end{aligned}
\qquad (5.56)
$$

Then the angular acceleration of any segment was calculated using the numerical differentiator of section 5.2. And the acceleration components in relation to the ground frame of reference are:

$$
\varepsilon_X = \frac{d\omega_X}{dt} \quad , \quad \varepsilon_Y = \frac{d\omega_Y}{dt} \quad , \quad \varepsilon_Z = \frac{d\omega_Z}{dt} \qquad (5.57)
$$

As the above angular velocities and accelerations are reflected in the ground frame of reference, they were transferred to the principal frame of reference by using the corresponding direction cosine matrix. The following relation shows that for the acceleration:

$$
\begin{bmatrix} \varepsilon_X^P \\ \varepsilon_Y^P \\ \varepsilon_Z^P \end{bmatrix} = \left[DCM_D\right]_{P \leftarrow G} \begin{bmatrix} \varepsilon_X^G \\ \varepsilon_Y^G \\ \varepsilon_Z^G \end{bmatrix} \qquad (5.58)
$$

5.5.3 Angular Displacements of the Body Joint

(i) Trunk Orientation

The trunk was considered to be represented by the line which connects the midpoint between the two shoulder markers and the midpoint between the hip joint centres. The trunk angular orientations were calculated relative to the ground frame of reference as:

$$
Trunk\ Extension\ (TE) = \arctan\left[\frac{(X_{RH}+X_{LH})-(X_{RSH}+X_{LSH})}{(Y_{RSH}+Y_{LSH})-(Y_{RH}+Y_{LH})}\right] \qquad (5.59)
$$

where: RH and LH are the positions of the right and left hip joint centres respectively, and RSH and LSH belong to the right and left shoulder marker respectively. If the " TE " angle is positive the trunk is considered to be extended, and if it is negative the trunk is flexed.

Simi'arly, the trunk mediolateral tilt is:

$$Trunk\ Tilt\ (TT) = \arctan\left[\frac{(Z_{RSH}+Z_{LSH}) - (Z_{RH}+Z_{LH})}{(Y_{RSH}+Y_{LSH}) - (Y_{RH}+Y_{LH})}\right] \quad (5.60)$$

The TT angle is positive when the left shoulder is higher than the right shoulder and this is a case of tilt to the right.

Torso rotation was calculated using the shoulder markers:

$$Torso\ Rotation\ (TR) = \arctan\frac{X_{RSH} - X_{LSH}}{Z_{RSH} - Z_{LSH}} \quad (5.61)$$

Torso rotation was considered to be positive to the left. i.e. Anticlockwise when looking from above.

(ii) Femur Angle

The femur angle was calculated from the position of the knee joint centre related to the hip joint centre. Only the femur flexion/extension angle is discussed in this thesis and it was calculated relative to the ground frame of reference as:

$$Femur\ FlexionAngle = \arctan\frac{X_K - X_H}{Y_H - Y_K} \quad (5.62)$$

Femur flexion was considered to be positive and femur extension to be negative.

(iii) Knee and Ankle Angular Orientations

The knee flexion angle was considered to be between the two principal axes of the thigh and shank, and it was obtained as:

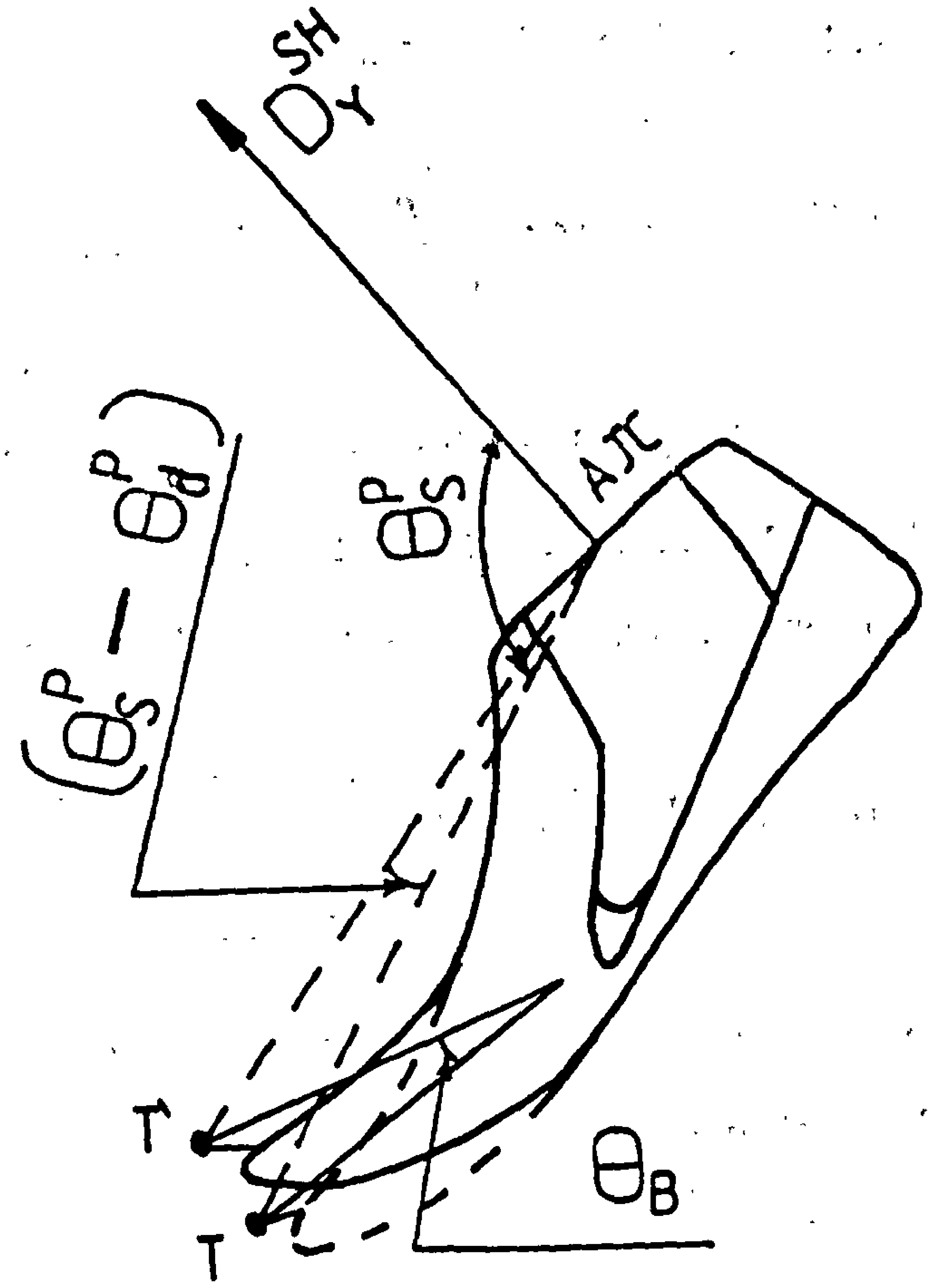


Figure 5.8 The dorsiflexion angle of the prosthetic foot.

$$Knee\ Flexion Angle = \arccos\left(\vec{D}_Y^{SH} \cdot \vec{D}_Y^{TH} \right) \tag{5.63}$$

To calculate the ankle dorsi/plantar flexion angle of the anatomical ankle, the angle θ between the foot X principal axis and the shank Y principal axis was first calculated:

$$\theta = \arccos\left(\vec{D}_X^{FT} \cdot \vec{D}_Y^{SH} \right) \tag{5.64}$$

the ankle dorsiflexion angle was then calculated by subtracting the dynamic value from the static value of that angle:

$$Ankle\ Dorsiflexion Angle = \theta_S - \theta_D \tag{5.65}$$

When $\theta_S > \theta_D$ the ankle is dorsiflexed and when $\theta_S < \theta_D$ the ankle is plantarflexed.

A similar procedure was used to find the prosthetic dorsiflexion angle. Firstly, the angle " θ^P " between the Y principal axis of the shank and the line which connects the ankle joint centre to the marker on the tip of the foot was calculated for the static and dynamic position (see fig. 5.8):

$$\theta^P = \arccos\left(\vec{D}_Y^{SH} \cdot \frac{(\vec{R}_{PT} - \vec{R}_{PA})}{\left|\vec{R}_{PT} - \vec{R}_{PA}\right|} \right) \tag{5.66}$$

the change in θ^P between the static and dynamic positions which represent the ankle dorsi/plantar flexion was then calculated as:

$$Prosthetic\ Ankle\ Dorsiflexion Angle = \theta_S^P - \theta_D^P \tag{5.67}$$

and this angle is positive when the foot is dorsiflexed.

Equation 5.67 actually determines the ankle dorsi/plantar flexion angle for the uniaxial foot only. For the SACH foot the ankle dorsi/plantar flexion angle is determined by angle θ_B caused by bending of the forefoot, as the ankle is rigid. Assuming that the distance from the ankle joint centre to the marker which is on the tip of the foot is approximately twice the distance from that marker to the bending centre of the foot, it follows that:

$$\theta_B = 2(\theta_S^P - \theta_D^P) \qquad (5.68)$$

5.5.4 Temporal Distance Parameters

As the subject walked on the force plates, the two force plates and the three TV cameras were operated simultaneously and synchronously at 50 Hz. Since only two force plates were used, the TV data were used in addition to the force plates data to calculate the temporal distance parameters.

The durations of stance phase for the left and right legs were calculated using the first and last active frames of the force plate data of the vertical force component. These frames corresponded to the heel strike and toe off of the foot respectively. Thus, if the left heel strike was at frame number F_{LHS} and the left toe off was at frame number F_{LTO} the duration of the left stance phase can be obtained as:

$$\textit{LeftStance Phase Duration} = 0.02(F_{LTO} - F_{LHS}) \quad [\,s\,] \qquad (5.69)$$

where 0.02 s is the duration of one frame.

For the stance phase duration of the right leg, it is possible to write:

$$\textit{RightStance Phase Duration} = 0.02(F_{RTO} - F_{RHS}) \quad [\,s\,] \qquad (5.70)$$

Equations 5.69 and 5.70 were calculated using the force plate data, but to calculate the duration of the gait cycle the frame number of the second left heel strike (F_{2LHS}) was determined from the TV data. This was by using the algorithm:

$$IF\ X_{(f+7)} - X_{(f)} < 4\ TV\ Units\ (1.2\ cm) : (\ f = f_{(LTO)}\ ,\ f_{(LTO+1)} \ldots\ldots) \quad (5.71)$$

Then $F_{2LHS} = f$ where, $X_{(f)}$ is the X coordinate of the left heel marker in the TV units at frame number f.

Thus, cycle duration can be calculated as:

$$Cycle\ Duration = 0.02\ (F_{2LHS} - F_{LHS}) \qquad [\ s\ ] \qquad (5.72)$$

Accordingly:

$$\begin{aligned} Step\ Width &= Z_{RHS+5}^{RAJC} - Z_{LHS+5}^{LAJC} \quad [\ m\ ] \\ RightStep\ Length &= X_{RHS}^{RAJC} - X_{LHS}^{LAJC} \quad [\ m\ ] \\ LeftStep\ Length &= X_{2LHS}^{LAJC} - X_{RHS}^{RAJC} \quad [\ m\ ] \\ Stride\ Length &= X_{2LHS}^{LAJC} - X_{LHS}^{LAJC} \quad [\ m\ ] \end{aligned} \qquad (5.73)$$

$$\begin{aligned} LeftDouble\ Support\ Duration &= 0.02(F_{LTO} - F_{RHS}) \quad [\ s\ ] \\ RightDouble\ Support\ Duration &= 0.02(F_{RTO} - F_{2LHS}) \quad [\ s\ ] \end{aligned} \qquad (5.74)$$

The subject's speed was calculated as the average of the speeds of RAJC, LAJC, RKJC, LKJC, RHJC, LHJC and the two shoulders marker. First, the speeds of all these points were calculated as:

$$V = \frac{X_{2LHS} - X_{LHS}}{Cycle\ Duration} \qquad [ms^{-1}] \qquad (5.75)$$

then the mean velocity of the subject was obtained as the average of the eight speeds of the above points.

5.6 Kinetics

5.6.1 Force Plate Data

The force plate data contain the 3D ground reaction forces (F_X , F_Y , F_Z) acting on the foot and the 3D moments (M_X , M_Y , M_Z) acting at the centre of the force plate. These forces and moments are measured in computer units

Table 5.1 The calibration factors of FP1 and FP2.

	FX	FY	FZ	MX	MY	MZ
FP1	-0.2843	-0.9923	0.1202	-0.2070	-0.0329	0.2068
FP2	0.2653	-1.0027	-0.1225	0.2070	-0.0329	-0.2117

The units are: [N/computer units] for the forces, and [Nm/computer units] for the moments.

and are related to the force plate frame of reference which is parallel to the ground frame of reference with its origin located at the centre of the force plate. Therefore,the data should be converted to SI units and should be transferred to the ground frame of reference.

The calibration factors of the force plate depend upon the setting of its charge and buffer amplifiers. The charge amplifiers were set to 50 mechanical unit per volt for F_X , F_Z , M_Y and 200 mechanical units per volt for F_Y , M_X and M_Z. The buffer amplifiers gains were set at 1 for F_X and F_Y, 2 for F_Z and M_Y and 1.25 for M_X and M_Z. This setting of the amplifiers was checked before every patient test to ensure the validity of the scaling factors.

Corresponding to the above settings of the charge and buffer amplifiers, calibration factors were supplied for force plate one (FP1) and for force plate two (FP2) as presented in table 5.1. It was reported by Kistler (see section 2.5.2) that an error of 0.05 percent and cross talk of ± 2 percent between F_X and F_Z, less than ± 3 percent to F_Y from F_X and F_Z, and less than ± 1 to F_X and F_Z from F_Y exist in the force plate with the above calibration factors. The validity of the calibration factors was checked every six months during the two year period of patient test for this work. The following explains the method of calculating the SI values of the forces and moments:

Force [N] = Calibration Factor x Force [computer units]

Moment [Nm] = Calibration Factor x moment [computer units]

Example:

$F_{FX} = -0.2843\ F_X$: F_X is in computer units and belong to FP1. And F_{FX} is the output of FP1 in SI units with respect to the force plate frame of reference.

The above procedure provides the forces F_F (F_X, F_Y, F_Z) and the moments M_F (M_X, M_Y, M_Z) in SI units related to the force plate frame of reference. To transfer these forces and moments to the ground frame of reference no change is applied on the forces of FP1 and FP2:

$$F_{GX} = F_{FX} \quad , \quad F_{GY} = F_{FY} \quad , \quad F_{GZ} = F_{FZ}$$

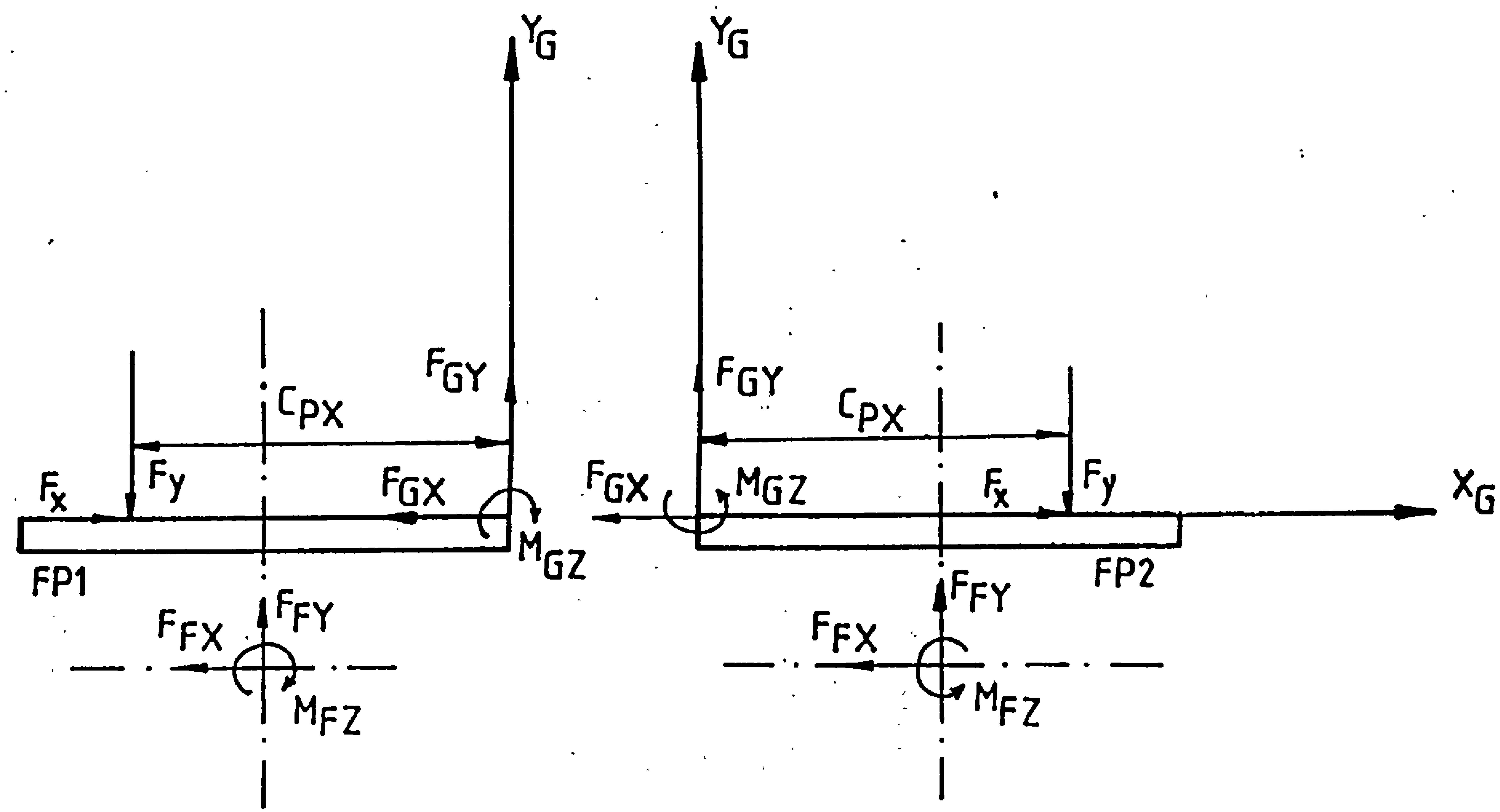


Figure 5.9 Centre of pressure determination.

But the moments will be changed to take the values of:

$$M_G = M_F + \vec{R} \times \vec{F}_F$$

Where R is the position vector of the centres of FP1 and FP2 related to the ground frame of reference and:

$$\begin{aligned} \vec{R}_1 &= [-0.303 \quad -0.04 \quad -0.11] \quad &for\ FP1 \\ \vec{R}_2 &= [\ 0.303 \quad -0.04 \quad 0.11\ ] \quad &for\ FP2 \end{aligned} \tag{5.76}$$

Example from FP2:

$$M_{GZ} = M_{FZ} + 0.303F_{FY} + 0.04F_{FX} \qquad [Nm]$$

Now all the forces F_G and the moments M_G are in SI units and are related to the ground frame of reference.

Calculation of the Position of Centre Of Pressure

The centre of pressure (CP) is the point between the floor and the foot in which the instantaneous reaction force is applied. The position of the CP (X, Y, Z) is of special importance in calculating the joints load. This position can be calculated in relation to the ground frame of reference by writing the equilibrium equation about the concerned axis at the force application point (see fig. 5.9):

$$\begin{aligned} CP_X &= \frac{M_{GZ}}{F_{GY}} \\ CP_Y &= 0 \\ CP_Z &= -\frac{M_{GX}}{F_{GY}} \end{aligned} \tag{5.77}$$

During progression of stance phase, a moment is generated about a vertical axis passing from the CP point. This is the friction moment (M_{CPY})

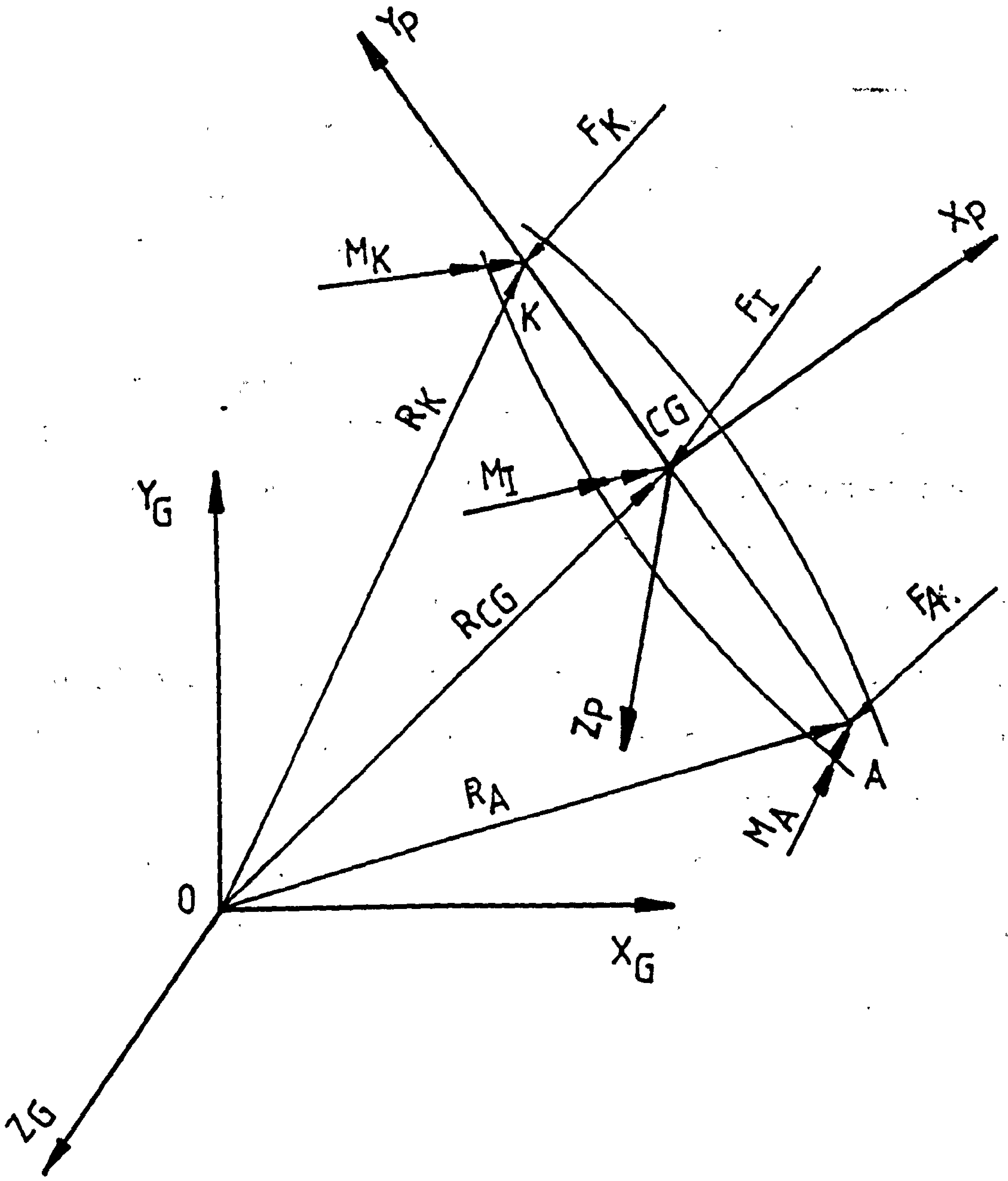


Figure 5.10 Resultant forces and moments applied on the shank segment.

between the foot and the ground and this is also important in the calculations of joint loads. This moment was obtained as:

$$M_{CPY} = M_{GY}+F_{GZ} \cdot CP_X-F_{GX} \cdot CP_Z \qquad (5.78)$$

5.6.2 The Joint Loads

So far the ground reaction forces and moments are determined relative to the ground frame of reference and at the centre of pressure. The positions of the body segments and joint centres are also determined from the TV data in relation to the principal axes of inertia and to the ground frame of reference. This is sufficient information to determine the inter-segmental loads at all joints. The same procedure was applied on all segments in order to calculate the inter-segmental loads and the following is an example showing the method of calculating the knee loads (fig. 5.10):

$$\overrightarrow{F_K} + \overrightarrow{F_I} + \overrightarrow{F_A} = 0 \qquad (5.79)$$

In this equation $\overrightarrow{F_K}$ is to be calculated. $\overrightarrow{F_I} = -m\ (\ a_{CG} + g\ ) : g = -9.806$ [ms^{-2}] in the vertical direction. $\overrightarrow{F_A}$ was calculated from the foot segment by applying the same procedure.

The equilibrium equation about the CG of the segment can be used to determine the knee moment $\overrightarrow{M_K}$:

$$\vec{M}_K+\overrightarrow{M_A}+\overrightarrow{MI_G}+\overrightarrow{F_A} \times (\vec{R}_A-\vec{R}_{CG})+\overrightarrow{F_K} \times (\vec{R}_K-\vec{R}_{CG}) = 0 \qquad (5.80)$$

Where:

$\overrightarrow{M_K}$ is to be calculated and $\overrightarrow{M_A}$ was found from the foot segment.

$\overrightarrow{MI_G}$ is the shank inertia moment expressed in the ground frame of reference and it was calculated as follows:

The segment moment of inertia $\overrightarrow{MI}$ about its principal axes of inertia can be obtained using "Euler equations" (Text book, Shames 1980 P. 794):

$$
\begin{aligned}
MI_X &= I_{XX}\varepsilon_X + \omega_Y\omega_Z(I_{ZZ}-I_{YY}) \\
MI_Y &= I_{YY}\varepsilon_Y + \omega_Z\omega_X(I_{XX}-I_{ZZ}) \\
MI_Z &= I_{ZZ}\varepsilon_Z + \omega_X\omega_Y(I_{YY}-I_{XX})
\end{aligned}
\tag{5.81}
$$

Where ε and ω are the segment angular accelerations and velocities in relation to the principal axes of inertia and were calculated in section 5.5.2.

The moment MI which is referred to the principal axes, can be related to the ground frame of reference by using the direction cosine matrix corresponding to the segment under consideration. Thus:

$$
MI_G = \left[DCM_D\right]^T_{P\leftarrow G} \; MI \tag{5.82}
$$

The knee forces $F_K(F_{KX}, F_{KY}, F_{KZ})$ and moments $M_K(M_{KX}, M_{KY}, M_{KZ})$ are related to the ground frame of reference and they can be transferred into the principal axes of the limb by multiplication in the direction cosine matrix $[DCM_D]_{P\leftarrow G}$.

5.7 Statistic

The variations among the temporal distance parameters of the normal subjects and the above-knee amputees, were determined using a t-test with no assumption. i. e. this test does not assume that the number of subjects is within the normal distribution.

Chapter Six

Results and Discussion

6.1 Introduction

This chapter presents the results obtained in this study for ten normal subjects, eight AK amputees wearing prostheses with quadrilateral (quad) sockets and three AK amputees wearing prostheses with ischial containment (IC) sockets. The experimental procedure of obtaining these results was described in chapter four and the analytical methods used for calculating the various gait parameters was presented in chapter five. Each gait parameter is the average of three different runs obtained from one day, except for the temporal-distance parameters' results of amputees with " normal alignment[1] " which were averaged for nine different runs obtained in three different days. However, because of some difficulties such as missing markers, it was not possible to have three successful runs for all alignment change sets of all subjects. Thus, for some of the alignment sets, data were averaged for two successful runs only; this will be pointed out whenever it is the case.

6.2 Subjects' Particulars

The particulars of the normal subjects and the AK amputees who were tested in this project are presented in tables 6.1 and 6.2 respectively. All subjects were numbered and coded in a manner that the code represents the patient, his side of amputation and the type of socket which was used on that patient. In the case of the normal subjects, the code represents the subject and the sex of that subject. The decoding is shown below tables 6.1 and 6.2. All subjects were males, this was controlled by the availability of the subjects and their willingness to perform the test. The normal subjects were between 28 and 34 years of age and most were colleagues of the author who was also one of the normal subjects. Although most of the patients were older than the normal subjects, the normal results are still of value to allow comparison of the gait patterns of the amputees with those of the normals, because Murray et al

[1] Normal alignment is the dynamic alignment which was set by the prosthetist for a patient before any alignment change sessions were undertaken.

Table 6.1 Particulars of the normal subjects.

Subject Number	Subject Code	Height [m]	Mass [kg]	Age [Years]	Sex
1	ALAM	1.730	73.95	34	Male
2	MUBM	1.755	67.55	30	Male
3	ZMCM	1.760	79.50	32	Male
4	LUDM	1.745	71.10	29	Male
5	SQEM	1.775	62.50	30	Male
6	MFFM	1.740	62.20	29	Male
7	SIGM	1.710	62.50	34	Male
8	SAHM	1.760	66.70	28	Male
9	MAIM	1.750	79.40	34	Male
10	NAJM	1.610	64.30	28	Male

Decoding:

The first three letters of the code belong to the subject identification.

The last letter of the code identifies the sex of the subject. M is a male subject.

(1964) studied sixty normal subjects in five groups of age (ranged from 20 to 65 years old), and they reported that no significant differences were found between the temporal-distance parameters, and the hip, knee, ankle, and trunk rotations of the first four groups which cover an age of 20 to 55 years. Also, two of the patients are of the same age group as the normal subjects and all patients are active. The activity scores of the amputees were calculated by the prosthetist who was involved in this work according to the method presented by Day (1981) and are shown in table 6.2.

6.3 Alignment Measurements

All alignment parameters of the test prostheses worn by the amputees were measured and are presented in table 6.3. The measurement method was explained in section 4.2 and the data presented are for normal alignment. Comparing the results with those presented by Zahedi et al (1986) and Yang Lang (1988), it is evident that the values measured in this study are broadly in the same range as those of the previous investigators. For instance, Zahedi et al presented an average of 0.35 $\mp$1.27 cm for knee set back and Lang obtained a range from -1 to 2 cm for the same parameter. The results of this study showed that a range from -1.2 to +0.9 cm was obtained for the knee set back parameter (fig. 4.12 and 3.33). Nine prostheses out of eleven were found to have their knees set back (the load line passes ahead of the knee) by not more than 0.9 cm and two prostheses only were found to have their knees set forwards. However, it should be noticed that patient DLAQ who had his knee set forwards by 1.2 cm was fitted with a safety uniaxial knee and his socket was aligned in 10.5 degrees of extension and set forwards by 4 cm. Thus, despite the fact that the prosthesis was fitted with a safety knee, the load line was passing ahead of the knee joint centre and the prosthesis was not unstable.

Examining the knee set out parameter, it was found that the results of ten patients were in a range from -1.2 to 0.5 cm (-0.1 to 0.1 cm from Yang Lang and -0.04 $\mp$1.11 cm from Zahedi et al), and only one patient (ILBI) had his knee joint centre set out by 3 cm. This patient had a relatively short stump

Table 6.2 Patients' Particulars.

Patient Number	Patient Code	Height [m]	Mass [kg]	Side Of Amputation	Sex	Age [Year]	Cause of Amputation	Activity Score (Day's 1981)
1	JLAQ	1.660	61.1	Left	Male	38	Trauma	+45
2	PLAQ	1.770	79.8	Left	Male	58	Trauma	+7
3	ILBQ	1.725	60.5	Left	Male	23	Trauma	+33
4	MRCQ	1.745	76.5	Right	Male	47	Trauma	+21
5	TRAQ	1.885	99.5	Right	Male	46	Trauma	+37
6	LRBQ	1.600	71.6	Right	Male	66	Osteoarthritis	-1
7	DLAQ	1.770	66.2	Left	Male	42	Trauma	+20
8	ELCQ	1.725	76.9	Left	Male	54	Sarcoma	+28

Patients 1-8 were fitted with quadrilateral socket under the above codes.

Patients 1,2 and 3 were also fitted with the Ischial Containment (IC) socket and for simplicity were given different numbers and codes as follows:

For the IC socket: patient number 1 was given the code **JLAI** and considered as patient number 9.
patient number 2 was given the code **PLAI** and considered as patient number 10.
patient number 3 was given the code **ILBI** and considered as patient number 11.

Decoding, Patient number 8 is taken as an example: **ELCQ**, the first letter **E** and the third letter **C** are the patient identity, the second letter **L** shows the side of amputation (left amputee), and the fourth letter may be **Q** (quadrilateral socket) or **I** (ischial containment socket). In this chapter the patient code has been extended to six letters such as **ELCQAF** in which **A** means visit number 1, and the last letter can be **F** (foot) or **S** (socket) or **K** (knee) alignment changes.

and such a large shift in knee set out can be attributed to an attempt to improve medio-lateral stability. It was noted that prostheses with IC sockets had the tendency to have a more pronounced knee set out than those of the quadrilateral sockets. This can also be related to an attempt to improve the medio-lateral stability of prostheses with IC sockets, however, it is not advised to report that prostheses with IC sockets are less stable in the medio-lateral plane than those with quad sockets, because three subjects only were fitted with IC sockets and more patient tests are suggested using the IC socket.

Excluding patient DLAQ, the quad sockets of all prostheses were aligned in flexion at a range from 6.4 to 10.6 degrees. This is within the range which was obtained by Lang in 1988 (0.4 to 10.1 degrees) and within the range of Zahedi et al in reported 1986 (-9.2 to 9.3 degrees). Socket flexion of subjects JLAQ, PLAQ and ILBQ was reduced by at least 3.9 degrees when their prostheses were fitted with IC sockets. This indicates that the IC sockets were aligned with less flexion than the quad sockets, however, it was reported by the prosthetist who aligned the prostheses that the IC sockets were not deliberately aligned in less flexion than the quad sockets.

Looking at the socket adduction parameter, it is clear that most of the quad sockets were in an abducted position while two of the IC sockets were in an adducted position and the third IC socket was neutral. This was related by the prosthetist of this work to the fact that he was able to adduct the IC socket more than the quad socket. Excluding subject LRBQ, a range from -9.8 to 11.8 degrees was obtained for the socket adduction parameter (-12.3 to 14.15 degrees in Zahedi at al and -10.0 to 0.2 degrees in Yang Lang). The range of this study is in agreement with that of Zahedi et al, and the small range which was obtained by Yang Lang (1988) can be related to the fact that he tested only four subjects. It is clear that the socket tilt has a greater variation in the ML than that in the AP plane. This confirms the suggestion of Zahedi et al (1986) that "the above-knee amputee can tolerate more variability in the ML direction than the AP". The ML tolerance can be related to the fact that the

Table 6.3 Results of alignment measurements for 11 AK prostheses (Normal Alignment).

Subject Code	Knee				Socket						Foot
	Height [cm]	Set Back [cm]	Tilt LSH [deg]	Set Out [cm]	Flexion [deg]	Rotated In [deg]	Adduction [deg]	Forward Set [cm]	Set Out [cm]	Height [cm]	Toe Out [deg]
JLAQ	35.0	0.4	0.2	-1.2	9.5	-3.8	-2.5	-1.1	-0.4	64.1	-9.6
PLAQ	39.6	0.8	2.3	-0.1	8.5	-14.4	-9.8	1.4	-1.1	69.8	-1.7
ILBQ	35.6	0.7	0.5	-0.3	6.4	-9.4	0.4	0.8	-2.4	71.3	-4.8
MRCQ	40.0	0.9	0.0	-0.2	8.0	-8.1	-2.6	-1.1	-0.6	70.9	-9.1
TRAQ	41.2	0.3	1.4	-0.6	10.6	2.0	-1.7	0.1	-0.9	72.2	-0.9
LRBQ	38.0	0.5	0.0	-0.2	7.0	10.7	-18.3	-1.0	-3.9	64.6	-5.9
DLAQ[1]	44.4	-1.2	2.6	-0.7	-10.5	-5.8	11.8	4.0	-0.3	67.0	1.4
ELCQ	38.8	0.4	0.0	0.4	7.8	-2.7	-5.3	0.2	0.2	70.3	-7.5
JLAI	34.9	0.6	-1.3	-0.3	5.6	-7.6	6.2	-0.9	2.3	60.9	-6.2
PLAI	40.5	-0.1	-1.1	0.5	4.3	0.0	0.0	-0.5	0.2	67.5	-6.0
ILBI	42.4	0.6	3.3	3.0	-1.1	-11.2	2.5	-0.2	-1.1	70.1	-1.1

Each of the above subjects was supplied with a SACH foot and a single axis knee.
(1) This subject was supplied with a uniaxial foot and a single axis Otto Book Safety knee.
LSH = Lateral Side Higher. ML = Medio-lateral. See fig. 3.33 for the definition of the above parameters.

knee mechanism is locked in the ML direction.

All subjects were found to have their feet rotated inwards, except the uniaxial foot which was rotated out by 1.4 degree. The toe out angle varied from -9.6 to +1.4 degrees, this was comparable to the results of Yang Lang (1988) who found that three subjects out of four had their feet rotated in and his data varied from -8 to +4.1 degrees.

Alignment Considerations:

In the following discussion, the foot is considered as a reference for the calculation of the alignment parameters. As mentioned in chapter four, alignment changes were applied at the foot, knee and socket. Changing the angular position of the foot by β degrees towards plantarflexion direction from its original position will cause socket flexion (by the same angle β) as seen in figure 6.1. If the foot is considered to be flat on the floor, the positions of the knee joint centre (KJC, point K) and the hip joint centre (HJC, point H) will also be shifted backwards. These shifts would be forwards if the foot angular changes were in the dorsiflexion direction. The magnitude of any shift in the position of KJC or HJC can be calculated as:

$$X = b \sin(\beta + \alpha) - b \sin \alpha \tag{6.1}$$

Where: X is the shift in the position of the point under consideration.

β is the foot angular change and it is positive when dorsiflexing the foot.

b is the distance between the AJC and the point under consideration.

α is the original angle (fig. 6.1) between a vertical line (y_a or parallel to it) and the line connecting the AJC to the point under consideration, and it is positive clockwise.

Changing the angular position of the socket will not affect the alignment parameters of the foot or the knee as the socket alignment changes will be conducted at the socket centre of rotation (SCR, point S in fig. 6.1). However, extending/flexing the socket will shift the HJC forwards/backwards by a

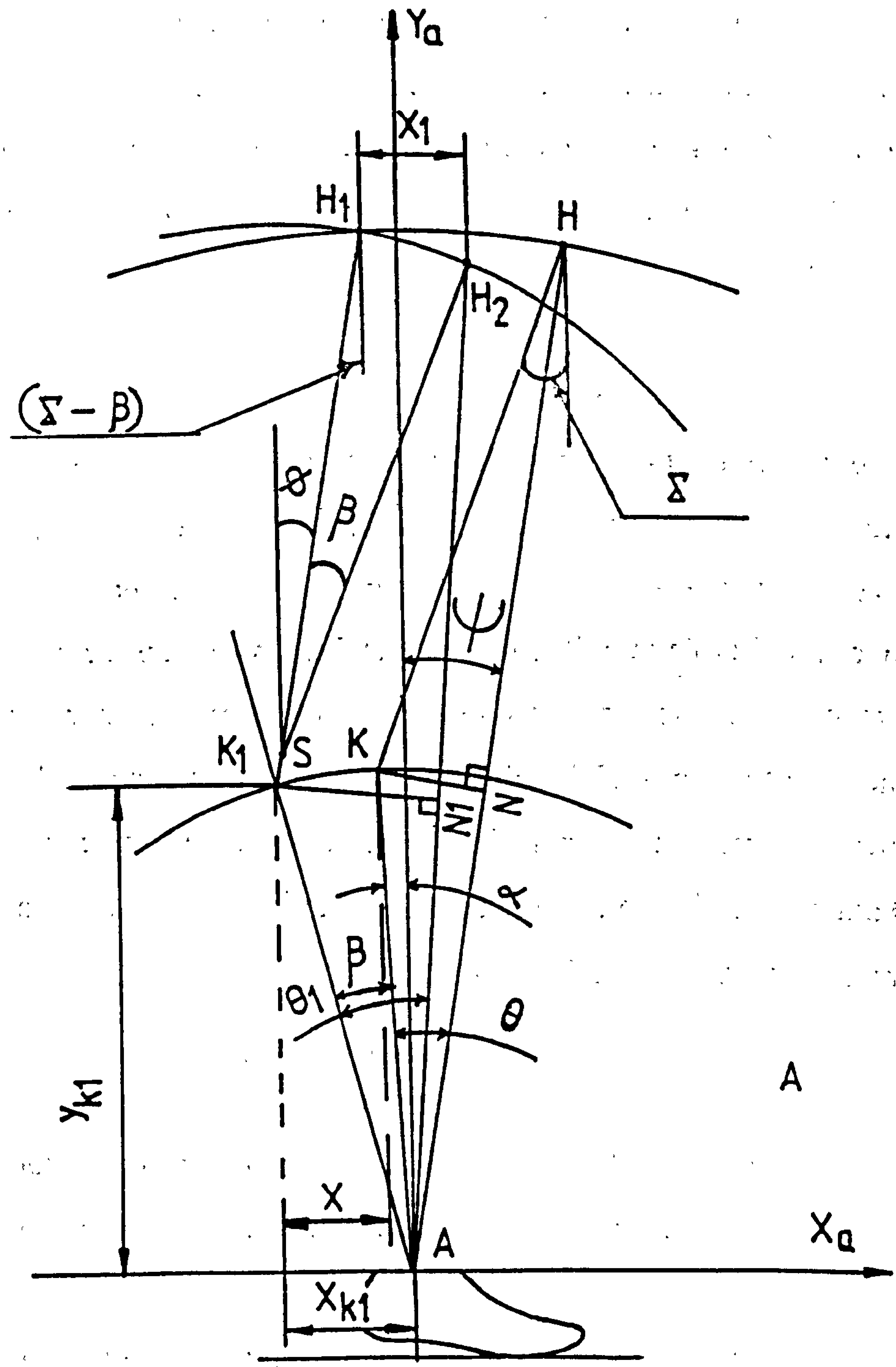


Figure 6.1 Calculation of the change in knee position relative to hip-ankle line due to equal and opposite changes in angular position of the foot and socket respectively. Refer to text section 6.3 for detailed explanation.

distance X1 which can also be calculated relative to the socket centre of rotation from equation 6.1 after replacing X by X1, b by L, α by Φ and β by γ, where L is the distance between the HJC and the socket centre of rotation, γ is the change in the socket angle and it is positive when extending the socket ($\gamma = \beta$ in fig. 6.1), and Φ is the original angle between the vertical line and the line connecting the HJC to the SCR and it is positive clockwise.

As mentioned, the above discussion is based on the assumption that the foot is considered as a reference for the calculation of the alignment parameters (see section 4.2.4) and the position of the ankle joint centre is fixed and not affected by the alignment changes. It is possible to consider the socket as a fixed reference for the calculation of the alignment parameters. This may be more reasonable than considering the foot as a reference because the socket is attached to the stump and together they form one entity. However, as the foot and the socket can be equally used without any geometrical preference, it was more convenient to consider the foot as a reference for measuring and calculating the alignment parameters and visualising the effect of the alignment changes on the whole prosthesis.

The knee alignment changes were carried out by shifting the KJC forwards and backwards from the original position without changing the angular configuration of the prosthesis. This was achieved by plantarflexing the foot and simultaneously extending the socket (or, dorsiflexing the foot and flexing the socket) by the same angle. To calculate the amount of shift of the KJC, equation 6.1 can be employed. However, as the knee stability is always controlled by the position of the KJC relative to the line which connects the HJC to the AJC, it was decided to consider the knee shift as the change in the distance between the KJC and the HA line (the line which connects the HJC to the AJC) and can be calculated as follows:

Referring to figure 6.1, the shift in the KJC away from the HA line which resulted from a simultaneous angular change (β) in the foot and socket can be calculated as:

$$a = K_1N_1 - KN$$

Where: " a " is the shift in the KJC relative to the HA line.

$KN = AK \sin \theta$ is the original distance between the KJC (K) and the HA line (before the two angular changes).

$K_1N_1 = AK \sin \theta_1$ is the distance between the KJC (K_1) and the H_2A line after the angular changes.

θ and θ_1 are the angles between the hip-ankle line and the knee-ankle line before (HA & KA) and after (H_2A & K_1A) the angular change respectively. θ is positive when the hip-ankle line passes behind the KJC, and θ_1 is positive when the H_2A line passes behind K_1.

The knee shift is forward when " a " is positive and backward when " a " is negative.

The original position of the HJC (point H) is known relative to the AJC from the static test (see chapter 4), and the KJC (point K) relative to the AJC (point A) was measured in section 4.2.2 using the coordinate measuring system. Thus, the distance KN can be found straightforwardly ($KN = AK \sin \theta$). To calculate the distance K_1N_1 after the angular change has been made, the new positions of the KJC (point K_1), the SCR (point S) and the HJC (point H_2) should be found relative to the AJC, which was not changed. The position of K_1 relative to the ankle frame of reference (y_a , x_a) can be found as:

$$x_{K1} = AK_1 \sin (\beta + \alpha)$$

$$y_{K1} = AK_1 \cos (\beta + \alpha)$$

Where: β is the angular change in the foot position, foot dorsiflexion taken as positive. α is the original angle between the y_a axis and the KA line, and it is positive clockwise.

A similar method can be employed for deriving the position relations of the other points relative to the AJC. However, the derivation of the coordinates of points H_1 (the position of the HJC after the foot change), H_2 , and S relative to the AJC, and the calculation of NK and N_1K_1 are shown in appendix A. A

Table 6.4 Temporal-Distance parameters of ten adult normal subjects.

Sub. Code	Velocity [m/s]	Cycle Duration [s]	Stride Length [m]	Step Width [m]	Left Step Length [m]	Right step Length [m]	Right Stance *	Left Stance *	Right Double Support *	Left Double Support *
ALAM	1.081 (0.025)	1.267 (0.031)	1.382 (0.013)	0.123 (0.016)	0.758 (0.025)	0.624 (0.035)	0.606 (0.013)	0.606 (0.017)	0.114 (0.006)	0.114 (0.006)
MUBM	1.032 (0.009)	1.360 (0.000)	1.380 (0.030)	0.115 (0.015)	0.696 (0.021)	0.684 (0.019)	0.609 (0.000)	0.638 (0.000)	0.130 (0.000)	0.130 (0.000)
ZMCM	1.099 (0.048)	1.253 (0.095)	1.356 (0.057)	0.128 (0.006)	0.705 (0.016)	0.650 (0.046)	0.634 (0.011)	0.623 (0.006)	0.136 (0.008)	0.136 (0.008)
LUDM	1.130 (0.053)	1.273 (0.012)	1.426 (0.074)	0.114 (0.012)	0.726 (0.045)	0.699 (0.039)	0.634 (0.011)	0.629 (0.018)	0.139 (0.014)	0.139 (0.014)
SQEM	1.315 (0.062)	1.020 (0.020)	1.330 (0.056)	0.105 (0.009)	0.682 (0.025)	0.657 (0.031)	0.603 (0.016)	0.596 (0.011)	0.109 (0.013)	0.109 (0.013)
MFFM	1.339 (0.018)	1.087 (0.012)	1.412 (0.027)	0.116 (0.023)	0.749 (0.030)	0.663 (0.004)	0.590 (0.015)	0.609 (0.017)	0.108 (0.001)	0.108 (0.001)
SIGM	0.945 (0.037)	1.413 (0.042)	1.338 (0.026)	0.089 (0.009)	0.702 (0.021)	0.636 (0.008)	0.619 (0.004)	0.628 (0.008)	0.130 (0.004)	0.130 (0.004)
SAHM	1.332 (0.047)	1.113 (0.012)	1.455 (0.088)	0.113 (0.014)	0.711 (0.064)	0.744 (0.025)	0.600 (0.006)	0.629 (0.004)	0.124 (0.001)	0.124 (0.001)
MAIM	1.142 (0.051)	1.233 (0.023)	1.447 (0.062)	0.080 (0.033)	0.768 (0.061)	0.679 (0.028)	0.617 (0.010)	0.633 (0.011)	0.133 (0.011)	0.133 (0.011)
NAJM[1]	1.160 (0.040)	1.130 (0.014)	1.297 (0.001)	0.108 (0.035)	0.655 (0.010)	0.642 (0.011)	0.626 (0.008)	0.635 (0.004)	0.139 (0.002)	0.139 (0.002)
Aver[2]	1.157 (0.135)	1.218 (0.125)	1.386 (0.064)	0.109 (0.021)	0.717 (0.045)	0.669 (0.042)	0.613 (0.017)	0.622 (0.017)	0.126 (0.013)	0.126 (0.013)

Each reading in this table is the average of 3 different runs carried out within one day. Figures between brackets are one standard deviation.
* The unit is: proportion of the cycle duration.
1 Data were averaged for 2 different runs carried out within one day.
2 Mean and SD were calculated for 29 different runs obtained for the above ten subjects.
See figure 2.21 for the definition of the temporal-distance parameters.

numerical example of calculation is also shown in appendix A for subject TRAQ. For this subject, it was found that by simultaneous plantarflexing of the foot and extending the socket by 3 degrees, the KJC was shifted backward relative to the line which connects the HJC to the AJC by 0.98 cm.

6.4 Temporal-Distance Parameters

6.4.1 Normal Subjects

Table 6.4 presents means and standard deviations of the temporal-distance parameters for the ten normal subjects of this study. The mean value of each parameter is the average of three different measurements carried out within one day. Since only three measurements were made the statistical confidence level would be low, however, the standard deviation (value between brackets) shows the closeness of the three readings from each other and from their mean. The temporal-distance parameters were defined for the normal subjects and the amputees in accordance with the method presented by Inman et al in 1981 (see chapter two of this thesis), and the method of calculation was discussed in chapter five.

Comparing the results of this study with those of previous work, it was found that the velocity (1.16 m/s) obtained for the subjects of this study was comparable to that obtained by Chao et al (1983) and Kadaba et al (1989) who reported a mean velocity for adult men of 1.2 m/s and 1.3 m/s respectively, however, the subjects of this study walked with a 24% slower than those of Murray et al (1980). This can only be referred to the gait pattern which was chosen by the subjects of this study, as the average of their heights (1.73 m) was approximately similar to that (1.75 m) of the subjects of Murray et al. It was also found that the mean stride length (1.39 m) was comparable to that presented by Chao et al (1983) who reported a (stride length)/(leg length) equal to 1.56 (1.4 m assuming leg length = 0.9 m), and Kadaba et al (1989) who obtained a stride length of 1.36 m for the normal subjects. However, the stride length obtained in this study was slightly shorter (11%) than that obtained by Murray et al (1964) and Murray et al (1980). This difference in the stride length can be attributed to the differences in the velocity between the subjects of Murray et al and the subjects of this study. The cycle duration (1.22 s), was 15% longer than that obtained by Murray et al (1980), and again the reason is referred to the differences in the velocity of subjects of this study and subjects of Murrayet al.

The mean step width of the normal subjects of this study was 0.109 ∓ 0.021 m. This is in agreement with that obtained by Murray et al (1964) who presented a mean of 8 ∓ 3.5 cm of the gait width parameter which was defined in similar manner to that of this study (distance between the right and left AJC). The results of this study are in disagreement with those obtained by Chodera and Levell (1972) and Akidele (1987). Chodera and Levell reported a mean of 6.02 ∓ 1.25 cm for the gait width which was measured between the right and left heels. Akidele presented a mean step width of 6.59 cm for normal subjects. However, Akidele defined the step width as the perpendicular distance between two parallel lines, one of which passes through the point of the right heel strike and parallel to the line of progression, and the other passes through the point of left heel strike and also parallel to the line of progression. As age and height have no effect on the step width (Murray et al 1964), the above differences in the step width between this study and the other work (Chodera 1972 and Akidele 1987) can partly be related to the different definitions adopted by the researchers for the gait width. This is confirmed by the fact that the results of this study were in agreement with those of Murray et al (1964) who adopted a similar definition of the gait width to that adopted in this study as shown above. The gait width differences can also be affected by the pattern of walking which is adopted by the different subjects.

The mean left step length measured in this study was 0.72 m and was comparable to that measured by Chao et al (1983) who obtained a mean step length of 0.73 m, and was 8% shorter than that reported by Murray et al (1964) for subjects of similar age. This can be explained as discussed before for the stride length. It was found that the mean right step length was 0.67 m and was

Table 6.5 Temporal-Distance parameters of 11 AK amputees. (Normal alignment).

Sub. CODE	Velocity [m/s]	Cycle Duration [s]	Stride Length [m]	Step Width [m]	Sound Step Length [m]	Prosth step Length [m]	Prosthetic Stance *	Sound Stance *	Prosthetic D Support *	Sound D Support *
JLAQ[1]	1.044 (0.052)	1.290 (0.043)	1.339 (0.031)	0.106 (0.016)	0.638 (0.025)	0.701 (0.014)	0.570 (0.008)	0.638 (0.017)	0.104 (0.010)	0.119 (0.016)
PLAQ	0.778 (0.039)	1.480 (0.062)	1.162 (0.048)	0.238 (0.021)	0.538 (0.032)	0.624 (0.032)	0.562 (0.020)	0.680 (0.027)	0.123 (0.011)	0.132 (0.023)
ILBQ	1.292 (0.044)	1.216 (0.030)	1.568 (0.023)	0.220 (0.044)	0.725 (0.041)	0.843 (0.020)	0.567 (0.010)	0.631 (0.012)	0.101 (0.012)	0.113 (0.003)
MRCQ[2]	1.222 (0.083)	1.140 (0.072)	1.364 (0.029)	0.162 (0.027)	0.710 (0.014)	0.663 (0.024)	0.557 (0.007)	0.670 (0.012)	0.121 (0.007)	0.123 (0.007)
TRAQ	1.048 (0.039)	1.402 (0.054)	1.463 (0.043)	0.136 (0.035)	0.732 (0.033)	0.731 (0.055)	0.573 (0.008)	0.660 (0.013)	0.099 (0.004)	0.148 (0.009)
LRBQ	0.642 (0.042)	1.713 (0.094)	1.091 (0.026)	0.147 (0.023)	0.457 (0.017)	0.634 (0.019)	0.534 (0.017)	0.670 (0.022)	0.081 (0.004)	0.134 (0.012)
DLAQ	0.907 (0.050)	1.420 (0.028)	1.281 (0.054)	0.189 (0.023)	0.569 (0.058)	0.712 (0.043)	0.542 (0.016)	0.693 (0.033)	0.109 (0.005)	0.139 (0.039)
ELCQ	0.798 (0.069)	1.462 (0.100)	1.179 (0.046)	0.230 (0.017)	0.514 (0.055)	0.665 (0.039)	0.556 (0.015)	0.679 (0.022)	0.129 (0.010)	0.120 (0.026)
Aver.	0.966 (0.226)	1.390 (0.178)	1.306 (0.161)	0.179 (0.048)	0.610 (0.106)	0.697 (0.070)	0.558 (0.016)	0.665 (0.021)	0.108 (0.016)	0.129 (0.012)
JLAI	1.132 (0.054)	1.267 (0.050)	1.439 (0.029)	0.106 (0.022)	0.680 (0.017)	0.759 (0.019)	0.572 (0.012)	0.625 (0.009)	0.100 (0.011)	0.112 (0.007)
PLAI	0.981 (0.030)	1.273 (0.033)	1.246 (0.019)	0.209 (0.023)	0.538 (0.023)	0.708 (0.015)	0.560 (0.007)	0.653 (0.008)	0.117 (0.008)	0.112 (0.009)
ILBI	1.123 (0.064)	1.347 (0.078)	1.516 (0.054)	0.167 (0.053)	0.673 (0.023)	0.843 (0.046)	0.569 (0.012)	0.631 (0.013)	0.112 (0.007)	0.103 (0.006)
Aver.	1.079 (0.085)	1.296 (0.045)	1.400 (0.139)	0.161 (0.052)	0.630 (0.080)	0.770 (0.068)	0.567 (0.006)	0.636 (0.015)	0.110 (0.008)	0.109 (0.005)

Each reading in this table is the average of 9 different runs carried out in 3 different days. Figures between brackets are one dtandard deviation.
** The unit is: proportion of the cycle duration.*
1 The data were averaged for 6 different runs carried out on 2 different days.
2 The data were averaged for 6 different runs carried out on 3 different days.
See figure 2.21 for the definition of the temporal distance parameters.

significantly ($P<0.01$) shorter than the mean left step length. This is in disagreement with the above mentioned studies and shows that the right and left legs of the normal subjects are not always symmetrical, and also suggests that to obtain the best results from any gait analysis test, the right and left legs should be treated separately and measured simultaneously.

The duration of the right (61%) and left (62%) stance phase and the double support time (12.6%) as percent of the gait cycle were in agreement with those presented by Murray et al (1964) who reported duration for stance phase and double support of 61% and 11% of the gait cycle respectively, Zuniga et al (1972) who reported 61% and 10% of the gait cycle for the stance phase and double support respectively, Chao et al (1983) who obtained 59% and 8.8% $\mp$ 1.9% of the gait cycle for the stance phase and double support respectively and Kadaba et al (1989) who reported a swing to stance ratio of 0.64. There were no significant differences ($P>0.1$) between the stance phase or the double support durations of the right and left sides. At the right side the swing phase existed for 39% of the gait cycle, and at the left side, swing phase accounted for 38% of the gait cycle.

6.4.2 Amputees with Normal Alignment

As mentioned in the previous chapter and in section 6.1 of this chapter, normal alignment is the dynamic alignment which was achieved by the prosthetist before conducting any experimental alignment change. The temporal-distance parameters of the amputees with normal alignment are shown in table 6.5. In this study, each patient was called to the laboratory three times on three different days. On each occasion the test procedure was started by obtaining three runs for the patient with his prosthesis in normal alignment. Therefore, each reading of those in table 6.5 presents the mean of nine different measurements carried out on three different days.

The average velocity (0.966 m/s for the quad subjects, and 1.079 m/s for the IC subjects) obtained in this study for the AK amputees, is in agreement with those presented by James and Oberg (1973), Godfrey et al (1975) and

Murray et al (1983), who reported an AK amputees' velocity of 0.94 m/s, 0.85∓0.18 m/s and 1.07 m/s respectively. The average stride length obtained in this study (1.31 m for the quad, and 1.4 m for the IC subjects) is also comparable to those obtained by the above listed investigators (1.29 m, 1.19∓0.2 m, and 1.42 m respectively). It was also found that the cycle duration obtained in this study (1.39 s for the quad, and 1.30 s for the IC subjects) was in agreement with those reported by the above listed investigators (1.41 s, 1.5∓0.24 s and 1.34 s respectively).

The average step widths obtained in this study were 17.9 cm and 16.1 cm for patients wearing quad and IC sockets respectively. The step width obtained in this study was comparable to those reported by Murray et al (1980) and Murray et al (1983) who reported a step width of 17.4 cm and 16.4 cm respectively. The average sound and prosthetic step length measured in this study for patients with quad sockets were 61.0 cm and 69.7 cm respectively. This is in agreement with that (61.7 cm for the sound and 67.5 cm for the prosthetic step length) presented by James and Oberg (1973), and slightly shorter than that (67 cm for the sound and 76 cm for the prosthetic step length) obtained by Murray et al (1983). As one would expect, significant differences ($P<0.05$) were found between the prosthetic and sound step lengths of the patients fitted with quad sockets. It was found that the prosthetic step length was longer than that of the sound leg. This can be explained by the fact that the patients were more comfortable, secure and stable when supported by the sound leg than when supported by the prosthesis. The short step length of the sound leg in comparison with the prosthetic leg, can also be related to the lack of foot plantarflexion on the prosthetic side at push off and prior to the sound heel strike. Although the sound step length of the patients fitted with IC sockets was slightly shorter than that of the prosthetic side, the difference was not significant ($P>0.05$).

The durations of stance phase (66.5% for the sound side, and 55.8% of the cycle duration for the prosthetic side) and double support time (12.9% for the sound side, and 10.8% of the cycle duration for the prosthetic side) for the subjects which were fitted with quad sockets, were comparable to those (65% and 57% of the gait cycle for the sound and prosthetic stance respectively, and 11% of the gait cycle for the sound and prosthetic double support) presented by James and Oberg (1973). The stance phase durations of the sound and prosthetic sides were also comparable to those (68% and 58% of the gait cycle for the sound and prosthetic stance phase respectively) presented by Murray et al (1980). The stance phase of the sound side was significantly longer ($P<0.01$) than that of the prosthetic side for patients fitted with the quad socket. The sound double support duration of these patients was also significantly ($P<0.05$) longer than that of the prosthetic side. This was expected and can be explained as discussed for the differences in the step length between the prosthetic and sound legs.

6.4.2.1 Comparison Between Quadrilateral and IC Sockets

The average stride length of patients fitted with quad sockets (1.306 m) was shorter than that of patients fitted with IC sockets (1.4 m), and the average cycle duration (1.39 s) of patients fitted with quad sockets was longer than the average cycle duration (1.29 s) of subjects fitted with IC sockets. The velocity of the subjects wearing IC sockets was therefore 10.5% higher than the velocity of the subjects wearing quadrilateral sockets. Two subjects (PLA and ILB) walked with a narrower based gait when fitted with IC sockets than when fitted with quad sockets. However, subject JLA with a long healthy stump, described by the prosthetist as a good walker, walked with the same step width when fitted with either socket. The prosthetic and sound step lengths of the patients wearing IC sockets were slightly longer than those of the patients wearing quad sockets but not significantly ($P>0.1$).

As discussed above, the stance phase of the sound side was significantly longer ($P<0.01$) than that of the prosthetic side for patients fitted with the quad socket, and the sound double support duration of these patients was also significantly ($P<0.05$) longer than that of the prosthetic side. For patients with

IC sockets, stance phase of the prosthetic side was noticeably shorter ($P=0.08$) than that of the sound side, and the prosthetic and sound double support had no significant differences ($P>0.1$). This indicates that patients fitted with IC sockets walked with more even parameters between the sound and prosthetic sides than those fitted with quad sockets, and this can be related to the fact that the patients were more comfortable with the IC socket than with the quad socket as reported by the three patients. However, since all of the above differences between the parameters of subjects wearing quad and subjects wearing IC sockets had no significant meaning, and because only three subjects wore IC sockets, definite conclusions cannot be drawn, thus, further testing for patients with IC sockets is suggested.

6.4.2.2 Comparison Between Normal Subjects and Amputees with Normal Alignment

As the differences between the parameters of subjects wearing quad and subjects wearing IC sockets were insignificant, the following comparison between the amputees and the normal subjects will not distinguish between the two groups of amputees. Although, some differences were found between stride length of the normal subjects and amputees with normal alignment, these differences were not significant ($P>0.1$). This can be attributed to the fact that the normal subjects and the amputees were of the same average height (1.73 m), and all amputees were active. The cycle duration of the normal subjects had no significant differences ($P>0.05$) from that of the amputees, although in general most normal subjects tended to walk faster than the amputees. This tendency was found with most subjects but on average, the differences between the velocity of the normal subjects and amputees were insignificant ($P>0.1$). This can be accounted by the fact that the normal subjects used in these series of tests are slightly slower walkers than that reported in the literature and the amputees are active.

As expected, normal subjects walked with a significantly ($P<0.05$) narrower step width than the amputees, which can be explained by the poor

Table 6.6 Average results for the effect of foot alignment changes on the temporal-distance parameters of AK amputees. The normal alignment results are also presented.

The Foot Change	Velocity [m/s]	Cycle Duration [s]	Stride Length [m]	Step Width [m]	Sound Step Length [m]	Prosth step Length [m]	Prosthetic Stance *	Sound Stance *	Prosthetic D Support *	Sound D Support *
Quadrilateral Socket (quad)										
NORM.AL	0.966 (0.226)	1.390 (0.178)	1.306 (0.161)	0.179 (0.048)	0.610 (0.106)	0.697 (0.070)	0.558 (0.016)	0.665 (0.021)	0.108 (0.016)	0.129 (0.012)
3DEG.DORS	0.950 (0.218)	1.417 (0.169)	1.305 (0.147)	0.156 (0.050)	0.615 (0.112)	0.683 (0.091)	0.549 (0.022)	0.671 (0.021)	0.107 (0.014)	0.122 (0.016)
6DEG.DORS	0.955 (0.223)	1.397 (0.175)	1.294 (0.166)	0.154 (0.053)	0.611 (0.103)	0.683 (0.089)	0.558 (0.018)	0.664 (0.023)	0.109 (0.017)	0.128 (0.014)
3DEG.PLAN	0.958 (0.212)	1.385 (0.177)	1.298 (0.155)	0.175 (0.052)	0.594 (0.093)	0.702 (0.078)	0.558 (0.025)	0.664 (0.026)	0.112 (0.020)	0.125 (0.017)
6DEG.PLAN	0.921 (0.239)	1.382 (0.163)	1.289 (0.158)	0.179 (0.057)	0.593 (0.089)	0.696 (0.082)	0.557 (0.015)	0.664 (0.026)	0.109 (0.018)	0.126 (0.013)
Ischial Containment Socket (IC)										
NORM.AL	1.079 (0.085)	1.296 (0.045)	1.400 (0.139)	0.161 (0.052)	0.630 (0.080)	0.770 (0.068)	0.567 (0.006)	0.636 (0.015)	0.110 (0.008)	0.109 (0.005)
3DEG.DORS	1.050 (0.108)	1.326 (0.035)	1.396 (0.168)	0.161 (0.048)	0.651 (0.086)	0.746 (0.087)	0.566 (0.003)	0.639 (0.022)	0.108 (0.016)	0.112 (0.009)
6DEG.DORS	1.009 (0.079)	1.370 (0.084)	1.391 (0.136)	0.149 (0.045)	0.638 (0.085)	0.753 (0.061)	0.554 (0.012)	0.640 (0.023)	0.100 (0.009)	0.108 (0.004)
3DEG.PLAN	1.022 (0.062)	1.351 (0.101)	1.389 (0.140)	0.161 (0.069)	0.632 (0.098)	0.758 (0.056)	0.573 (0.016)	0.635 (0.039)	0.112 (0.011)	0.108 (0.011)
6DEG.PLAN	0.982 (0.072)	1.374 (0.065)	1.370 (0.139)	0.157 (0.045)	0.615 (0.095)	0.755 (0.056)	0.580 (0.002)	0.633 (0.032)	0.117 (0.016)	0.111 (0.016)

Each reading in this table is the average of 8 subjects in the case of quad socket, and for three subjects in the case of IC socket.
Figures between brackets are one standard deviation.
* The unit is: proportion of the cycle duration.

medio-lateral stability of the amputees in comparison with the normal subjects. The left step length of the normal subjects was found to be significantly longer ($P<0.05$) than the sound step length of the amputees. The prosthetic step length of the amputees had no significant differences from the right (or left) step length of the normal subjects. However, the prosthetic step length of the amputees was noticeably longer ($P=0.07$) than that of the right side of the normal subjects. Duration of the stance phase of the normal subjects was significantly ($P<0.01$) longer than that of the prosthetic side of amputees with normal alignment, and the duration of the sound stance phase of the amputees was significantly longer ($P<0.01$) than stance phase of the normal subjects. This is evidence that the patient spends more time on his sound leg than that on the prosthetic leg as he feels more secure and stable. The prosthetic double support time of the amputees was significantly ($P<0.01$) shorter than the double support time of the normal subjects, but the sound double support displayed no difference from the normal subjects. This confirms the above statement that the amputee relied on his sound side rather than the prosthetic side, because the prosthetic side does not give the patient the comfortable support or the secure stability of a normal limb. The long double support duration of the sound side in comparison to that of the prosthetic side, can also be related to the fact that the sound leg has to stay on the floor for a longer time after the prosthetic heel strike than that of the prosthetic leg after the sound heel strike, in order to ensure the stability of the prosthetic leg before leaving the floor.

6.4.3 The Effect of Alignment Changes on the Temporal-Distance Parameters

6.4.3.1 Effect of Foot Changes

Table 6.6 presents the average temporal-distance parameters obtained for the amputees after changes were made to the alignment of the foot. The individual results of each subject were also calculated corresponding to each alignment change, however, because of the space limitation, only the results of the extreme change (6 degrees foot changes) are presented here in tables 6.7

Table 6.7 Temporal-Distance parameters of 11 AK amputees. (6 Degrees foot dorsiflexion change, from the normal position)

Sub. Code	Velocity [m/s]	Cycle Duration [s]	Stride Length [m]	Step Width [m]	Sound Step Length [m]	Prosth step Length [m]	Prosthetic Stance *	Sound Stance *	Prosthetic D Support *	Sound D Support *
JLAQ	0.986 (0.048)	1.353 (0.064)	1.325 (0.017)	0.088 (0.008)	0.641 (0.017)	0.684 (0.003)	0.582 (0.012)	0.655 (0.008)	0.112 (0.003)	0.140 (0.016)
PLAQ	0.807 (0.019)	1.487 (0.042)	1.213 (0.054)	0.219 (0.020)	0.573 (0.026)	0.640 (0.029)	0.549 (0.010)	0.668 (0.018)	0.106 (0.013)	0.124 (0.018)
ILBQ	1.330 (0.045)	1.193 (0.031)	1.561 (0.018)	0.230 (0.039)	0.738 (0.017)	0.823 (0.017)	0.577 (0.013)	0.621 (0.012)	0.099 (0.003)	0.115 (0.003)
MRCQ[1]	1.202 (0.027)	1.150 (0.014)	1.384 (0.009)	0.130 (0.003)	0.752 (0.005)	0.632 (0.014)	0.547 (0.007)	0.692 (0.028)	0.120 (0.001)	0.137 (0.023)
TRAQ[1]	1.021 (0.002)	1.430 (0.014)	1.440 (0.014)	0.113 (0.028)	0.666 (0.014)	0.774 (0.029)	0.566 (0.006)	0.655 (0.016)	0.097 (0.001)	0.138 (0.021)
LRBQ[1]	0.686 (0.069)	1.700 (0.113)	1.161 (0.051)	0.108 (0.020)	0.496 (0.003)	0.665 (0.048)	0.541 (0.006)	0.657 (0.002)	0.082 (0.005)	0.128 (0.008)
DLAQ	0.834 (0.030)	1.493 (0.023)	1.223 (0.038)	0.161 (0.017)	0.513 (0.035)	0.710 (0.034)	0.533 (0.013)	0.670 (0.009)	0.115 (0.007)	0.101 (0.016)
ELCQ[1]	0.773 (0.004)	1.370 (0.014)	1.044 (0.018)	0.185 (0.070)	0.508 (0.001)	0.535 (0.018)	0.569 (0.036)	0.691 (0.007)	0.137 (0.012)	0.137 (0.032)
Aver.	0.955 (0.223)	1.397 (0.175)	1.294 (0.166)	0.154 (0.053)	0.611 (0.103)	0.683 (0.089)	0.558 (0.018)	0.664 (0.023)	0.109 (0.017)	0.128 (0.014)
JLAI	1.092 (0.024)	1.313 (0.031)	1.444 (0.046)	0.107 (0.006)	0.699 (0.021)	0.744 (0.024)	0.565 (0.017)	0.625 (0.007)	0.095 (0.011)	0.110 (0.011)
PLAI[1]	0.934 (0.013)	1.330 (0.042)	1.237 (0.036)	0.197 (0.011)	0.541 (0.012)	0.696 (0.024)	0.541 (0.027)	0.667 (0.000)	0.111 (0.014)	0.111 (0.014)
ILBI	1.002 (0.019)	1.467 (0.042)	1.492 (0.028)	0.144 (0.069)	0.674 (0.034)	0.818 (0.023)	0.556 (0.013)	0.628 (0.013)	0.094 (0.003)	0.103 (0.014)
Aver.	1.009 (0.079)	1.370 (0.084)	1.391 (0.136)	0.149 (0.045)	0.638 (0.085)	0.753 (0.061)	0.554 (0.012)	0.640 (0.023)	0.100 (0.009)	0.108 (0.004)

Each reading in this table is the average of 3 different runs carried out within one day. Figures between brackets are one standard deviation.

* The unit is: proportion of the cycle duration.

1 The data of this subject were averaged for 2 different runs carried out within one day.

and 6.8, and the results of the other change (3 degrees foot changes) are shown in tables A1 and A2 of appendix A. Each value depicted in these results was averaged for three different runs obtained in one day. The standard deviation is also presented, shown between brackets.

On average, changing the alignment of the foot by 3 and 6 degrees towards dorsiflexion or plantarflexion, resulted in no significant (P>0.1) changes in the recorded temporal-distance parameters. However, trends were noticed in the data. For example, most of the patients walked slightly more slowly when the foot was dorsiflexed than with normal alignment. This was caused by the slight increase in the cycle duration for most patients as the dorsiflexed prostheses became less stable in comparison with those of the normal alignment. It was also found that the patients tend to walk more slowly with a plantarflexed foot than with the normal alignment. This was caused by a short stride length which resulted from the shorter sound step length in the plantarflexed position. As the foot was plantarflexed, the prosthesis became more stable and more power was required to initiate knee flexion just before toe off. This may leave the patient with less power to spend in order to shift his body forwards and in this case the shift will be shorter than that of the normal alignment. Although this effect was noticed in changing the foot alignment 3 or 6 degrees, no noticeable differences were found in the temporal-distance parameters of these two alignment changes. The foot alignment changes had similar effect on the temporal-distance parameters of patients whether they wore quad or IC sockets.

6.4.3.2 Effect of Socket Changes

The average temporal-distance parameters corresponding to the socket alignment changes are presented in table 6.9 for the eleven AK amputees and regarding the quad and IC sockets. As was done in the previous section, the individual results of each subject are presented in tables 6.10 and 6.11 for the extreme cases (6 degrees socket changes), and the other cases (3 degrees socket changes) are shown in tables A3 and A4 of appendix A. The results presented

Table 6.8 Temporal-Distance parameters of 11 AK amputees. (6 Degrees foot plantarflexion change, from the normal position).

Sub. Code	Velocity [m/s]	Cycle Duration [s]	Stride Length [m]	Step Width [m]	Sound Step Length [m]	Prosth step Length [m]	Prosthetic Stance *	Sound Stance *	Prosthetic D Support *	Sound D Support *
JLAQ	1.060 (0.027)	1.253 (0.042)	1.328 (0.009)	0.105 (0.016)	0.615 (0.010)	0.713 (0.016)	0.576 (0.005)	0.639 (0.006)	0.115 (0.005)	0.115 (0.012)
PLAQ[1]	0.858 (0.003)	1.420 (0.028)	1.255 (0.005)	0.228 (0.016)	0.587 (0.026)	0.667 (0.021)	0.556 (0.011)	0.674 (0.003)	0.118 (0.012)	0.125 (0.002)
ILBQ	1.264 (0.017)	1.227 (0.023)	1.536 (0.053)	0.236 (0.031)	0.721 (0.028)	0.815 (0.028)	0.572 (0.001)	0.620 (0.002)	0.096 (0.002)	0.112 (0.002)
MRCQ[1]	1.136 (0.092)	1.240 (0.057)	1.385 (0.011)	0.197 (0.007)	0.680 (0.018)	0.705 (0.007)	0.548 (0.013)	0.666 (0.015)	0.111 (0.005)	0.119 (0.006)
TRAQ[1]	1.008 (0.026)	1.450 (0.014)	1.417 (0.027)	0.113 (0.009)	0.645 (0.015)	0.771 (0.042)	0.565 (0.004)	0.653 (0.026)	0.095 (0.001)	0.136 (0.021)
LRBQ	0.642 (0.057)	1.727 (0.061)	1.079 (0.082)	0.121 (0.047)	0.482 (0.021)	0.598 (0.061)	0.530 (0.023)	0.687 (0.008)	0.080 (0.003)	0.148 (0.026)
DLAQ	0.889 (0.013)	1.367 (0.042)	1.213 (0.032)	0.208 (0.021)	0.484 (0.015)	0.729 (0.022)	0.548 (0.017)	0.702 (0.013)	0.130 (0.004)	0.135 (0.026)
ELCQ	0.786 (0.019)	1.373 (0.061)	1.100 (0.031)	0.228 (0.012)	0.527 (0.023)	0.573 (0.009)	0.560 (0.023)	0.670 (0.017)	0.129 (0.006)	0.115 (0.019)
Aver.	0.921 (0.239)	1.382 (0.163)	1.289 (0.158)	0.179 (0.057)	0.593 (0.089)	0.696 (0.082)	0.557 (0.015)	0.664 (0.026)	0.109 (0.018)	0.126 (0.013)
JLAI	1.056 (0.031)	1.373 (0.042)	1.466 (0.044)	0.118 (0.009)	0.703 (0.035)	0.764 (0.011)	0.579 (0.013)	0.622 (0.004)	0.105 (0.010)	0.110 (0.007)
PLAI[1]	0.913 (0.040)	1.310 (0.014)	1.211 (0.062)	0.206 (0.000)	0.515 (0.036)	0.696 (0.025)	0.579 (0.017)	0.669 (0.018)	0.135 (0.023)	0.128 (0.012)
ILBI[1]	0.976 (0.054)	1.440 (0.085)	1.433 (0.035)	0.148 (0.013)	0.628 (0.016)	0.806 (0.019)	0.583 (0.024)	0.609 (0.013)	0.110 (0.006)	0.096 (0.006)
Aver.	0.982 (0.072)	1.374 (0.065)	1.370 (0.139)	0.157 (0.045)	0.615 (0.095)	0.755 (0.056)	0.580 (0.002)	0.633 (0.032)	0.117 (0.016)	0.111 (0.016)

Each reading in this table is the average for 3 different runs carried out within one day. Figurs between brackets are one standard deviation.
* The unit is: proportion of the cycle duration.
1 Data of this subject were averaged for 2 different runs carried out within one day.

Table 6.9 Average results for the effect of socket alignment changes on the temporal-distance parameters of AK amputees. The normal alignment results are also presented.

The Socket Change	Velocity [m/s]	Cycle Duration [s]	Stride Length [m]	Step Width [m]	Sound Step Length [m]	Prosth step Length [m]	Prosthetic Stance *	Sound Stance *	Prosthetic D Support *	Sound D Support *
Quadrilateral Socket (quad)										
NORM.AL	0.966 (0.226)	1.390 (0.178)	1.306 (0.161)	0.179 (0.048)	0.610 (0.106)	0.697 (0.070)	0.558 (0.016)	0.665 (0.021)	0.108 (0.016)	0.129 (0.012)
3DEG.FLEX	0.959 (0.252)	1.396 (0.191)	1.307 (0.197)	0.182 (0.060)	0.619 (0.104)	0.687 (0.099)	0.570 (0.026)	0.672 (0.041)	0.109 (0.018)	0.148 (0.048)
6DEG.FLEX	0.940 (0.266)	1.384 (0.187)	1.264 (0.243)	0.184 (0.073)	0.622 (0.113)	0.641 (0.142)	0.565 (0.027)	0.676 (0.039)	0.103 (0.015)	0.152 (0.034)
3DEG.EXT	0.938 (0.247)	1.433 (0.221)	1.308 (0.170)	0.167 (0.049)	0.603 (0.120)	0.704 (0.081)	0.562 (0.023)	0.664 (0.031)	0.111 (0.024)	0.130 (0.024)
6DEG.EXT	0.922 (0.249)	1.472 (0.228)	1.311 (0.167)	0.189 (0.054)	0.587 (0.115)	0.724 (0.073)	0.562 (0.017)	0.664 (0.027)	0.115 (0.027)	0.125 (0.011)
Ischial Containment Socket (IC)										
NORM.AL	1.079 (0.085)	1.296 (0.045)	1.400 (0.139)	0.161 (0.052)	0.630 (0.080)	0.770 (0.068)	0.567 (0.006)	0.636 (0.015)	0.110 (0.008)	0.109 (0.005)
3DEG.FLEX	1.018 (0.034)	1.313 (0.070)	1.341 (0.111)	0.162 (0.063)	0.627 (0.072)	0.714 (0.059)	0.579 (0.012)	0.650 (0.023)	0.117 (0.007)	0.127 (0.029)
6DEG.FLEX	0.982 (0.080)	1.331 (0.050)	1.296 (0.151)	0.169 (0.050)	0.637 (0.081)	0.667 (0.071)	0.572 (0.031)	0.652 (0.027)	0.107 (0.012)	0.132 (0.035)
3DEG.EXT	1.007 (0.126)	1.348 (0.114)	1.358 (0.133)	0.182 (0.075)	0.596 (0.068)	0.762 (0.077)	0.561 (0.013)	0.656 (0.026)	0.121 (0.004)	0.110 (0.015)
6DEG.EXT	0.993 (0.132)	1.349 (0.097)	1.332 (0.164)	0.179 (0.064)	0.579 (0.082)	0.753 (0.087)	0.580 (0.009)	0.649 (0.036)	0.132 (0.012)	0.112 (0.017)

Each reading in this table is the average of 8 subjects in the case of quad socket, and for three subjects in the case of IC socket.
Figures between brackets are one standard deviation.
* The unit is: proportion of the cycle duration.

Table 6.10 Temporal-Distance parameters of 11 AK amputees. (6 Degrees socket extension change, from the normal position).

Sub. Code	Velocity [m/s]	Cycle Duration [s]	Stride Length [m]	Step Width [m]	Sound Step Length [m]	Prosth step Length [m]	Prosthetic Stance *	Sound Stance *	Prosthetic D Support *	Sound D Support *
JLAQ	1.065 (0.060)	1.307 (0.061)	1.389 (0.033)	0.095 (0.027)	0.634 (0.017)	0.755 (0.029)	0.567 (0.012)	0.638 (0.012)	0.111 (0.006)	0.111 (0.006)
PLAQ	0.756 (0.036)	1.513 (0.046)	1.165 (0.025)	0.264 (0.032)	0.497 (0.030)	0.668 (0.006)	0.557 (0.009)	0.696 (0.20)	0.135 (0.006)	0.131 (0.013)
ILBQ	1.282 (0.024)	1.233 (0.012)	1.584 (0.034)	0.226 (0.033)	0.721 (0.034)	0.864 (0.019)	0.575 (0.005)	0.628 (0.008)	0.106 (0.010)	0.112 (0.001)
MRCQ[1]	1.242 (0.049)	1.140 (0.057)	1.401 (0.012)	0.151 (0.020)	0.723 (0.061)	0.678 (0.048)	0.552 (0.003)	0.664 (0.020)	0.121 (0.006)	0.113 (0.018)
TRAQ	0.869 (0.028)	1.700 (0.053)	1.432 (0.051)	0.168 (0.026)	0.659 (0.043)	0.773 (0.058)	0.566 (0.007)	0.640 (0.025)	0.081 (0.002)	0.136 (0.017)
LRBQ	0.617 (0.014)	1.753 (0.042)	1.102 (0.043)	0.167 (0.033)	0.464 (0.026)	0.638 (0.020)	0.534 (0.032)	0.662 (0.050)	0.079 (0.002)	0.128 (0.019)
DLAQ	0.855 (0.031)	1.473 (0.050)	1.259 (0.020)	0.207 (0.004)	0.572 (0.011)	0.687 (0.016)	0.554 (0.007)	0.688 (0.034)	0.121 (0.015)	0.134 (0.027)
ELCQ[1]	0.689 (0.006)	1.660 (0.141)	1.156 (0.083)	0.235 (0.025)	0.426 (0.008)	0.730 (0.075)	0.590 (0.008)	0.693 (0.058)	0.162 (0.022)	0.133 (0.045)
Aver.	0.922 (0.249)	1.472 (0.228)	1.311 (0.167)	0.189 (0.054)	0.587 (0.115)	0.724 (0.073)	0.562 (0.017)	0.664 (0.027)	0.115 (0.027)	0.125 (0.011)
JLAI	1.123 (0.018)	1.247 (0.023)	1.414 (0.045)	0.118 (0.050)	0.640 (0.022)	0.773 (0.023)	0.579 (0.001)	0.642 (0.003)	0.126 (0.002)	0.111 (0.002)
PLAI[1]	0.860 (0.014)	1.360 (0.000)	1.144 (0.027)	0.246 (0.048)	0.486 (0.006)	0.657 (0.021)	0.572 (0.010)	0.688 (0.031)	0.145 (0.000)	0.130 (0.020)
ILBI[1]	0.997 (0.043)	1.440 (0.113)	1.439 (0.040)	0.173 (0.014)	0.612 (0.008)	0.828 (0.048)	0.589 (0.007)	0.617 (0.009)	0.124 (0.010)	0.096 (0.007)
Aver.	0.993 (0.132)	1.349 (0.097)	1.332 (0.164)	0.179 (0.064)	0.579 (0.082)	0.753 (0.087)	0.580 (0.009)	0.649 (0.036)	0.132 (0.012)	0.112 (0.017)

Each reading in this table is the average of 3 different runs carried out within one day. Figures between brackets are one standard deviation.
* The unit is: proportion of the cycle duration.
1 Data were averaged for 2 different runs carried out within one day.

Table 6.11 Temporal-Distance parameters of 11 AK amputees. (6 Degrees socket flexion change, from the normal position).

Sub. Code	Velocity [m/s]	Cycle Duration [s]	Stride Length [m]	Step Width [m]	Sound Step Length [m]	Prosth step Length [m]	Prosthetic Stance *	Sound Stance *	Prosthetic D Support *	Sound D Support *
JLAQ	0.993 (0.021)	1.353 (0.012)	1.330 (0.023)	0.063 (0.016)	0.647 (0.003)	0.683 (0.026)	0.568 (0.011)	0.631 (0.007)	0.092 (0.008)	0.121 (0.008)
PLAQ[1]	0.793 (0.019)	1.430 (0.014)	1.152 (0.028)	0.281 (0.025)	0.547 (0.012)	0.605 (0.016)	0.559 (0.004)	0.724 (0.017)	0.110 (0.001)	0.186 (0.012)
ILBQ	1.282 (0.035)	1.260 (0.040)	1.615 (0.007)	0.162 (0.040)	0.759 (0.020)	0.856 (0.014)	0.573 (0.007)	0.620 (0.010)	0.099 (0.006)	0.109 (0.003)
MRCQ[1]	1.211 (0.021)	1.110 (0.014)	1.336 (0.043)	0.205 (0.006)	0.709 (0.029)	0.627 (0.014)	0.540 (0.006)	0.708 (0.009)	0.124 (0.002)	0.142 (0.002)
TRAQ	1.047 (0.038)	1.373 (0.061)	1.451 (0.040)	0.208 (0.019)	0.715 (0.041)	0.736 (0.023)	0.579 (0.003)	0.665 (0.007)	0.101 (0.004)	0.158 (0.007)
LRBQ	0.619 (0.033)	1.720 (0.040)	1.060 (0.079)	0.129 (0.056)	0.477 (0.047)	0.583 (0.033)	0.529 (0.001)	0.686 (0.004)	0.080 (0.002)	0.146 (0.004)
DLAQ[2]	----------	----------	----------	----------	----------	----------	----------	----------	----------	----------
ELCQ[1]	0.632 (0.095)	1.440 (0.057)	0.902 (0.040)	0.242 (0.014)	0.503 (0.065)	0.399 (0.035)	0.610 (0.014)	0.697 (0.089)	0.116 (0.005)	0.204 (0.070)
Aver.	0.940 (0.266)	1.384 (0.187)	1.264 (0.243)	0.184 (0.073)	0.622 (0.113)	0.641 (0.142)	0.565 (0.027)	0.676 (0.039)	0.103 (0.015)	0.152 (0.034)
JLAI	0.930 (0.029)	1.333 (0.031)	1.270 (0.013)	0.121 (0.055)	0.648 (0.009)	0.622 (0.007)	0.606 (0.003)	0.665 (0.017)	0.113 (0.009)	0.172 (0.014)
PLAI[1]	0.941 (0.067)	1.280 (0.085)	1.160 (0.038)	0.221 (0.008)	0.551 (0.002)	0.629 (0.040)	0.546 (0.003)	0.670 (0.011)	0.115 (0.003)	0.116 (0.018)
ILBI[1]	1.074 (0.045)	1.380 (0.000)	1.459 (0.077)	0.166 (0.015)	0.711 (0.015)	0.749 (0.062)	0.564 (0.030)	0.621 (0.010)	0.093 (0.010)	0.107 (0.010)
Aver.	0.982 (0.080)	1.331 (0.050)	1.296 (0.151)	0.169 (0.050)	0.637 (0.081)	0.667 (0.071)	0.572 (0.031)	0.652 (0.027)	0.107 (0.012)	0.132 (0.035)

Each reading in this table is the average of 3 different runs carried out within one day. Figures between brackets are one standard deviation.
* The unit is: proportion of the cycle duration.
1 Data were averaged for 2 different runs carried out within one day.
2 The alignment unit did not allow any change beyond 3 degrees of socket flexion.

Table 6.12 Average results for the effect of knee forward and backward shifts on the temporal-distance parameters of AK amputees. The normal alignment results are also presented.

The Knee Shifts	Velocity [m/s]	Cycle Duration [s]	Stride Length [m]	Step Width [m]	Sound Step Length [m]	Prosth step Length [m]	Prosthetic Stance *	Sound Stance *	Prosthetic D Support *	Sound D Support *
Quadrilateral Socket (quad)										
NORM.AL	0.966 (0.226)	1.390 (0.178)	1.306 (0.161)	0.179 (0.048)	0.610 (0.106)	0.697 (0.070)	0.558 (0.016)	0.665 (0.021)	0.108 (0.016)	0.129 (0.012)
0.8 cm FOR	0.965 (0.214)	1.390 (0.180)	1.311 (0.161)	0.168 (0.054)	0.626 (0.094)	0.684 (0.083)	0.549 (0.029)	0.675 (0.032)	0.184 (0.012)	0.134 (0.014)
1.6 cm FOR	0.970 (0.174)	1.335 (0.112)	1.284 (0.150)	0.169 (0.050)	0.649 (0.059)	0.636 (0.112)	0.559 (0.027)	0.693 (0.046)	0.105 (0.012)	0.162 (0.049)
0.8 cm BACK	0.940 (0.241)	1.438 (0.212)	1.312 (0.167)	0.184 (0.058)	0.590 (0.105)	0.721 (0.081)	0.555 (0.026)	0.670 (0.036)	0.114 (0.022)	0.126 (0.015)
1.6 cm BACK	0.934 (0.233)	1.458 (0.202)	1.324 (0.146)	0.186 (0.058)	0.593 (0.098)	0.731 (0.064)	0.556 (0.028)	0.663 (0.040)	0.116 (0.023)	0.118 (0.011)
Ischial Containment Socket (IC)										
NORM.AL	1.079 (0.085)	1.296 (0.045)	1.400 (0.139)	0.161 (0.052)	0.630 (0.080)	0.770 (0.068)	0.567 (0.006)	0.636 (0.015)	0.110 (0.008)	0.109 (0.005)
0.8 cm FOR	0.949 (0.141)	1.389 (0.094)	1.309 (0.149)	0.162 (0.050)	0.619 (0.059)	0.660 (0.050)	0.574 (0.023)	0.666 (0.035)	0.110 (0.004)	0.144 (0.052)
1.6 cm FOR	0.943 (0.109)	1.374 (0.136)	1.285 (0.100)	0.148 (0.026)	0.640 (0.048)	0.645 (0.089)	0.578 (0.028)	0.689 (0.056)	0.102 (0.012)	0.180 (0.092)
0.8 cm BACK	1.058 (0.079)	1.317 (0.006)	1.404 (0.112)	0.166 (0.059)	0.613 (0.061)	0.791 (0.068)	0.575 (0.016)	0.636 (0.025)	0.121 (0.011)	0.105 (0.000)
1.6 cm BACK	1.054 (0.056)	1.322 (0.043)	1.405 (0.125)	0.178 (0.067)	0.611 (0.062)	0.794 (0.074)	0.569 (0.013)	0.644 (0.023)	0.122 (0.009)	0.106 (0.004)

Each reading in this table is the average of 8 subjects in the case of quad socket, and for three subjects in the case of IC socket.
Figures between brackets are one standard deviation.
* The unit is: proportion of the cycle duration.

are the mean and standard deviation of three measurements obtained in one day. As the socket was flexed by 3 and 6 degrees or extended by 3 and 6 degrees, again, no significant differences (P>0.1) were found between the temporal-distance parameters corresponding to socket alignments changes and those of normal alignment. However, some trends were noticed which generally, were similar to those which resulted from the foot alignment changes. Flexing the socket showed similar effects to those which resulted from dorsiflexing the foot, and extending the socket exhibited similar effects to those which resulted from plantarflexing the foot. No noticeable differences were found between the effect of socket alignment changes on quad and IC wearers.

6.4.3.3 Effect of Knee Shifts

The means and standard deviations of the temporal-distance parameters of different knee positions relative to the hip-ankle line are shown in table 6.12 for AK amputees regarding the quad and IC sockets. Results of each subject are presented in tables 6.13 and 6.14 for the extreme cases (1.6 cm knee shifts) and the other cases (0.8 cm knee shifts) are shown in tables A5 and A6 of appendix A. The different knee positions were found by shifting the KJC forwards and backwards from its original position which corresponded to the normal alignment. By shifting the knee forwards, most subjects tend to decrease their velocity. However, the change in the velocity was not significant (P>0.1) and was not exhibited by all patients. The reduction in the velocities of the amputees was obtained by a slight increase in the cycle duration, or by slight decrease in the stride length (or by both) as the prosthetic knee became less stable than that of the normal alignment. By shifting the KJC backwards relative to the hip-ankle line, the subjects' velocity was also decreased, but not significantly (P>0.1). The decrease in the velocity, was mainly caused by increasing the time of the cycle duration, and it can be related to the fact that by shifting the knee backward more power will be needed to maintain velocity and initiate knee flexion before the prosthetic toe

Table 6.13 Temporal-Distance parameters of 11 AK amputees. (1.6 cm knee forwards shift, from the normal position).

Sub. Code	Velocity [m/s]	Cycle Duration [s]	Stride Length [m]	Step Width [m]	Sound Step Length [m]	Prosth step Length [m]	Prosthetic Stance *	Sound Stance *	Prosthetic D Support *	Sound D Support *
JLAQ	0.960 (0.029)	1.340 (0.040)	1.295 (0.009)	0.084 (0.032)	0.647 (0.015)	0.648 (0.019)	0.578 (0.015)	0.676 (0.013)	0.093 (0.009)	0.176 (0.021)
PLAQ	0.747 (0.041)	1.500 (0.069)	1.106 (0.030)	0.246 (0.030)	0.616 (0.010)	0.490 (0.028)	0.562 (0.010)	0.693 (0.021)	0.105 (0.005)	0.163 (0.026)
ILBQ[1]	1.197 (0.014)	1.300 (0.028)	1.562 (0.025)	0.185 (0.018)	0.739 (0.027)	0.823 (0.002)	0.568 (0.001)	0.614 (0.002)	0.091 (0.002)	0.106 (0.002)
MRCQ	1.185 (0.032)	1.133 (0.012)	1.328 (0.034)	0.144 (0.033)	0.679 (0.024)	0.650 (0.048)	0.526 (0.009)	0.688 (0.015)	0.121 (0.001)	0.110 (0.011)
TRAQ[1]	0.999 (0.043)	1.300 (0.028)	1.298 (0.057)	0.152 (0.016)	0.690 (0.011)	0.608 (0.046)	0.568 (0.020)	0.697 (0.015)	0.106 (0.002)	0.174 (0.007)
LRBQ[2]	----------	----------	----------	----------	----------	----------	----------	----------	----------	----------
DLAQ	0.898 (0.006)	1.393 (0.050)	1.268 (0.017)	0.178 (0.020)	0.562 (0.009)	0.706 (0.019)	0.519 (0.026)	0.718 (0.022)	0.099 (0.004)	0.152 (0.055)
ELCQ	0.803 (0.013)	1.380 (0.035)	1.133 (0.029)	0.197 (0.017)	0.609 (0.014)	0.524 (0.022)	0.591 (0.016)	0.768 (0.053)	0.119 (0.010)	0.253 (0.055)
Aver.	0.970 (0.174)	1.335 (0.112)	1.284 (0.150)	0.169 (0.050)	0.649 (0.059)	0.636 (0.112)	0.559 (0.027)	0.693 (0.046)	0.105 (0.012)	0.162 (0.049)
JLAI[3]	0.820	1.520	1.292	0.119	0.690	0.602	0.610	0.753	0.091	0.286
PLAI[1]	0.982 (0.032)	1.250 (0.014)	1.182 (0.027)	0.170 (0.040)	0.595 (0.001)	0.586 (0.028)	0.567 (0.016)	0.654 (0.041)	0.102 (0.010)	0.134 (0.035)
ILBI	1.027 (0.028)	1.353 (0.012)	1.381 (0.021)	0.154 (0.053)	0.634 (0.017)	0.747 (0.004)	0.558 (0.012)	0.660 (0.007)	0.112 (0.008)	0.121 (0.008)
Aver.	0.943 (0.109)	1.374 (0.136)	1.285 (0.100)	0.148 (0.026)	0.640 (0.048)	0.645 (0.089)	0.578 (0.028)	0.689 (0.056)	0.102 (0.012)	0.180 (0.092)

Each reading in this table is the average of 3 different runs carried out within one day. Figures between brackets are one standard deviation.
* The unit is: proportion of the cycle duration. (1) Data were averaged for 2 different runs carried out within one day.
(2) The prosthesis with this alignment was unstable, therefore the patient was not allowed to walk with it. (3) Only one successful run was obtained.

Table 6.14 Temporal-Distance parameters of 11 AK amputees. (1.6 cm knee backward shift, from the normal position).

Sub. Code	Velocity [m/s]	Cycle Duration [s]	Stride Length [m]	Step Width [m]	Sound Step Length [m]	Prosth step Length [m]	Prosthetic Stance *	Sound Stance *	Prosthetic D Support *	Sound D Support *
JLAQI	1.019 (0.026)	1.360 (0.020)	1.391 (0.028)	0.096 (0.045)	0.651 (0.004)	0.740 (0.025)	0.584 (0.010)	0.618 (0.004)	0.111 (0.007)	0.106 (0.008)
PLAQ	0.674 (0.046)	1.707 (0.101)	1.173 (0.027)	0.263 (0.028)	0.523 (0.018)	0.649 (0.017)	0.561 (0.049)	0.687 (0.019)	0.144 (0.029)	0.116 (0.007)
ILBQ	1.323 (0.039)	1.173 (0.012)	1.553 (0.036)	0.212 (0.035)	0.708 (0.025)	0.846 (0.024)	0.592 (0.015)	0.626 (0.013)	0.117 (0.001)	0.117 (0.001)
MRCQ[1]	1.165 (0.016)	1.240 (0.028)	1.410 (0.011)	0.198 (0.044)	0.651 (0.018)	0.759 (0.029)	0.532 (0.001)	0.667 (0.007)	0.111 (0.002)	0.103 (0.009)
TRAQ[1]	0.948 (0.035)	1.570 (0.042)	1.437 (0.006)	0.101 (0.012)	0.714 (0.029)	0.722 (0.023)	0.578 (0.020)	0.623 (0.008)	0.088 (0.002)	0.126 (0.014)
LRBQ[1]	0.657 (0.041)	1.710 (0.099)	1.134 (0.016)	0.215 (0.002)	0.476 (0.006)	0.658 (0.010)	0.525 (0.027)	0.672 (0.055)	0.081 (0.005)	0.128 (0.024)
DLAQ	0.902 (0.003)	1.387 (0.031)	1.273 (0.009)	0.212 (0.010)	0.503 (0.010)	0.770 (0.018)	0.517 (0.004)	0.735 (0.030)	0.133 (0.006)	0.133 (0.039)
ELCQ	0.787 (0.010)	1.513 (0.012)	1.221 (0.028)	0.193 (0.036)	0.519 (0.016)	0.701 (0.012)	0.557 (0.023)	0.683 (0.025)	0.139 (0.008)	0.113 (0.039)
Aver.	0.934 (0.233)	1.458 (0.202)	1.324 (0.146)	0.186 (0.058)	0.593 (0.098)	0.731 (0.064)	0.556 (0.028)	0.663 (0.040)	0.116 (0.023)	0.118 (0.011)
JLAI[1]	1.061 (0.006)	1.350 (0.014)	1.451 (0.034)	0.108 (0.042)	0.661 (0.025)	0.790 (0.009)	0.584 (0.006)	0.621 (0.017)	0.117 (0.022)	0.102 (0.001)
PLAI[1]	0.994 (0.051)	1.270 (0.071)	1.263 (0.014)	0.241 (0.004)	0.541 (0.012)	0.722 (0.026)	0.558 (0.009)	0.666 (0.007)	0.132 (0.004)	0.109 (0.006)
ILBI	1.106 (0.034)	1.347 (0.050)	1.501 (0.016)	0.184 (0.069)	0.631 (0.014)	0.870 (0.002)	0.566 (0.010)	0.644 (0.022)	0.117 (0.004)	0.107 (0.010)
Aver.	1.054 (0.056)	1.322 (0.043)	1.405 (0.125)	0.178 (0.067)	0.611 (0.062)	0.794 (0.074)	0.569 (0.013)	0.644 (0.023)	0.122 (0.009)	0.106 (0.004)

Each reading in this table is the average for 3 different runs carried out within one day. Figures between brackets are one standard deviation.
* The unit is: proportion of the cycle duration.
1 Data were averaged for 2 different runs carried out within one day.

off.

A noticeable decrease in the prosthetic step length obtained when the KJC was shifted forwards. This change was not very pronounced when the KJC was shifted by 0.8 cm, but it was significant ($P<0.05$) when the shift was 1.6 cm. This was associated with a significant ($P=0.05$) increase in the sound stance phase time as the prosthesis became less stable and the patient tended to rely on his sound side. Shifting the KJC backwards has resulted in a significant ($P<0.05$) decrease of the sound side double support time. This can be explained as follows: as the KJC was shifted backwards the prosthetic knee became more stable, the patient's confidence increased and he loaded the prosthesis with reduced help from the sound side which was free to leave the ground earlier than with normal alignment. The knee shifts had similar effect on the temporal-distance parameters of subjects wearing quad and IC sockets.

6.5 Kinetics and Kinematics of the Normal subjects

Ten normal subjects were tested and their data were analysed. All results will be discussed but due to space limitations, only the results of five subjects are presented in this chapter. The results of the other five subjects are shown in appendix B. The ground reaction forces are presented relative to the ground frame of reference. The ankle and knee moments were calculated relative to the shank frame of reference, while the hip moments were calculated relative to the femur frame of reference (shank and femur frames of reference were discussed in chapter five). The ground reaction force in the medio-lateral direction (FPZ), the medio-lateral (MX) and transverse (MY) moments of the lower joints (ankle, knee and hip) of the left leg were multiplied by -1 in order to present them in a form more convenient for comparison between right and left legs. In order to compare the results of different subjects, the ground reaction forces were normalised to the subject's body weight, and the joint moments were normalised to the product of the subject's body weight and height. Thus, the effect of different body weights and different subject heights on the results will be reduced. Ground reaction forces and joint moments were

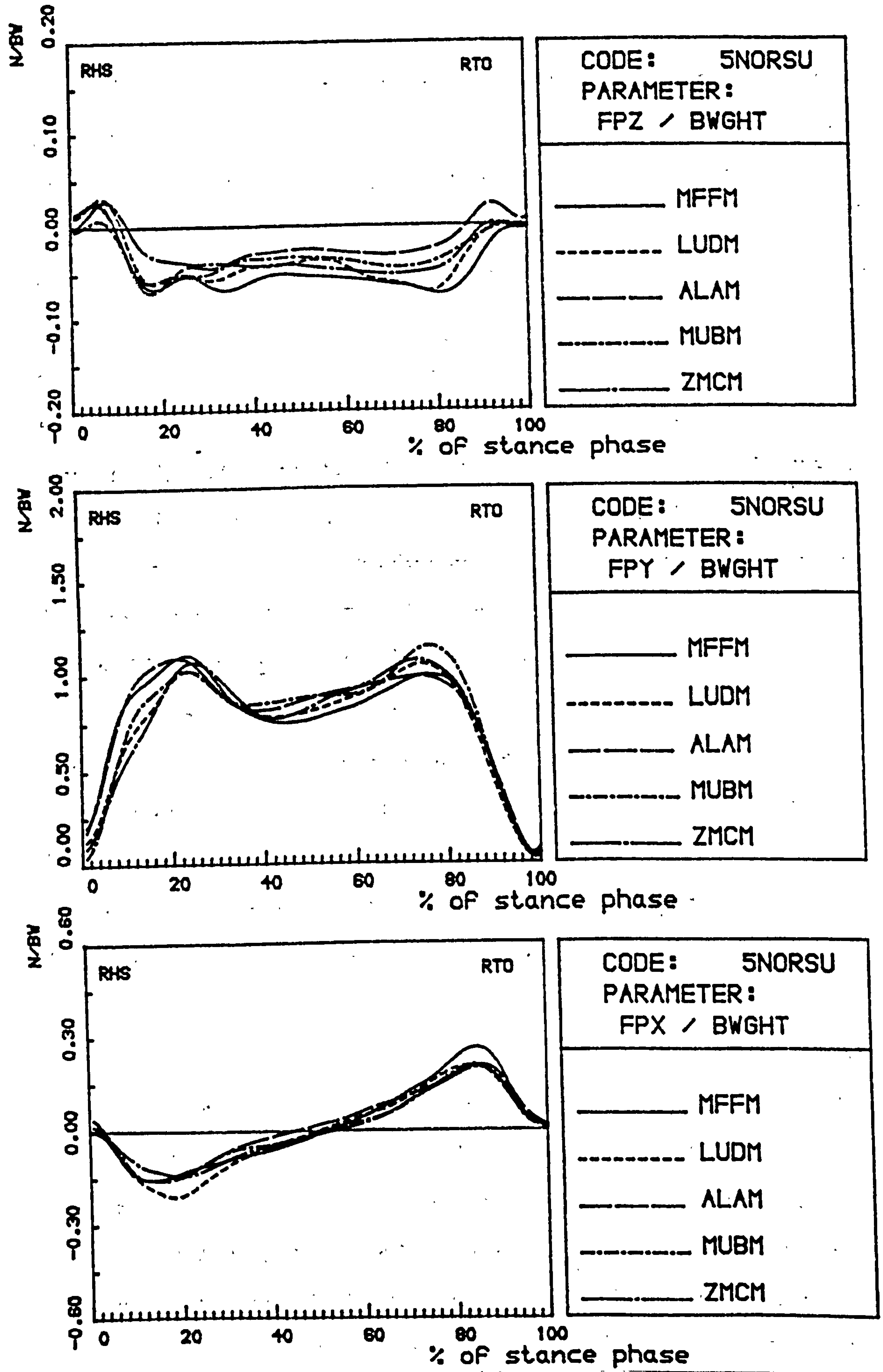


Figure 6.2 Ground reaction forces with time for five normal subjects. (Right leg).

also normalised to 100% of stance phase and 100% of the gait cycle respectively.

6.5.1 Ground Reaction Forces

The ground reaction forces acting on the foot of the right and left legs of five normal subjects are shown in figures 6.2 and 6.3 respectively. The results of each subject presented in these graphs were averaged for three different runs obtained within one day. The results of this study were comparable to those presented by previous researchers such as Chao et al (1983) and Winter (1984) (see chapter 2). No remarkable differences were found among the different subjects and between the right and left legs. The fore-and-aft force (FPX) showed the typical pattern which can be identified by the braking force and the push off force. It is found that the average magnitude of the braking force is 19% (ranging from 13% to 29%) of body weight, and the average magnitude of the push off force is 21% (ranging from 17% to 27%) of body weight. The magnitude of the braking force of the left leg (21% of body weight) was larger than that of the right leg (16% of the body weight). This is attributed to the fact that the left step length was significantly ($P<0.01$) longer than that of the right leg. Subject SIGM (shown in appendix B) has shown a large fore-and-aft force in his left foot in comparison to that of the other subjects. Against expectation, this subject walked more slowly (see table 6.4) than the others, therefore, this can only be explained as odd behaviour. As expected, subject MFFM who walked faster than the other subjects has shown the largest ground reaction force in the direction of progression.

The vertical force (FPY) of all subjects displayed the typical two peak and trough pattern. The first peak found just after heel strike as the leg was accepting the load, and the average magnitude of this peak was 109% (ranging from 100% to 123%) of the body weight. The second peak occurred at the push off phase and had an average magnitude of 109% (ranging from 100% to 116%) of the body weight.

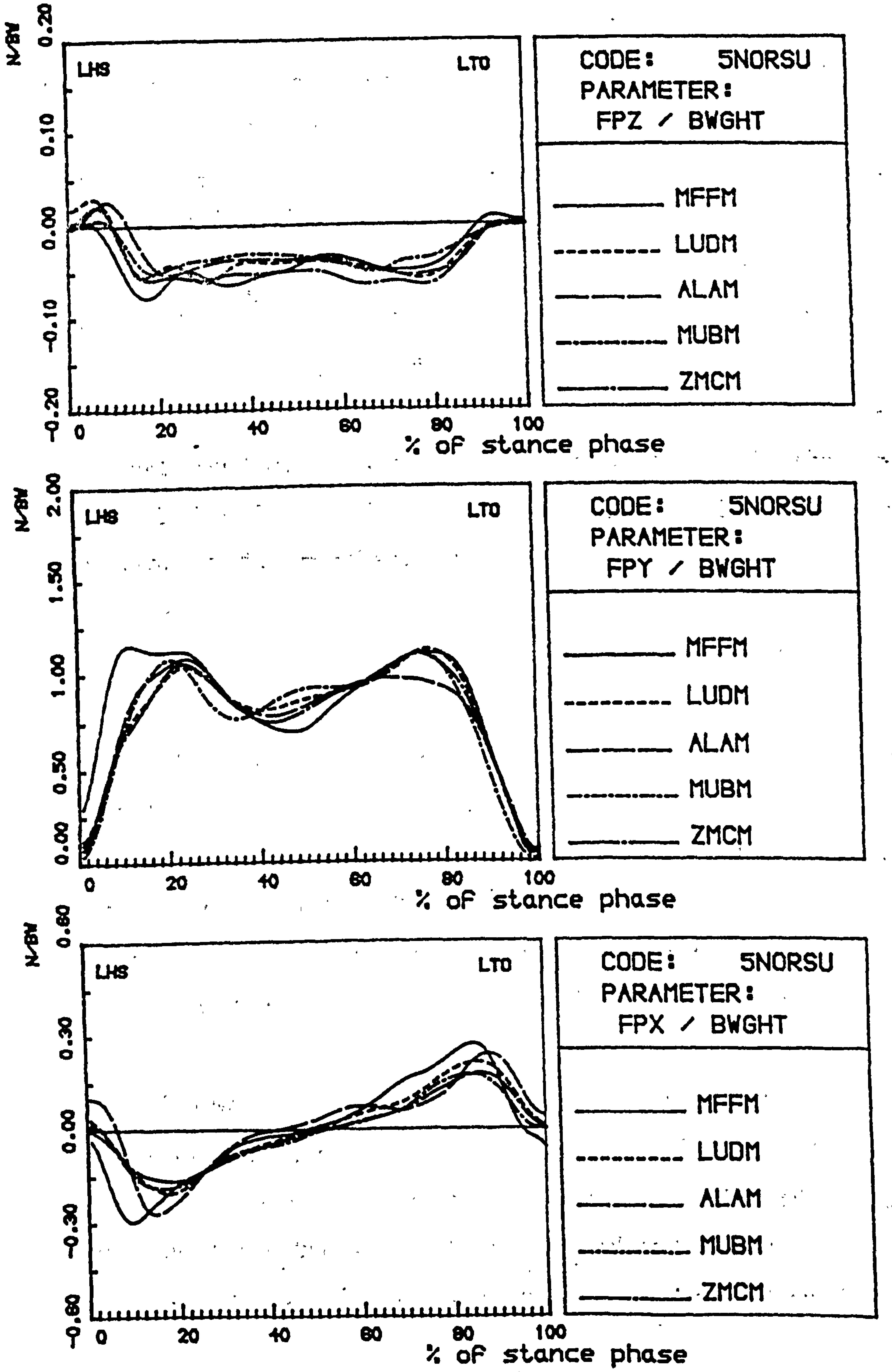


Figure 6.3 Ground reaction forces with time for five normal subjects. (Left leg).

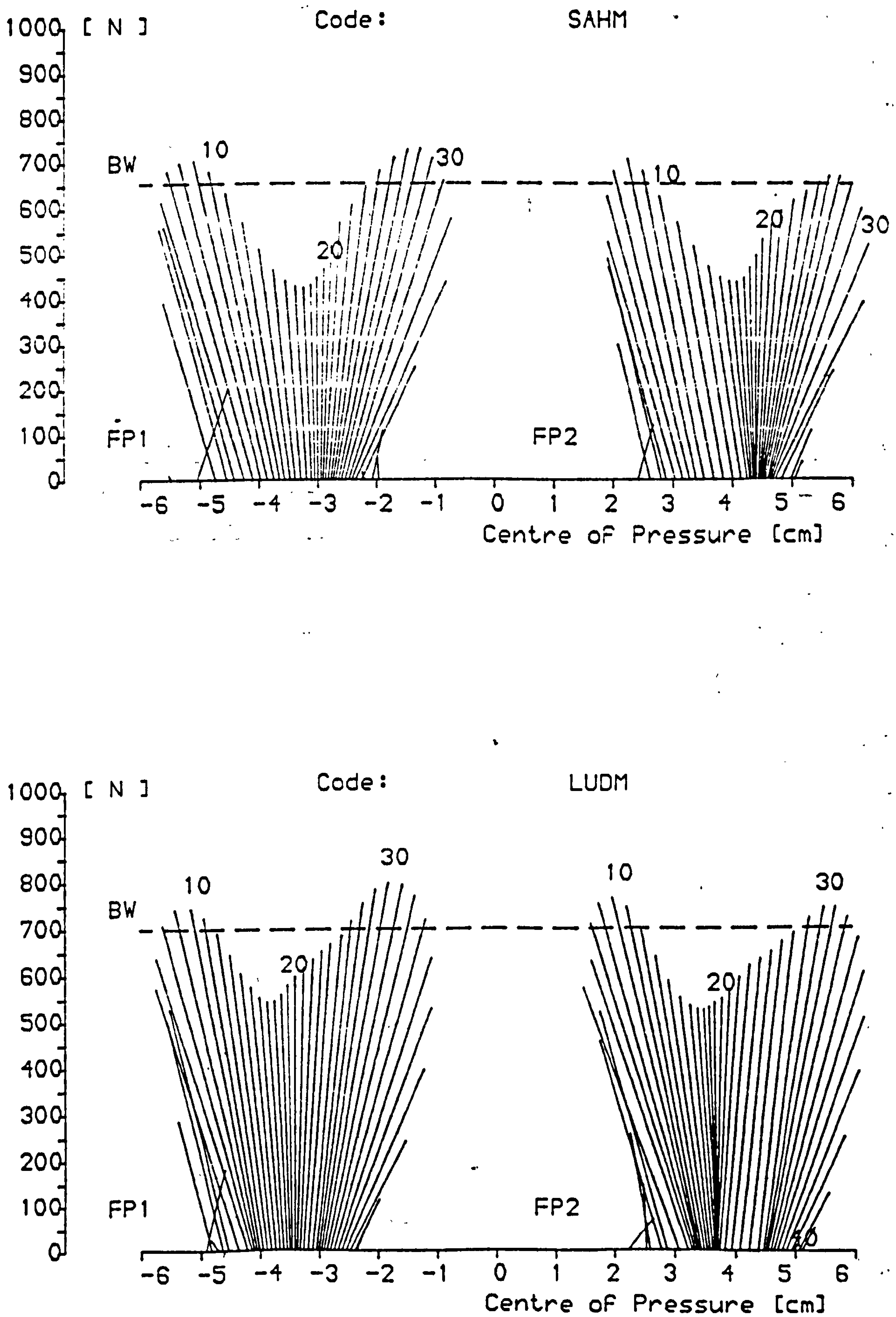


Figure 6.4a "Butterfly" diagrams of two normal subjects. FP1 corresponds to the left leg and FP2 to the right leg forces.

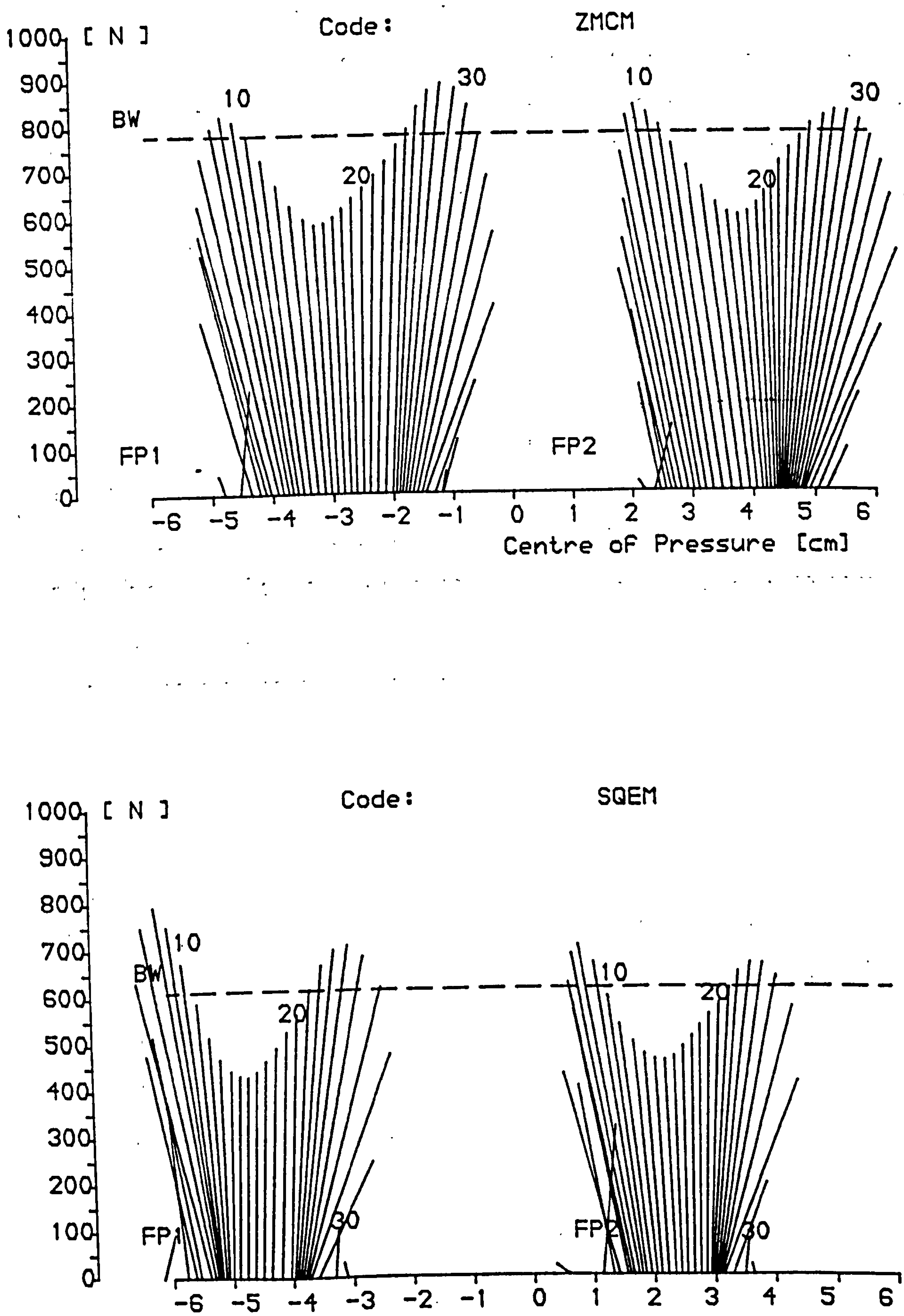


Figure 6.4b "Butterfly" diagrams of two normal subjects (different subjects from those shown in figure 6.4a). FP1 corresponds to the left leg and FP2 to the right leg forces.

The medio-lateral force (FPZ) found to be abducting the foot during the period from heel strike until about 11% of the stance phase, and was adducting the foot for the rest of the stance phase. No noticeable differences between the right and left legs were seen on the medio-lateral force. However, FPZ showed some variations among subjects which is thought to be related to the individual way of controlling medio-lateral stability.

Figures 6.4a and 6.4b show "Butterfly" vector diagrams for four of the normal subjects. In addition to the presentation of FPX and FPY, the vector diagram shows the progression of the centre of pressure with respect to time over the stance phase. The vector diagram of subject SAHM which was typical of most subjects, shows that the centre of pressure travels relatively quickly from heel strike to about 41% of the stance phase. For the remainder of the stance phase the centre of pressure progressed slowly as the foot was in the push off phase. The left leg of subject LUDM showed uniform travel of the centre of pressure over 82% (from 18% until toe off) of the stance phase. This reflects a smooth translation of the body mass over the left leg. Subjects SQEM and SAHM had a deeper trough (peak to trough distance) in their "Butterfly" diagrams than those of subjects ZMCM and LUDM, reflecting a higher speed of walking for subjects SQEM and SAHM (1.32 & 1,33 m/s) in comparison to that of subjects ZMCM and LUDM (1.10 & 1.13 m/3).

In summary, similarity in the patterns and magnitudes of the ground reaction forces was found between the results of this study and those obtained by other researchers. Most of the subjects investigated have shown the expected pattern of forces, however, some differences among subjects and between the right and left legs were found. These differences suggest that symmetry between the right and left legs cannot always be assumed.

6.5.2 Angular Displacements of the Lower Limb Joints

The anterior-posterior (A/P) angular displacements of the right and left leg joints of five normal subjects are shown in figures 6.5 and 6.6 respectively. The results of this study are comparable to those reported by Eberhart et al

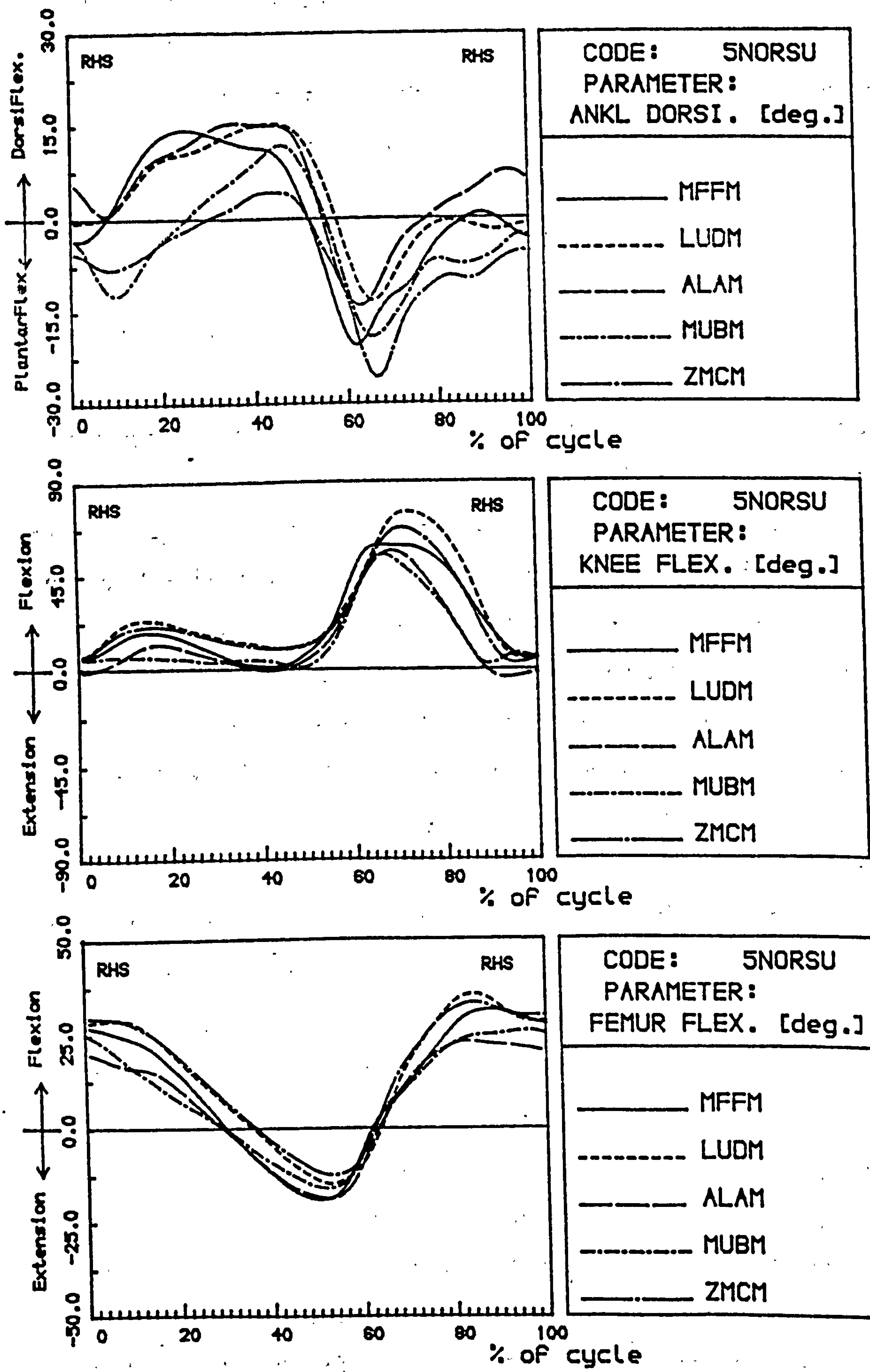


Figure 6.5 A/P angular displacement of the lower limb joints with time for five normal subjects. (Right leg).

(1954), Murray et al (1964), Winter (1984) and Kadaba et al (1989) (see chapter 2). At heel strike, the femur was generally flexed between 25 and 35 degrees. Thereafter, the femur began extending to reach its maximum extension just before toe off when femur flexion started to take place, this flexion action was terminated at about 84% of the gait cycle as the femur moved towards extension so that the swing phase will be terminated. Femur flexion/extension angle (the angle between the femur and the vertical line, as calculated in chapter 5) has shown consistency in the pattern among the various subjects and between the right and left legs. The knee flexion/extension angle (angle between the thigh and the shank, as calculated in chapter 5) of most subjects showed two flexion events. The first event was of approximately 10 degrees of knee flexion and occurred at approximately 14% of the gait cycle after heel strike. This allowed a smooth transfer of the body mass over the foot. The second event was of approximately 67 degrees of knee flexion and appeared during swing phase. The right knee of subject MUBM did not show the first flexion event and the knee was locked in full extension during stance phase (like a prosthetic knee). Variations in the knee flexion/extension angle were noted between subjects; this can be related to the speed variations among subjects. However, there was no obvious relationship between these variations and the speed of the subjects.

All subjects exhibited a roughly similar pattern of ankle dorsiflexion/plantarflexion angle, and the results are in the ranges of those obtained by Murray et al (1964) and Kadaba et al (1989). At heel strike 15 out of 20 feet were found to be plantarflexed by an average of 3.5 degrees (ranging from 0 to 10 degrees) in order to maintain a soft heel strike. The 5 remaining feet were found to be dorsiflexed by an average of 4.1 degrees (ranging from 2 to 8.5 degrees). This is in agreement with Inman et al (1981), they reported that "at heel strike the ankle angle is usually near the standing position, but the ankle may be slightly dorsiflexed or somewhat plantarflexed, depending on the person". However, the overall average of the data for the 20 feet showed that

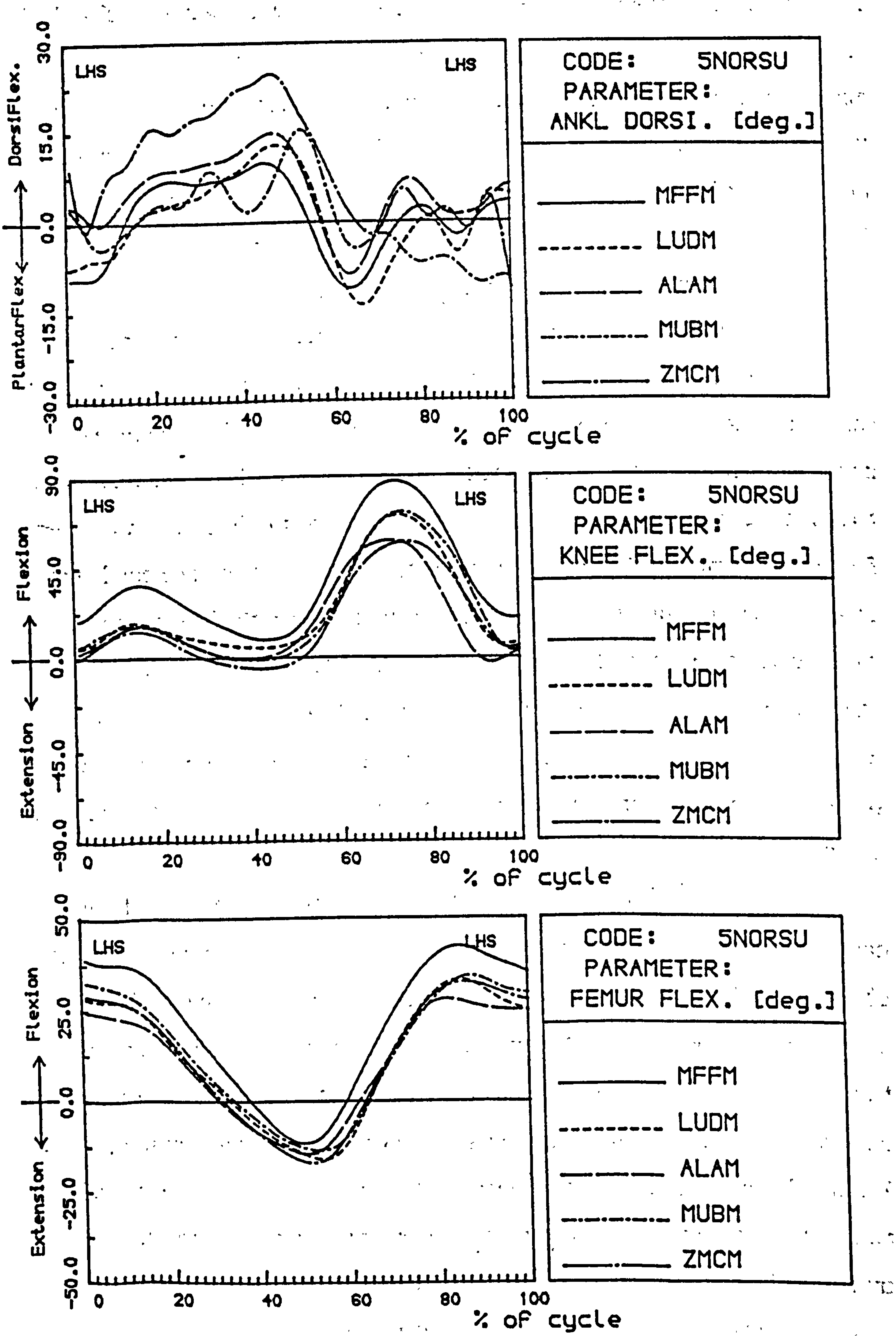


Figure 6.6 A/P angular displacement of the lower limb joints with time for five normal subjects. (Left leg).

at heel strike the foot was plantarflexed by 1.6 degrees (overall average of 1.5 and 3 degrees were obtained by Murray et al (1964) and Kadaba et al (1989) respectively). After heel strike the plantarflexion was slightly increased ensuring a gradual transfer of load from the heel to the front of the foot. When the foot was in full contact with the floor at (6% to 8% of the gait cycle), foot dorsiflexion started to build up as the foot is in contact with the floor and the body is moving forwards. Foot dorsiflexion reached its peak at approximately 45% of the gait cycle (72% of stance phase) when the heel started to rise and the foot plantarflexion action started to take place. The magnitude of this dorsiflexion peak varied among subjects and an average of 14.3 degrees (ranging from 4 to 20 degrees) was found for 18 out of 20 feet (average of 3.3 and 10 degrees obtained by Kadaba et al and Murray et al respectively). The two remaining feet exhibited large dorsiflexion angles (25 degrees on the left foot of subject MUBM and 27 degrees on the left foot of subject NAJM) compared to the other subjects. Foot plantarflexion reached its peak at 62% of the gait cycle which equates to the toe off. An average of 12.3 degrees (range from 0 to 26 degrees) was found for the plantarflexion peak in 19 feet out of 20 (average of 19 and 21.4 degrees were obtained by Kadaba et al and Murray et al respectively), and the remaining foot (right foot of subject NAJM) exhibited a peak of 35 degrees (Inman et al 1981 reported such a large foot plantarflexion angle, it was 27 and 30 degrees for subjects walking with speed of 1 m/s and 1.5 m/s respectively). Although, the average magnitude of the foregoing foot dorsiflexion and plantarflexion peaks differ from those obtained by the other mentioned researchers, the difference between these two peaks (26.6 degrees) is comparable to that obtained by those researchers (22.3 and 31.4 degrees obtained by Kadaba et al and Murray et al respectively). The variations found in the foot angle among subjects can only be related to variations in speed of the subjects and the pattern of walk which was chosen by the subjects.

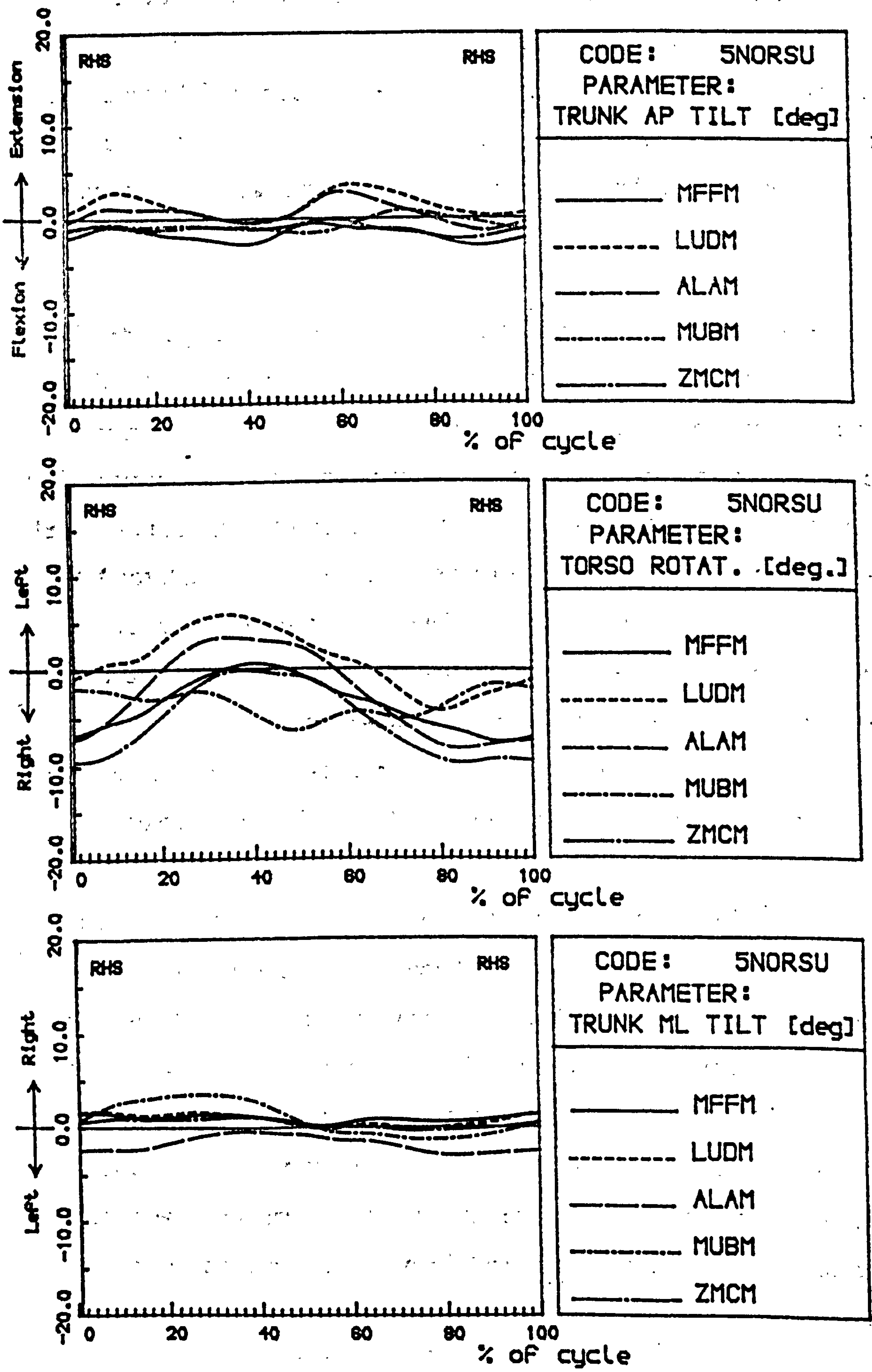


Figure 6.7 A/P and M/L angular displacements of the trunk and transverse rotational displacement of the torso with time for five normal subjects.

6.5.3 Angular Displacements of the Trunk

Figure 6.7 represents the angular displacements of the trunk of five normal subjects. Most subjects displayed similar trunk tilt patterns in the ML plane but the magnitude varied among subjects. During the gait cycle, a complete wave of oscillation was shown by the trunk of most subjects in the ML plane. The amplitude of this oscillation was approximately 2 degrees of tilt to the right side and 2 degrees of tilt to the left side (i.e. trunk tilt was ± 2 degrees). In figure 6.7 the tilt is positive when the trunk is tilted to the right (i.e. the left shoulder is higher than the right shoulder). During the gait cycle, the trunk was always tilted to the supportive side in order to shift the body centre of mass towards the supporting leg. It was found that the trunk was tilted to the right for approximately 50% of the gait cycle, this coincided with the interval from right heel strike to left heel strike whereupon the trunk tilt direction was reversed. As will be discussed later (section 6.6.3), trunk tilt is more pronounced with pathological gait than normal gait. Subject MUBM who walked with a stiff knee during his right stance phase (similar to a prosthetic knee) exhibited the largest trunk tilt among subjects.

In the transverse plane, subjects have shown a complete cycle of oscillation in torso rotation during the gait cycle, and the average difference between any two successive peaks of that cycle was 7.2 degrees (average of 5.2 and 9 degrees obtained by Murray et al 1964 and Inman et al 1981), ranging from 4 to 10 degrees. It is believed that the variations in torso rotation among subjects are related to the variations in the velocity, push off force and stride length of the subjects. However, all subjects exhibited the same pattern of rotation except subject MUBM (the subject with a stiff knee) who showed an inconsistent pattern of rotation of noticeably smaller magnitude in comparison to the other subjects. In general, the maximum torso rotation to the right[2] was achieved just before the right heel strike, and as the subject progressed through

[2] Rotation to the right is a clockwise rotation when one is looking against Y axis of the ground frame of reference.

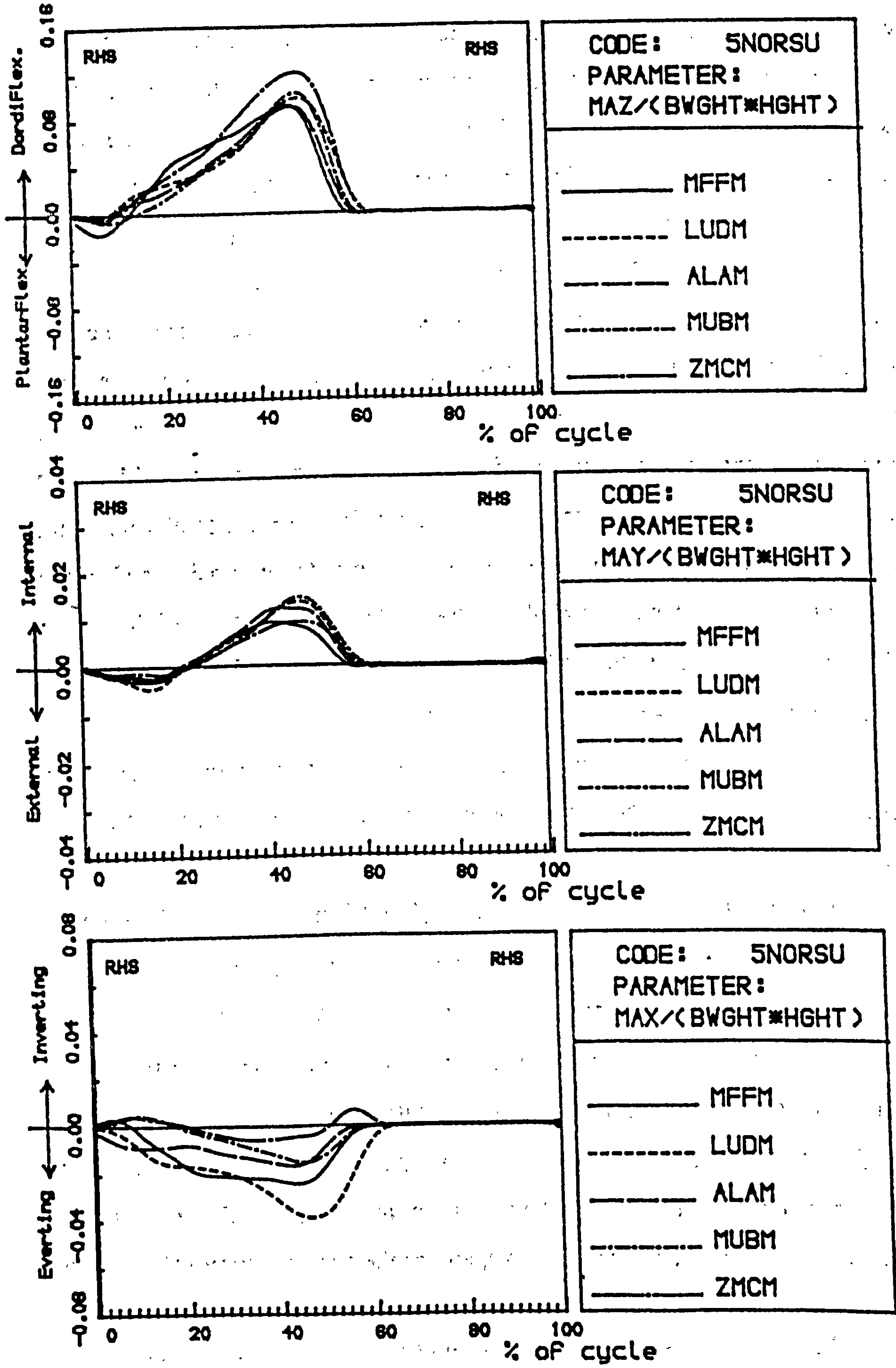


Figure 6.8 Ankle joint moments with time for five normal subjects. (Right leg).

right stance phase, rotation to the left was building up to reach its maximum magnitude at about 40% of the gait cycle which equates with 75% of the left swing phase (or 10% of the gait cycle before the left heel strike).

In the A/P plane, most subjects showed two cycles of trunk oscillations during the gait cycle. The oscillations were always about a base line which was either flexed, extended or neutral dependent on the individual. In general, the trunk A/P tilt was approximately ± 2 degrees about the base line, however, subjects SQEM and MAIM showed very little oscillation about that line. Taking subject LUDM as an example, it was found that after the right heel strike, the trunk continued extending from its base line to reach its first peak of trunk extension (approximately 2 degrees) at about 14% of the gait cycle which corresponded to the left toe off when the trunk began flexing. The trunk flexion reached its maximum magnitude (approximately 2 degrees) at about 42% of the gait cycle which corresponded to 8% of the gait cycle before the left heel strike. The other two peaks of the trunk A/P tilt were extension at the right toe off (approximately 60% of the gait cycle) and flexion at 94% of the gait cycle which is equivalent to 80% of the right swing phase. Thus, there were two peaks of trunk extension and two peaks of trunk flexion during the gait cycle. The two extension peaks occurred at the left and right toe off, and the two flexion peaks occurred at 80% of swing phase of each leg.

In summary, the trunk angular movements patterns were found to be consistent amongst the subjects, and they were very descriptive and meaningful.

6.5.4 The Ankle Joint Moments

The moments of the ankle joint in the A/P, transverse and M/L planes of five normal subjects are presented in figures 6.8 and 6.9 for the right and left legs respectively. The results of each subject were averaged for three runs carried out within one day. The results obtained in this study had similar patterns and comparable magnitudes to those presented by Winter (1984), Paul (1986) and Kadaba et al (1989). Most subjects showed a plantarflexing moment from heel strike until about 12% of the gait cycle. The average

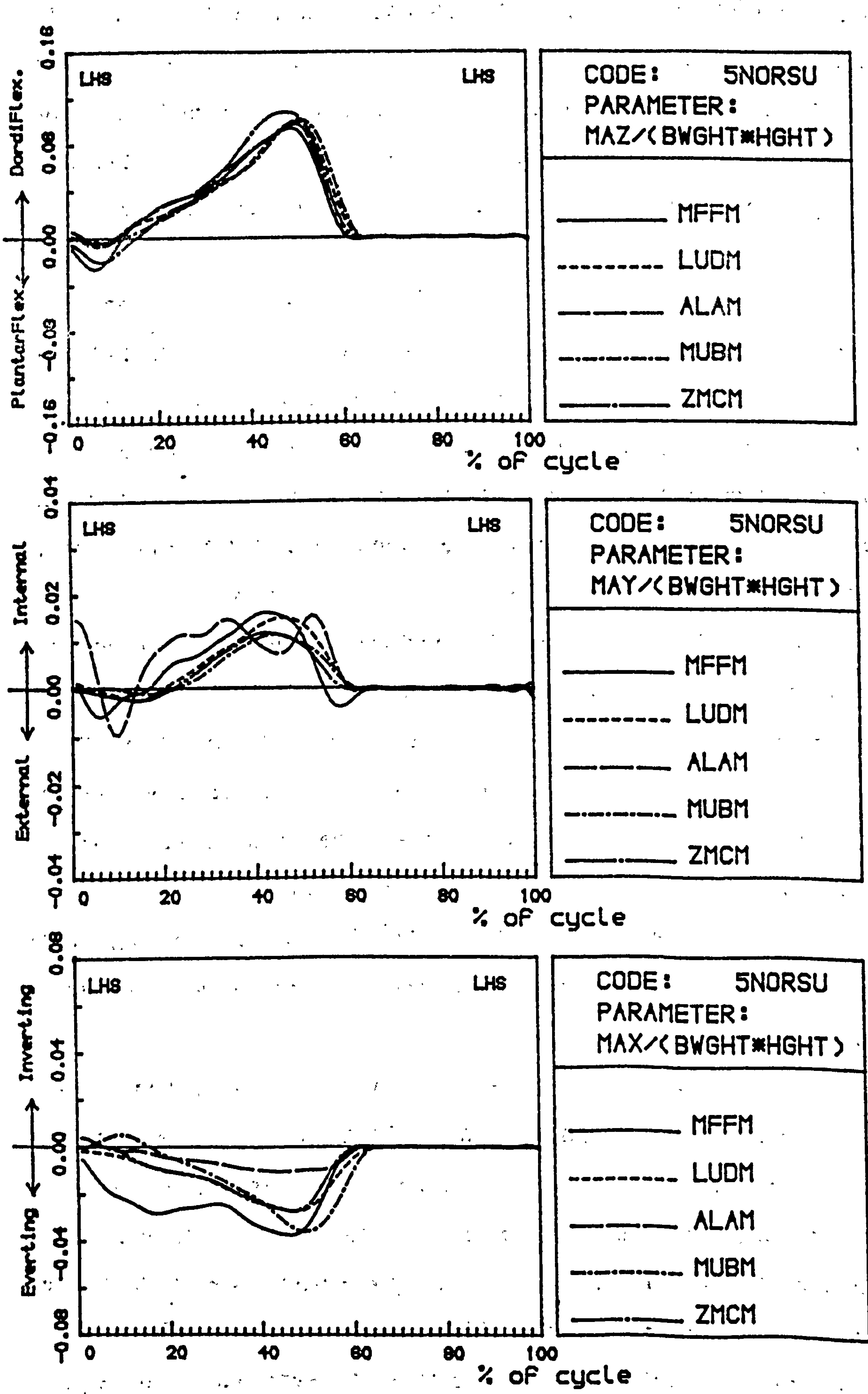


Figure 6.9 Ankle joint moments with time for five normal subjects. (Left leg).

magnitude of this moment was 13 Nm (ranging from 0 to 43 Nm), similar to that presented by Paul (1986), but more pronounced than those of the other mentioned researchers who reported a very small plantarflexing moment. The plantarflexing moment is resisted by the pretibial muscles and that avoids "slap" of the forefoot on the ground and controls the transformation of the body mass over the supporting leg. The dorsiflexing moment dominated the rest of the stance phase and the average of its maximum value is 113 Nm (113 Nm, 100 Nm, and 90 Nm obtained by Winter (1984), Paul (1986) and Kadaba et al (1989) respectively) ranging from 50 to 142 Nm, excluding the left ankle of subject MAIM which was subjected to a dorsiflexing moment of 194 Nm). This dorsiflexing moment occurred at about 50% of the gait cycle corresponding to the foot push off. No noticeable differences among subjects or between the right and left legs were found in the A/P moment of the normal subjects.

The pattern of moment which was obtained in this study for the ankle in the transverse plane (MAY), was comparable to that obtained by Kadaba et al (1989). The ankle joint exhibited a negative moment of 4.5 Nm (3.2 Nm in Kadaba et al) at heel strike which tended to rotate the foot externally and lasted until 22% of the gait cycle. During the rest of stance phase, the ankle was subjected to a positive moment of 13 Nm (about 11 Nm in Kadaba et al) ranged from 8 to 19 Nm. This moment tended to rotate the foot internally, so as rotational stability was maintained in the transverse plane as the body was shifted forwards.

In the coronal plane, all subjects exhibited an eversion moment (MAX), suggesting that the AJC was always medial to the centre of pressure during stance phase. The magnitude of this moment was on average 30 Nm (33 Nm in Kadaba et al, and 10 Nm in Paul), ranging from 7 to 55 Nm. The magnitude of this moment is influenced by the magnitude of the ground reaction force in the medio-lateral (FPZ) and vertical (FPY) directions, and by the medio-lateral offset between the centre of pressure and the ankle joint

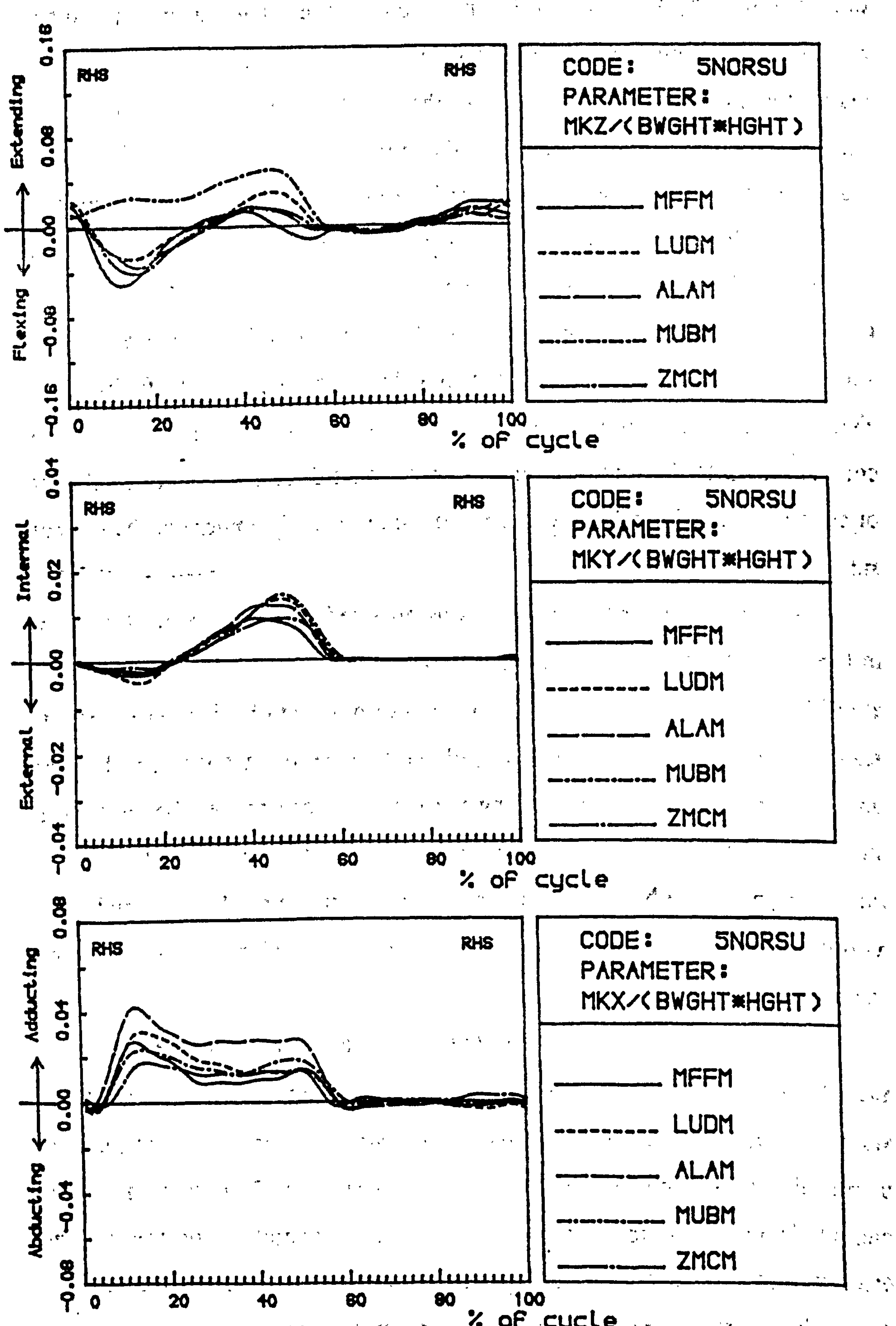


Figure 6.10 Knee joint moments with time for five normal subjects. (Right leg).

centre. This offset is usually small (0 to 4 cm) but the large difference between FPZ and FPY has made it an effective factor in generating the medio-lateral moment, and its variation among subjects has caused some variations in the MAX among the subjects.

6.5.5 The Knee Joint Moments

Figures 6.10 and 6.11 show the knee joint moments of the right and left legs respectively for five normal subjects. The moment patterns of the results obtained in this work are comparable to those presented by Eberhart et al (1954), Winter (1984), Paul (1986) and Kadaba et al (1989), and the magnitude of the moments of this work are in the range obtained by the other listed researchers. In the sagittal plane (MKZ), all subjects showed the typical lock/unlock/lock/unlock moment pattern during stance phase, except for the right leg of subject MUBM who walked with a locked knee during the stance phase. The first positive peak kept the knee in extension to maintain stability from heel strike until about 7% (6.4% obtained by Inman et al 1981) of the gait cycle. The average magnitude of this peak found to be 20.9 Nm (ranging from 6.6 to 38 Nm), and this is in the range obtained by the above researchers (average of 15.5 Nm, 33.3Nm and 6 Nm obtained by Winter, Paul and Kadaba et al respectively). The second positive peak started at about mid-stance (30% of the gait cycle) and lasted until about 54% of the gait cycle when the knee starts flexing in preparation for toe off. The average magnitude of this peak was found to be 22 Nm (20, 17 and 16 Nm obtained by Winter, Paul and Kadaba et al respectively) and ranged from 4 Nm to 68 Nm, but this is excluding the left knee of subject SIGM which exhibited a moment of 90 Nm. Between these two periods of extension, the knee was subjected to a flexing moment as it was flexed to absorb energy and the vector of the ground reaction force passed behind the knee joint centre. The average magnitude of this flexing moment was found to be 47 Nm (38.8, 33 and 44 Nm obtained by Paul, Winter and Kadaba et al respectively), and ranged from 19.4 Nm to 74 Nm. The average value of the flexing moment does not include the moments

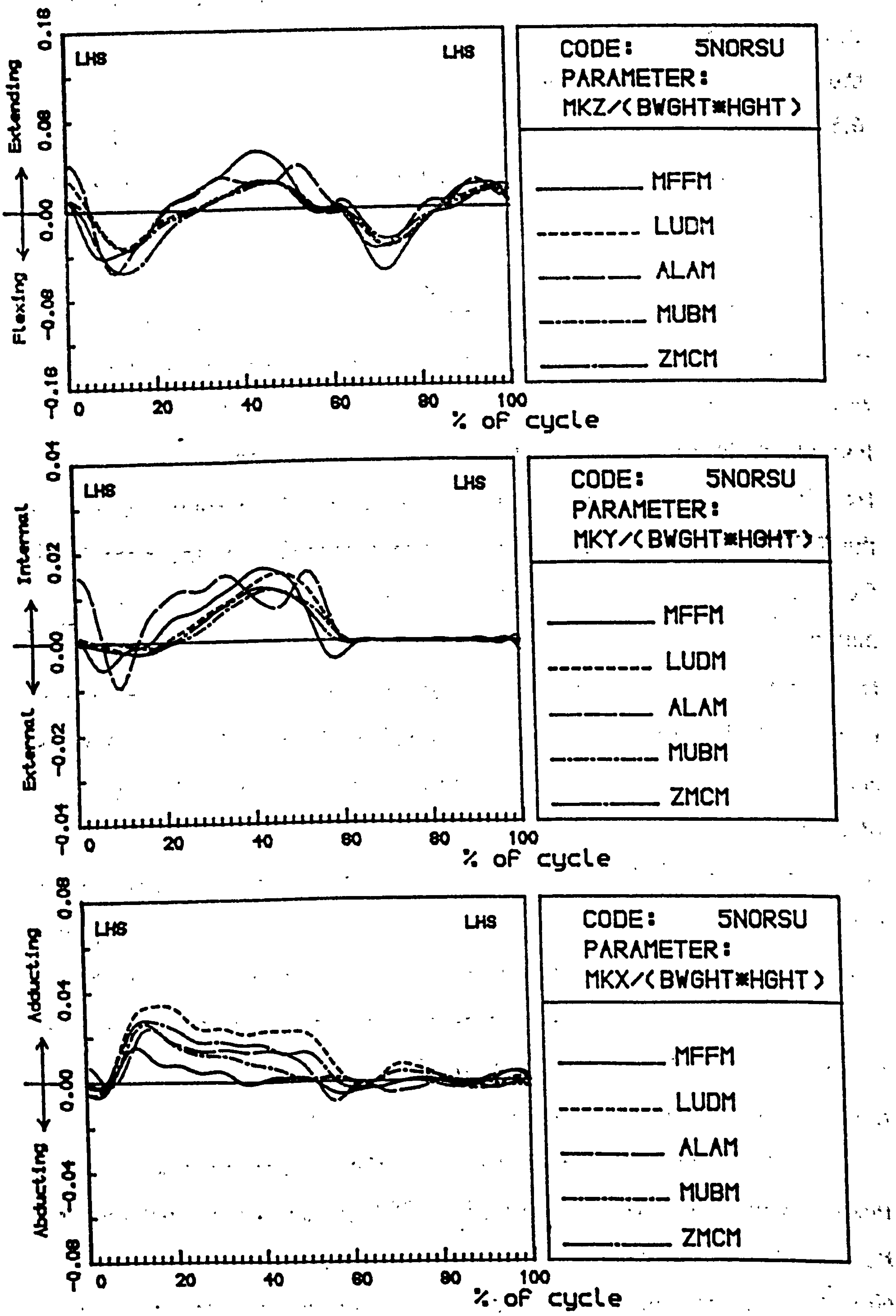


Figure 6.11 Knee joint moments with time for five normal subjects. (Left leg).

of the left knee of subjects SAHM and SIGM who exhibited large moments of 90 and 116 Nm respectively. The A/P knee moments of all subjects were also comparable to that obtained by Winter (1984), Paul (1986) and Kadaba et al (1989) during the swing phase. The A/P knee moment of the right leg was found to be smaller than that of the left leg during swing phase. This can be referred to the fact that the left leg moved faster than the right leg during swing phase and can be explained as follows:

It was found (section 6.4.1 and table 6.4) that the subjects walked with a left step length longer than the right step length. As the two legs had approximately the same time duration of swing phase and the left leg had to travel a longer distance during that time, the velocity of the left leg will be faster than that of the right leg during swing phase. Since the velocity of the left leg was larger than that of the right leg, the inertia moment generated at the left knee will also be larger than that generated at the right knee. Generally, the A/P knee moment has a flexing peak corresponding to the maximum knee flexion angle during the swing phase, and peak extending moment which corresponded to the knee extension angle when the foot is in its furthest point forward during the swing phase. The magnitude of these two peaks is influenced by the velocity of the subject (Winter 1984) and by the individual pattern of walk adopted by the subject.

In the transverse plane, the knee joint exhibited a similar moment (MKY) in pattern and magnitude to that exhibited by the ankle joint (MAY). This was expected as the knee and ankle joints belong to the same body segment (the shank) and their moments were expressed in its frame of reference. Thus, and as the shank should be subjected to a uniform torque, MKY and MAY would be the same.

In the coronal plane, all subjects exhibited a similar pattern of knee moment which tends to adduct the shank in order to maintain the medio-lateral stability. The magnitude of this knee adducting moment varied among subjects from 8 Nm and 50 Nm (average of 29 Nm). This is in agreement with that of

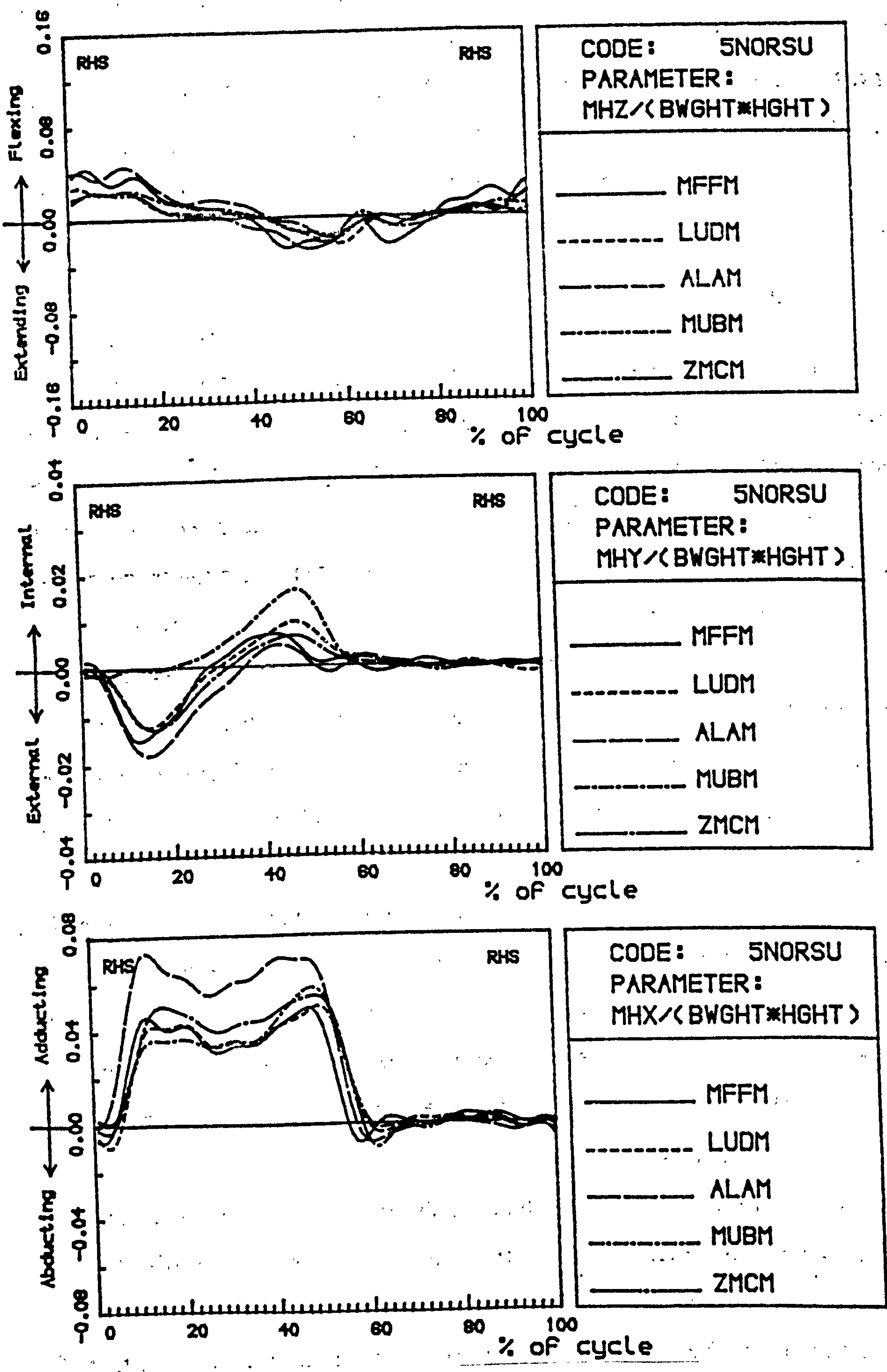


Figure 6.12 Hip joint moments with time for five normal subjects. (Right leg).

Paul (1986) and Kadaba et al (1989) who obtained an average value for this moment of 30 Nm and 25 Nm respectively. The variations of this moment among subjects, can be referred to the variations in the offset between the KJC and the vector of the ground reaction force, in the medio-lateral plane. This offset is affected by the medio-lateral force (FPZ) which influences the inclination of the vector of the ground reaction force in the medio-lateral direction.

6.5.6 The Hip Joint Moments

The hip joint moments of five normal subjects are presented in figures 6.12 and 6.13 for the right and left legs respectively. The hip joint moments obtained in this study were in agreement with those presented by Eberhart et al (1954), Winter (1984), Paul (1986) and Kadaba (1989) which are presented in chapter two. In the A/P plane, most subjects maintained a flexing moment at the hip joint for the duration from heel strike to about mid-stance as the force vector passed ahead of the HJC. Thereafter, an extending moment started to build up as the force vector passed behind the HJC and lasted until the instant of toe off. In most subjects, the peak flexing moment during stance phase occurred at approximately 12% of the gait cycle (20% of the stance phase) as the vertical ground reaction force (FPY) approached its first peak. The maximum value of this flexing moment was in average 41 Nm (compared with the average of: Winter 43 Nm, Paul 58 Nm and Kadaba et al 33 Nm), and varied among subjects from 23 Nm to 65 Nm (excluding subject MFFM and SIGM). These variations among subjects can be mainly related to the variations among the velocities of the subjects. Subject MFFM showed a large flexing moment at the left hip (93 Nm), although, this subject was the fastest among all subjects, this high flexing moment can be referred to the large FPY which was exhibited by this subject (see fig. 6.3). The hip extending moment during stance phase reached its maximum value at approximately 50% to 54% of the gait cycle (80 % to 87% of the stance phase), which coincides with the second peak of the FPY. The maximum value of this extending moment was

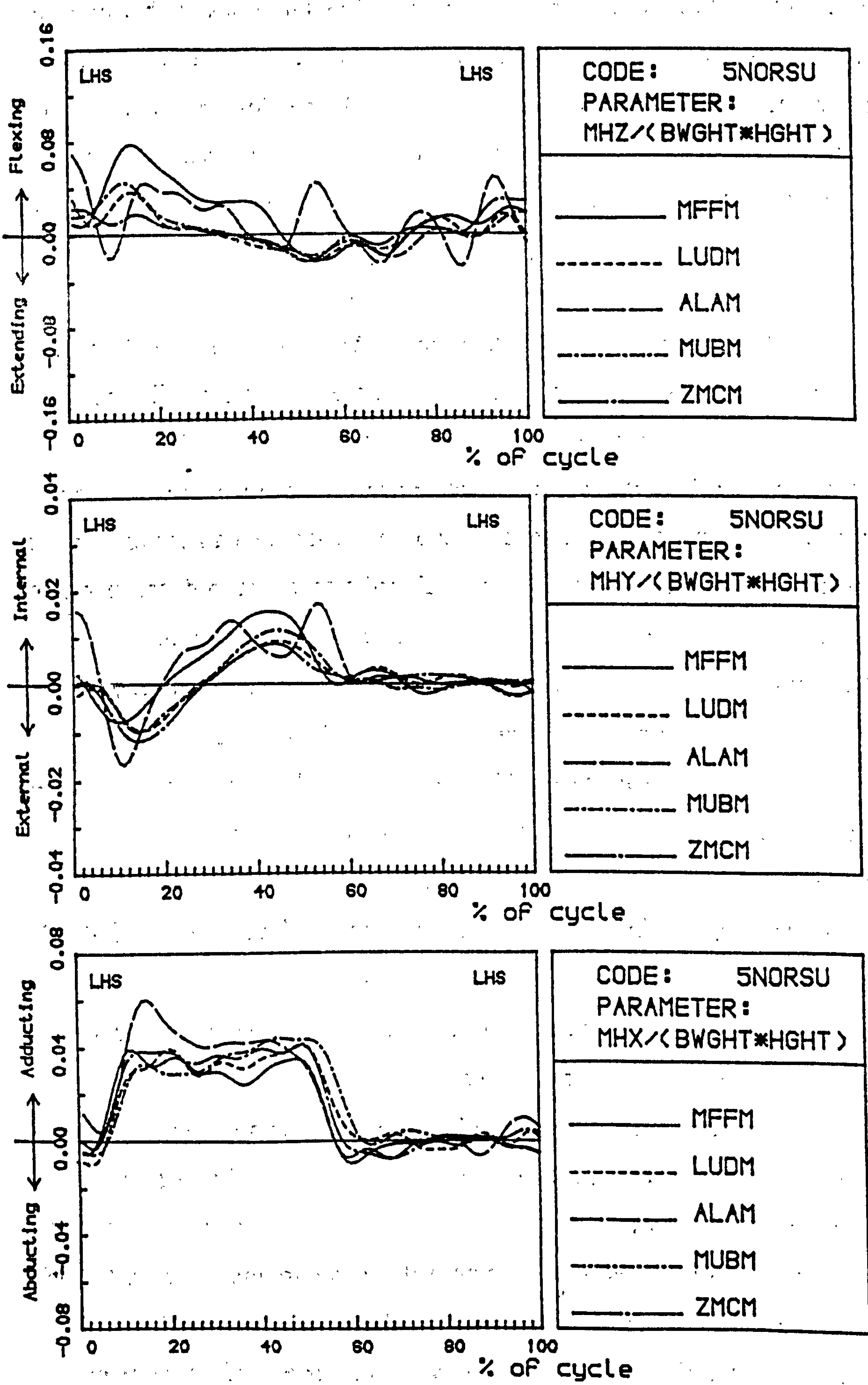


Figure 6.13 Hip joint moments with time for five normal subjects. (Left leg).

in average 35 Nm (compared with the average of: Winter 28 Nm, Paul 83 Nm and Kadaba at al 92 Nm), and varied among the subjects from 18 Nm to 70 Nm (excluding subject SIGM) depending on the velocity of the subjects . The maximum moment of this study was 15% and 24% smaller than that obtained by Paul and Kadaba et al respectively. This can be related to the fact that the subjects of this study walked relatively slowly when compared to the other normal subjects (see section 6.4.1). The minimum value of this extending moment (18 Nm) was obtained for the right hip of only two subjects (SIGM and SAHM), one of them SIGM was found to have the lowest velocity. The odd A/P hip moment pattern exhibited by the left leg of subject SIGM can only be related to the fact that the left leg of this subject is 1.5 cm shorter than his right leg. As a result, the duration of the left flexing moment after heel strike was very small (5% of the gait cycle) because the subject can easily shift his body mass over the short leg. This subject also developed the habit of producing a large fore-and-aft force with his left foot (FPX in Appendix B) particularly at push off which caused the large flexing moment at the push off time. During swing phase, most subjects exhibited a comparable moment in pattern and magnitude to those obtained by the above researchers. The extending moment exhibited during the first half of swing phase and its maximum value (approximately 18 Nm) occurred when the thigh was at its furthest point behind the HJC. A flexing moment dominated the second half of the swing phase, and its maximum value (approximately 21 Nm) occurred when the thigh was at its furthest point ahead of the HJC. During swing phase the hip moment was more pronounced at the left hip than that of the right hip. This can be referred to the fact that the left step length was longer than that of the right leg, and can be explained as shown in the previous section for the MKZ during swing phase.

In the transverse plane, a moment tending to externally rotate the thigh was obtained for the duration from heel strike to mid-stance. Thereafter, a moment tending to rotate the thigh internally was maintained for the rest of stance phase. The pattern of this moment is influenced by the pattern of the trunk rotation in the transverse plane. The average magnitude of the external rotating moment is 16 Nm (10 Nm and 5 Nm obtained by Paul (1986) and Kadaba et al (1989) respectively), and varied among subjects from 11 to 22 Nm. The average magnitude of the internal rotating moment is 9 Nm (13.6 Nm and 7 Nm obtained by Paul and Kadaba et al respectively), and ranged from 4.4 to 19 Nm.

In the coronal plane, all subjects have exhibited an adduction moment at the hip joint during the stance phase, and no differences were found between the moment of the right and left legs. The moment obtained in this study was comparable in pattern and magnitude (average of 50 Nm) to those obtained by other researchers (average of 60 Nm, 53 Nm and 51 Nm obtained by Andriacchi and Strickland (1985), Paul (1986) and Kadaba et al (1989) respectively), and varied among subjects from 35 to 71 Nm. The above obtained pattern of hip moment in the coronal plane, maintained the upright position of the trunk in the M/L direction throughout the stance phase. The variations in the M/L hip moment among subjects are related to the variations in the body medio-lateral movement of the subjects. This can be clearly seen in subject ALAM who had the largest moment. Subject ALAM had a different M/L trunk movement from the other subjects, his trunk moved laterally to the left side during right stance (see fig. 6.7). This may have forced the HJC to move laterally to the right in order to bring the body centre of gravity over the foot. Thus, the medio-lateral offset between the HJC and the centre of pressure increased and generated the high adduction moment.

In summary, the results obtained in this study for the normal subjects were comparable to those presented by other researchers. Results from various subjects were consistent but some individual variations were noted. Speed and the walking style can affect the kinetics and kinematics of the body and its lower limb joints.

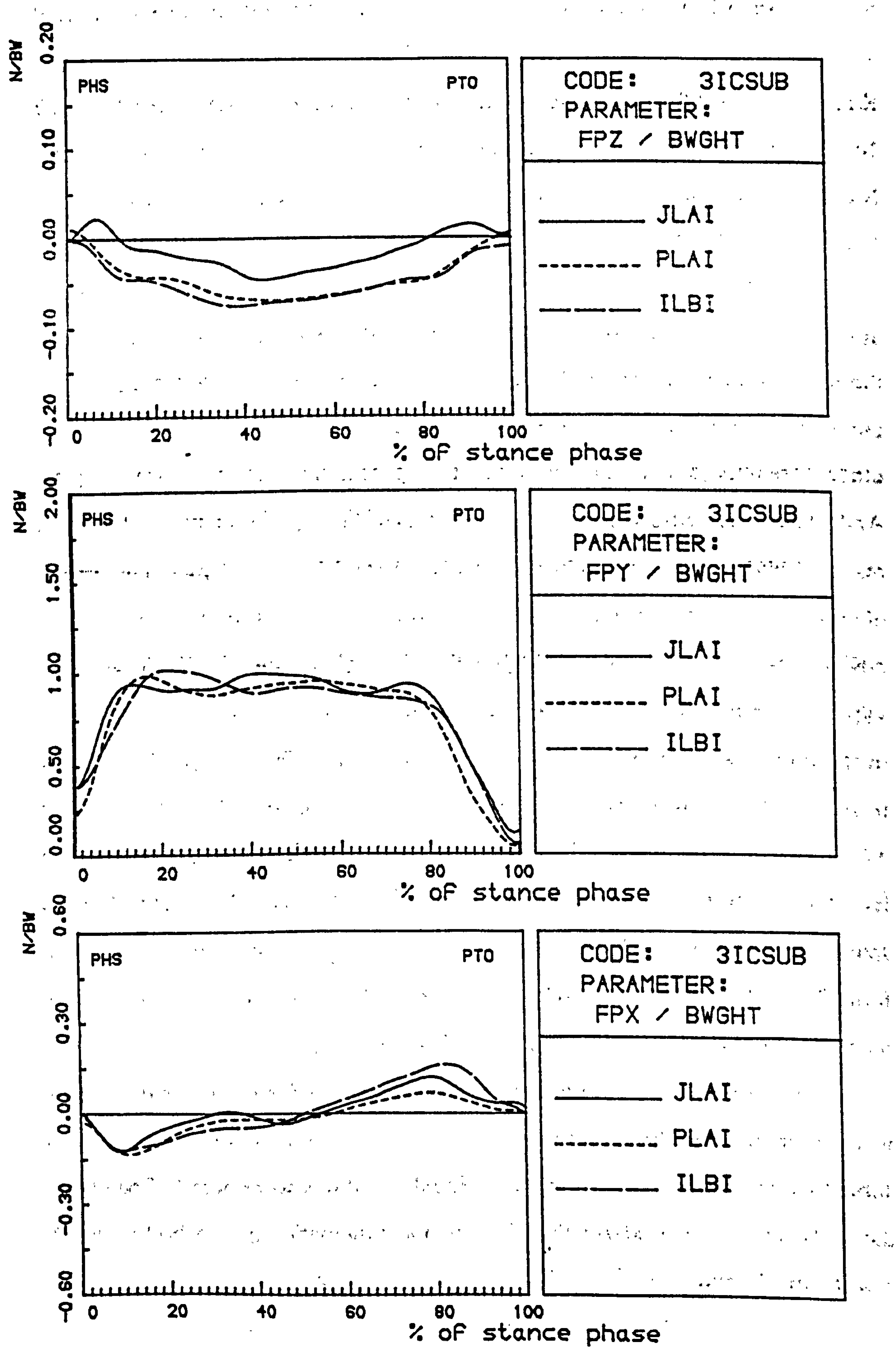


Figure 6.14 Ground reaction forces on the prosthetic leg with time for three AK amputees fitted with IC sockets. (Normal alignment).

6.6 Kinetics and Kinematics of the Amputees with Normal Alignment

Two series of tests will be discussed according to the type of socket which was used to fit the amputees. The first series was undertaken on eight AK amputee subjects who were fitted with quadrilateral (quad) sockets. The second series was conducted on three AK amputees who were fitted with ischial containment (IC) sockets. The subjects' particulars are presented in section 6.2 of this chapter. All subjects were supplied with SACH feet and uniaxial (Otto Bock) constant friction knee mechanisms with extension assist, except subject DLAQ who was supplied with a uniaxial foot and uniaxial safety knee mechanism (see section 4.1). Because so many parameters were considered, innumerable graphs were produced for the eleven patients, therefore, only results of the sound and prosthetic sides for five subjects wearing quadrilateral sockets, and the results of the prosthetic side of the three subjects who wore IC sockets are presented in this chapter. The results of the sound and prosthetic sides of the remaining three patients who wore quad sockets and the sound side of the patients wearing IC sockets are presented in appendix C. All results were averaged for three runs obtained during one day. The ground reaction forces were normalised to the body weight and all the joint moments were normalised to the product of body weight and subject height. The medio-lateral ground reaction force (FPZ), and the joint moments in the transverse and coronal planes of the left leg were multiplied by -1 in order to make the comparison between right and left legs more convenient. The above information regarding data averaging, normalising and left/right leg comparison is applicable to all results which will be presented in this chapter.

6.6.1 The Ground Reaction Forces

Figure 6.14 shows the ground reaction forces applied to the prosthetic side for the three subjects wearing IC sockets. The ground reaction forces of the prosthetic and sound sides of five subjects wearing quad sockets are presented in figures 6.15 and 6.16 respectively. Results from the sound side of the subjects wearing IC sockets, and the left and right sides of the three

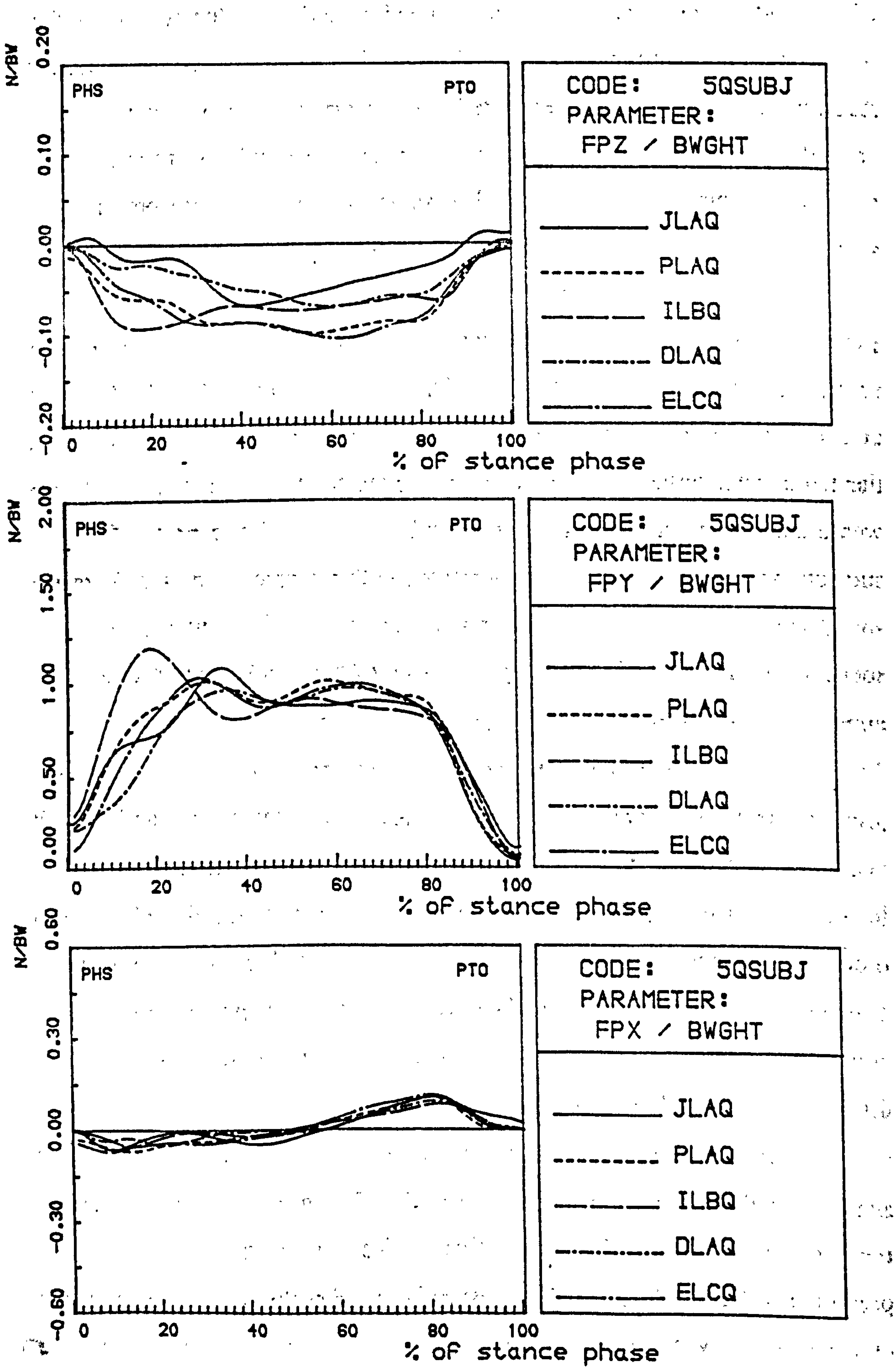


Figure 6.15 Ground reaction forces on the prosthetic leg with time for five AK amputees fitted with quadrilateral sockets. (Normal alignment).

remaining subjects wearing quad sockets are shown in appendix C. The results of this study were comparable to those obtained by Cunningham (1950, shown in fig. 3.42) and Akidele (1987, shown in fig. 3.48). In general, the ground reaction forces of all amputees were larger at the sound side than that at the prosthetic side. Medio-lateral force (FPZ) of the sound side (average of 6% of body weight) was slightly larger than that of the prosthetic side (average of 4% of body weight), and also larger than that of normal subjects (average of 5% of body weight). No noticeable differences were found in the pattern of FPZ between the sound side of the amputees and the normal subjects or between the prosthetic side of subjects fitted with quad and IC sockets. The FPZ pattern of the prosthetic side was slightly different from that of the normal subjects. For the normal subjects, the FPZ was abducting the foot during the first 11% of the stance phase, this abduction period was reduced to about 6% of the stance phase at the sound side, and disappeared at the prosthetic side. This can be attributed to the poor medio-lateral stability of the prosthetic side in comparison to the normal subjects. The individual variations in FPZ were more noticeable among the amputees than among the normal subjects.

Vertical ground reaction force (FPY) of the amputees' sound side displayed the typical two peaks which appear just after heel strike and at push off. The average magnitude of the first peak is 119% (ranging from 103% to 147%) of body weight, and the average magnitude of the second peak is 109% (ranging from 103% to 127%) of body weight. Most patients exhibited a "vaulting" action which was characterised by a third peak in the sound FPY pattern which appeared at about mid-stance (this peak is shown clearly later in the "Butterfly" diagram, figures 6.17 and 6.18) as the patient rose on his sound side to provide clearance for his swinging prosthetic toe. This "vaulting" action can also be caused by a prosthesis which is too long or by a loose suspension system. It was found (fig 6.18a) for the three subjects fitted with quad and IC sockets, that the "vaulting" action was greater when the subjects were fitted with quad than with IC sockets. This may suggest that the suction suspending

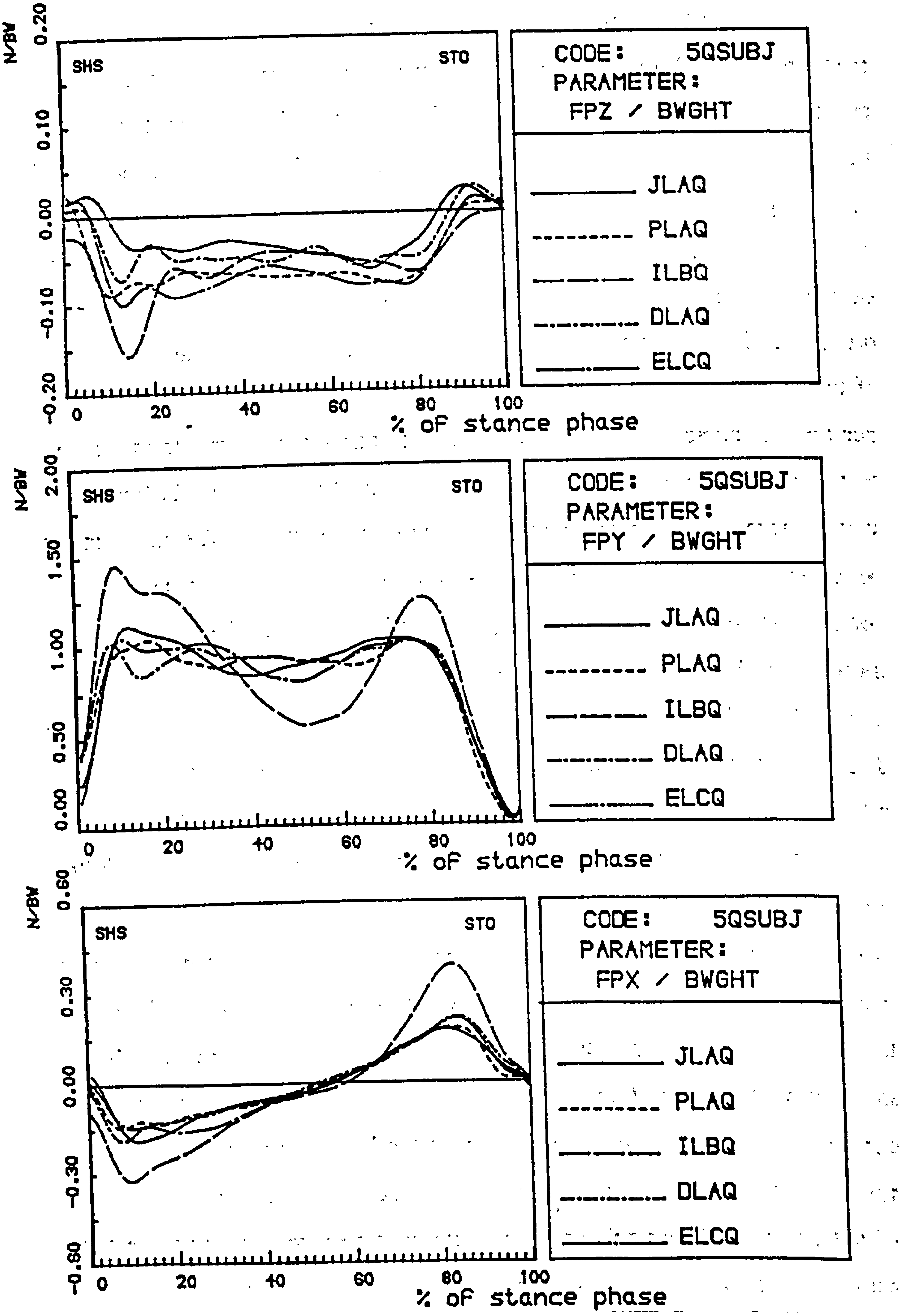


Figure 6.16 Ground reaction forces on the sound leg with time for five AK amputees fitted with quadrilateral sockets. (Normal alignment).

method was more effective with the IC socket than that with the quad socket as it is believed that the IC socket takes the shape of the stump, unlike the quad socket which has a different shape from the residual stump. It was found that the first peak of the FPY of the normal subject occurred at 22% of the stance phase. In the sound side of the AK amputee, this peak appeared at only 12% of the stance phase, this is due to the fact that the patient preferred to load his sound leg quickly, after heel strike, as this would lead to a more stable, comfortable and secure situation. At the prosthetic side, the average magnitude of the first vertical peak was 103% (ranging from 95% to 112%) of body weight, and the average magnitude of the push-off peak was 91% (ranging from 75% to 100%) of body weight. The small magnitude of the push off peak can be attributed to the lack of active plantarflexion in the prosthetic leg. For subjects with quad sockets (fig. 6.15), the first peak of the prosthetic vertical force occurred at about 33% of the stance phase. The delay in the occurrence of this peak could be due to an uncomfortable prosthesis or to a patient's unwillingness to trust the prosthetic leg. It should be noted, that this delay in the appearance of the first vertical peak of the prosthetic leg, may be caused by insufficient ischial bearing of the quadrilateral socket at the heel strike as suggested by Lehneis (1985), this is also supported by the fact that for subjects fitted with IC sockets, this peak occurred at 16% of the stance phase.

In the A/P direction (FPX), the braking force of the sound side of the amputees (average magnitude of 17% of body weight, ranged from 10% to 27%) was slightly smaller that of the normal subjects (average magnitude of 19% of body weight). The push off force of the sound side (average of 21% of body weight) was comparable to that of the normal subjects (average of 21% of body weight). Small variations were found among the subjects except for subject ILBQ whose pattern was substantially different from the others. On the prosthetic side, the fore-and-aft force was found to be much smaller than that of the sound side. For subjects fitted with quad sockets, braking force of the prosthetic side was equal to 40% (66% for subjects with IC sockets) of that of

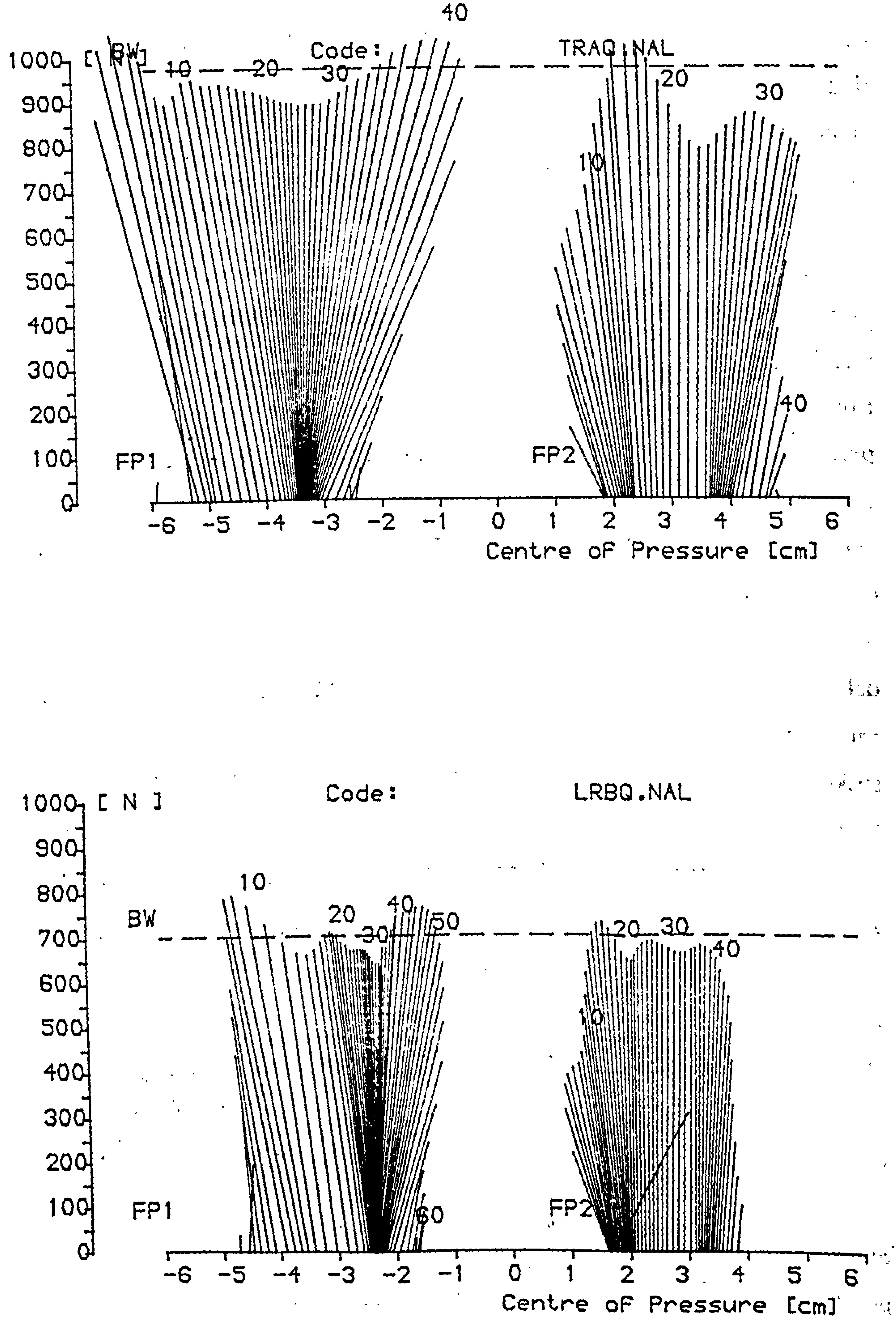
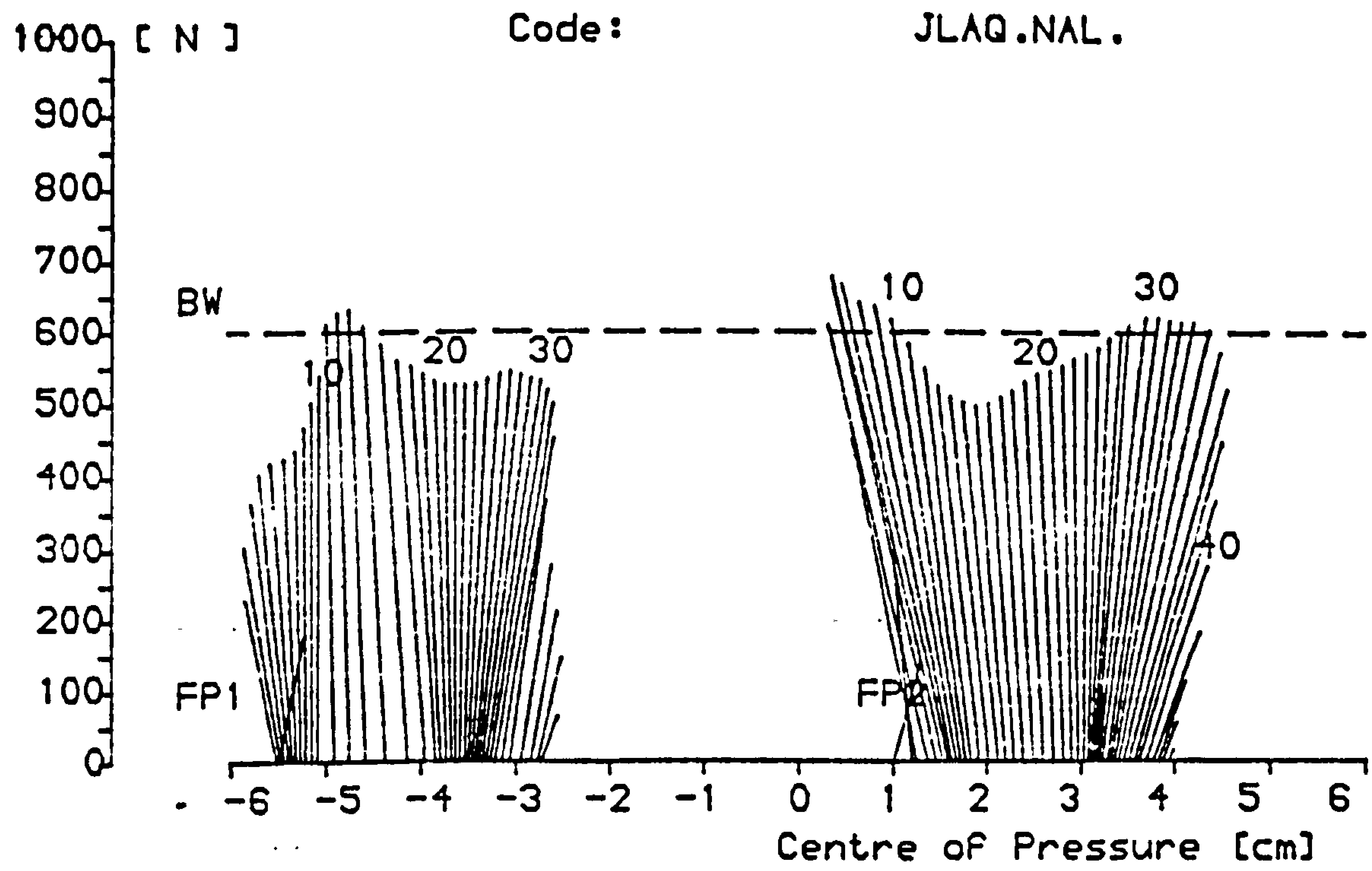


Figure 6.17 "Butterfly" diagrams for two AK amputees fitted with quadrilateral sockets. FP1 corresponds to the sound side and FP2 to the prosthetic side forces. (Normal alignment).



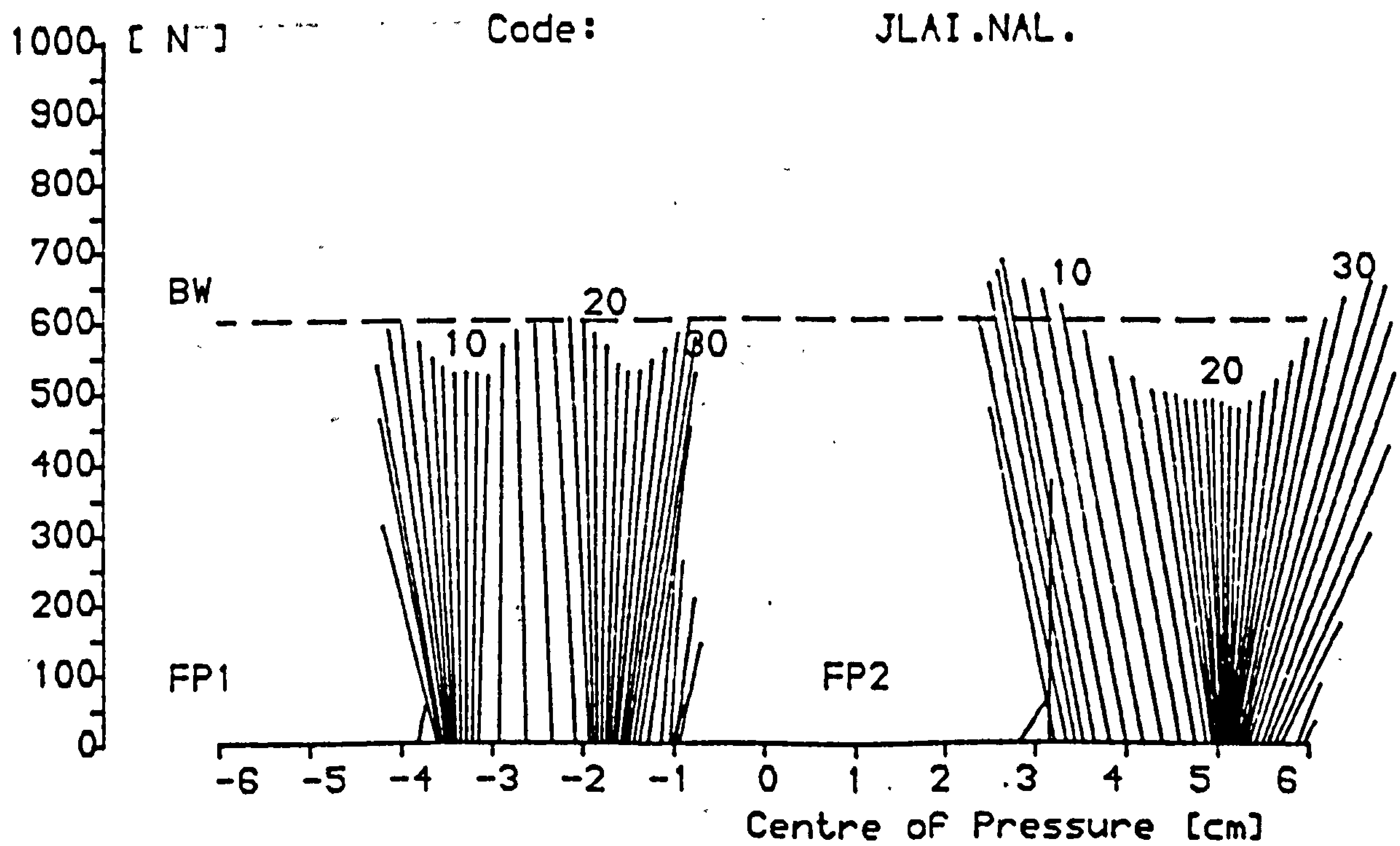


Figure 6.18. Comparison between the "Butterfly" diagrams of a subject fitted with quadrilateral socket (top diagram) and with IC socket (bottom diagram). FP1 corresponds to the prosthetic leg and FP2 to the sound leg forces. (Normal alignment).

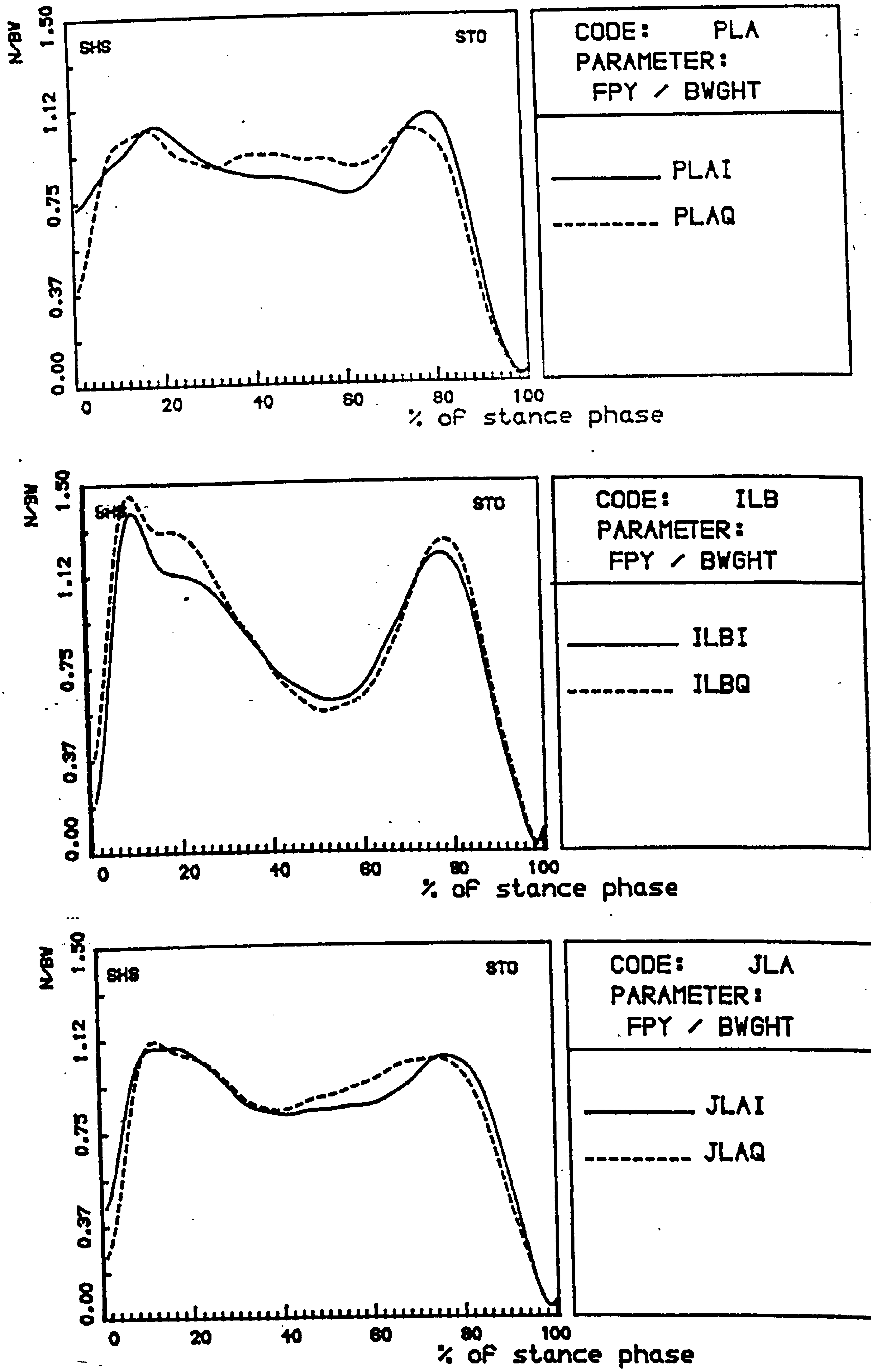


Figure 6.18a Effect of the sockets' type on the "vaulting" action of the sound leg for three above-knee amputees.

the sound side, and the push off force found to be 50% (43% for subjects with IC sockets) of that of the sound side. The small FPX applied on the prosthetic leg in comparison with that on the sound leg can be explained as follows:

The application of a large braking force on the prosthetic foot will generate a high flexing moment on the prosthetic knee. Under this high flexing moment the knee may collapse because of the absence of an extensor mechanism and the need of not having an over stabilised knee alignment. Some patients do not trust the prosthetic leg even if it contained a stabilised knee as in the case of (subject DLAQ) and they prefer not to apply large braking forces on their prostheses. The push off force of the prosthetic leg was also smaller than that of the sound leg. This can be related to the lack of active plantarflexion, and is compensated partially by the hip on the prosthetic side (providing the prosthetic push off force seen in the graph) and partially by the sound leg (providing the large push off force seen at the sound leg).

The braking and generally the push off force of the prosthetic leg was slightly larger for patients wearing IC sockets than that for patients wearing quad sockets. This suggests that the IC socket was more comfortable than the quad socket, in fact this was reported by the three subjects who were fitted with quad and IC sockets. It can also be referred to the fact that subjects fitted with IC sockets walked slightly faster than subjects fitted with quad sockets.

Figure 6.17 shows the "Butterfly" diagram of the sound and prosthetic legs of two AK amputees (TRAQ and LRBQ) fitted with quadrilateral sockets. The "vaulting" effect was noticeable on the sound side and the delay in loading on the prosthetic leg was also evident (33% and 31% of stance phase for TRAQ and LRBQ respectively). The centre of pressure of the sound side travelled slowly during the period between mid-stance and end of the push off, because the sound foot would wait for the prosthetic heel strike and would also wait until the stability of the prosthesis is ensured on the ground. For the prosthetic leg, the centre of pressure moved slowly from heel strike until about 33% of the stance phase, and also during the push off period. This occurred

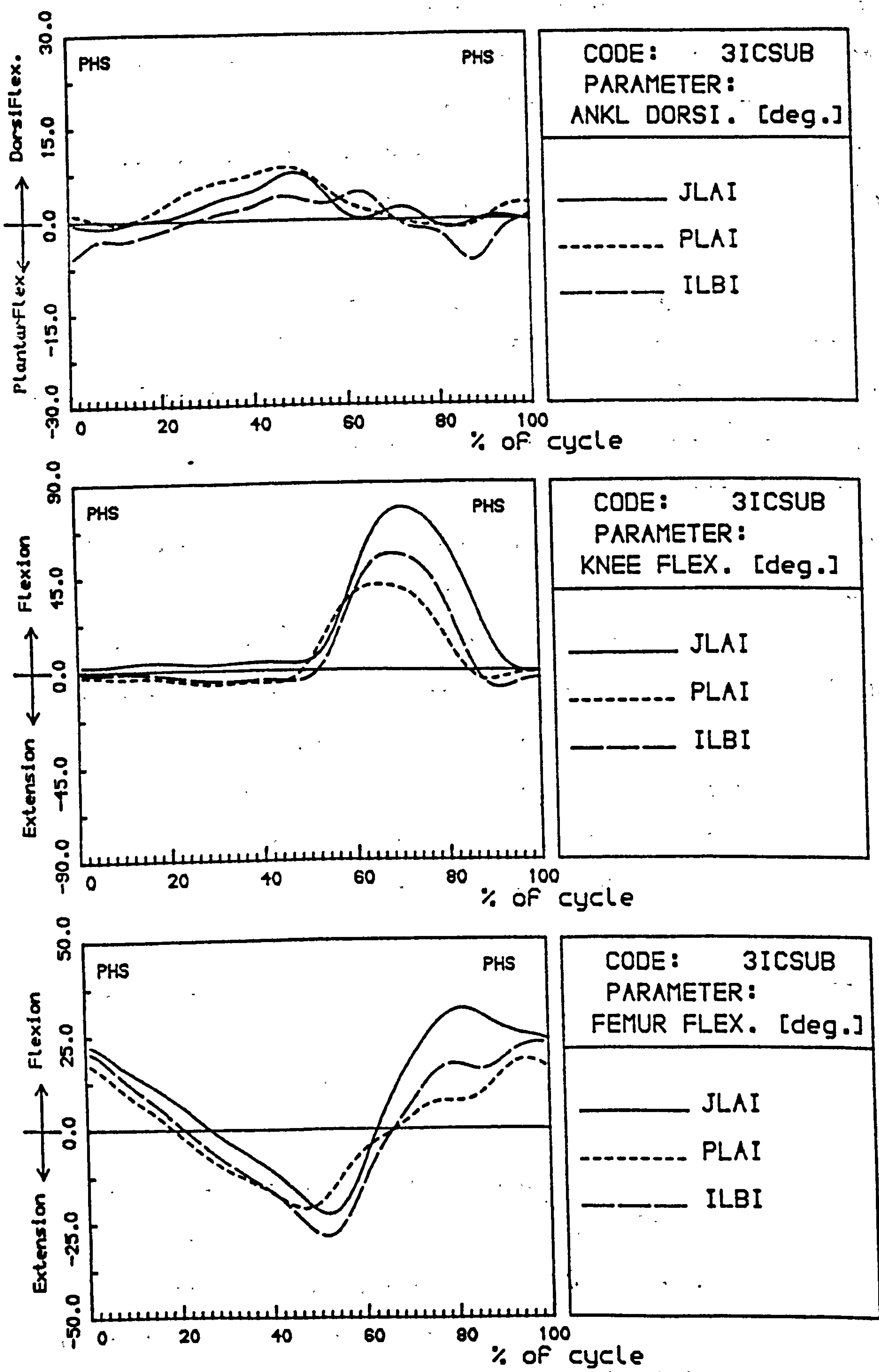


Figure 6.19 A/P angular displacement of the lower limb joints with time for the prosthetic leg for three AK amputees fitted with IC socket. (Normal alignment).

for the period after heel strike because the patient was reluctant to load his prosthesis during that period as he did not trust the stability of the prosthesis and therefore, the progress of stance phase and the centre of pressure will be slowed down. For the push off period, the slow movement of the prosthetic centre of pressure occurred because the subject's body rotated around the foot metatarsal part which was in contact with the ground helping the process of generating the push off force. This slow movement of the centre of pressure during the push off period also existed in the normal subjects. Figure 6.18 shows a "Butterfly" comparison for a subject fitted with quad and IC sockets. The first peak of the "Butterfly" of the prosthetic leg, occurred earlier with the IC socket (16% of the stance phase) than that of the quad socket (31% of the stance phase), and the "vaulting" action disappeared from the sound side with the IC socket. This suggests that the IC socket was more comfortable and had a better suspension system than the quad socket. As a matter of fact all 3 subjects reported that the IC socket felt more comfortable than the quad socket. It should be reported here, that subject MRCQ was converted by the prosthetist of this project from quad to IC user. After wearing the IC socket for five months he reported that the IC socket was more comfortable and the suspension system (suction system) was so successful that this socket felt as if it was a "part of him".

It should be mentioned that some patients (JLAI and LRBQ) showed "vaulting action" at the prosthetic side. This can be explained as another push off phase exerted after heel rise, by the hip of the prosthetic side in order to help the original push off phase which occurred before toe off.

6.6.2 Angular Displacements of the Lower Limb Joints

The A/P angular displacements of the lower limb joints are shown in figure 6.19 for the prosthetic side of the three subjects who wore IC sockets. Figures 6.20 and 6.21 present the A/P angular displacements of the lower limb joints for the prosthetic and sound sides of five AK amputees wearing quad sockets. Results of the other three remaining subjects who were fitted with

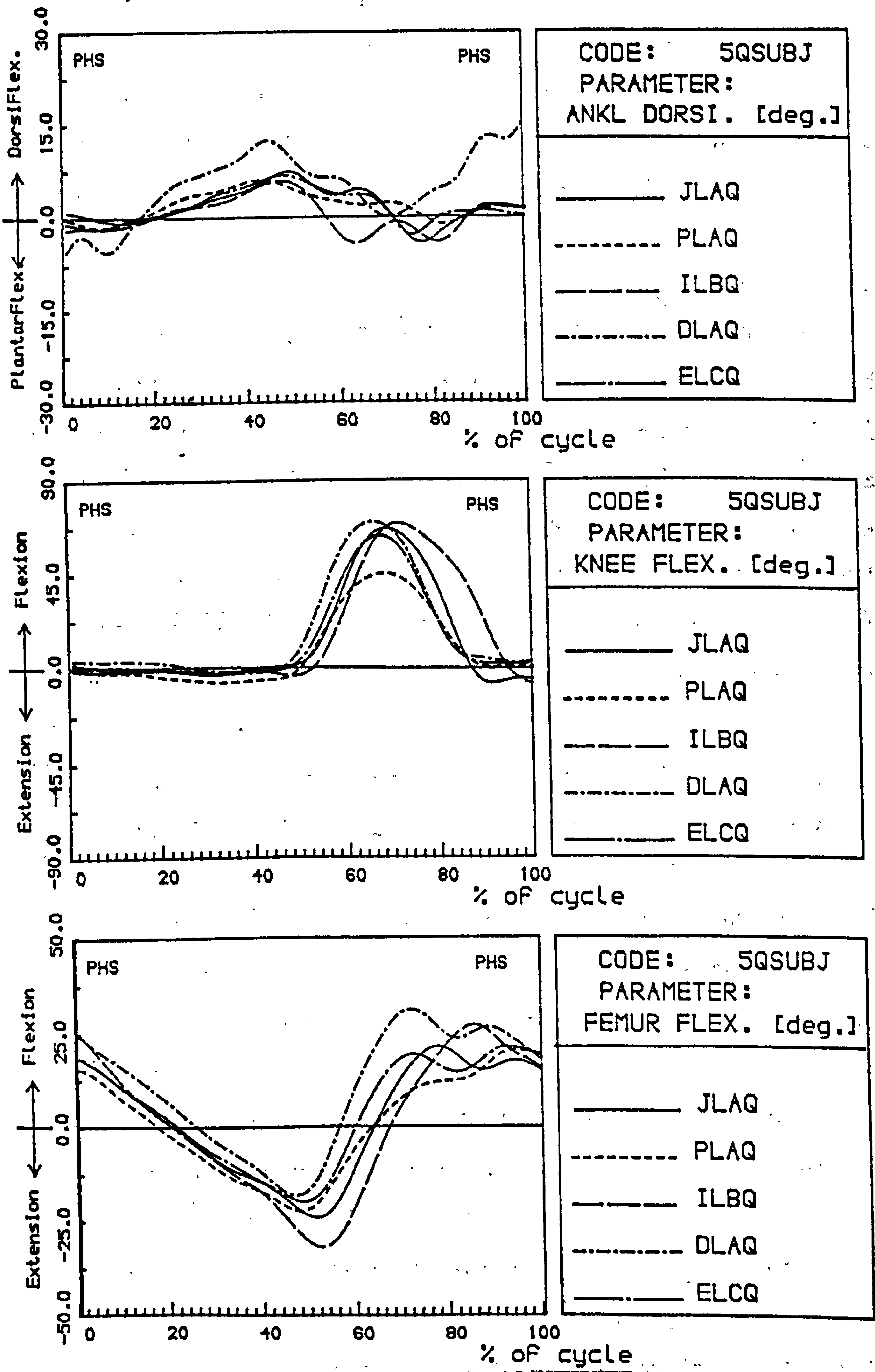


Figure 6.20 A/P angular displacement of the lower limb joints with time for the prosthetic leg for five AK amputees fitted with quad sockets. (Normal alignment).

quad sockets, and of the sound side of the subjects fitted with IC sockets are presented in appendix C. The ankle dorsiflexion/plantarflexion angle of the sound side was comparable in pattern and magnitude to that presented by Goh (1982, fig. 3.45) and generally to that of the normal subjects which were presented in this thesis (fig. 6.5). However, some differences were found between the sound side of the amputees and the normal subjects, in that most patients showed a smaller dorsiflexion angle during the first half of the stance phase, and a larger plantarflexion angle during the push off. This can be attributed to the fact that the amputee has to plantarflex the sound foot in order to achieve a long prosthetic step length. However, no significant (P=0.07) differences were found between the step length of normal subjects and that of the prosthetic leg of the amputees (section 6.4.2.2). This is related to the fact that the push off force of the prosthetic leg was smaller than that of the normal subjects. Subjects ILBQ and ELCQ showed a noticeable plantarflexion (13 to 14 degrees) just after mid-stance (34 % of the gait cycle) which coincided with the "vaulting action". This plantarflexion angle was larger (about 22 degrees) with subject ILBI which is attributable to a habit developed by that subject.

At the prosthetic ankle, all SACH feet showed a similar behaviour over the gait cycle (except subject LRBQ) and the results of this study are comparable to those presented by Goh (1982) and Krebs and Tashman (1985, fig. 3.47). The prosthetic ankle showed a slight plantarflexion angle after heel strike and a noticeable dorsiflexion angle (2 to 7 degrees) at the push off (compare 5.4 in Goh, and 9 in Krebs and Tashman). The variations in the dorsiflexion and plantarflexion angle of the prosthetic ankle can be attributed to the stiffness of the prosthetic heel and to the flexibility of the prosthetic fore foot respectively, and to the load applied on the foot. Most patients had a slight foot plantarflexion angle (3 to 7 degrees) just after toe off. This can be explained as a foot spring back occurring when the load was removed from the foot, and the magnitude and time of occurrence of this plantarflexion movement are controlled by the stiffness of the fore foot. Subject DLAQ

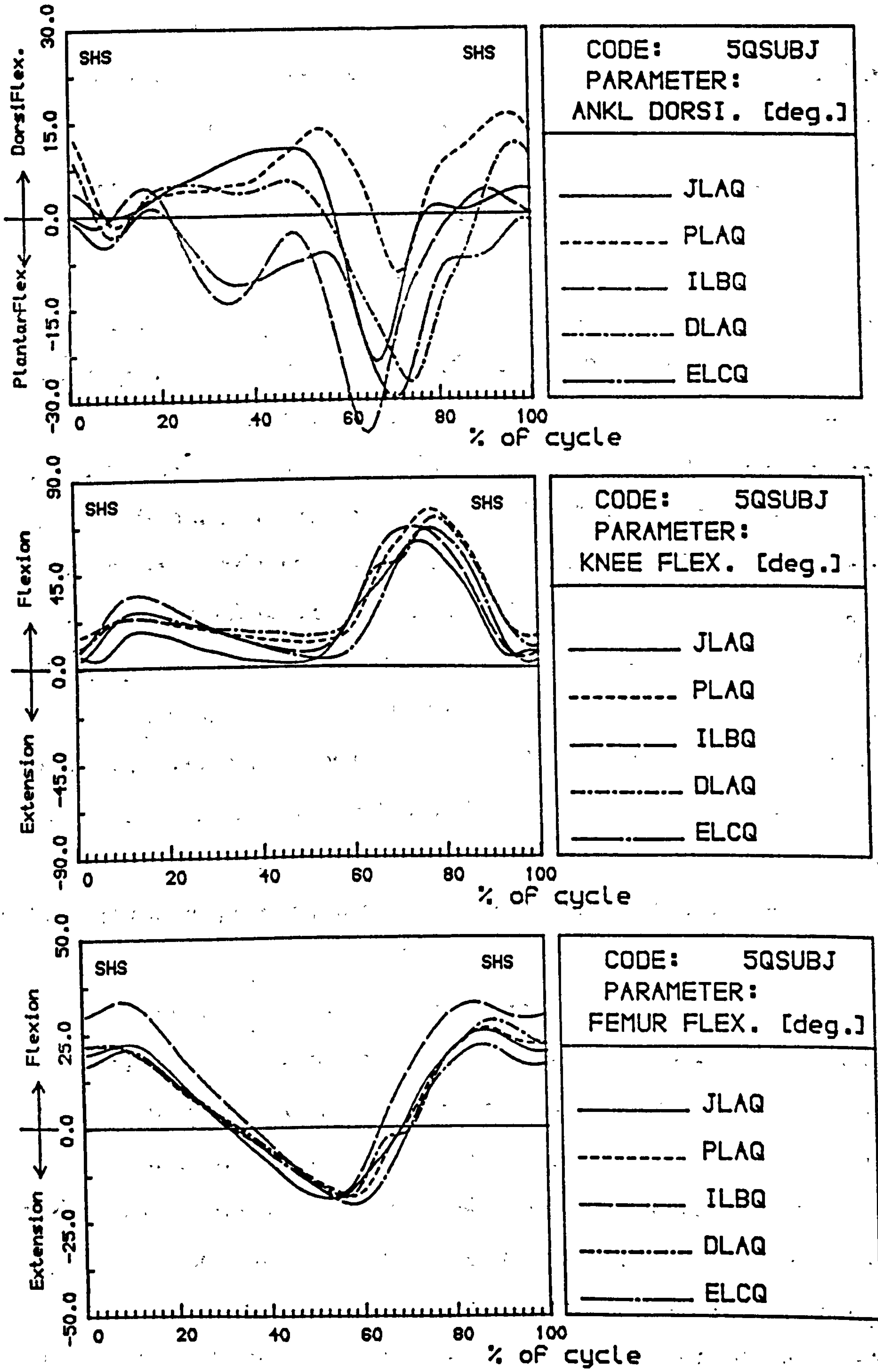


Figure 6.21 A/P angular displacement of the lower limb joints for the sound leg with time for five AK amputees fitted with quad sockets. (Normal alignment).

showed a larger foot dorsiflexion angle during the push off phase of his prosthetic side than all other subjects. This can be referred to the uniaxial type of foot which was supplied to this subject, and such a difference in performance between the uniaxial and SACH foot was also found by Goh (1982).

The A/P knee and femur angular displacement of the sound and prosthetic sides obtained in this study for the AK amputees, are comparable to those presented by Eberhart (1947), Murray et al (1980), Goh (1982) and Krebs and Tashman (1985). No differences were found between displacements of the sound leg and those of the normal subjects. The prosthetic knee showed no flexion during the stance phase as the knee was maintained in full extension in order to ensure the stability of the prosthesis. The amplitude of the knee flexion angle during swing phase for the sound and prosthetic legs of most subjects is approximately the same, except subjects LRBQ and PLAI who showed a relatively small knee flexion during the swing phase of the prosthetic leg. These two subjects exerted a markedly small push off force at the prosthetic leg in comparison to the other amputees. Comparing with the normal subjects or the sound side of the amputees, the femur of the prosthetic leg showed no further flexion after heel strike and it began extending immediately after heel strike. This can be attributed to the lack of knee flexion during the stance phase. At heel strike the femur of the prosthetic leg was flexed by 20 degrees, while the femur of the sound side was flexed by 24.5 degrees. The femur of the prosthetic leg was flexed by approximately 20 degrees during swing phase, and was held in that flexed situation for approximately 17% of the gait cycle. Holding the femur in flexion can be attributed to the fact that the thigh of the prosthetic leg has to wait for the swinging foot which remains in the air for a longer time period than that the sound (or the normal) leg, since it is usually travelling for a longer distance (longer prosthetic step length), and also because of the lack of quadriceps section over the knee which normally bring the shank forwards . No

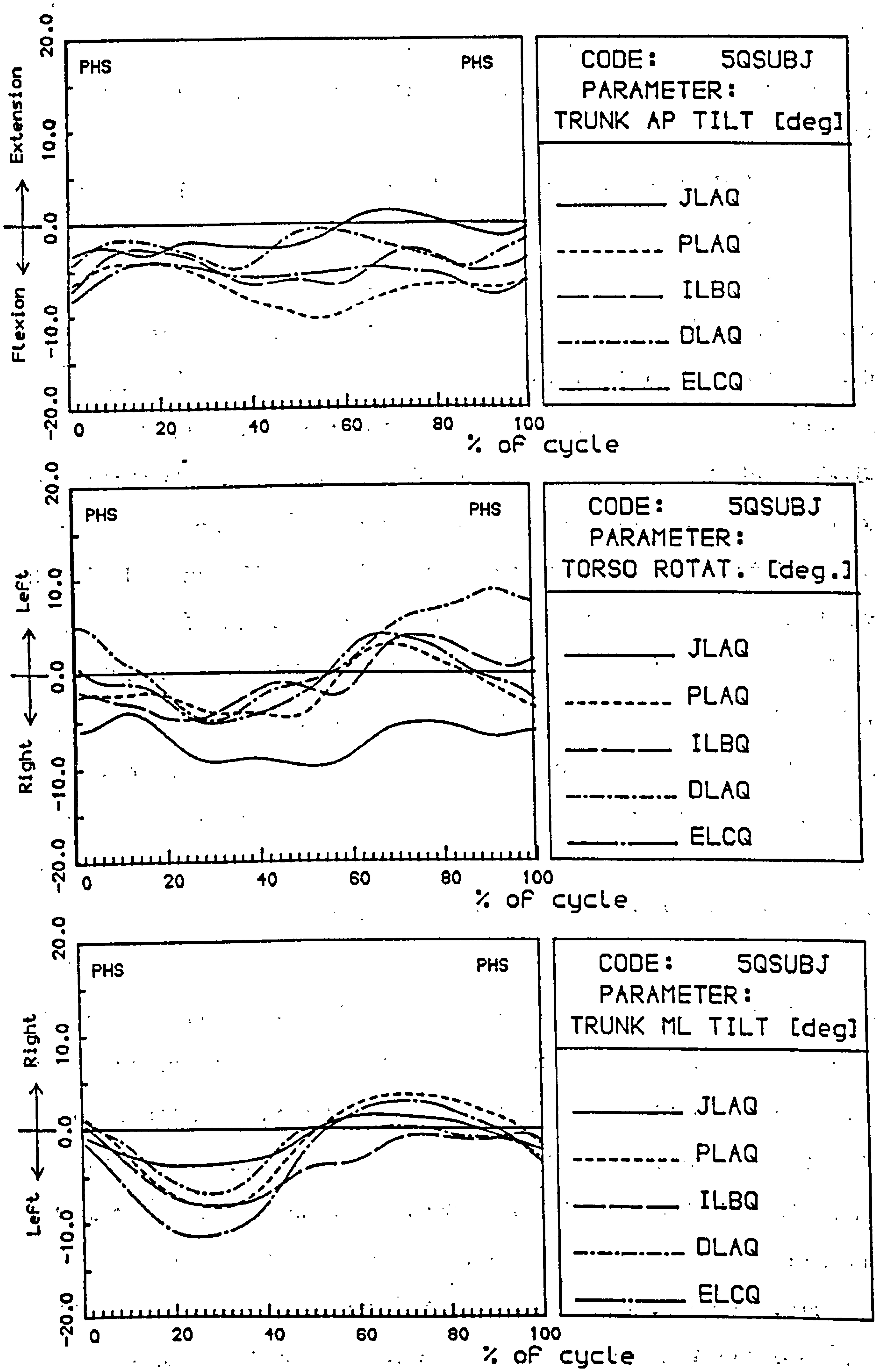


Figure 6.22 A/P and M/L angular displacements of the trunk and transverse rotational displacement of the torso with time for five AK amputees fitted with quadrilateral sockets. (Normal alignment).

differences were found between the quad and the IC sockets with respect to the A/P displacement of the lower limb joints.

6.6.3 Angular Displacements of the Trunk

Figures 6.22 and 6.23 present the results obtained in this study for the trunk angular displacements of five AK amputees fitted with quad sockets and three AK amputees fitted with IC sockets respectively. Results for the remaining three subjects whom were fitted with quad sockets are presented in appendix C. Comparing the results with those of the normal subjects (fig. 6.7), the M/L trunk tilt of the amputees had a similar pattern to that of the normal subjects, i.e. the trunk was always tilted towards the supporting leg. However, the magnitude of the tilt was larger for the amputees than for the normal subjects (2 degrees), and it was also larger towards the prosthetic side (about 7 degrees) than that towards the sound side (about 3.5 degrees). This can be related to the difficulties faced by the amputees in maintaining medio-lateral stability as the thigh adductors were weakened by the amputation. It should be mentioned that no noticeable differences were found between the trunk M/L displacements of subjects wearing quad and IC sockets. This does not agree with Sabolich (1985) who reported that with the IC socket "everything is stabilised in the ML direction", and since the socket is not abducted the patient does not have to lean toward the prosthetic leg during its stance phase, and thus, the patient would walk with a normal gait.

The A/P trunk displacements of the amputees also had a similar pattern to that of the normal subjects. However, the magnitude of this displacement (approximately ± 3.3 about a base line) was larger for the amputees than for the normal subjects (approximately ± 2 about a base line), and more variations existed among the amputees than among the normal subjects. These variations could be related to the degree of control which can be effected by the amputee, on the prosthesis to achieve A/P stability.

In the transverse plane, most patients showed a similar pattern for torso rotation to that of normal subjects, and the difference between two successive

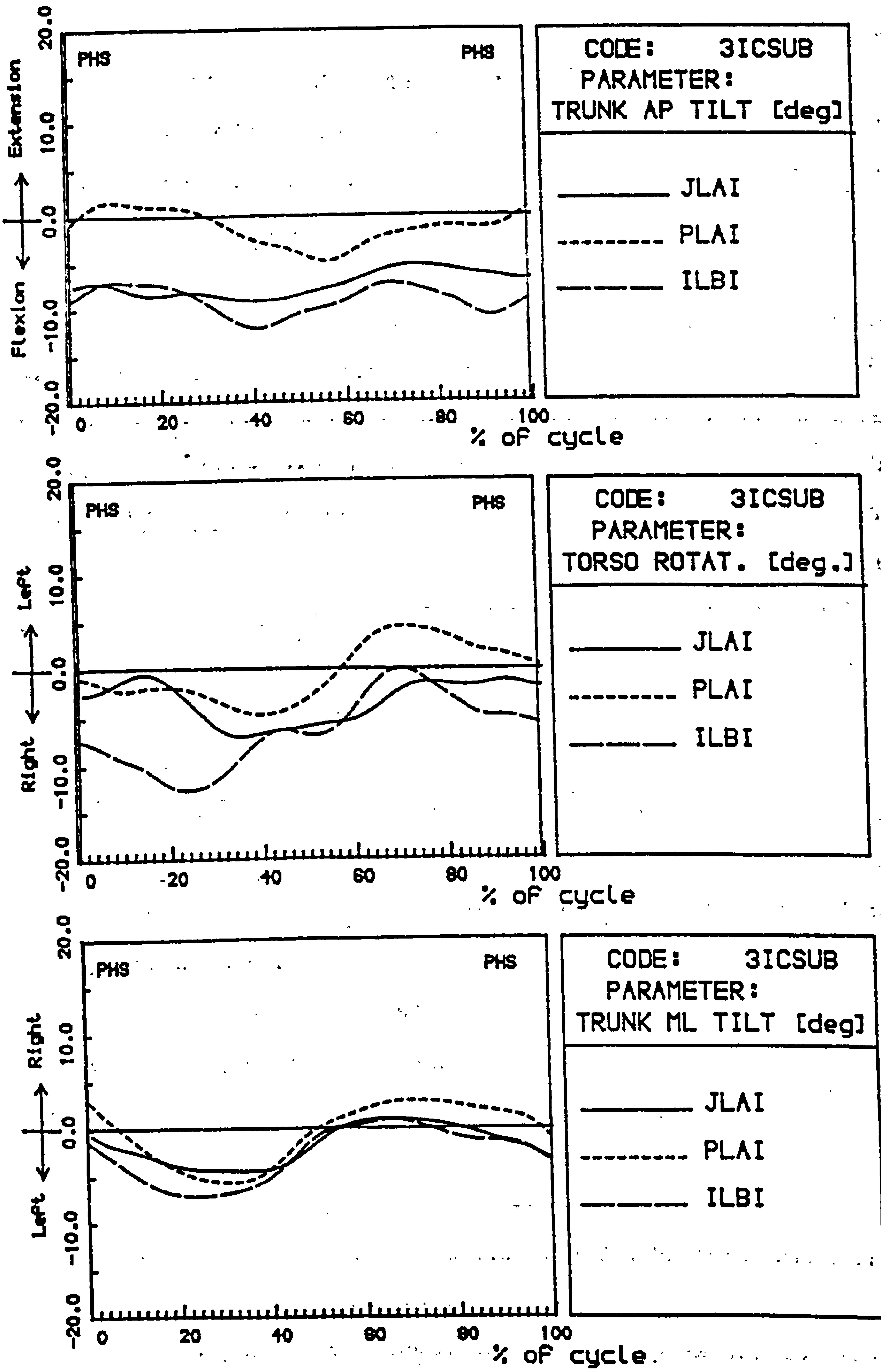


Figure 6.23 A/P and M/L angular displacements of the trunk and transverse rotational displacement of the torso with time for three AK amputees fitted with IC sockets. (Normal alignment).

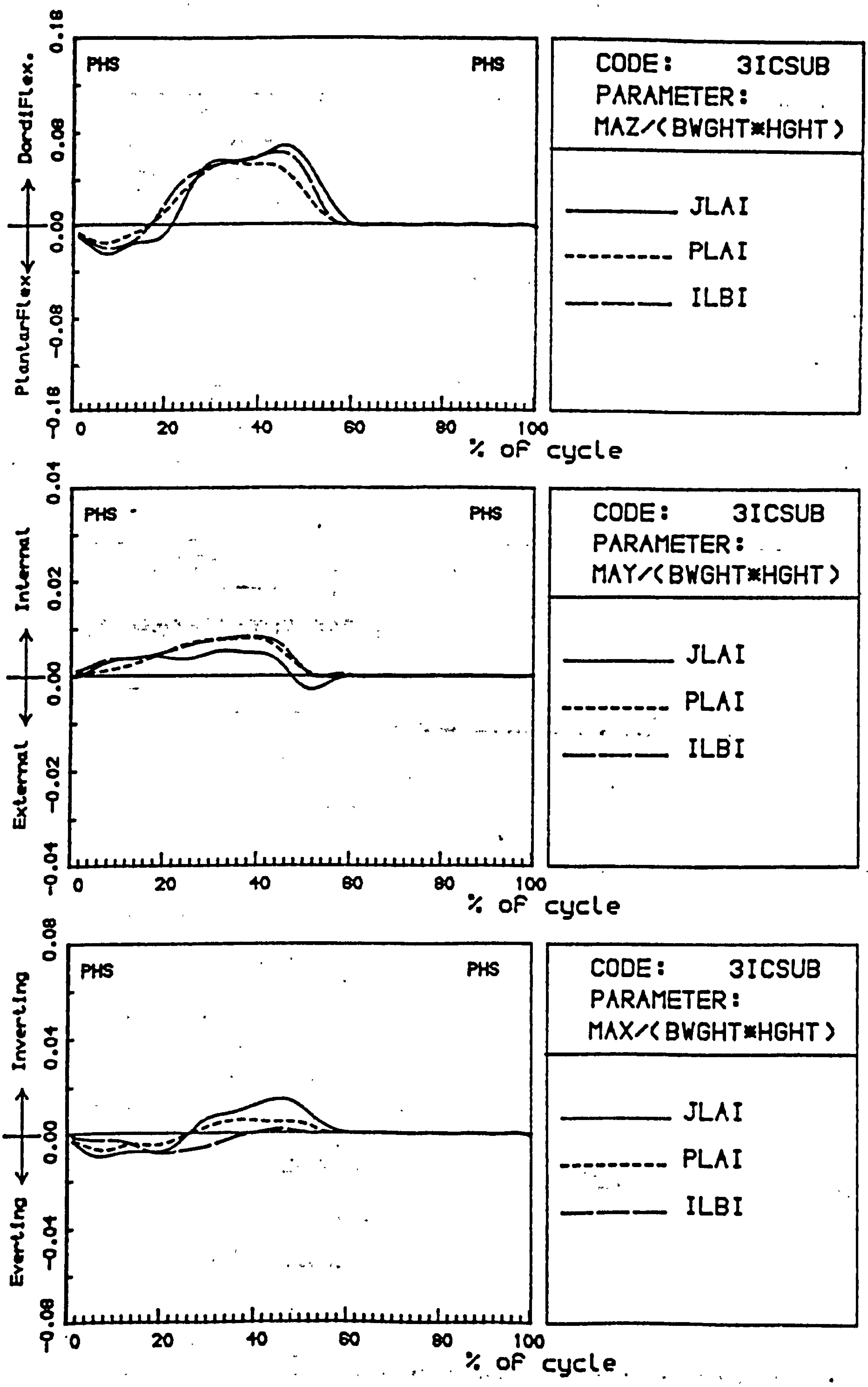


Figure 6.24 Prosthetic ankle joint moments with time for three AK amputees fitted with IC sockets. (Normal alignment)

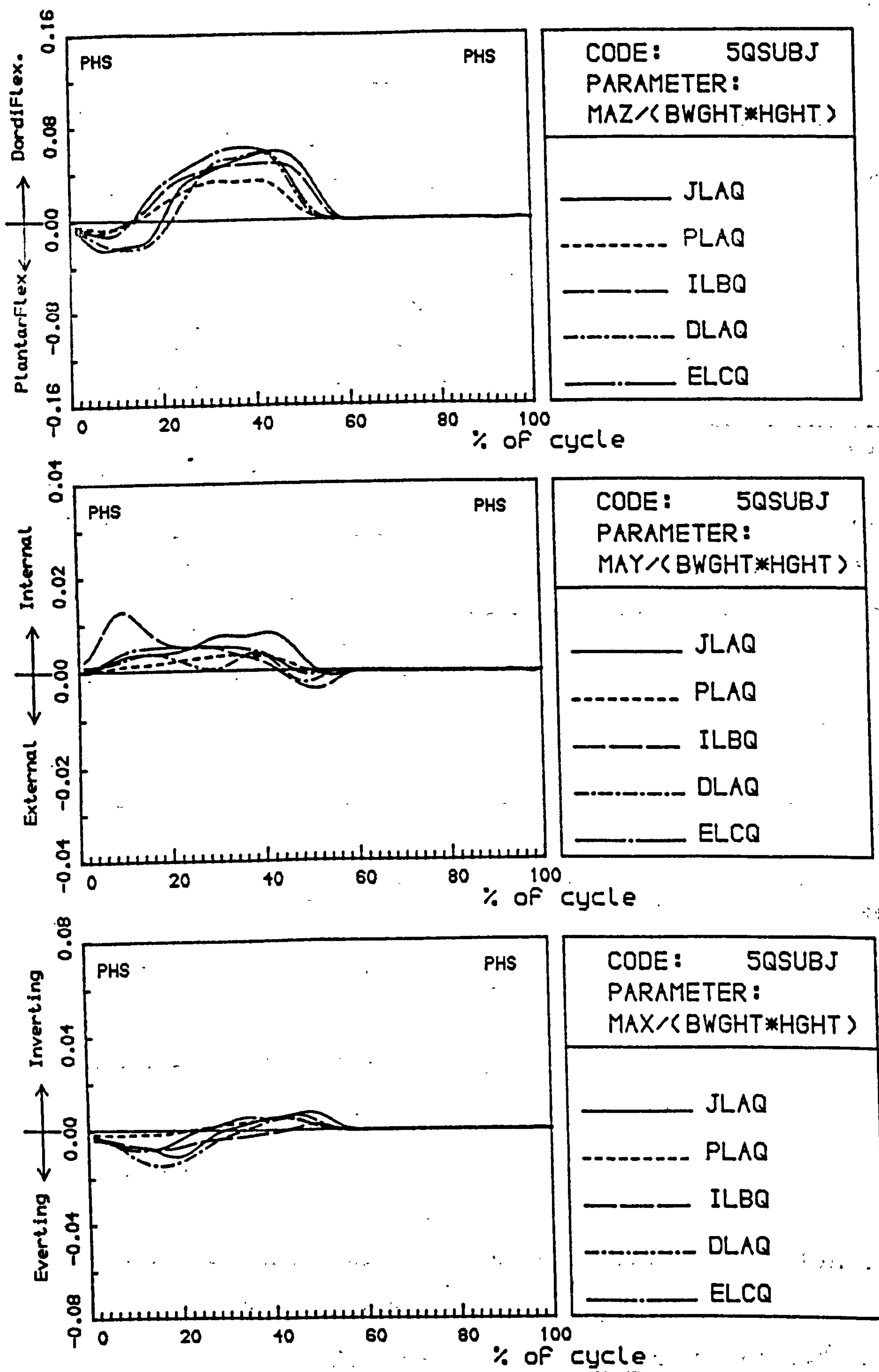


Figure 6.25 Prosthetic ankle joint moments with time for five AK amputees fitted with quadrilateral sockets. (Normal alignment)

peaks of torso rotation was approximately 10 degrees (also 10 degrees for the normal subjects). This can be attributed to the fact that the patients are active and had approximately similar velocities to that of the normal subjects. Most patients showed less torso rotation to the prosthetic side (about 5 degrees) than to the sound side (about 7 degrees), and patients PLAI and TRAQ showed a very small rotation to the prosthetic side (see fig. 6.22). The difference in torso rotation between the sound and prosthetic sides can be attributed to the difference between the sound and prosthetic step lengths (see table 6.5). Some differences were found between the pattern of torso rotation for subjects wearing quad and the IC sockets, however, these differences were inconsistent.

6.6.4 The Ankle Joint Moments

The ankle joint moments of the prosthetic side for the three amputees fitted with IC sockets are shown in figure 6.24, for the normal alignment. The ankle moments of five subjects wearing quad sockets are shown in figures 6.25 and 6.26 for the prosthetic and sound side respectively. In the A/P plane, moments of the sound leg were comparable to those presented by Goh (1982) and to those obtained in this study for the normal subjects. However, the pattern of the A/P moment of the sound leg of most subjects increased sharply after foot flat and formed a different pattern shape from that of the normal subjects which increases gradually between foot flat and push off. At the prosthetic side, the A/P ankle moment (MAZ) showed some differences from that of the sound leg and of the normal subjects. The plantarflexing moment lasted much longer (22% to 44% of stance phase) than that of the normal subjects (11% to 16% of stance phase). This can be attributed to difficulties faced by the patient in transferring his body weight over the prosthesis as a result of the inactive ankle joint, and that the subject had to spend a longer time on the heel area causing a longer duration of the plantarflexing moment. The inactive ankle joint of the prosthetic leg also caused a reduction in the magnitude of the dorsiflexing moment at the push-off, and this was noticed in all patients in comparison to the normal subjects. The dorsiflexing moment of

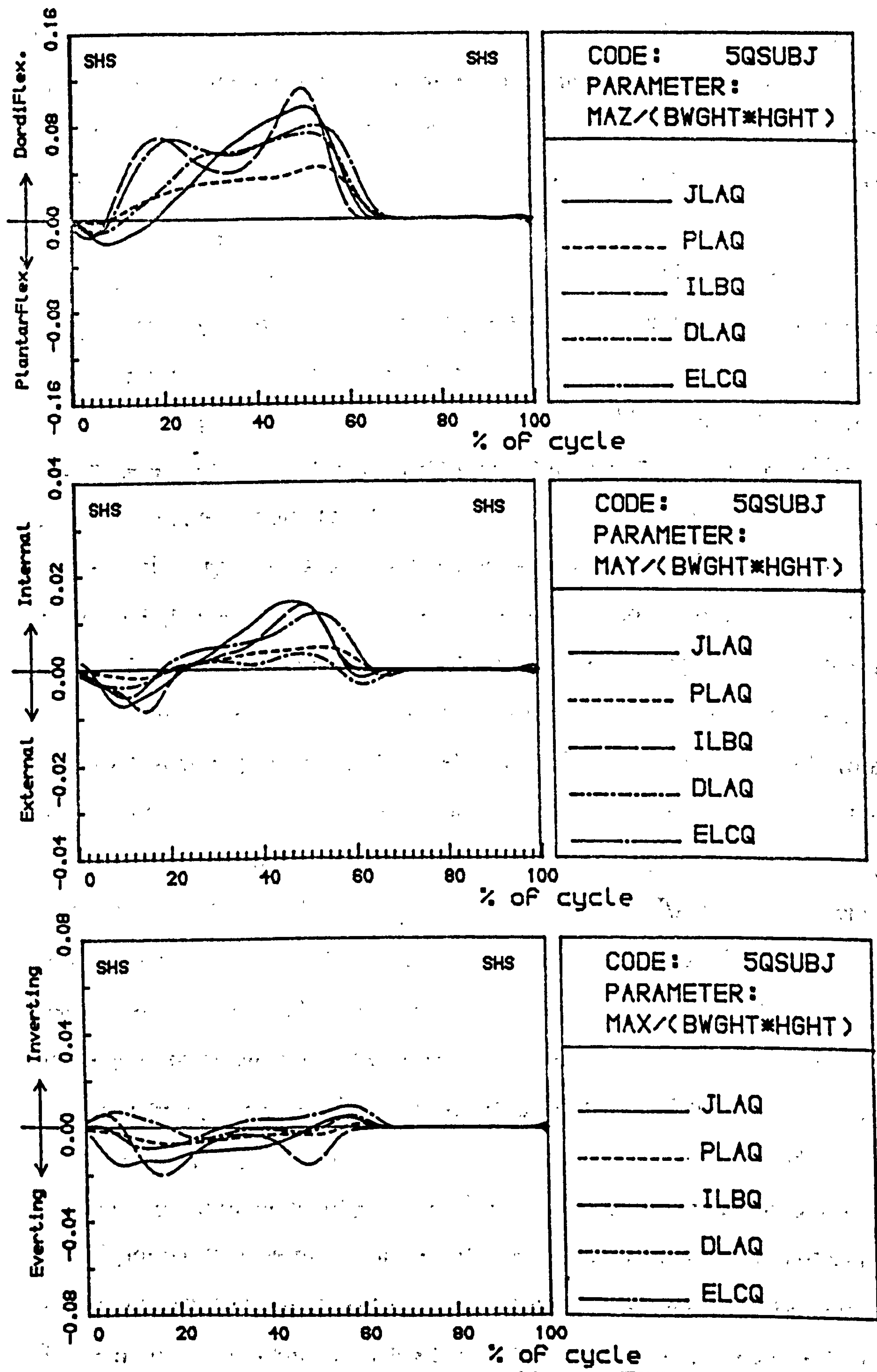


Figure 6.26 Ankle joint moments on the sound leg with time for five AK amputees fitted with quadrilateral sockets. (Normal alignment)

the prosthetic ankle was approximately 69% of that of the normal subject which was found to have an average value of 113 Nm. No differences were found between the A/P ankle moment of subjects wearing IC and quad sockets.

In the transverse plane, most amputees exhibited a similar pattern of moment (MAY) at their sound ankle to those of the normal subjects. For the duration from heel strike until approximately 33% (33% for the normal subjects) of the stance phase, the sound ankle was subjected to a rotation moment which tended to rotate the ankle externally, the average magnitude of this moment is 4 Nm (4.5 for the normal subjects), and varied among subjects from 1.6 to 10 Nm. Thereafter, an internally rotating moment dominated the rest of the stance phase, the average magnitude of this moment is 12 Nm (13 Nm for the normal subjects), and ranged from 4 to 19 Nm. Subjects TRAQ and MRCQ only, exhibited a different pattern of ankle moment on their sound side regarding the transverse plane. At the prosthetic ankle, all subjects showed a similar pattern of moment in that an internally rotating moment, dominated the stance phase, this pattern is comparable to that obtained by Goh (1982) shown in figure 3.46. For most subjects, the peak magnitude of this moment was smaller (about 50%) than that of the sound ankle. The small magnitude of the prosthetic MAY during stance phase can be related to the small step length which was undertaken by the sound leg when the prosthetic leg was on the ground.

In the M/L plane, the ankle joint moments of the sound and prosthetic legs of the AK amputees were different in pattern and magnitude from those of the normal subjects. Also, no consistency among subjects was found in the pattern of the M/L moment of the sound leg. However, on the prosthetic side, all the left amputees exhibited a similar pattern of moment, in a manner that the ankle of most subjects was subjected to an everting moment (approximately 10 Nm) from heel strike until about 50% of the stance phase. Thereafter, an inverting moment (approximately 8 Nm) dominated the rest of the stance phase. The right amputees again exhibited a consistent pattern of moment at

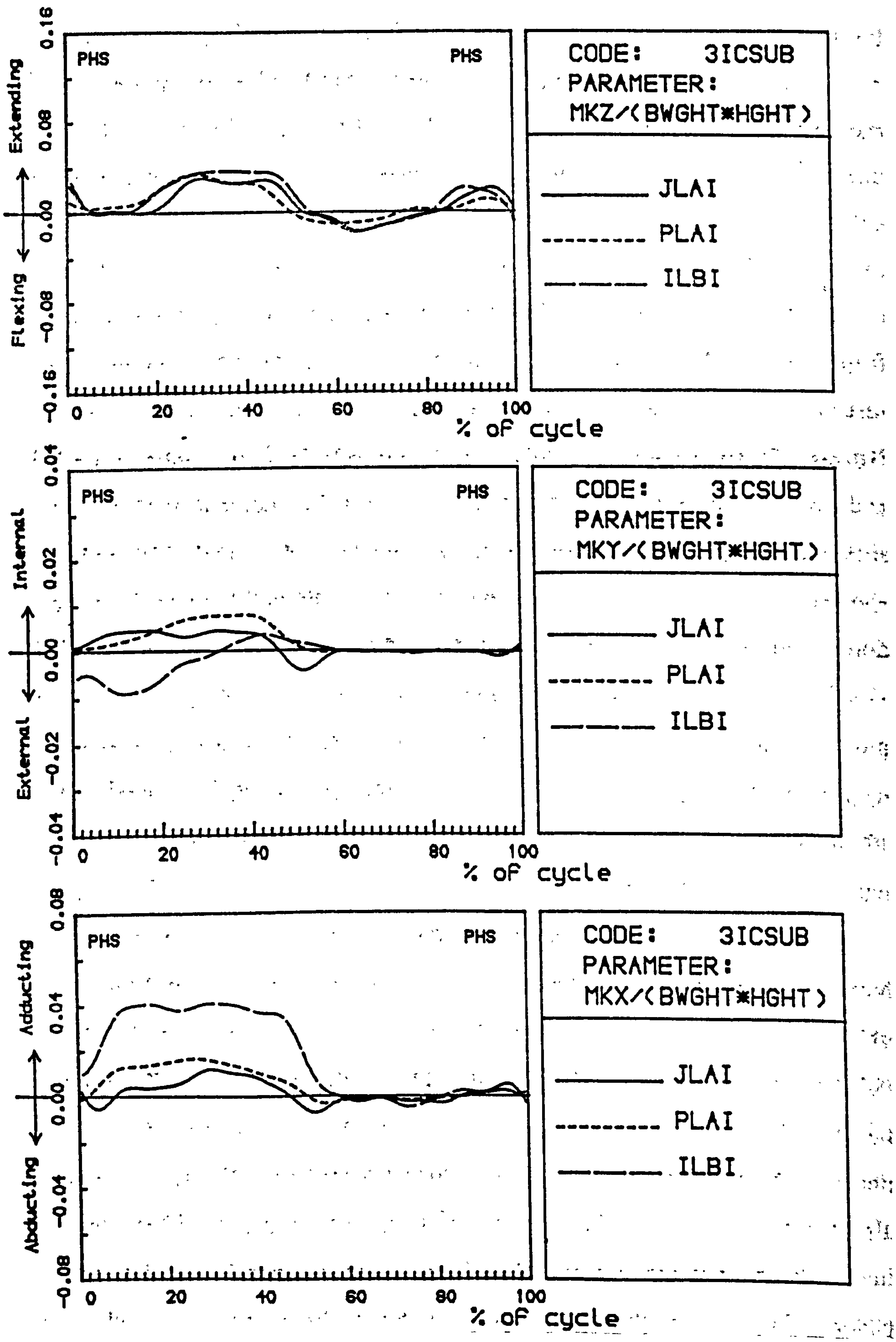


Figure 6.27 Prosthetic knee joint moments with time for three AK amputees fitted with IC sockets. (Normal alignment)

their prosthetic ankle in the M/L plane, and this pattern was comparable to that of normal subjects. The magnitude of this moment was 2.6 times larger than that of the prosthetic leg of the left amputees, and 1.6 times larger than that of normal subjects. The difference in the ankle's M/L moment pattern among subjects and between the right and left amputees can be related to the foot alignment in the M/L direction which dictates whether the subject lands medially or laterally on the foot.

6.6.5 The Knee Joint Moments

Figure 6.27 shows the knee joint moments of the prosthetic side of the three subjects whom were fitted with IC sockets. Moments of the prosthetic and sound knee of five subjects fitted with quad sockets are presented in figures 6.28 and 6.29 respectively. The results of this study were comparable to those obtained by Goh (1982, shown in fig. 3.47). In the A/P plane, most patients showed a similar pattern of moment on their sound side, and this pattern was comparable to that obtained in this study for normal subjects (shown in fig. 6.10 and 6.11). Two periods of extending moment, and a period of flexing moment were found at the sound side during stance phase. The flexing moment lasted from about 4% to about 22% of the gait cycle, and had an average magnitude of 31 Nm (47 Nm for the normal subjects), ranging from 19 to 61 Nm. The first period of extending moment lasted from heel strike until 4% of the gait cycle (7% for the normal subjects), and its average magnitude is 26 Nm (20.9 Nm for the normal subjects), ranging from 13 to 64 Nm. The second period of extending moment lasted from about 22% of the gait cycle until toe off, and the average magnitude of this moment is 32 Nm (22 Nm for the normal subjects), ranging from 16 to 65 Nm. The reasons behind this pattern of moment have been discussed in section 6.5.5 for the normal subjects. During swing phase, the A/P moment of the sound leg was also comparable to that of normal subjects with no noticeable differences.

On the prosthetic side, all subjects showed a consistent and similar pattern of moment in the A/P plane. The stance phase was dominated by an

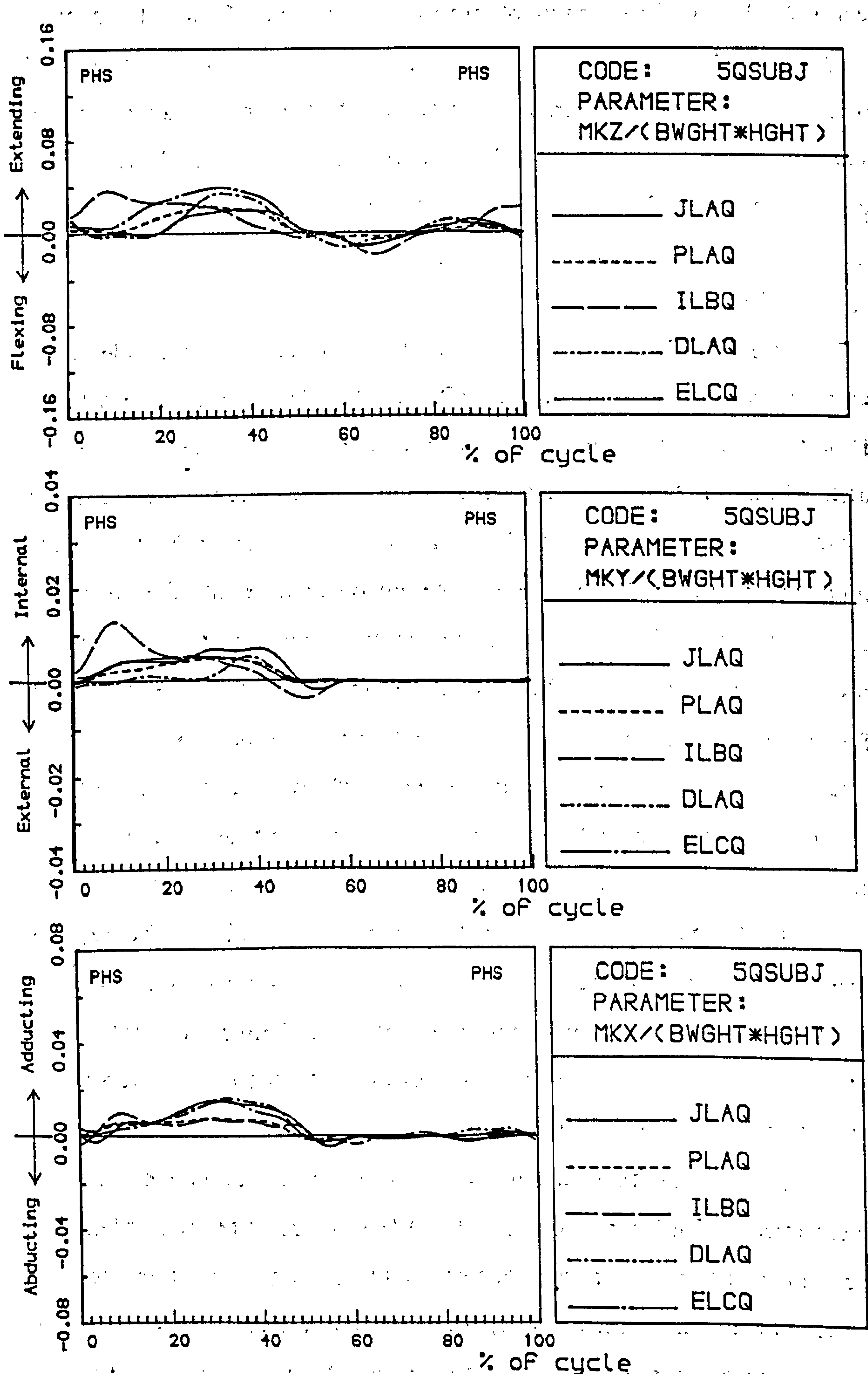


Figure 6.28 Prosthetic knee joint moments with time for five AK amputees fitted with quadrilateral sockets. (Normal alignment)

extending moment (average value of 38 Nm, ranged from 22.6 to 48 Nm) in order to maintain knee stability because a flexing moment cannot be supported by a plain uniaxial knee. Subjects MRCQ and LRBQ showed a small knee flexing moment (about 5 Nm) at the prosthetic knee during stance phase. Since the patients maintained stability during that period of knee flexing moment, it is believed that this knee flexing moment was compensated by pin friction of the knee axis. Subject DLAQ also showed a small knee flexing moment during the prosthetic stance phase, but this can be accepted as the patient used a uniaxial safety knee. During the swing phase, the prosthetic knee exhibited a typical pattern of knee moment in the A/P plane. During the first half of the swing phase, the prosthetic knee was subjected to a flexing moment with the maximum value coincident with the largest knee flexion as the acceleration was maximum. The second half of the swing phase was dominated by an extending moment at the prosthetic knee, and the maximum extending moment corresponded to the instant when the knee was in full extension. The magnitude of the prosthetic knee moment during swing phase was about 50% smaller than that of the sound leg. This can be related to the differences in the mass between the prosthetic and the anatomical shanks.

In the transverse plane, the knee moments of the prosthetic and sound sides were similar to their respective ankle joint moments. This was expected, as the moments of the knee and ankle joints were expressed in the shank frame of reference, and both of them belong to the shank segment which would act as a rigid body and would be subjected to a uniform torque. The transverse moment of the prosthetic knee had the tendency of rotating the knee internally in order to maintain the stability of the prosthesis.

In the medio-lateral plane, all subjects showed an adduction knee moment which helped in maintaining medio-lateral stability during the stance phase. The pattern and magnitude (27.6 Nm) of this moment which was obtained for the sound side, was comparable to that of the normal subjects (29 Nm). The prosthetic knee moments were always adducting the shank, and the

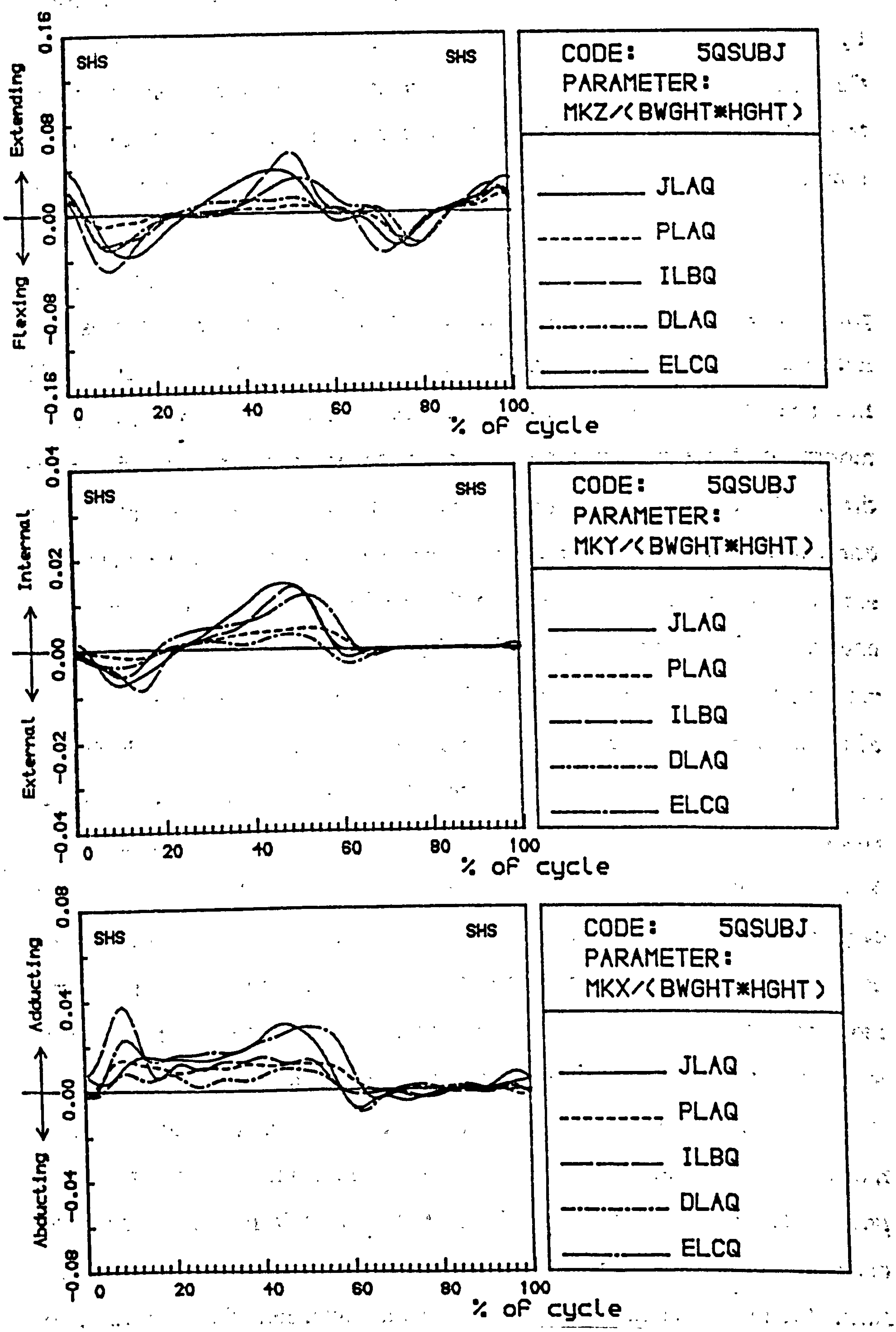


Figure 6.29 Knee joint moments on the sound leg with time for five AK amputees fitted with quadrilateral sockets. (Normal alignment)

magnitude of this moment was noticeably smaller (about 50%) than that of the sound side. This can be related to the fact that the M/L tilt of the amputee's trunk was greater towards the prosthetic side than that of the sound side. Thus, the M/L offset between the knee joint centre and the vector of the ground reaction force, will be small on the prosthetic side in comparison to that of the sound side.

6.6.6 The Hip Joint Moments

Figure 6.30 presents the hip joint moments obtained in this study for the prosthetic side of three AK amputees fitted with IC sockets. Hip moments for five AK amputees fitted with quadrilateral sockets are shown in figures 6.31 and 6.32 for the prosthetic and sound side respectively. It was found that patterns of the hip moments obtained in this study for the sound leg are comparable to those obtained by Goh (1982, shown in fig. 3.47), and to those obtained in this study for the normal subject (fig. 6.12). In the A/P plane, the sound hip of most subjects was subjected to a flexing moment from heel strike until approximately mid-stance when the moment reversed to an extending moment ana lasted until the end of the stance phase. The average magnitude of the hip flexing moment is 63.6 Nm (41 Nm obtained in this study for the normal subjects, and 100 Nm obtained by Goh 1982 for the sound side of an AK amputee), ranging from 38 Nm to 107 Nm. This excluding patient MRCQ who exhibited a hip flexing moment of 207 Nm. The difference in the magnitude of the hip flexing moment between the sound leg of the amputees and the normal subjects, is related to the fact that the sound leg of the amputees have larger FPY and smaller FPX than those of the normal subjects. In some patients such as TRAQ and ILBQ, the hip moment of the sound leg was dominated by a flexing moment during stance phase. This can be related to the style of walking which was adopted by these patients, or to the large push off force (200 N and 266 N respectively) which was exhibited by these subjects in comparison to the other subjects (about 150 N). The average magnitude of the hip extending moment of the sound side is 24 Nm (35 Nm

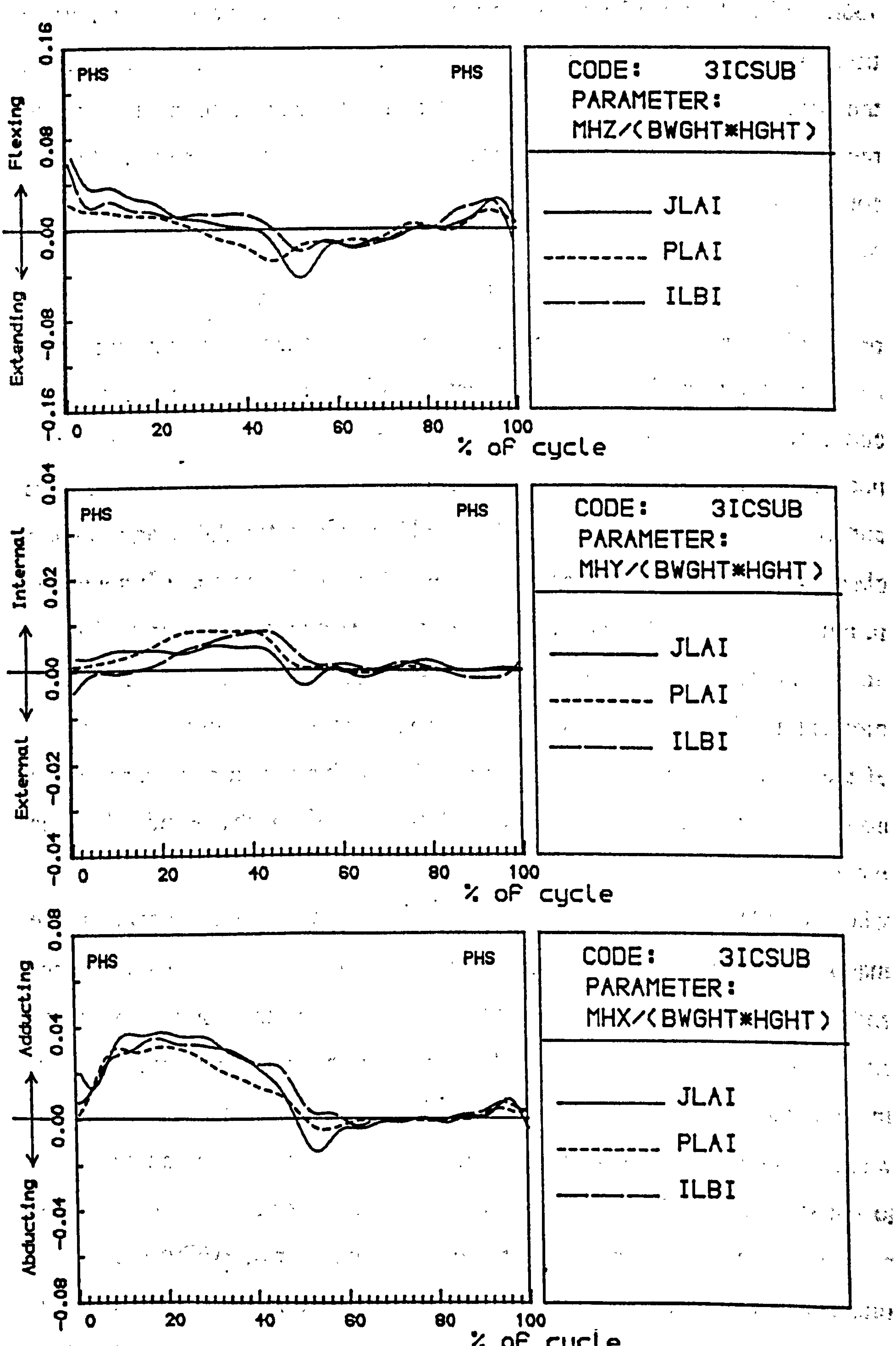


Figure 6.30 Hip joint moments on the prosthetic leg with time for three AK amputees fitted with IC sockets. (Normal alignment).

obtained in this study for the normal subjects, and 50 Nm obtained by Goh 1982 for the sound side of AK amputees), ranged from 0 to 83 Nm. During the second half of swing phase, the hip moment of the sound leg of most amputees was larger than that of the normal subjects. This can be related to the higher mass of the sound leg in comparison to that of normal subjects, as the sound leg of an amputee is always well built.

The M/L hip moment of the sound leg was always adducting the femur in order to maintain the body's M/L stability during the stance phase. This moment is comparable to that of the normal subjects in pattern and magnitude.

In the transverse plane, the hip moment of the sound leg of the majority of patients showed a tendency to rotate the thigh externally from heel strike until about mid-stance. Thereafter, the stance phase was dominated by a moment which tended to rotate the thigh internally in order to provide the stability of the leg in the transverse plane. This moment was also comparable in pattern and magnitude to that of the normal subjects.

At the prosthetic hip, most patients showed a consistent pattern of moment in the A/P plane, and was comparable to that of the sound leg. However, the magnitude of this moment was smaller than that of the sound leg especially at the heel strike and during the swing phase. At heel strike, the hip moment of the prosthetic leg was approximately 50% of that of the sound leg. This can be related to the fact that the vertical ground reaction force of the prosthetic leg is smaller than that of the sound leg. Also, at heel strike, the flexion of the femoral remnant (20 degrees) is smaller than that of the femur of the sound leg (24.5 degrees). During swing phase, the prosthetic hip moment was also smaller (about 50%) than that of the sound hip. This can be related to the differences between the mass of the prosthetic leg and that of the sound leg.

In the transverse plane, the hip moment of the prosthetic side tended to rotate the thigh internally throughout the stance phase in order to maintain stability in that plane. In general the magnitude of this moment (average of 8.2

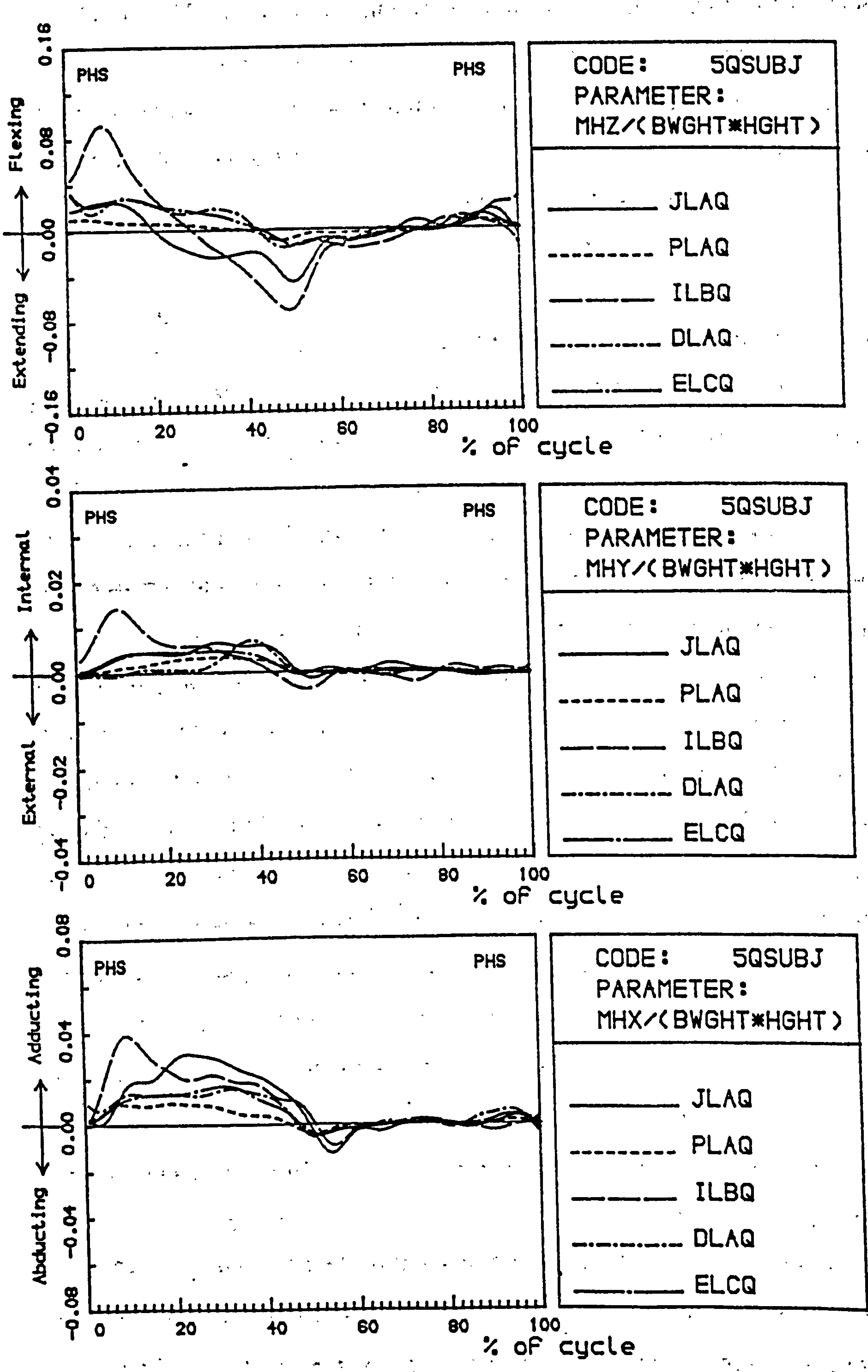


Figure 6.31 Hip joint moments on the prosthetic leg with time for five AK amputees fitted with quadrilateral sockets. (Normal alignment).

Nm) was slightly smaller than that of the sound leg (9 Nm). This can be attributed to lack of supportive muscles in the prosthetic leg, and to the fact that the sound step length was shorter than that of the prosthetic leg.

In the M/L plane, all subjects exhibited an adducting moment with an average value of 39 Nm (50 Nm for the sound leg and for the normal subjects), ranging from 15.5 to 93 Nm. No abducting moment was found following heel strike at the prosthetic hip because the weak adductors cannot compensate for an abduction moment. However, in the literature, this abducting moment was obtained by some researchers (Goh 1982), and was not obtained by others (Yang Lang 1988), and it is believed that this moment may appear at the prosthetic leg, and its magnitude increases with increasing the stability of the prosthesis as presented by Zahedi et al (1988).

It should be mentioned that no noticeable differences were found between the pattern of the hip moment of subjects fitted with IC and quad sockets. However, the M/L hip moment of the prosthetic leg was generally larger for the subjects fitted with IC than that of subjects fitted with quad sockets. This could be related to the fact that the subjects fitted with the IC sockets had adducted sockets whereas subjects fitted with quad sockets had abducted sockets (see table 6.3). The adducted socket position will reduce, or may reverse, the effect of the vertical ground reaction force in producing a hip abducting moment. i.e. the vertical ground reaction force may produce a hip adducting moment if the socket is adducted, depending on the foot position relative to the hip joint centre.

6.7 Effect of Alignment Changes on the Kinetic and Kinematic parameters of Above-Knee Amputees

In general, no noticeable differences were found between the gait parameters of subjects fitted with quadrilateral (quad) or ischial containment (IC) sockets, also, no noticeable differences were found between the effect of alignment changes on the gait parameters of subjects fitted with either of the above two type of sockets. Therefore, the data of subjects fitted with the two

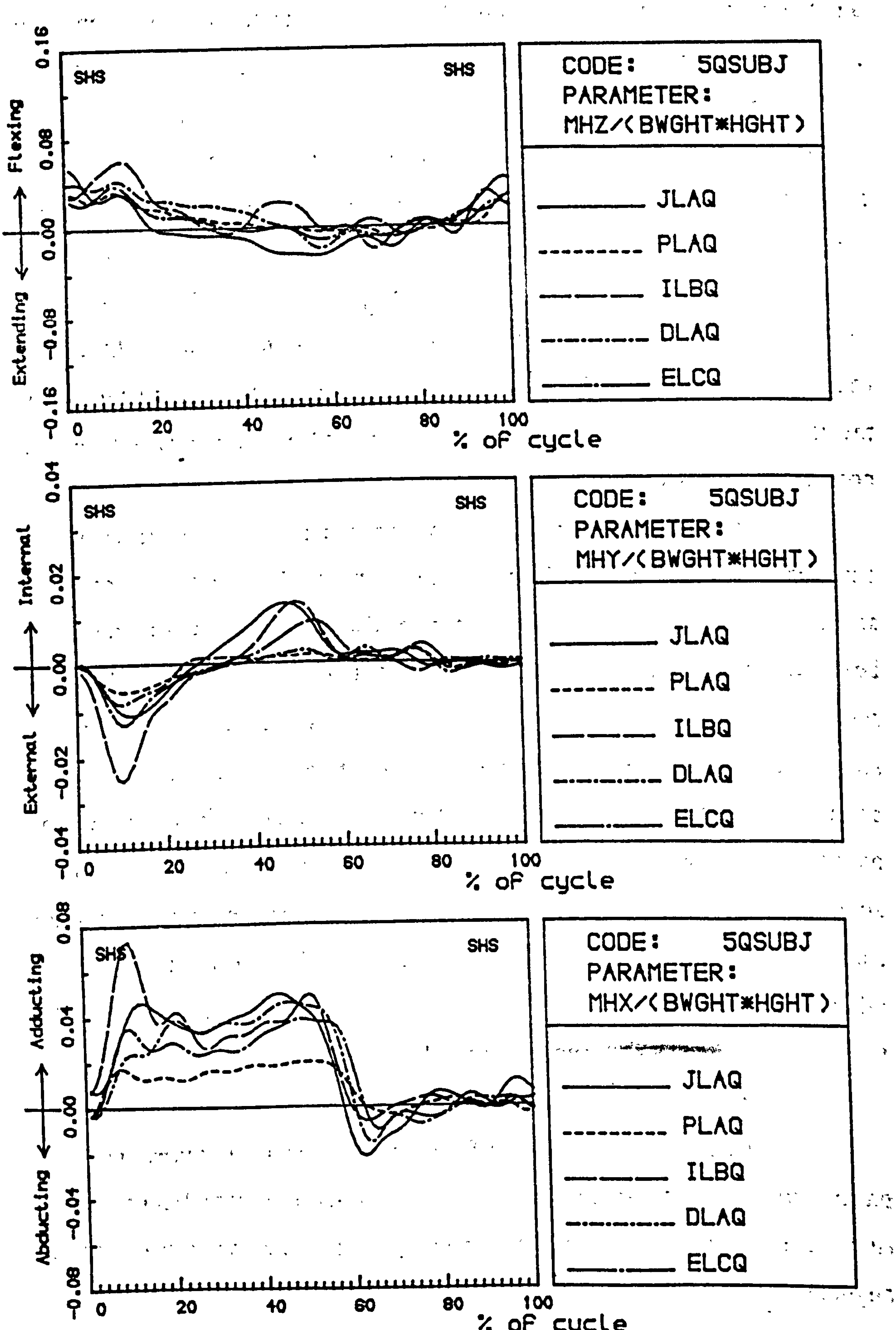


Figure 6.32 Hip joint moments on the sound leg with time for five AK amputees fitted with quadrilateral sockets. (Normal alignment).

socket types are discussed together to study the effects of alignment changes on the gait parameters of above-knee amputees, and differences between these effects on the gait parameters of subjects fitted with quad and IC sockets will be discussed when found.

In order to study the effect of alignment changes on the gait parameters, the curves representing gait parameters of each deliberate mal-alignment, were compared with their respective curves corresponding to the normal alignment. Since none of the current available statistical tests can be applied to the results (personal communication, and consultation with the Statistics Department), and applying an unestablished statistical test may produce misleading conclusion, the comparison was made by calculating the percentage difference between the normal alignment and the mal-alignment curves.

Many parameters were considered in order to assess the effect of alignment changes on the kinetic and kinematic parameters of the AK amputees, and therefore many graphs were produced. These graphs cannot all be presented here for reasons of space limitation, therefore, samples of the results are included in this section to enhance and clarify the discussion, and all the noticeable effects of the alignment changes on the different gait parameters are also tabulated and presented in this section. The complete set of results for all patients, and the set of computer programs which were developed and used for the calculation of the gait parameters, can be obtained from the author or from Mr Solomonidis of Strathclyde University.

6.7.1 Effect of Alignment Changes on the Angular Displacements of the Trunk

6.7.1.1 Effect of the Foot Changes

Table 6.15 shows the results for the effect of alignment changes on the trunk angular displacements for all subjects, and figures 6.33 and 6.34 show the effect of the foot alignment changes on the trunk angular displacements of two representative subjects.

It was found that the trunk angular displacement in the AP plane was

Table 6.15 Effect of alignment changes on the angular displacements of the trunk.

Trunk AP Tilt (Flexion)	Torso Rotation	Trunk ML tilt	Trunk AP Tilt (Flexion)	Torso Rotation	Trunk ML tilt
Effect of the Foot Changes					
6 Degrees Change Toward Dorsiflexion			6 Degrees Change Toward Plantarflexion		
+ in 8 amp (1.5 - 10 deg) - in 2 amp. (slightly) = in 1 amputee	Inconsistent variations by a range from 0 to 5 degrees	= during stance phase. - slightly in some subjects during PSP.	- in 5 amp. (1.5 - 3 deg) + in 4 amp. (3.3 -7.2 deg) = in 2 amputees	Inconsistent variations by a range from 0 to 5 degrees	= during stance phase. + slightly in some subjects during PSP.
Effect of Socket Changes					
6 Degrees Change Toward Flexion			6 Degrees Change Toward Extension		
+ in 8 amp. by a range from 2 to 4.7 degrees. = in 3 amputees.	Inconsistent trunk changes.	Inconsistent trunk changes.	- slightly in 5 amputees. = in 6 amputees.	Inconsistent trunk changes.	Inconsistent trunk changes.
Effect of the Knee Shift					
1.8 cm Forward Shift			1.8 cm Backward Shift		
+ in 9 amp. by a range from 1.7 to 10.6 deg. - slightly in 2 amputees.	Inconsistent changes range from 4 to 10 degrees.	- slightly in 6 amp during PSP + slightly in 2 amp. = in 3 amputees.	- in 6 amp. by (1-3.3 deg) + in 3 amp. by max. 3 deg = in 2 amputees.	Inconsistent changes range from 4 to 10 degrees.	+ slight in 5 amp. during PSP = in 2 amputees. - slightly in 4 amp.

+ Increased relative to the normal value. = No Change amp. amputee
- Decreased relative to the normal value. PSP Prosthetic Swing Phase.

affected more by foot dorsiflexion than plantarflexion changes. This is related to the fact that the patient may lose knee stability when dorsiflexing the foot, and therefore, he/she has to change the trunk angular position in order to maintain stability. Throughout stance phase of the prosthetic leg, when the foot was dorsiflexed from the normal position, eight subjects out of eleven showed an increase in the trunk flexion angle, one subject (JLAIAF) showed no change and the two remaining subjects (PLAIAF, LRBQAF) showed a decrease in the trunk flexion position. The amount of change which was exhibited by the eight subjects varied among them from 1.5 degrees in subject PLAQAF to 10 degrees in subject MRCQAF when the prosthetic foot was dorsiflexed by 6 degrees (in increments of 3 degrees). However, the changes had a consistent trend among subjects, therefore, they are meaningful and can be explained as follows:

In the following discussion and throughout the rest of this chapter, the vector of the ground reaction force is considered to be acting on the foot centre of pressure and passing through the body CG, ignoring the offset between the force vector and the body CG, which can be caused by the moment resulting from the angular acceleration of the body CG.

Comparing the two alignments (i.e. normal and the misaligned prosthesis) at the same values of foot-floor angles, and as seen in figure 6.35, for the period from heel strike until mid-stance and figure 6.36 for the period from mid-stance until toe off, dorsiflexing the foot will shift the knee joint centre forward from its original position K to a new position Kd, and the hip joint centre will also be shifted from H to Hd. If the trunk angular position did not change, the body centre of gravity would move from CG to CG1, this would change the vector of the ground reaction force from R which passes ahead of the KJC (point K), to R1 which may pass ahead, on or behind the new position of the KJC (point Kd) depending on the original configuration of the prosthesis and on the amount of change when dorsiflexing the foot. Thus, if the new force vector R1 passes behind the KJC (point Kd), the prosthesis

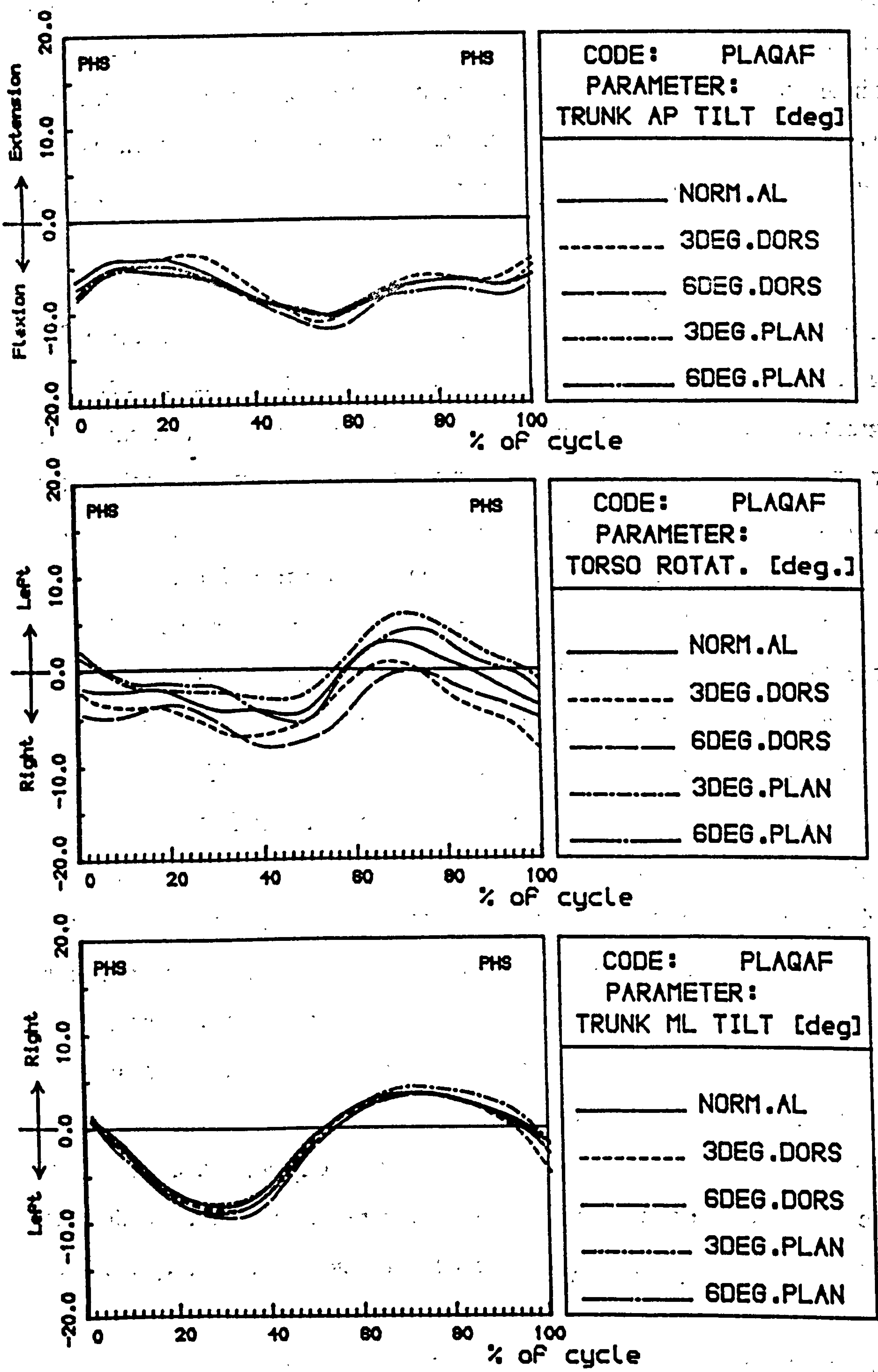


Figure 6.33 Effect of foot alignment changes on the variation with time of the AP and ML angular displacements of the trunk, and on the transverse rotation of the torso for an AK amputee fitted with a quadrilateral socket.

will be unstable and therefore, the patient will flex the trunk forward by an angle θd, shifting the body centre of gravity from CG1 to CGd, and changing the vector of the ground reaction force from R1 to Rd which passes ahead of the KJC (point Kd) and maintains the stability of the prosthesis. The foregoing explanation is seen to be true on the eight subjects who showed an increase in the trunk flexion angle when the foot was dorsiflexed from the original position. If the ground reaction force vector R1 passed ahead of the KJC (point Kd), the trunk angular position may not need to be changed because stability of the prosthesis exists and the patient will not be forced to compensate for the lost stability. However, the patient still has the option of changing the trunk position toward flexion, or even extension as long as the stability is maintained. This explanation is applicable for the patient who did not change his trunk position, and for the two patients who extended their trunk position when the foot was dorsiflexed from the normal position.

It should be mentioned, that the patient can compensate for the lost stability at the prosthetic knee when the foot was dorsiflexed, by either flexing the trunk forwards, or by maintaining the original position of the trunk and exerting force in the hip extensors. This will produce a positive force at the heel in the direction of progression. Therefore, the fore-and-aft force (FPX), which is the sum of the decelerating force (negative force) and the hip extensors force (positive force), would be smaller than that when the foot was in normal alignment, so that the inclination of the vector of the ground reaction force will be changed to pass ahead of the KJC (point Kd) and maintain the stability of the prosthetic knee. The patient may also strike the floor in a manner that may produce a positive FPX acting at the heel and contribute in maintaining the knee stability and accelerating the body forwards. However, it is evident that the two foregoing possibilities of compensation for the lost stability were not adopted by the subjects of this study, and had they done so, trunk forward flexion would not be obtained in eight subjects out of eleven as discussed above.

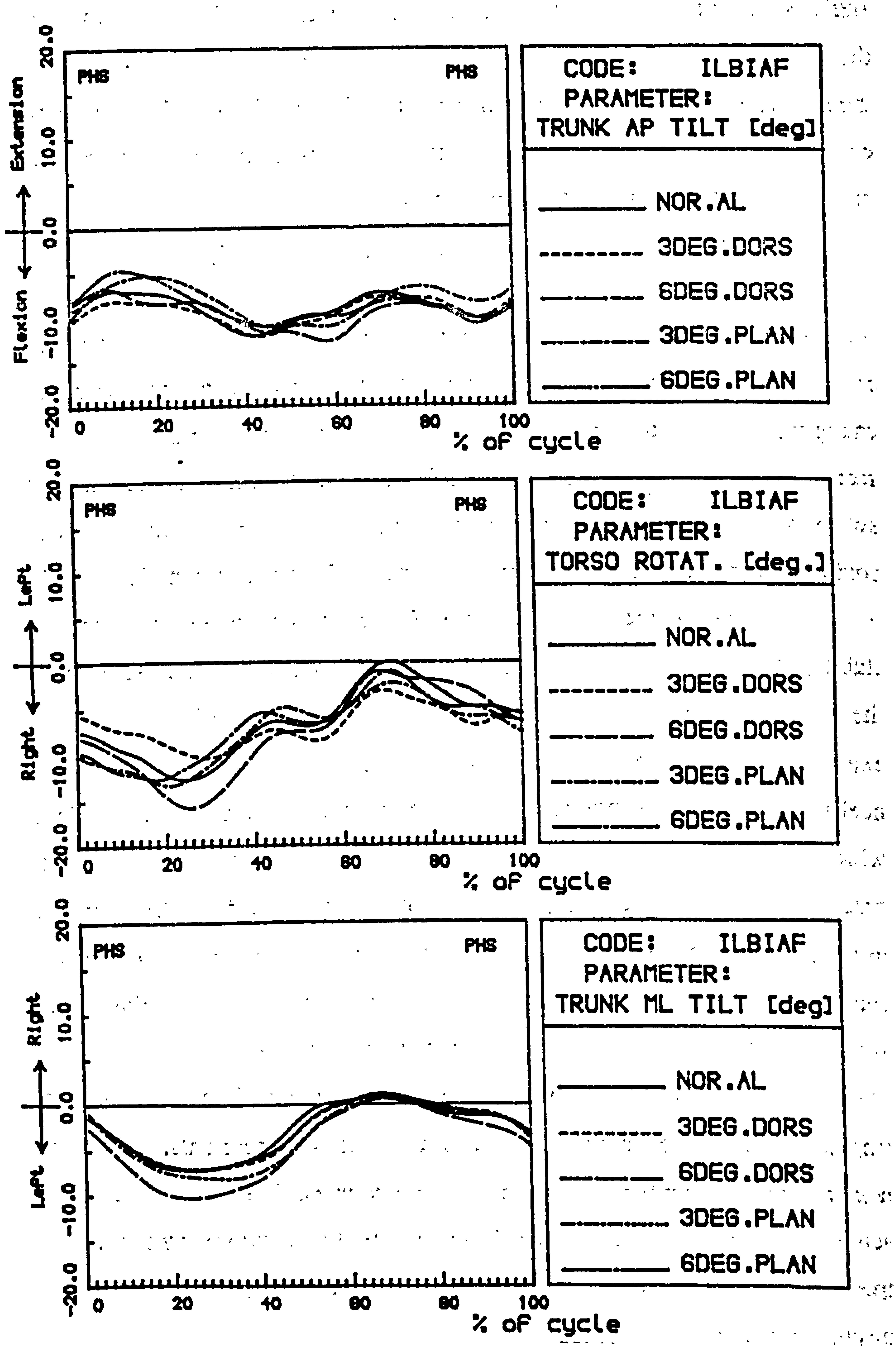


Figure 6.34 Effect of foot alignment changes on the variation with time of the AP and ML angular displacements of the trunk, and on the transverse rotation of the torso for an AK amputee fitted with an IC socket.

When the foot was plantarflexed by 6 degrees (in increments of 3 degrees) from the normal position, five subjects (MRCQAF, JLAQAF, JLAIAF, ILBIAF, TRAQAF) extended the trunk from its normal position by a range from 1.5 to 3 degrees, two subjects (PLAIAF, PLAQAF) showed no change in the angular attitude of the trunk, and the four remaining subjects flexed the trunk from its normal position by a range from 3.3 to 7.2 degrees. These changes can be explained as follows, (see fig. 6.35 and 6.36):

Plantarflexing the foot from the original position will shift the KJC from K to Kp and the HJC from H to Hp (at the same values of foot-floor angles). If the trunk angular position did not change, the body centre of gravity will move from CG to CG2 and the vector of the ground reaction force will change from R to R2. This will subject the hip joint to a high flexing moment at heel strike, and will render the knee joint over stabilised causing difficulties in initiating knee flexion prior to swing phase. Therefore, the patient would extend the trunk backwards by an angle θP shifting the body centre of gravity from CG2 to CGp, and changing the vector of the ground reaction force from R2 to Rp in order to reduce the hip flexing moment at heel strike, and ease the initiation of knee flexion prior to the swing phase. This is the case with the five subjects who extended their trunks when the foot was plantarflexed. Since plantarflexing the foot does not reduce the stability of the prosthesis, the patient still has the option of flexing the trunk (the case of four subjects), or maintaining the normal trunk position (the case of two subjects), as long as the patient is able to support the large hip flexing moment at heel strike and to overcome the highly stabilised knee prior to swing phase.

In the transverse plane, all subjects showed variations in the torso angular position over the gait cycle with the foot alignment changes. These variations varied among subjects from 0 to 5 degrees. However, the changes were inconsistent and no trend of change was found, therefore, no conclusion can be stated regarding the effect of foot alignment changes on the torso angular rotation in the transverse plane.

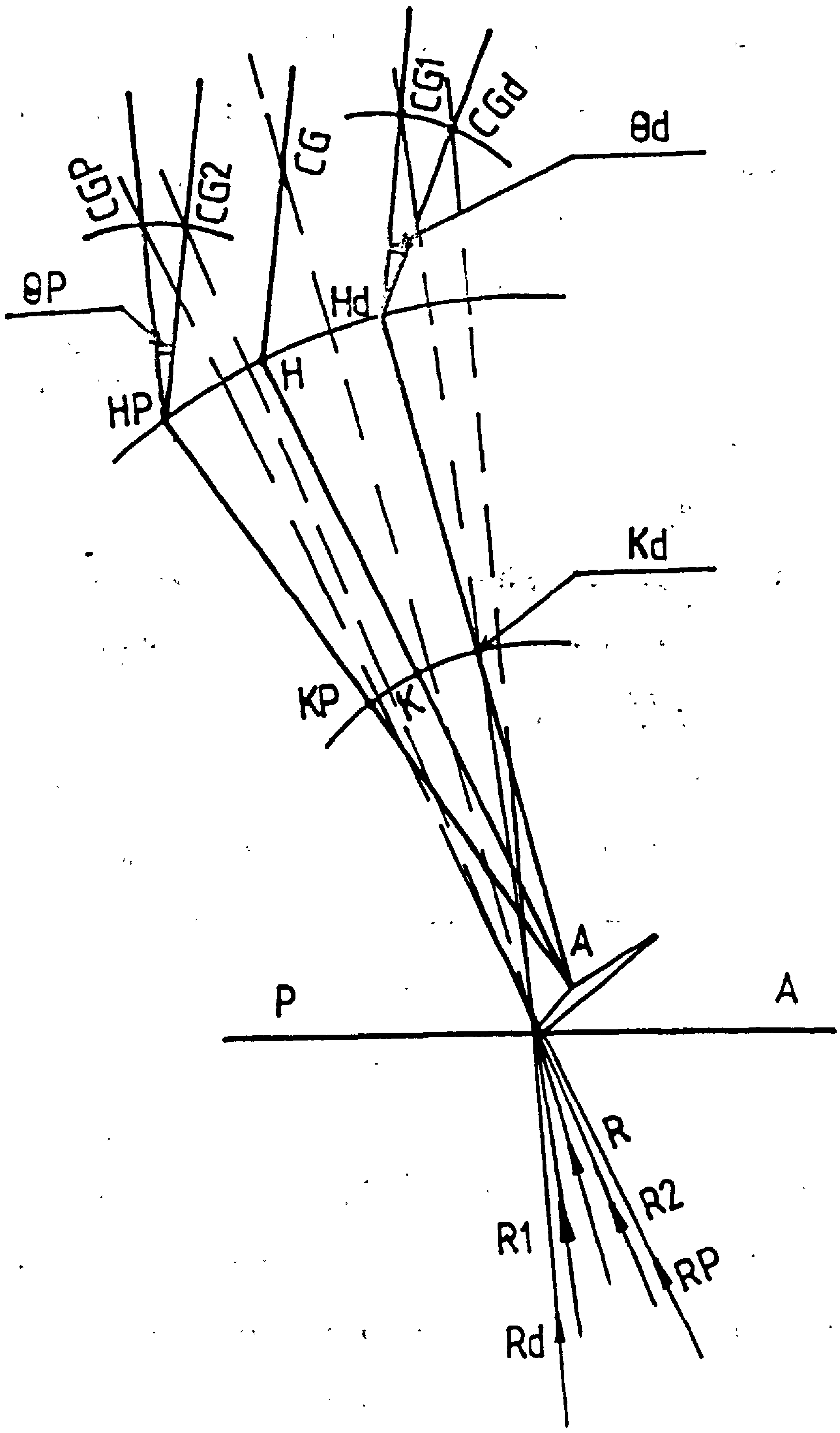


Figure 6.35 Effect of foot alignment changes on the orientation of the trunk and the segments of the prosthesis. The orientation of the ground reaction force vector with respect to the joint centres of the limb is also shown. **(from heel strike until mid-stance of the prosthetic leg).**

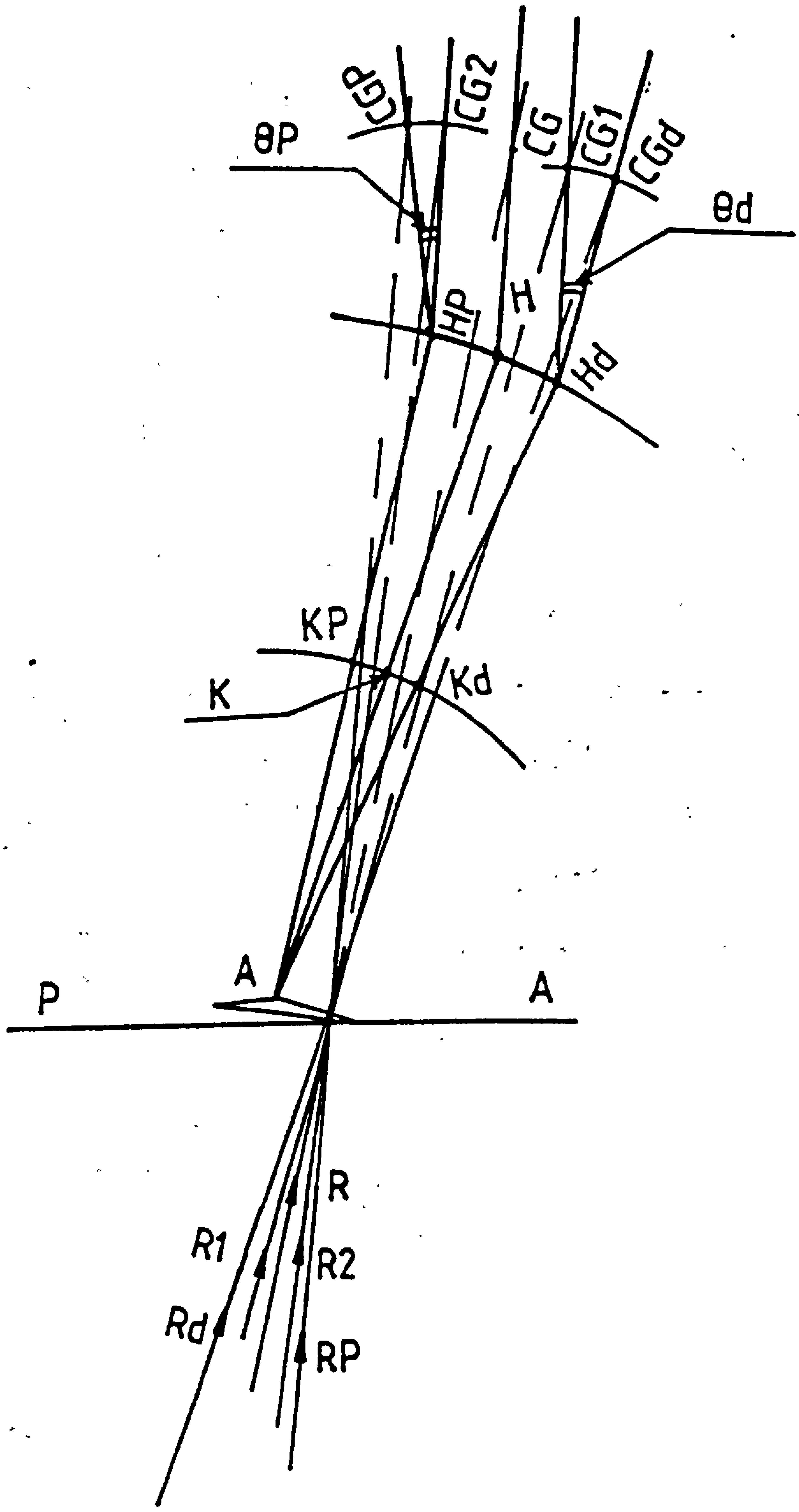


Figure 6.36 Effect of foot alignment changes on the orientation of the trunk and the segments of the prosthesis. The orientation of the ground reaction force vector with respect to the joint centres of the limb is also shown. (from mid-stance until toe off of the prosthetic leg).

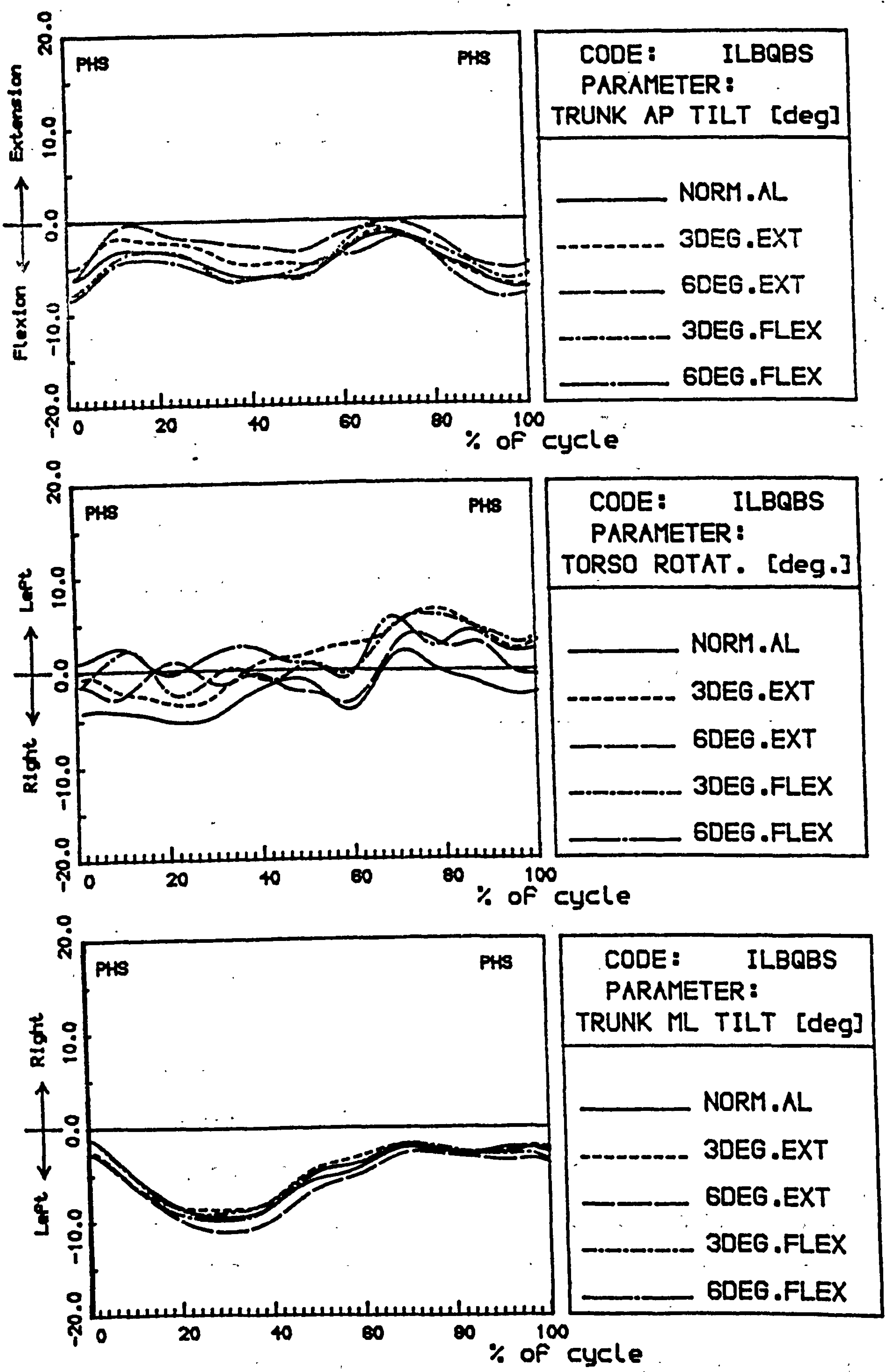


Figure 6.37 Effect of socket alignment changes on the variation with time of the AP and ML angular displacements of the trunk, and on the transverse rotation of the torso for an AK amputee fitted with a quadrilateral socket.

In the medio-lateral plane, in general the pattern and magnitude of the trunk ML angular tilt did not change with foot alignment changes. This can be related to the fact that changing the AP angular inclination of the foot is not supposed to change the ML angular configuration of the prosthesis. However, it was found that some patients showed a slight trend to increase/decrease the trunk ML tilt, towards the sound leg during swing phase as the prosthetic foot was plantar/dorsiflexed from the normal position. This trend of change is clearly seen in subject TRAQAF and is related to the fact that plantar/dorsiflexing the foot, might increase/decrease the hip-toe distance. Therefore, these patients increased the trunk ML tilt to the sound leg in order to provide ground clearance for the swinging foot when the foot changes were toward plantarflexion. It should be noted that the trunk ML tilt changes were more noticeable when plantarflexing than those arising when dorsiflexing the foot, since dorsiflexing the foot does not extend the hip-toe length and therefore, the subject will not be forced to change the attitude of the trunk.

6.7.1.2 Effect of Socket Alignment Changes

The effect of socket alignment changes on the angular displacements of the trunk is shown in table 6.15, and figures 6.37 and 6.38 show this effect on two representative subjects. In general, since the socket alignment changes were confined to the AP plane, the trunk AP displacements were more affected by these changes than the trunk ML and torso transverse displacements. As the socket was flexed by 6 degrees (in increments of 3 degrees) from its normal position, three subjects (PLAIBS, DLAQBS, MRCQBS) showed no change in the trunk flexion position, while the eight remaining subjects exhibited an increase in the trunk flexion relative to its normal flexion angle by a range from 2 to 4.7 degrees. These trunk changes were more pronounced when the socket was changed by 6 degrees than that by 3 degrees, and were consistent during stance phase only. The reason for these changes in the trunk flexion angle is explained as follows (see fig. 6.39):

In the original position of the prosthesis (normal alignment), as the

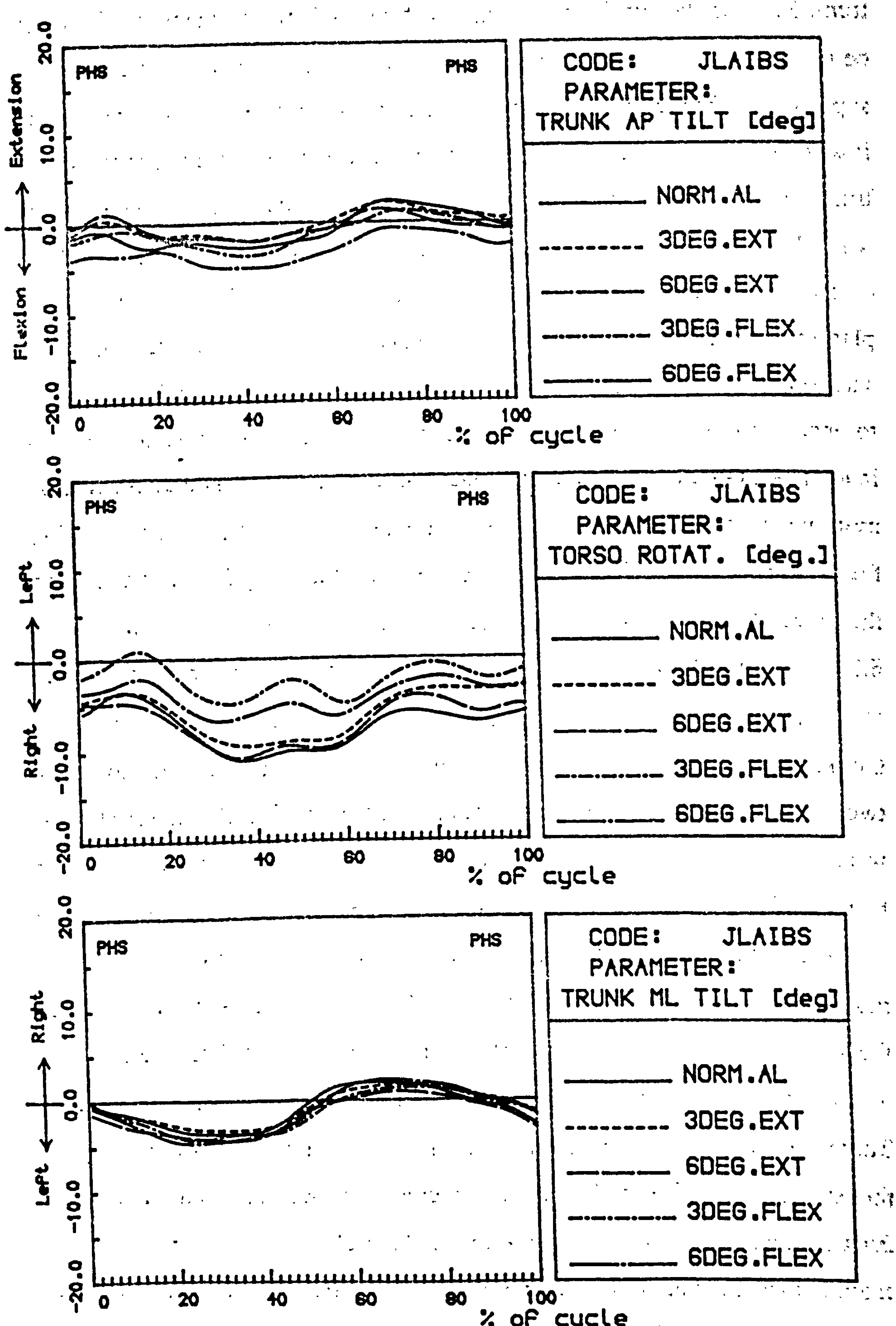


Figure 6.38 Effect of socket alignment changes on the variation with time of the AP and ML angular displacements of the trunk, and on the transverse rotation of the torso for an AK amputee fitted with an IC socket.

subject loads the prosthesis, the ground reaction force vector R will act on the foot centre of pressure, passing ahead of the KJC (K) to the body centre of gravity (CG). Flexing the socket from its original position will force the HJC to be shifted backward from H to HF, and the body CG will also be shifted from CG to CG1 (considering the foot held at the same angle with the floor). This will change the position of the ground reaction force vector from R to R1, passing through CG1, and may pass ahead, on, or behind the KJC, depending on the original position of the KJC and on the amount of the socket flexion change. If R1 passes ahead or on the KJC, the knee is still stable and no changes in the attitude of the trunk will be necessary. This applies to the three subjects who showed no trunk angular changes with the socket flexion change. If R1 passes behind the KJC, the knee joint (i.e. the prosthetic knee) will be unstable and therefore, the subject will have to flex the trunk forwards shifting the body centre of gravity from CG1 to CGF, and changing the ground reaction force vector from R1 to RF, in order to maintain the stability of the prosthesis. This applies to the eight subjects who changed the trunk angular position when the socket was flexed from its normal position. The variations in the trunk angular changes between subjects (ranged from 2 to 4.7 degrees) may correspond to the degree of knee instability caused by flexing the socket, and to the amount of CG shift (the CG1-CGF distance) which was achieved by the subject.

Extending the socket from its original position will force the HJC to move from H to HE and the body centre of gravity will be shifted from CG to CG2 (see fig. 6.39). This will change the ground reaction force vector from R to R2 passing further ahead of the KJC which may well be over stabilised. Therefore, the subject will extend the trunk backward by an angle θE shifting the body CG from CG2 to CGE, and changing the ground reaction force vector from R2 to RE passing close (but ahead) to the KJC in order to keep the body erect, reducing the hip flexing moment during the first half of stance phase, and facilitating the initiation of knee flexion prior to swing phase (see fig. 6.40).

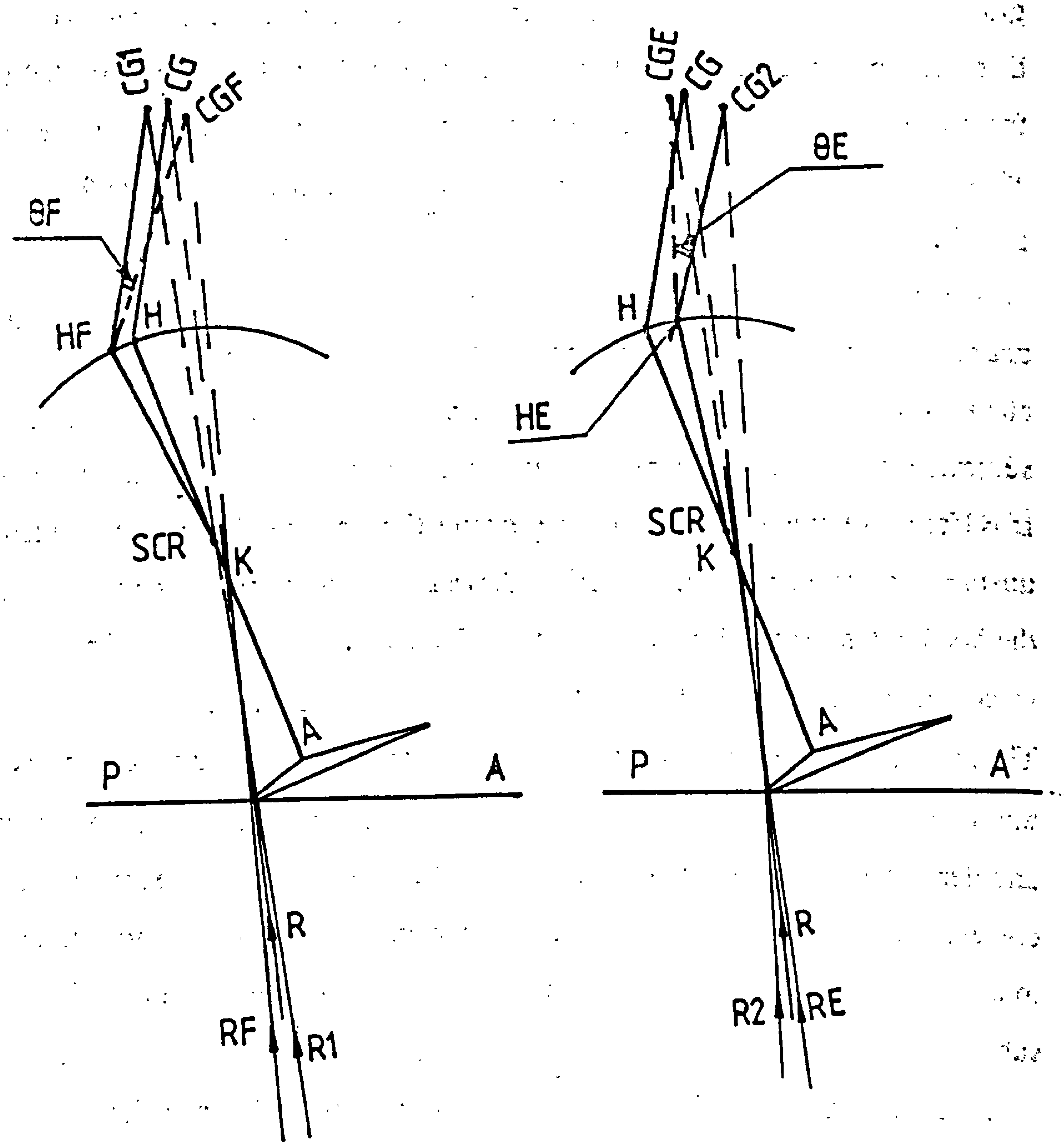


Figure 6.39 Effect of socket alignment changes on the orientation of the trunk and the segments of the prosthesis. The orientation of the ground reaction force vector with respect to the joint centres of the limb is also shown. **(from heel strike until mid-stance of the prosthetic leg).**

The foregoing explanation is applicable on five subjects (TRAQBS, ILBQBS, JLAIBS, ELCQBS, DLAQBS) of this study who showed a trend of extending their trunks backward when the socket was extended from the normal position. The six remaining subjects showed no change in the trunk AP position when the socket was extended from the normal position. This may be attributed to the fact that the prosthesis did not lose stability when extending the socket, and therefore, the patient would not choose to change the attitude of the trunk as long as the stability is maintained and the subject is physically able to operate the prosthesis. Another reason for the unchanged trunk angular position in six subjects, when the socket was extended, is that extending the trunk by an angle θE will increase the hip extending moment at push off (see fig. 6.40). This may force the posterior brim of the extended socket to dig into the gluteus maximus muscle or under the ischial tuberosity of the patient causing pain and discomfort (as reported by subjects PLAIBS and ILBIBS).

In the transverse (TORSO ROTATION) and medio-lateral (TRUNK ML TILT) planes, all subjects showed inconsistent changes with the socket alignment changes, in that, the trunk changes resulting from flexing the socket did not have an opposite trend to those which resulted from extending the socket, and the trunk changes which resulted from changing the socket angle by 3 and 6 degrees were not in the same direction. This is related to the fact that changing the socket alignment in the AP plane will not change the alignment of the prosthesis in the transverse and ML planes, and its effect will not be seen in the torso rotation and ML tilt of the trunk displacements unless the AP alignment changes are large. Thus, no conclusion can be drawn regarding the torso angular changes in the transverse and the trunk changes in the ML planes with the socket alignment change. This finding does not agree with Yang Lang (1988) who stated that extending the socket increases the trunk lateral tilt toward the sound leg during prosthetic swing phase. However, Yang Lang tested two subjects only whereas eleven patients were tested in this study.

6.7.1.3 Effect of the Knee Shifts

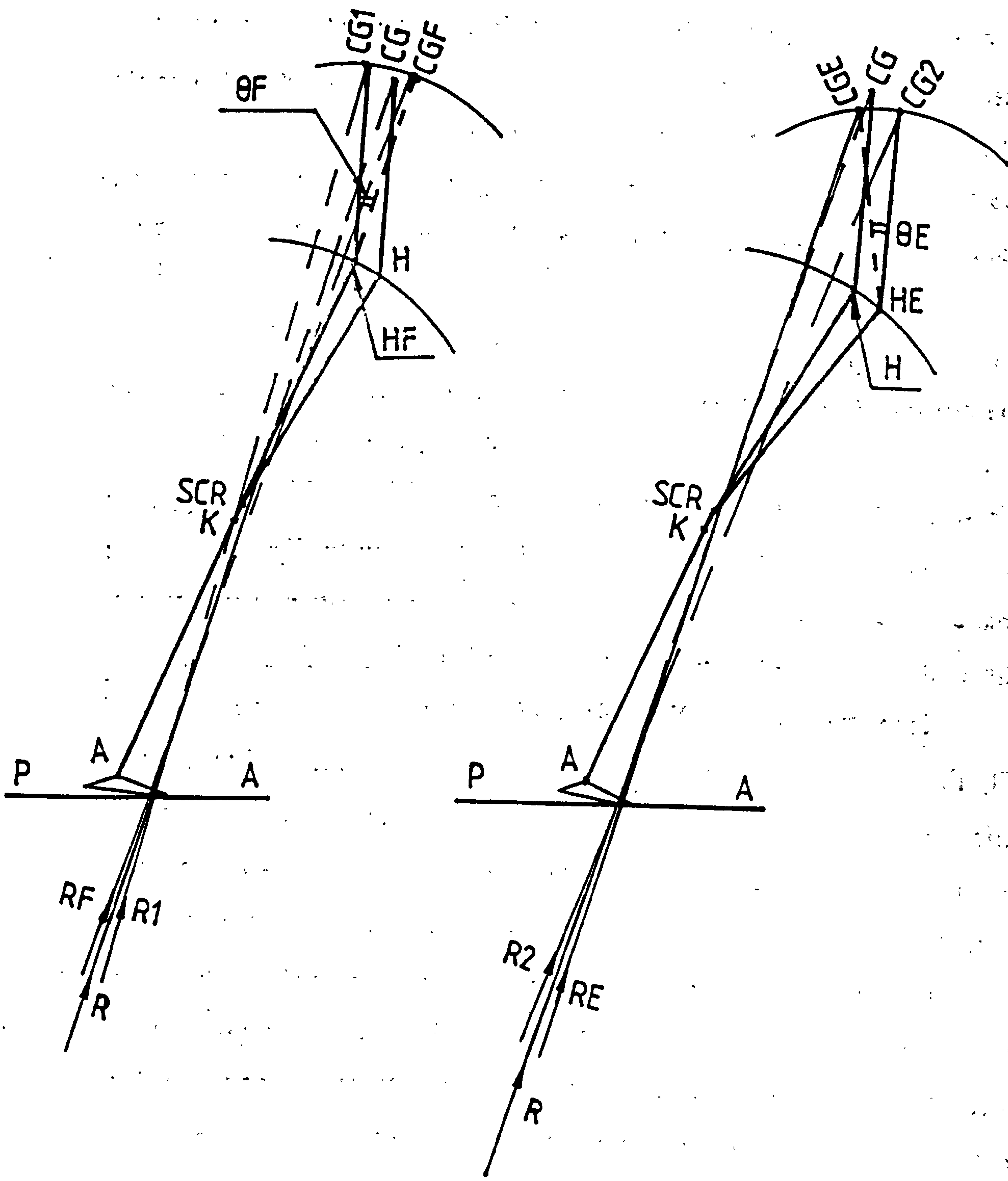


Figure 6.40 Effect of socket alignment changes on the orientation of the trunk and the segments of the prosthesis. The orientation of the ground reaction force vector with respect to the joint centres of the limb is also shown. (from mid-stance until toe off of the prosthetic leg).

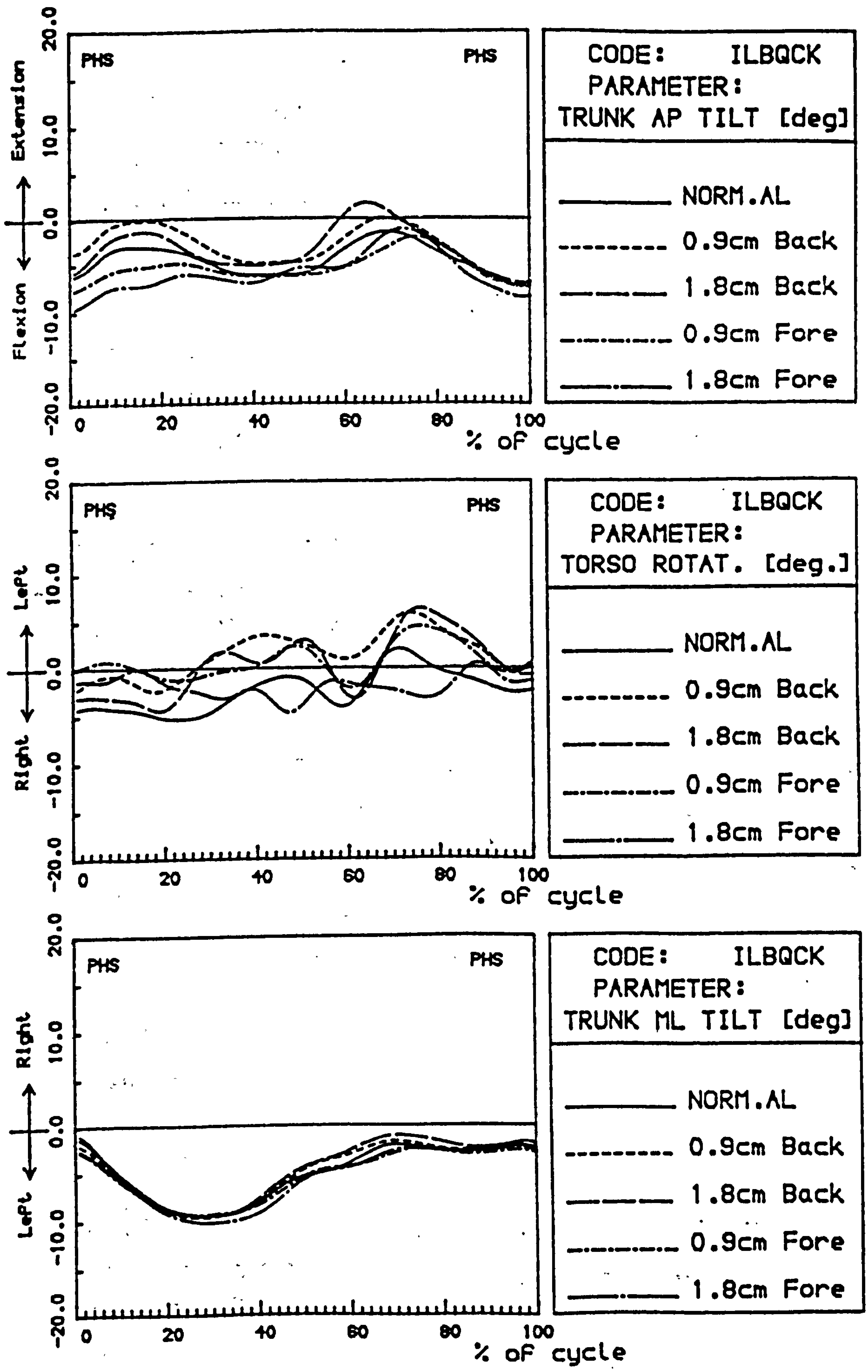


Figure 6.41 Effect of knee shifts on the variation with time of the AP and ML angular displacements of the trunk, and on the transverse rotation of the torso for an AK amputee fitted with a quadrilateral socket.

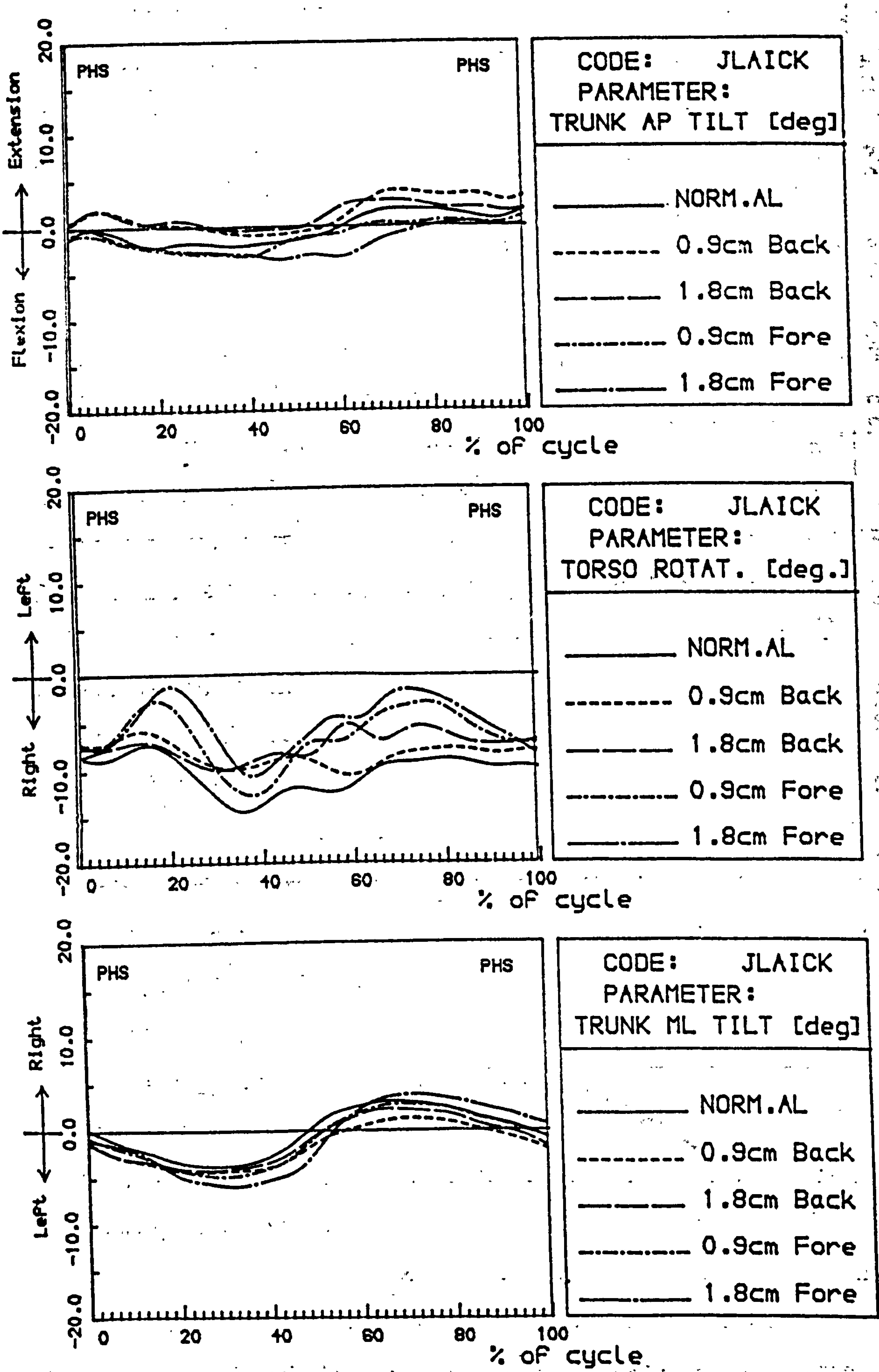


Figure 6.42 Effect of knee shifts on the variation with time of the AP and ML angular displacements of the trunk, and on the transverse rotation of the torso for an AK amputee fitted with an IC socket.

The knee alignment changes were achieved by shifting the KJC position forwards or backwards relative to the hip-ankle line. It was found that knee alignment changes resulted in noticeable changes to the trunk angular displacements, and these trunk changes were more pronounced and consistent than those which resulted from the foot or socket alignment changes. As discussed above (section 6.3), knee forward shift was achieved by dorsiflexing the foot and flexing the socket simultaneously by an equal angle, and similarly, knee backward shift was achieved by plantarflexing the foot and extending the socket simultaneously by the same angle. This procedure of changing the alignment results in shifts in the knee forward/backward and causes no change in the angular orientations of the socket and foot relative to each other.

Table 6.15 shows the effect of knee alignment changes on the angular displacements of the trunk, and figures 6.41 and 6.42 show this effect on two representative subjects. In the AP plane (TRUNK AP TILT), shifting the KJC forwards by 1.8 cm (in increments of 0.9 cm) from the normal position, resulted in a noticeable and consistent change in the trunk AP angular displacement. The trunk flexion angle increased in nine subjects out of eleven and slightly decreased in the two remaining subjects (PLAICK, LRBQCK) in comparison to that of the normal alignment. The increase in the trunk flexion angle ranged from 1.7 to 10.6 degrees, and was generally more pronounced and consistent during stance than swing phase. This behaviour of the trunk was undertaken by the patient in order to maintain the stability of the prosthesis which was affected by shifting the KJC forwards relative to the hip-ankle line, and it can be explained using the kinetic and kinematic data as follows:

Referring to figures 6.43 and 6.44, shifting the KJC forwards relative to the hip-ankle line, will change the KJC from its normal position K to a new position KF. If the trunk angular position did not change, the body centre of gravity will move from CG to CG1, and therefore the vector R1 of the ground reaction force will pass behind (possibly on or ahead) the KJC (point KF) and the prosthesis will be unstable. Therefore, the patient will flex his/her trunk

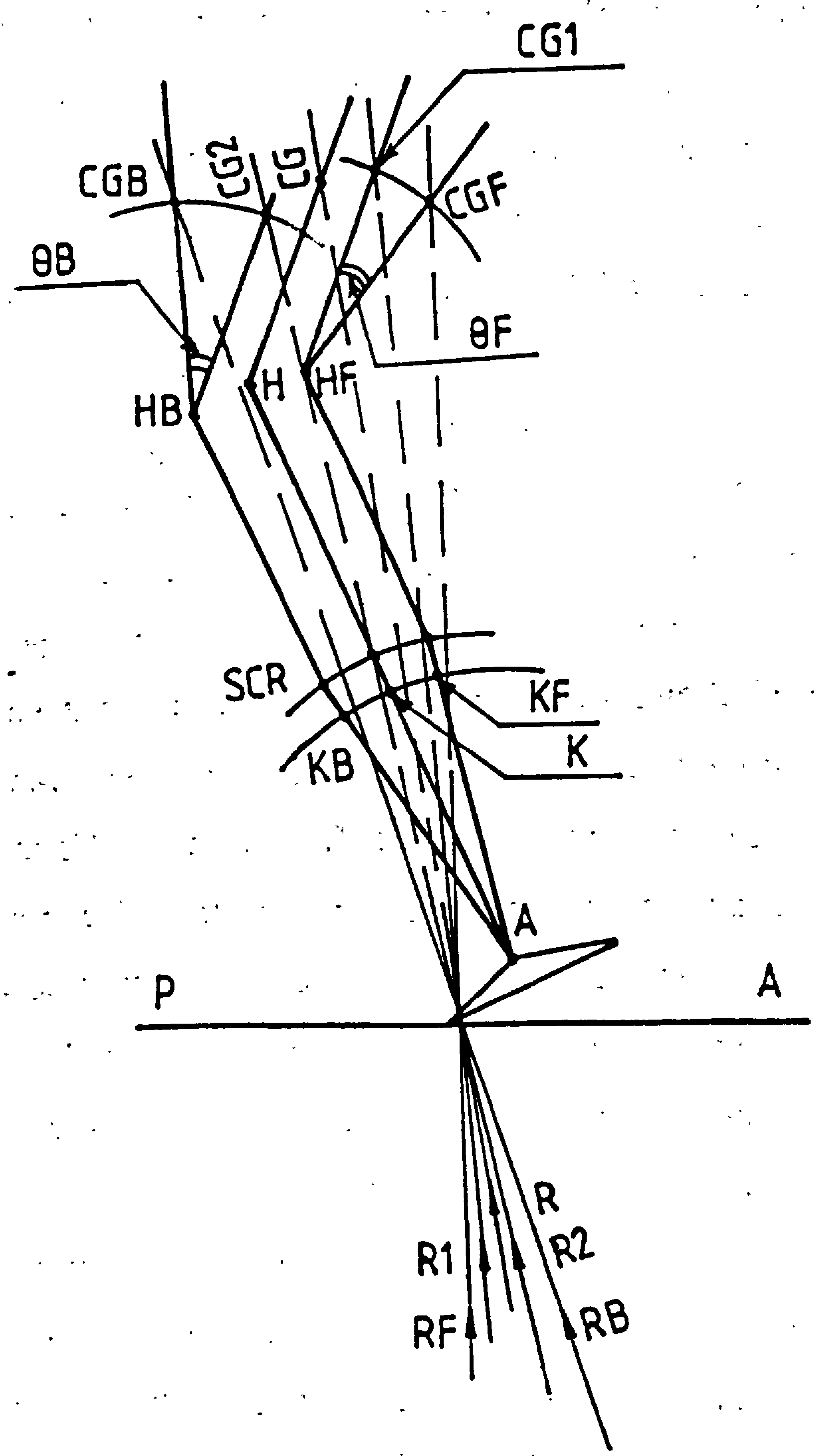


Figure 6.43 Effect of knee shifts on the orientation of the trunk and the segments of the prosthesis. The orientation of the ground reaction force vector with respect to the joint centres of the limb is also shown. (from heel strike until mid-stance of the prosthetic leg).

forwards by an angle θF shifting the body CG from CG1 to CGF, and changing the ground reaction force vector from R1 to RF which passes ahead of the KJC (KF) and maintains the stability of the prosthesis. This explanation is applicable to the nine subjects who showed an increase in the trunk flexion angle associated with shifting the KJC forwards.

The two subjects who showed a decrease in the trunk flexion angle associated with a forward shift of the KJC, displayed a situation whereby the ground reaction force vector R1 passed ahead of the KJC (KF), and the patient still has the ability to flex or extend the trunk. This would be found if the trunk was normally held in a position of large flexion (subject LRBQCK), or if the original position of the KJC was far behind the load line, and therefore, the subject can tolerate some knee forward shift without endangering the stability of the knee joint and thus, without changing the position of his trunk. This would be the case of subject PLAICK as he extended his trunk only when the knee was shifted forward by 0.9 cm, and he had to flex his trunk slightly when the knee was shifted forwards by 1.8 cm. It should be noted that the magnitude of θF is affecting the AP moment of the hip joint, however, this will be discussed later with the effect of alignment changes on the hip joint moments.

As the KJC was shifted backwards by 1.8 cm (in increments of 0.9 cm) relative to the hip-ankle line, the trunk flexion angle increased by a maximum of 3 degrees in three subjects (PLAICK, ILBICK, DLAQCK), did not change in two subjects (PLAQCK, LRBQCK), and decreased during the gait cycle in the six remaining subjects by a range from 1 degree in subject JLAQCK to 3.3 degrees in subject ILBQCK. This trunk behaviour can be explained using figure 6.43 for the duration from heel strike until mid-stance, and figure 6.44 for the duration from mid-stance until toe off. Shifting the KJC backward relative to the hip-ankle line from K to KB will also shift the HJC from H to HB, and the body centre of gravity from CG to CG2. This will subject the hip joint to a large flexing moment at heel strike, and create difficulties in initiating

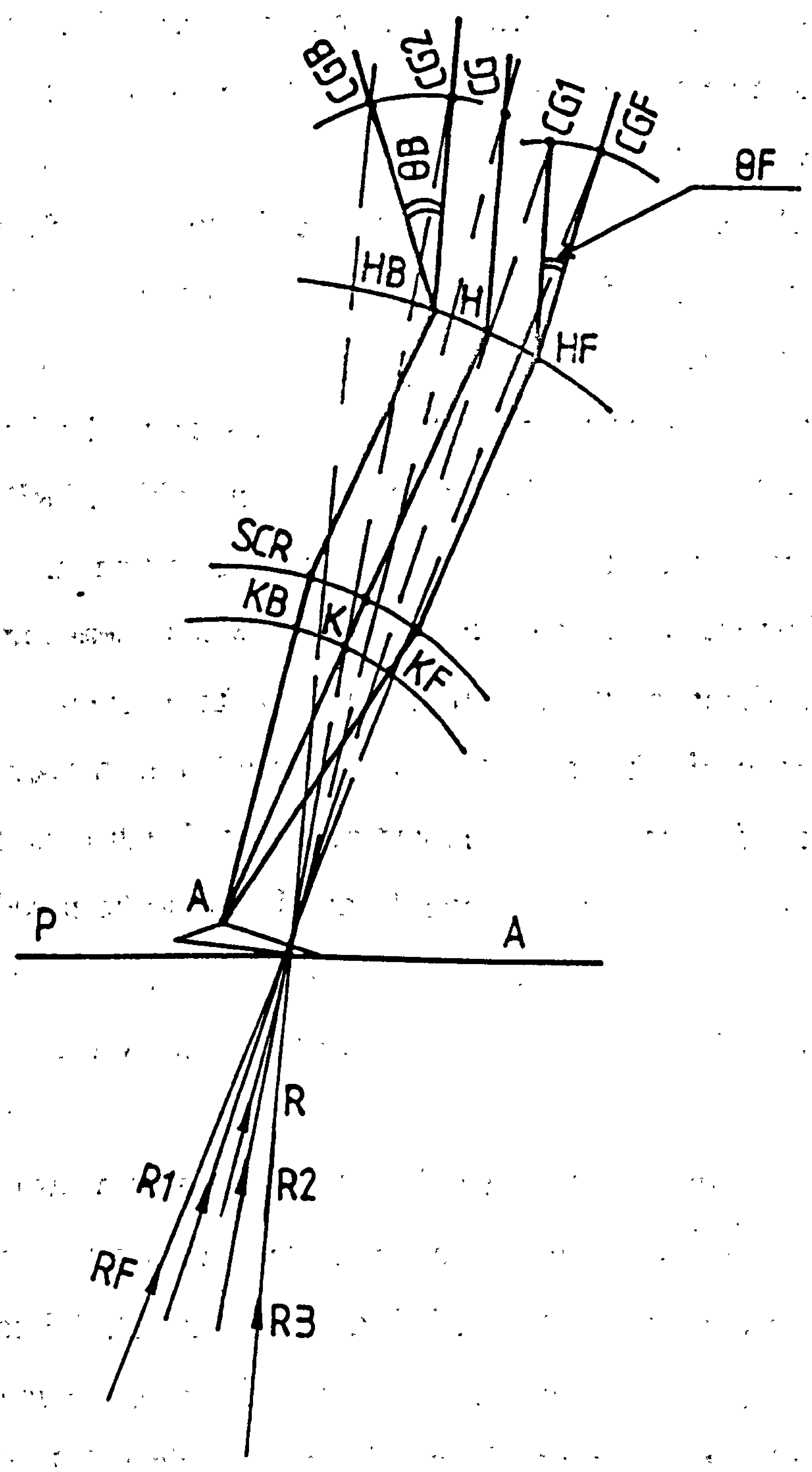


Figure 6.44 Effect of knee shifts on the orientation of the trunk and the segments of the prosthesis. The orientation of the ground reaction force vector with respect to the joint centres of the limb is also shown. (from mid-stance until toe off of the prosthetic leg).

knee flexion prior to swing phase as the load line (R2) will be passing far ahead of the KJC (point KB). Therefore, the patient would extend his/her trunk backwards by an angle θB shifting the body CG from CG2 to CGB and changing the ground reaction force vector from R2 to RB in order to reduce the AP hip moment at heel strike and facilitate the initiation of knee flexion prior to the swing phase. The foregoing explanation is applied to the six subjects who extended the trunk position with shifting the KJC backwards. As can be seen in figure 6.43, shifting the KJC and extending the trunk backwards will increase the fore-and-aft braking force (compare vector RB with R), thus, the subject will encounter a higher resistance when the knee is shifted backwards than that when the knee is in the normal position. This was described by subject PLAICK by saying that the prosthesis was pushing him backwards. Therefore, as the stability still maintained, the subject may flex the trunk forwards or maintain the original trunk flexion in order to avoid the high resistance during the first half of stance phase. This explanation is applied to the subjects who flexed or did not change the trunk position with shifting the KJC backwards. This explanation does not include subject DLAQCK as he was supplied with a uniaxial foot, and the knee backwards shift was compensated by a large change in the foot dorsiflexion angle. Thus, this subject should not be counted with those who flexed the trunk with shifting the KJC backwards, i.e only two subjects (PLAICK and ILBICK) can be counted in flexing the trunk with shifting the KJC backwards. It should be mentioned, that extending the trunk when shifting the KJC backwards may force the posterior brim of the socket to dig into the gluteus maximus muscle giving the patient pain and discomfort. Although no complaint was reported, this can still be a reason for flexing the trunk forwards, when the KJC is shifted backwards, in order to relief pain, since the subjects were describing the function of the prosthesis after conducting the trunk adjustments.

In the ML direction (TRUNK ML TILT), the AP knee alignment changes resulted in small and inconsistent changes in the trunk angular position during stance phase. However, some subjects showed a meaningful trend of change in the trunk position during swing phase. When the KJC was shifted backwards from the normal position, five subjects (ILBQCK, PLAICK, PLAQCK, TRAQCK, ELCQCK) increased the trunk ML tilt toward the sound leg during the prosthetic swing phase, two subjects (JLAQCK and DLAQCK) showed no change and the four remaining subjects showed a decrease in the trunk ML tilt toward the sound leg during the prosthetic swing phase. When the knee was shifted forwards, six subjects (ILBQCK, MRCQCK, PLAQCK, LRBQCK, ILBICK, PLAICK) slightly decreased the trunk ML tilt towards the sound leg during the prosthetic swing phase, three subjects (JLAQCK, JLAICK, DLAQCK) showed no change, and the two remaining subjects showed a slight increase in the trunk ML tilt toward the sound leg during the prosthetic swing-phase. The increase/decrease in the trunk ML tilt toward the sound leg which was exhibited by most subjects during swing phase when the KJC was shifted backward/forward, is due to the fact that shifting the KJC backward/forward increases/decreases the hip-toe distance of the prosthesis during swing phase. Therefore, the subject is forced to lean over the sound leg during the prosthetic swing phase in order to clear the way for the swinging foot and prevent stumbling. The change in the hip-toe distance is affected by the original orientation of the prosthesis and by the amount of shift in the KJC, and the patient may tolerate some change in that distance without having to change the trunk ML behaviour. This explains why some subjects did not change the trunk ML angle with the knee alignment changes, and some subjects even decreased/increased the trunk ML tilt toward the sound leg during the prosthetic swing phase when the KJC was shifted backward/forward.

In the transverse plane (TORSO ROTATION), changes in torso rotation were found in most subjects with knee alignment changes, however, torso changes were inconsistent and followed no certain pattern of change. The magnitude of these changes varied among subjects from 4 to 10 degrees. These changes are related to the changes in the fore-and-aft force which were

Table 6.16 Effect of alignment changes on the angular displacements of the lower joints of the prosthetic leg[1].

Knee Flexion Angle	Ankle Dorsiflexion Angle	Thigh Angle	Knee Flexion Angle	Ankle Dorsiflexion Angle	Thigh Angle
Effect of the Foot Changes					
6 Degrees Change Toward Dorsiflexion			6 Degrees Change Toward Plantarflexion		
No noticeable changes.	= During swing phase. - in 8 amp (2-6 deg) at push off. = in 3 amp.	= During swing phase. + TFA at HS, and - TEA at push off, (1 - 3.5 deg)*	= During stance phase. - Slightly during swing phase.	= During swing phase. + in 7 amp (1.5-3.5 deg) at push off. = in 4 amp.	= During swing phase. - TFA at HS, and + TEA at push off, (1 - 3 deg)*
Effect of Socket Changes					
6 Degrees Change Toward Flexion			6 Degrees Change Toward Extension		
Increased slightly during swing phase. = During stance phase.	= During swing phase. + in most amp (0 - 3 deg) at push off.	= During swing phase. + TFA at HS, and - TEA at push off, (2 - 6 deg)*	= During swing phase. = During stance phase.	= During swing phase. - in most amp (0 - 3 deg) at push off.	= During swing phase. - TFA at HS, and + TEA at push off, (2 - 6 deg)*
Effect of the Knee Shift					
1.8 cm Forward Shift			1.8 cm Backward Shift		
= During stance phase. + slightly in 5 & = in 6 amp during swing phase.	Inconsistent variations.	= During swing phase. + TFA at HS, and - TEA at push off, (1.5 - 6 deg)*	= During stance phase. - slightly in 5 & = in 6 amp during swing phase.	Inconsistent variations.	= During swing phase. - TFA at HS, and + TEA at push off, (1.5 - 6 deg)*

+ Increased, - Decreased, = No change relative to the normal position.

TFA Thigh flexion angle. TEA Thigh extension angle. HS Heel strike. amp: Amputee

1 No noticeable changes were found on the sound leg. ** The change was found in all subjects.*

found on the sound and prosthetic legs with knee alignment changes, as an increase in the push off force was expected to result in an increase in the torso rotation. However, as these changes were inconsistent, no conclusion can be drawn.

6.7.2 Effect of Alignment Changes on the Angular Displacements of the Lower limb Joints

6.7.2.1 Effect of Foot Alignment Changes

Figures 6.45 and 6.46 show the effect of the foot alignment changes on the AP angular displacements of the lower joints of the sound and prosthetic legs respectively, for one subject, and table 6.16 shows a summary of the results for this effect for all subjects.

It was found that the angular displacements of the sound leg joints were not affected by the foot alignment changes. Small and inconsistent variations were found in all subjects, and the change exhibited in the ankle dorsiflexion angle of subject PLAQAF was only found in this subject. Therefore, no conclusion can be drawn regarding the effect of foot alignment changes on the angular displacements of the lower limb joints of the sound leg.

On the prosthetic side, no noticeable changes were found in the knee flexion angle, however, some patients displayed a slight decrease in the knee flexion angle during swing phase when foot alignment changes toward plantarflexion were made. This corresponds to the decrease in the push off force which was found at the prosthetic leg when the foot alignment changes were toward plantarflexion.

It was found that at the prosthetic leg, foot alignment changes were compensated by changes in the ankle[3] and thigh angles during stance phase, but no changes were found in the ankle and thigh angles during swing phase. When the foot was dorsiflexed by 6 degrees (in increments of 3 degrees) from

[3] For the prostheses which were fitted with SACH feet, the changes in the ankle angle are in fact changes in the bending angle of the forefoot because the ankle is solid, see section 5.5.3 for the calculation of this angle.

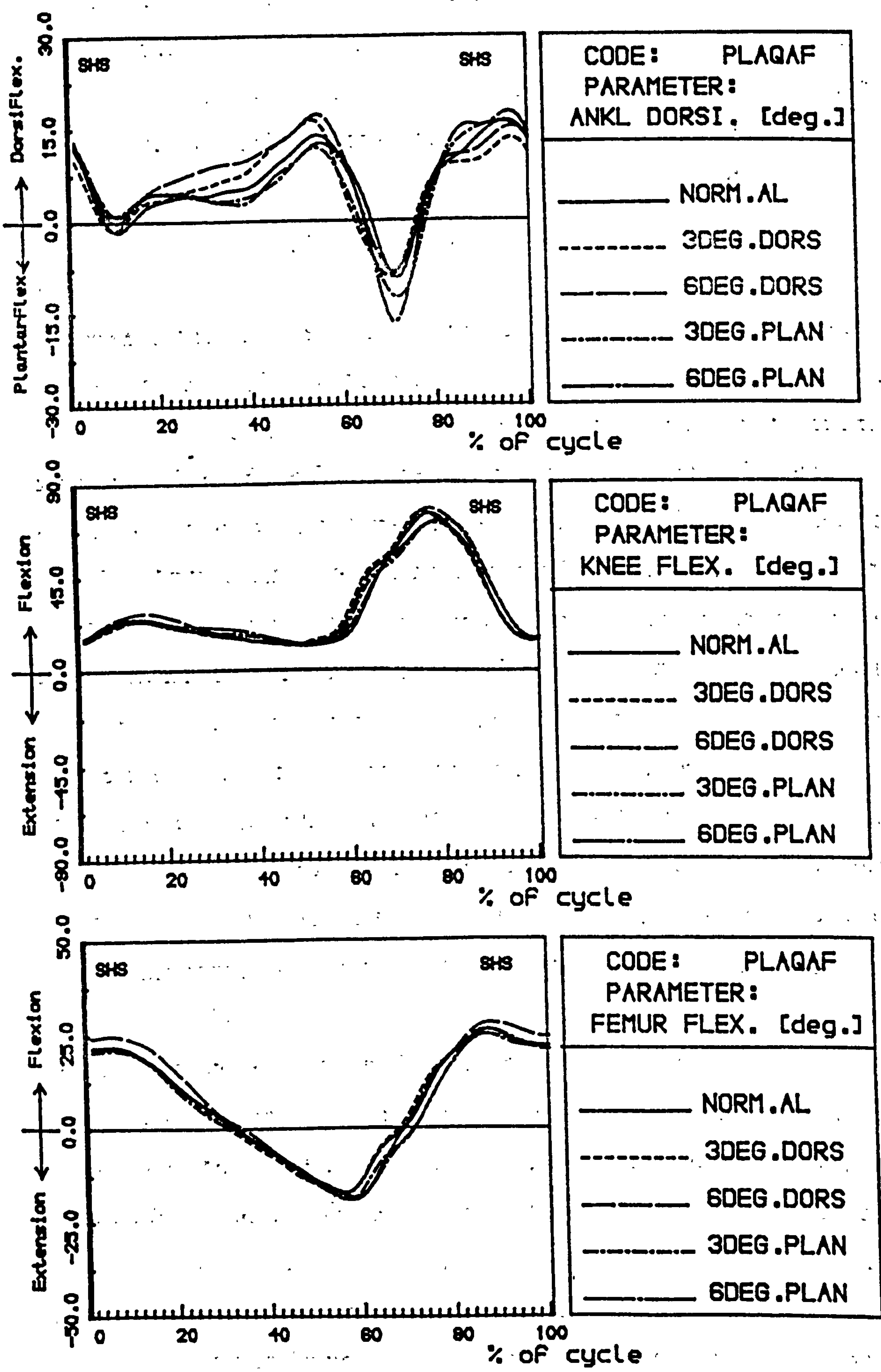


Figure 6.45 Effect of foot alignment changes on the variation with time of the angular displacements of the sound lower limb joints for an AK amputee.

its normal position, the ankle dorsiflexion angle decreased at push off in eight subjects by an average of 3.6 degrees (ranging from 2 to 6 degrees) in comparison to that when the foot was in normal alignment, and the three remaining subjects (PLAQAF, JLAQAF, ELCQAF) showed no changes in the ankle dorsiflexion angle. Plantarflexing the foot by 6 degrees (in increments of 3 degrees) from the normal position, increased the ankle dorsiflexion angle at push off in seven subjects by an average of 2.3 degrees (ranging from 1.5 to 3.5 degrees), and the four remaining subjects (MRCQAF, ELCQAF, ILBQAF, DLAQAF) showed no changes. Another interesting change was found in the ankle angle parameter with foot alignment changes, that is, in the time duration of the ankle plantarflexion after heel strike. This however, will be discussed later with the ankle joint moments. The above finding of the decrease/increase in the ankle angle at push off when the foot was dorsi/plantarflexed from the original position, may be related to the fact that the subject was compensating for the foot alignment changes. However, the patients have no control on the ankle angle because they were all fitted with SACH feet (except subject DLAQAF), and the changes measured in the ankle angle were in fact changes in the bending angle of the forefoot. The change in the bending angle of the forefoot depends on the stiffness of the foot, and on the moment applied about the foot bending centre which is mainly generated by the vertical component of the ground reaction force. However, the foot stiffness was not changed because the foot was not changed, and the vertical ground reaction force (FPY) was found to have no noticeable changes at push off with the foot alignment changes. Therefore, the above changes in the ankle angle can only be related to changes in the distance between the FPY and the foot centre of bending. The anterior part of the prosthetic foot (which is equivalent to the anatomical toes) may not be fully supportive at push off when the foot changes were toward dorsiflexion, and the subject was more supported by the prosthetic toes when the foot changes were toward plantarflexion than that when the foot was in normal alignment. This is

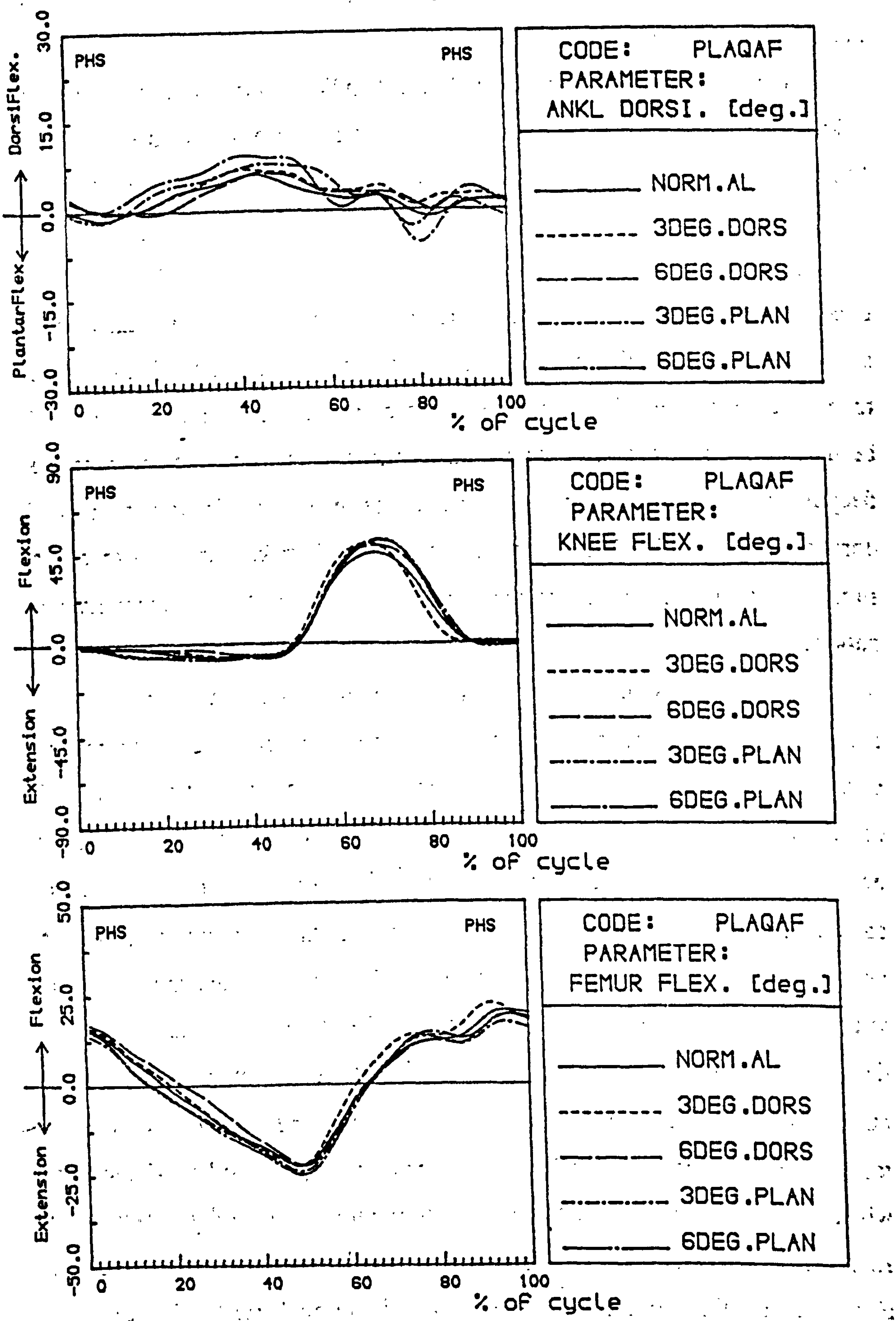


Figure 6.46 Effect of foot alignment changes on the variation with time of the angular displacements of the prosthetic lower limb joints for an AK amputee.

supported by the fact that the ankle dorsiflexing moment showed a slight decrease/increase at push off as the foot was dorsiflexed/plantarflexed from its normal position (see section 6.7.4.1).

On the prosthetic side, it was found that dorsiflexing the foot from its original position by 6 degrees (in increments of 3 degrees) increased the thigh flexion angle at heel strike by a range from 1 to 3.5 degrees, and decreased the thigh extension angle at push off by a range from 1 to 3.5 degrees. Plantarflexing the foot from its original position by 6 degrees (in increments of 3 degrees) decreased the thigh flexion angle at heel strike by a range from 1 to 3 degrees, and increased the thigh extension angle at push off by a range from 1 to 3 degrees, in comparison to that when the foot was in normal alignment. These adaptations were made by the subject in order to compensate for the angular changes of the foot and for lost stability. The mechanism of compensation for the dorsiflexion changes at heel strike and at push off is explained below. A similar explanation can be applied to the plantarflexion changes:

Referring to figure 6.47, dorsiflexing the foot by an angle θd will shift the KJC from K to K1, the HJC from H to H1 and the body centre of gravity will also be shifted from CG to CG1. This will increase the height of the HJC (point H1, by about 2 cm) relative to the ground, and will change the vector of the ground reaction force from R to R1 which passes behind the KJC (K1) causing instability of the prosthesis. Therefore, the amputee will put the prosthetic foot further forwards (or the HJC backwards) so that the HJC (H1) will move downwards from H to Hd (distance H-Hd is exaggerated to demonstrate the point) and reduces the height gained. This will change the thigh flexion angle from Ψ to Ψd, and it is clear that the value of Ψd is larger than Ψ, and this case was obtained in the above discussed results. The subject will also flex the trunk forwards to restore stability. At push off phase, the HJC of the prosthetic side loses height by dorsiflexing the foot (compare points H and Hd in fig. 6.36), therefore, the patient will shift the HJC backwards

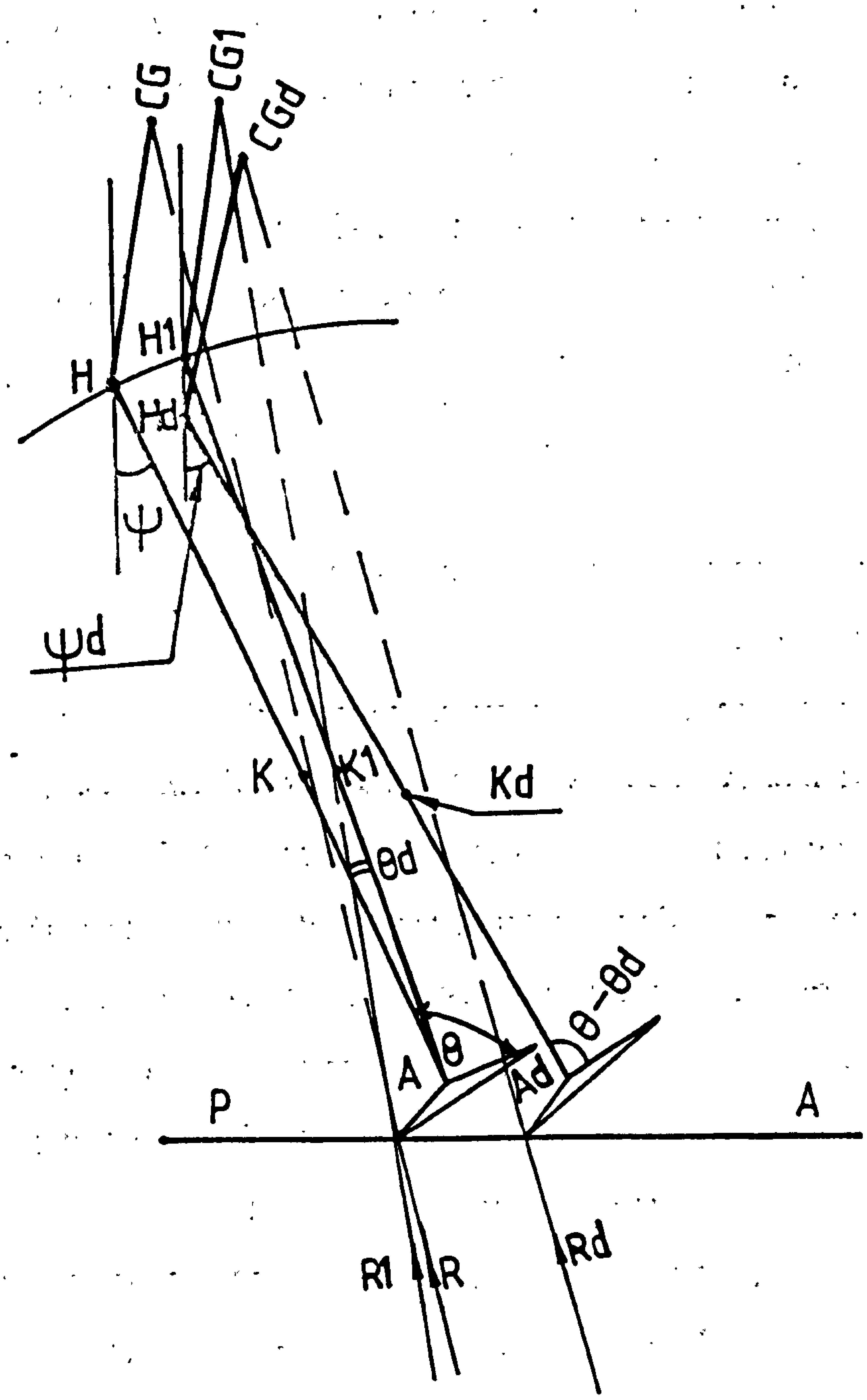


Figure 6.47 Effect of foot alignment changes on the orientation of the segments cf the prosthesis with respect to the vertical line and to the ground. The orientation of the ground reaction force vector with respect to the joint centres of the limb is also shown. **(from heel strike until mid-stance of the prosthetic leg).**

(toward the point H) in order to compensate for the lost height. This will decrease the thigh extension angle at push off, and this is applicable to the above discussed results.

6.7.2.2 Effect of the Socket Alignment Changes

The effects of socket alignment changes on the angular displacements of the lower limb joints in the AP plane, are shown in table 6.16, and figures 6.48 and 6.49 show this effect on the prosthetic and sound sides respectively, for a representative subject.

On the prosthetic side, no noticeable changes were found in the ankle dorsiflexion, knee flexion or in the thigh flexion angle during swing phase with socket alignment changes. However, the knee flexion angle of most patients showed a tendency towards a slight increase during swing phase as the alignment changes were towards flexing the socket. This is attributed to the increase in the prosthetic push off force (FXP) which was found with flexing the socket from its original position (see section 6.7.3.2). During stance phase, flexing/extending the socket from its normal position by 6 degrees (in increments of 3 degrees), resulted in flexing/extending the prosthetic knee angle in all subjects by 6 degrees (proportional changes can be assumed) from its normal position (the knee angle was defined as the angle between shank and thigh, see section 5.5.3). This change in the knee angle is in fact not a compensating action, because during stance phase the prosthetic knee is always held in full extension, as it is a plain uniaxial knee, and the patient cannot control the prosthetic knee angle during stance phase. Thus, the resulting changes in the prosthetic knee angle during stance phase noted are a reflection of the changes which were conducted at the socket.

When the socket was flexed/extended from its original position, the flexion angle of the prosthetic thigh at heel strike was increased/decreased, while the extension angle of the prosthetic thigh at push off was decreased/increased in comparison to that when the socket was in normal alignment. When the socket was changed by 6 degrees (in increments of 3

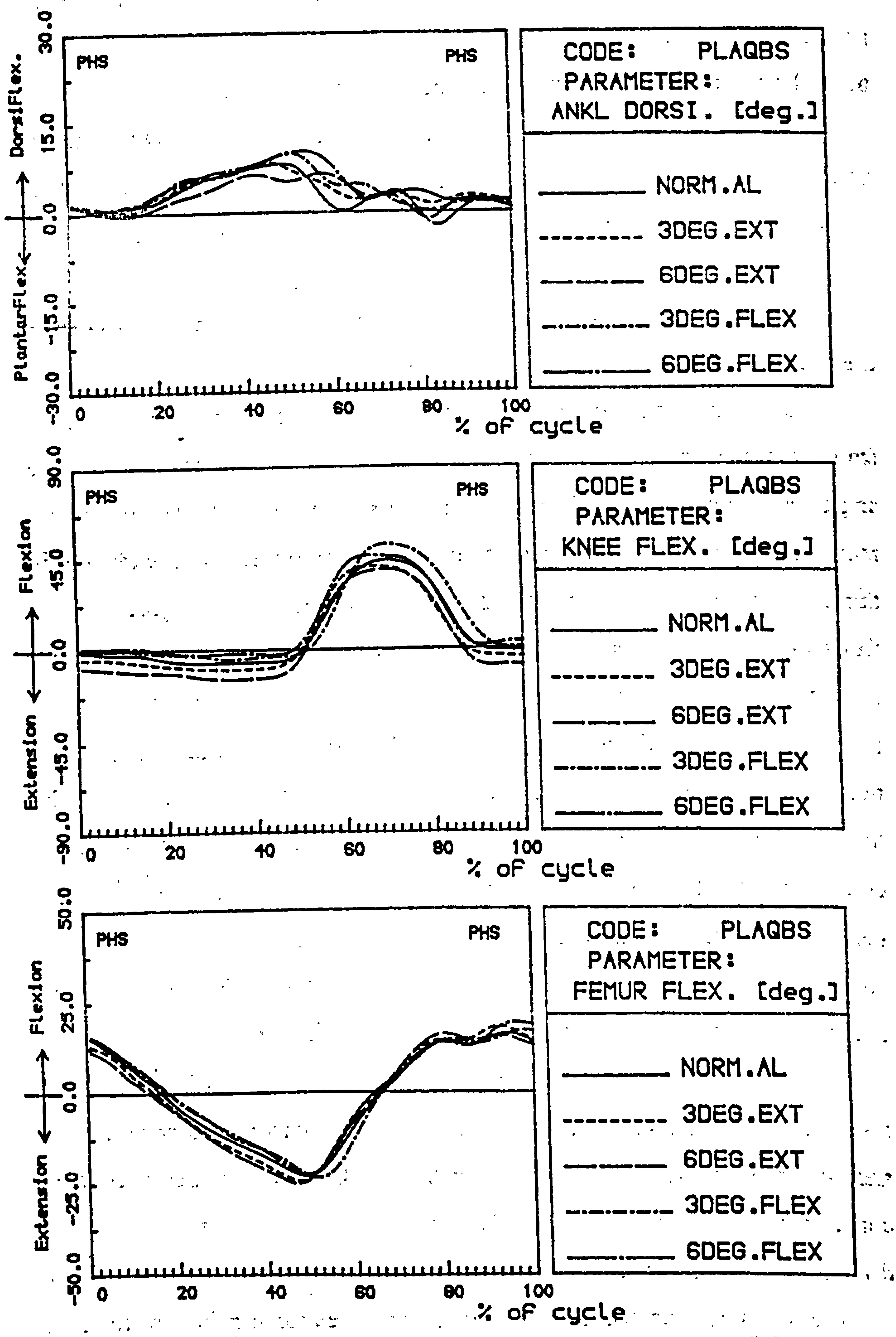


Figure 6.48 Effect of socket alignment changes on the variation with time of the angular displacements of the prosthetic lower limb joints for an AK amputee.

degrees) in either direction, the change in the thigh angle varied between subjects from 2 to 6 degrees (proportional changes can be assumed for each subject). It was also found, that in most subjects flexing/extending the socket by 6 degrees (in increments of 3 degrees) increased/decreased the ankle dorsiflexion angle at push-off by a range from zero to 3 degrees (proportional changes can be assumed for each subject). The changes in the thigh and ankle angle are influenced by the changes in the socket angle, and are also influencing each other. This mechanism of the changes in the thigh and ankle angles with the socket alignment changes is explained as follows, considering the socket flexion change as an example:

By flexing the socket by an angle ΨF (3 and 6 degrees in this study) from its normal position, the subject will have to flex his stump by the same angle (ΨF) in order to fit his stump in the flexed socket, and maintain similar angulation between the foot and the ground to that when the socket was in normal alignment, during the events of stance phase. If the subject failed to flex the stump by the angle ΨF and flexed it by an angle Ψ1 only (where Ψ1 is smaller than ΨF), the foot angular position would be changed with the ground in order to compensate for the differences in the thigh angle, and as such, the foot would act as if it had been plantarflexed by an angle equal to the difference between ΨF and Ψ1 (ΨF-Ψ1). This explains why the ankle dorsiflexion angle of most subjects behaved in a similar manner when the socket was flexed to that when the foot was plantarflexed. If the changes in the thigh angle are similar to the changes in the socket angle (subject TRAQBS is a clear example), the ankle angle would not be changed.

On the sound side, no noticeable changes were found in the angular displacements of the joint with the socket alignment changes, however, small and inconsistent variations were found but they cannot be considered conclusive.

6.7.2.3 Effect of the Knee Shifts

Figures 6.50 and 6.51 show the effect of knee forward/backward shifts

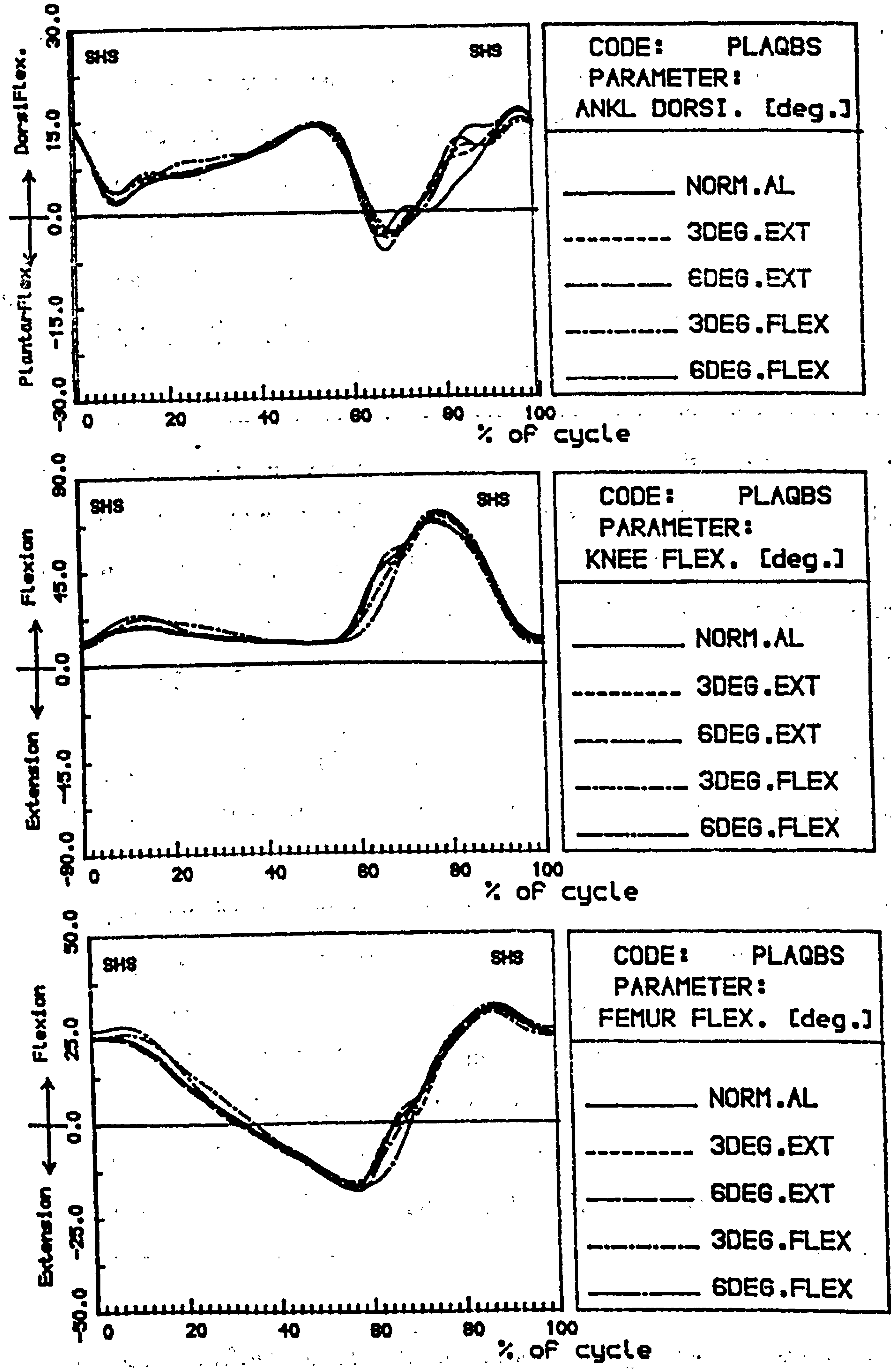


Figure 6.49 Effect of socket alignment changes on the variation with time of the angular displacements of the sound lower limb joints for an AK amputee.

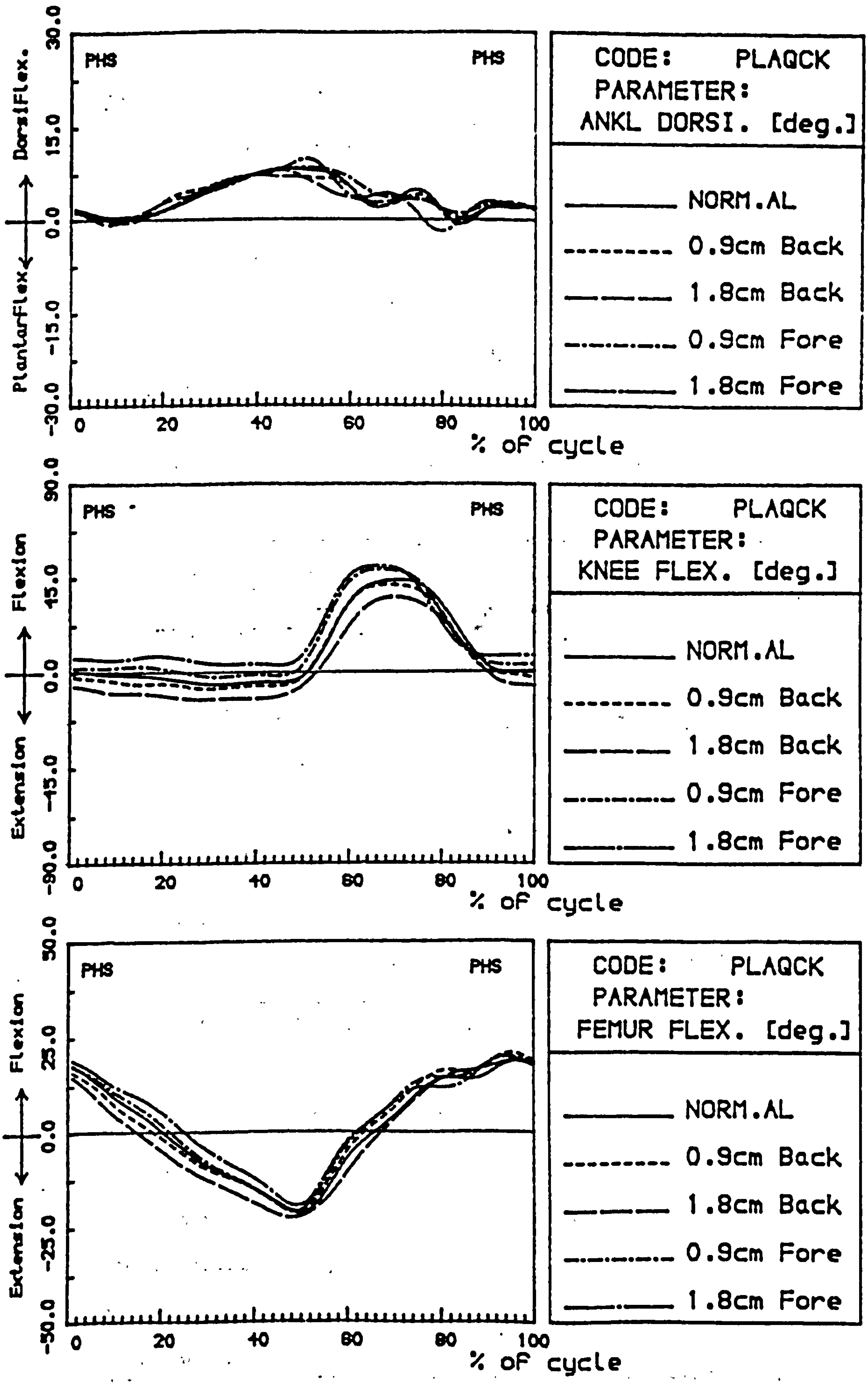


Figure 6.50 Effect of knee shifts on the variation with time of the angular displacements of the prosthetic lower limb joints for an AK amputee.

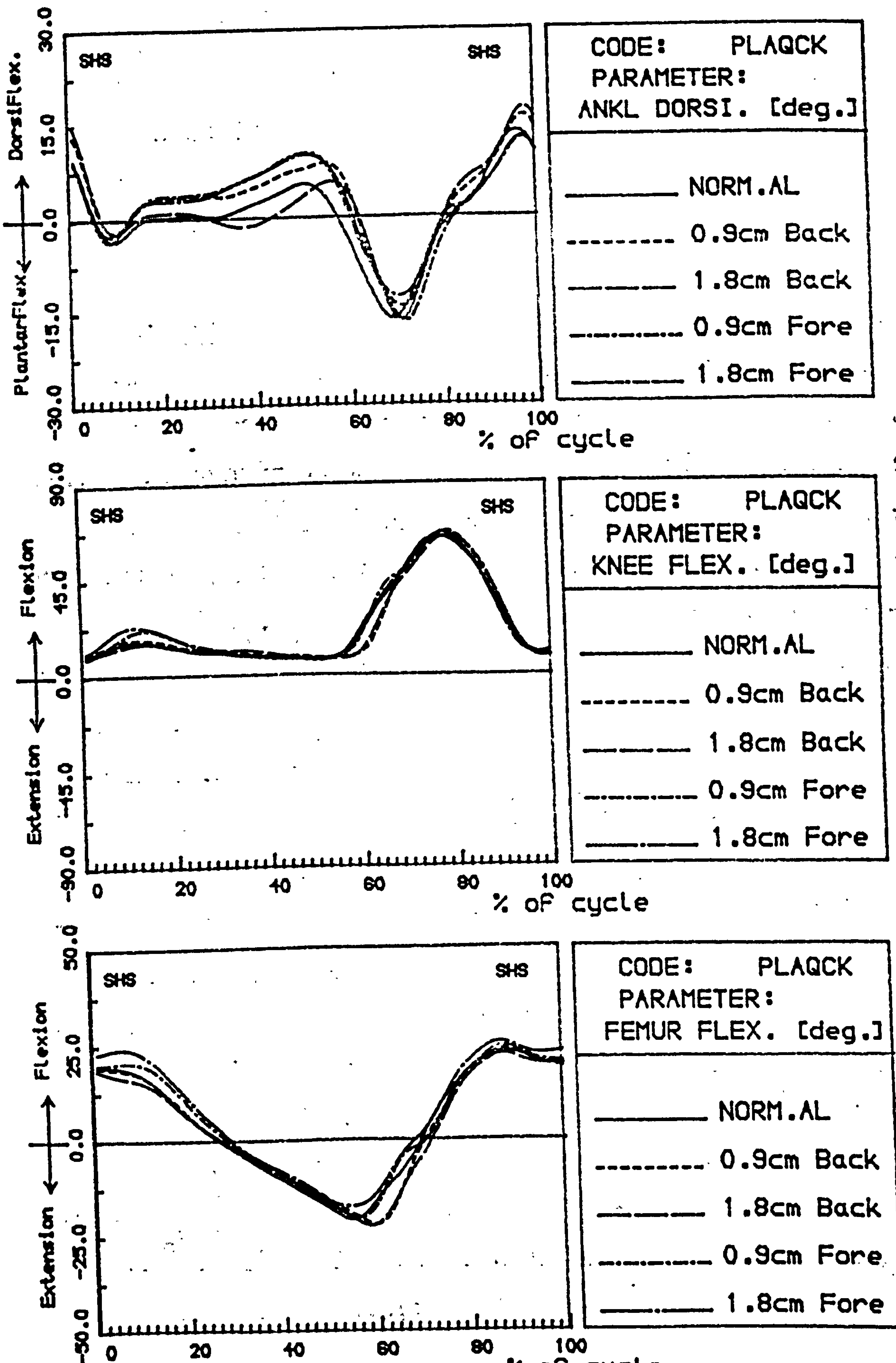


Figure 6.51 Effect of knee shifts on the variation with time of the angular displacements of the sound lower limb joints for an AK amputee.

on the angular displacements of the prosthetic and sound leg joints in the A/P plane respectively for one representative subject, and table 6.16 shows a summary for all patients.

On the prosthetic side, the knee angle was changed in all subjects during stance phase. Shifting the KJC forwards/backwards relative to the hip-ankle line resulted in an increase/decrease in the knee flexion angle. This change in the knee angle is equal to the angular change which was applied on the socket in order to achieve the knee shift. As mentioned before in this chapter, the knee shifts were achieved by a simultaneous change at the prosthetic foot and socket by a certain angle. For example, shifting the KJC forwards by 0.9 cm (fig. 6.50) was conducted by flexing the socket and dorsiflexing the foot simultaneously by 3 degrees. As the knee is always locked in full extension during stance phase, the above obtained changes in the knee angle were in fact a measure of the angular changes at the socket and they have no compensatory mechanism. During swing phase, five subjects (TRAQCK, ILBQCK, PLAICK, PLAQCK, ELCQCK) showed a tendency of increasing the knee flexion angle as the KJC was located anteriorly. This is related to the increase in the push off force (FXP) which was found at the prosthetic leg as the KJC was shifted forwards. The six remaining subjects showed inconsistent variations in the knee flexion angle during swing phase with the knee shifts.

Shifting the KJC forwards/backwards does not affect the socket and foot orientations relative to the ground, because the foot angular changes were compensated by suitable socket angular changes when the knee shifts were conducted, therefore, the subjects have no angular changes in the orientation of the prosthesis to compensate for. However, shifting the KJC forwards/backwards from its normal position resulted in slight variations in the ankle angle of some subjects, and in a noticeable change in the thigh flexion angle of all subjects. Shifting the KJC forwards by an increment of 0.9 cm, increased the thigh flexion angle at heel strike and decreased the thigh extension angle at push off by a range from 1.5 to 3 degrees per shift. This

Table 6.17 Effect of alignment changes on the fore-and-aft ground reaction farce (FPX) of the prosthetic leg.

Transition Time (TT)	Braking Force (FXB)	Push Off Force (FXP)	Transition Time (TT)	Braking Force (FXB)	Push Off Force (FXP)
Effect of the Foot Changes					
6 Degrees Change Toward Dorsiflexion			6 Degrees Change Toward Plantarflexion		
- in 8 amp (38% - 67%)* = in 3 amputees	- in 5 amp (16% - 87%). + in 3 amp (14% - 22%). = in 1&inconsis. in 2 amp	+ in 3 amputees by (10% - 26%). = in 8 amputees.	= in 8 amputees. + in 3 amputees by an average of 37%	+ in 9 amp (10% - 100%) - in 2 amp by an average of 16.5%.	- in 6 amp (10.5%-29%). + in 1 amp by 18%. = in 4 amputees.
Effect of Socket Changes					
6 Degrees Change Toward Flexion			6 Degrees Change Toward Extension		
- in 4 amp (6% - 30%). = in 7 amputees.	- in 10 amp (14% - 66%). + in 1 amp by 57%.	+ in 10 amputees by (10% - 130%). = in 1 amputee.	+ in 1 amputee by 8%. - in 1 amputee by 29% = in 9 amputees.	+ in 10 amputees by a range from 15% to 100%. + in 1 amputee by 350%.	- in 8 amputees by a range from 8% to 35%. = in 3 amputees.
Effect of the Knee Shift					
1.8 cm Forward Shift			1.8 cm Backward Shift		
- in 6 amputees by a range from 26% to 59%. = in 5 amputees.	- in 10 amputees by a range from 33% to 157%. + in 1 amputee by 143%.	+ in 8 amputees by a range from 6% to 31%. = in 3 amputees.	+ in 4 amputees by a range from 8% to 73%. = in 7 amputees.	+ in 10 amputees by a range from 10% to 198%. - in 1 amputee by 22%.	- in 9 amp (9% - 29%). = in 1 amputee. Inconsis change in 1 amp

+ Increased, - Decreased, = No change relative to the normal value. amp: amputee

* All changes are in percentage of the normal value. *See figure 6.52 for the illustration of terms TT, FXB and FXP.*

is explained as follows:

Although, shifting the KJC forwards did not change the orientation of the prosthesis, the height of the HJC which is represented by the point HF (see fig. 6.43) increased in comparison to that of the normal alignment (point H). Therefore, the subject was forced to put his foot further forward (or his HJC further backward) in order to reduce the gained height. This increased the thigh flexion angle and caused a slight change in the ankle dorsiflexion angle during stance phase. The change in the foot and thigh angles which were obtained from shifting the KJC forwards, is similar to that obtained from dorsiflexing the foot and was discussed in section 6.7.2.1.

On the sound side, no noticeable changes were found in the angular displacements of the lower joints with the knee shifts. However, some variations were noted in the thigh angle of subjects JLAICK, TRAQCK, MRCQCK, PLAICK, ELCQCK and JLAQCK. In these cases the thigh angle slightly increased as the KJC was shifted forwards. This may be due to the fact that some subjects would compensate by thigh angular changes on the prosthetic and sound legs in order to have a uniform gait.

6.7.3 Effect of Alignment Changes on the Ground Reaction Forces

The effects of alignment changes on the fore-and-aft ground reaction force (FPX) of the prosthetic and sound legs are shown in tables 6.17 and 6.18 respectively. The effect on the vertical ground reaction force (FPY) is also shown in table 6.19 for the prosthetic and sound legs.

6.7.3.1 Effect of the Foot Changes

Figures 6.52 and 6.53 show the effect of foot alignment changes on the ground reaction force of the prosthetic and sound legs respectively for a typical subject, and tables 6.17, 6.18 and 6.19 show a summary of the results of all subjects.

- **The Effect on the Fore-and-aft Force (FPX):**

To make the discussion more effective, the fore-and-aft ground reaction force (FPX) was characterised by three parameters (see fig. 6.52), thus, the

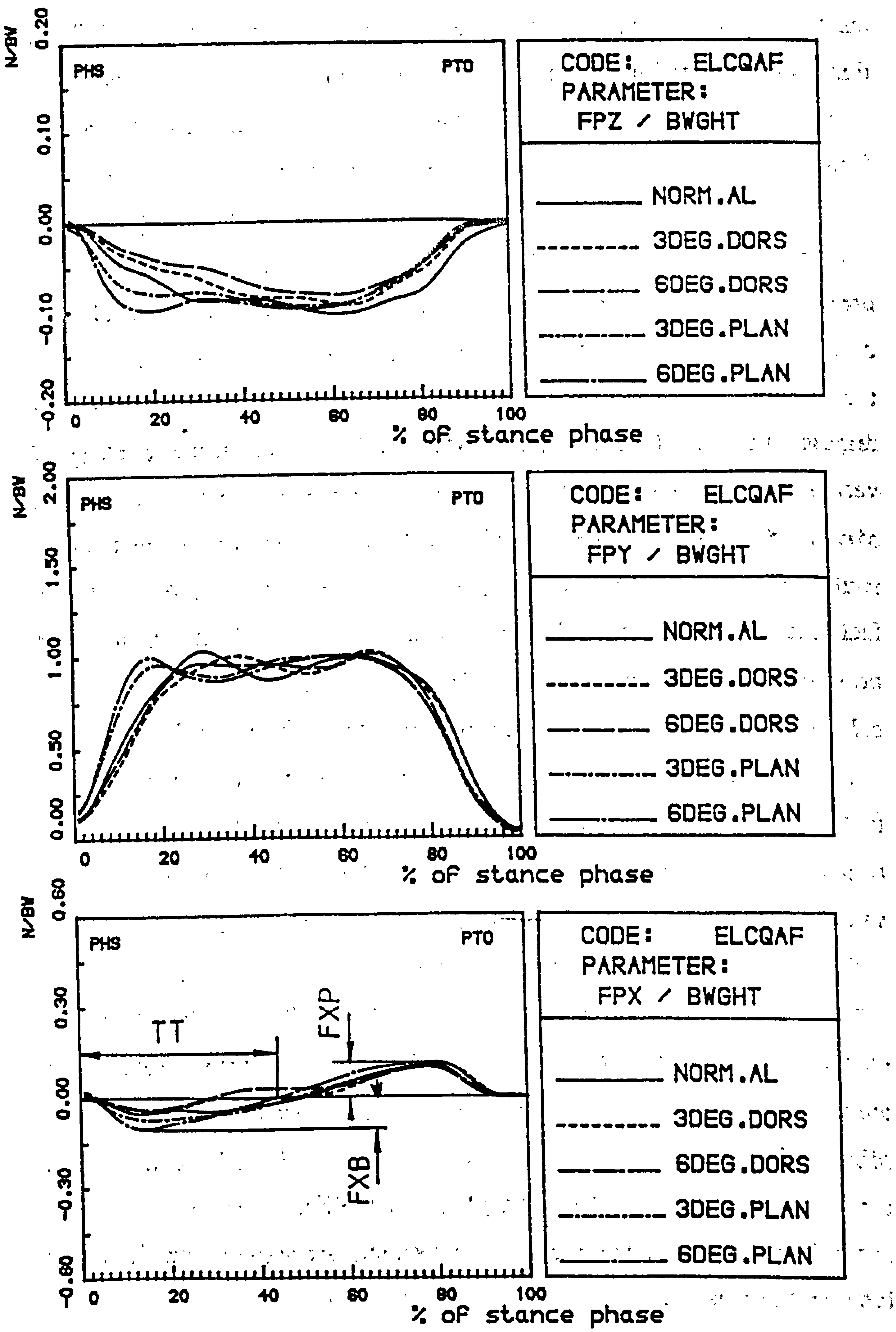


Figure 6.52 Effect of foot alignment changes on the variation with time of the ground reaction forces of the prosthetic leg for an AK amputee.

changes in the FPX with the alignment changes can be studied in detail. The parameters are: the braking force (FXB), the push off force (FXP) and the transition time (TT) during which the FPX will appear to be negative (i.e the duration of the braking force).

On the prosthetic leg, noticeable changes were found in the pattern of FPX with foot alignment changes. The pattern became more comparable to that of the normal subjects, when the foot was plantarflexed from the normal position. This was achieved by an increase in the magnitude of the braking force (FXB) and in the transition time (TT). When the foot was dorsiflexed by 6 degrees (in increments of 3 degrees) from the normal position, the TT of the prosthetic leg decreased in eight patients and did not change in the three remaining patients (LRBQAF, TRAQAF, ILBIAF). This decrease in the TT was also reported by Mizrahi et al (1986) and Yang Lang (1988). The maximum decrease in the TT value with dorsiflexion of the foot varied individually among subjects from 38% (subject PLAQAF) to 67% (subject ILBQAF) of that obtained when the foot was in normal alignment. It should be noticed that the change in the value of TT was not proportional to the change in the foot angle, and generally, a similar change in the TT value was obtained when the foot was dorsiflexed by 3 and 6 degrees. This is attributed to the fact that the subject has to spend a certain time on the heel area in order to allow shifting his body mass forwards. This time is influenced by the velocity of the subject which was found to have no significant (P>0.1) changes with the foot alignment changes. The reason of the decrease in the value of TT will be discussed later in this section along with the changes in the magnitudes of the braking force (FXB) and the push off force (FXP). When the foot changes were towards plantarflexion, and against the expectation, eight subjects showed no change in the TT value of the prosthetic leg and the three remaining subjects (ILBQAF, TRAQAF,LRBQAF) only showed an increase in the TT value by an average of 37% of the value obtained when the foot was in normal alignment.

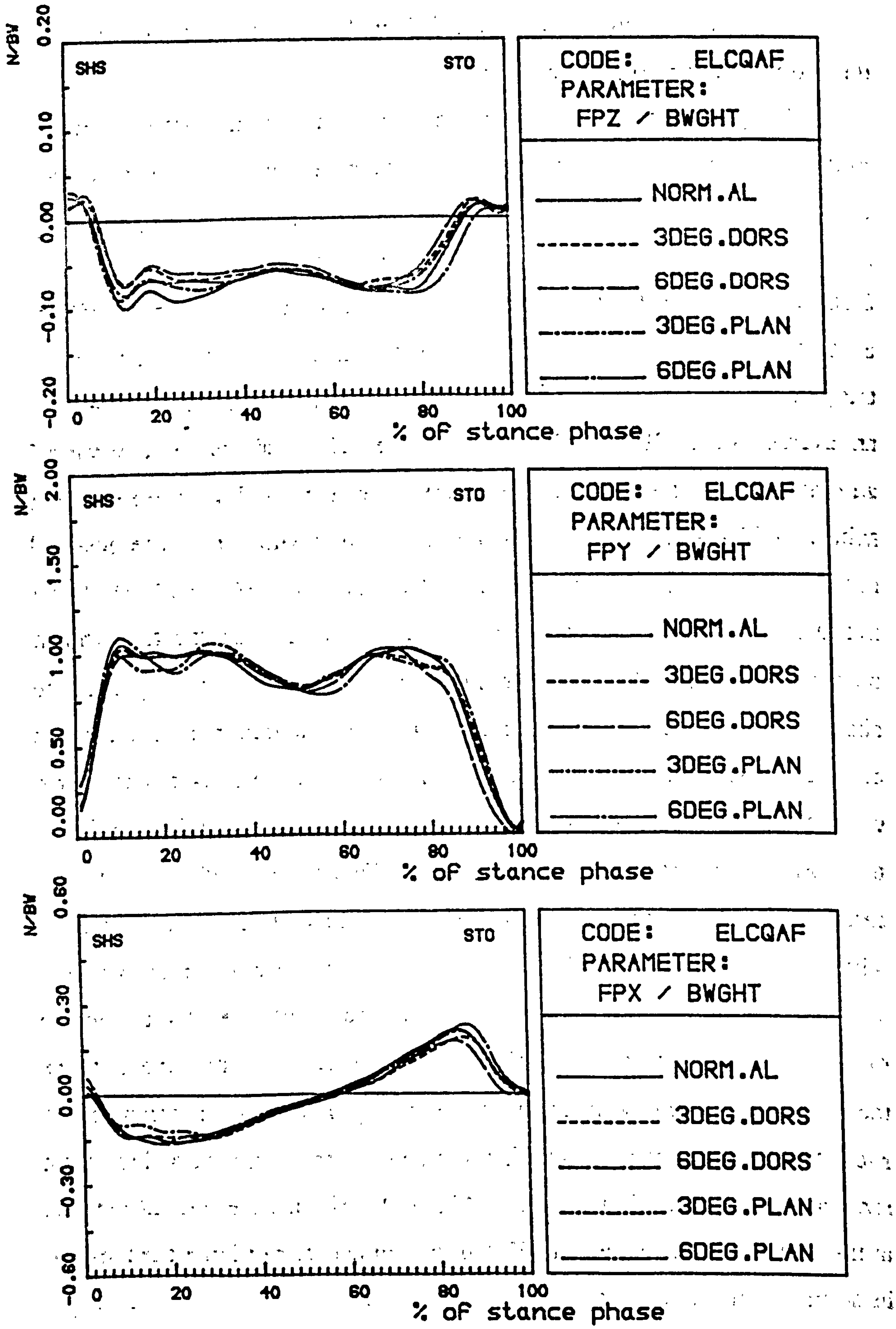


Figure 6.53 Effect of foot alignment changes on the variation with time of the ground reaction forces of the sound leg for an AK amputee.

On the sound leg, no changes were found in the TT value with foot alignment changes, and the small variation which was obtained (maximum of 7% of that when the foot was in normal alignment) is considered to be within the individual step to step variations.

On the prosthetic leg, the magnitude of the fore-and-aft force (FPX) showed noticeable changes with foot alignment changes. When the foot was dorsiflexed by 6 degrees (in increments of 3 degrees) from the normal position, the braking force (FXB) decreased in five subjects (PLAIAF, MRCQAF, ELCQAF, ILBIAF, DLAQAF) by an average of 43% (ranging from 16% to 87% and proportional changes can be assumed) of that of the normal alignment, and increased in three subjects (PLAQAF, JLAIAF, LRBQAF) by an average of 17.3% (ranging from 14% to 22%) of that of the normal alignment, one subject (JLAQAF) showed no change, and the two remaining subjects (ILBQAF, TRAQAF) showed inconsistent variations. The push off force (FXP) increased in three subjects (MRCQAF, PLAQAF, PLAIAF) by an average of 20% (ranged from 10% to 26%) of that when the foot was in normal alignment, and showed no changes in the eight remaining subjects. The above finding agrees with Yang Lang (1988) who reported insignificant changes in the FXP and significant changes in the FXB as the foot was changed from plantarflexion to dorsiflexion. When the foot was plantarflexed by 6 degrees (in increments of 3 degrees) from the normal position, the braking force (FXB) of the prosthetic leg increased in nine subjects by an average of 45% (ranging from 10% to 100% and proportional changes can be assumed) of that when the foot was in normal alignment, and decreased by an average of 16.5% of that of the normal alignment in the two remaining subjects (LRBQAF, PLAIAF). Plantarflexing the foot by 6 degrees (in increments of 3 degrees) decreased the push off force (FXP) of the prosthetic leg in six subjects (MRCQAF, TRAQAF, ILBQAF, DLAQAF, ILBIAF, PLAIAF) by an average of 16.5% (ranging from 10.5% to 29% and proportional changes can be assumed) of that when the foot was in normal alignment. Only one subject

Table 6.18 Effect of alignment changes* on the fore-and-aft ground reaction force (FPX) of the sound leg.

Transition Time (TT)	Braking Force (FXB)	Push Off Force (FXP)	Transition Time (TT)	Braking Force (FXB)	Push Off Force (FXP)
Effect of the Foot Changes					
6 Degrees Change Toward Dorsiflexion			6 Degrees Change Toward Plantarflexion		
No noticeable changes.	No noticeable changes.	A trend of decrease was found by max. of 23%.	No noticeable changes.	- in 6 amputees by a range from 15% to 41%. = in 5 amputees.	A trend of increase was found by max. of 28%.
Effect of Socket Changes					
6 Degrees Change Toward Flexion			6 Degrees Change Toward Extension		
No noticeable changes.	A trend of slight increase.	A trend of slight decrease	No noticeable changes.	A trend of slight decrease	A trend of slight increase
Effect of the Knee Shift					
1.8 cm Forward Shift			1.8 cm Backward Shift		
No noticeable changes.	+ in 9 amp (5% - 43%). = in 1 amputee. inconsis change in 1 amp	- in 8 amp (20% - 60%). = in 2 amputees. + in 1 amputee by 85%.	No noticeable changes.	- in 9 amputees by a range from 8% to 55%. = in 2 amputees.	+ in 7 amp (11% - 48%). = in 3 amputees. - in 1 amputee by 22%.

+ Increased, - Decreased, = No change relative to the normal value. amp: amputee.

* All changes are in percentage of the normal value.

(PLAQAF) showed an increase in the FXP, by 18% of that of the normal alignment, and the four remaining subjects showed no change.

To study the mechanism of the above changes in the fore-and-aft force with the foot alignment changes, the kinematic and kinetic data acquired were used and figures 6.35 and 6.36 were drawn to illustrate the discussion.

Suppose a prosthesis in its normal alignment is represented by the AJC (A), KJC (K) and HJC (H), the body centre of gravity by CG, at heel strike (fig. 6.35), the vector of the ground reaction force R passes ahead of the KJC and the prosthesis is stable. Dorsiflexing the foot will shift the KJC form K to Kd and the HJC from H to Hd. If the trunk angular position did not change, the ground reaction force will be represented by the vector R1 which may pass behind the KJC (Kd) causing instability to the prosthesis. Therefore, the patient will flex his trunk forwards shifting the body centre of gravity from CG2 to CGd, in this case, the ground reaction force vector will be represented by Rd which passes ahead of Kd and restores the stability of the prosthesis. Comparing vector R with Rd (fig. 6.35), it is clear that the inclination of vector Rd to the vertical line in the AP plane is smaller than that of vector R, thus, the fore-and-aft (the braking force) component of vector Rd is smaller than that of vector R with the condition that the absolute values of vectors Rd and R are equal ($|R| = |Rd|$). This explanation is appropriate to the five subjects who showed a reduction in the FXB when the foot was dorsiflexed from the normal position. The reason that three subjects showed an increase in the FXB when dorsiflexing the foot, is believed to be the possibility that the dorsiflexion which was applied on the foot, did not cause instability to the prosthesis, and the subjects did not have to flex their trunks in order to compensate for the lost stability. This is supported by the fact that (referring to these 3 subjects) subject LRBQAF extended his trunk slightly, subject JLAIAF showed no change and subject PLAQAF exhibited very little trunk flexion as the foot was dorsiflexed from the normal position.

The above explanation which showed that the braking force (FXB), i.e.

the horizontal component of vector Rd was smaller than that of vector R, proves that the subject encountered a lower resistance when the foot was dorsiflexed than that when the foot was in normal alignment. This explains why eight subjects had a decrease in the transition time (TT) when the foot was dorsiflexed from the normal position.

The differences among subjects noted in the changes in FXB and TT as a result of foot alignment changes, can be related to the changes in the inclination of the ground reaction force vector (GRFV). The changes in the inclination of the GRFV were in turn due to changes in the attitude of the trunk adopted by the subjects in order to ensure adequate stability of the prosthetic knee.

Applying a similar analysis on the push off force FXP (see fig. 6.36) shows that dorsiflexing the foot would increase the FXP. However, only three subjects followed this mechanism of change, and the changes in the FXP of the remaining eight subjects were not noticeable (see discussion above in this section and table 6.17). This is attributed to the fact that during the push off phase, the foot alignment changes performed did not endanger the stability of the prosthesis, and therefore, the subjects did not have to change the trunk position in order to compensate for lost stability.

Considering the foot changes in the plantarflexion direction, and again referring to figures 6.35 and 6.36 and following a similar procedure of analysis to that for the changes in the dorsiflexion direction, it can be seen that plantarflexing the foot will change the vector of the ground reaction force from R to RP (the GRFV passes through the body CG, ignoring the offset between the body CG and GRFV caused by the body angular acceleration). This would increase the magnitude of FXB as indeed was found in nine subjects and decrease the value of FXP as was obtained for six subjects. The increase in the magnitude of FXB would increase the value of TT, however, this was found for three subjects only and eight subjects showed no change in the TT value when the foot was plantarflexed from the normal position. This is believed to be due to the fact that most subjects were fit enough to overcome the resultant high resistance without delay in the TT. Furthermore, the TT cannot be delayed by more than a limited time, this time is influenced by the velocity of the subject which was found have no significant ($P>0.1$) changes with the foot alignment changes. Therefore, the subjects were forced to make the effort and overcome the resultant high resistance without delaying in the TT.

The differences among subjects in the changes in FXB and FXP, and the reason that four subjects showed no change in the FXP with plantarflexing the foot from the normal position, are attributed to the fact that plantarflexing the foot would increase the stability of the prosthesis, therefore, the subject may adopt a compensatory mechanism and change the vector of the ground reaction force from R to Rp, however, the subject is not forced to adopt that change as the stability of the prosthesis is maintained.

The decrease in the prosthetic push off force when the foot is plantarflexed from the original position, resulted in a noticeable decrease in the braking force of the sound leg. This decrease was found on the sound side of six subjects (DLAQAF, ILBIAF, MRCQAF, ILBQAF, JLAQAF, PLAIAF) and ranged from 15% to 41% of that when the foot was in normal alignment. The push off force of the sound leg showed a trend of increase/decrease when the prosthetic foot was plantarflexed/dorsiflexed from the normal position. The maximum increase was 28% (subject JLAQAF) and the maximum decrease was 23% (subject DLAQAF) of that when the foot was in normal alignment. This increase/decrease in the push off force of the sound leg is controlling the braking force of the prosthetic leg which generally increased/decreased with plantarflexion/dorsiflexion of the prosthetic foot from the normal position. It should be mentioned, that no noticeable differences were found between the effect of foot alignment changes on the fore-and-aft force of quad and IC socket wearers, therefore, they were discussed as one group of patients.

Table 6.19 Effect of alignment changes on the vertical ground reaction force (FPY).

Effect of the Foot Changes		
	6 Degrees Change Toward Dorsiflexion	6 Degrees Change Toward Plantarflexion
Prosthetic Leg	- the first vertical peak in 6 amp. by 11% to 66% of the normal value[1].	+ the first vertical peak in 5 amp. by 28% to 42% of the normal value.
Sound Leg	A trend of slight increase/decrease in the first/second vertical peak.	A trend of slight decrease/increase in the first/second vertical peak.
Effect of Socket Changes		
	6 Degrees Change Toward Flexion	6 Degrees Change Toward Extension
Prosthetic Leg	- the first vertical peak in 8 amp. by 9% to 64% of the normal value.	No noticeable changes.
Sound Leg	No noticeable changes.	No noticeable changes.
Effect of the Knee Shift		
	1.8 cm Forward Shift	1.8 cm Backward Shift
Prosthetic Leg	- the first vertical peak in 8 amp. by 9% to 117% of the normal value. - slightly the occurrence time in the trough of some amputees.	+ the first vertical peak in 5 amp. by 13% to 29% of the normal value. + slightly the occurrence time in the trough of some amputees.
Sound Leg	A trend of slight increase/decrease in the first/second peak.	A trend of slight decrease/increase in the first/second peak.

+ Advanced   - Delayed   amp. amputee.

1 Normal value: is the occurrence time of the first vertical peak corresponding to that when the prosthesis was in normal alignment.

The Effect on the Vertical Force (FPY):

As the vertical ground reaction force is typically characterised by two peaks and a trough, this will be employed in addition to the phase of occurrence of these three parameters to study the effect of the alignment changes on the vertical ground reaction force.

On the prosthetic leg, most subjects exhibited variations in the vertical ground reaction force with the alignment changes. However, these variations were noticeable and consistent on the time of occurrence of the first vertical peak only, in that plantarflexing/dorsiflexing the prosthetic foot advanced/delayed the time of occurrence of the first vertical peak, and as a result, the occurrence of the trough was also advanced/delayed in most subjects (see fig. 6.52 and table 6.19). Dorsiflexing the prosthetic foot by 6 degrees (in increments of 3 degrees) delayed the occurrence of the first vertical peak in six subjects (JLAQAF, JLAIAF, ILBQAF, ELCQAF, MRCQAF, PLAIAF) by a range from 11% to 66% of that when the foot was in normal alignment. Although the changes are small relative to the stance phase, they are quite large relative to their value which was obtained when the foot was in normal alignment. For instance, subject ELCQAF showed a delay of 8% of stance phase, this is equal to 27% of the time of occurrence which was obtained for the first vertical peak when the foot was in normal alignment. This delay is attributed to the fact that dorsiflexing the foot will reduce the stability of the prosthesis, and will increase the foot-floor angle at heel strike, therefore, the patient was dwelling for a longer time on his heel to ensure the stability of the prosthesis and full contact between the foot and the floor before transferring load on to it. The reason that the delay time varied among subjects and did not exist in some subjects may be related to the degree of instability which was felt by the subject when dorsiflexing the foot. The reverse of the above explanation can be adopted to explain the advanced occurrence of the first vertical peak in the prosthetic ground reaction force, which was found in five patients (DLAQAF, TRAQAF, PLAQAF, JLAQAF, ELCQAF) when the foot was plantarflexed, and ranged from 28% to 42% of the normal value. The above advance/delay in the occurrence of the first vertical peak is also related to the time which is needed from heel strike until the foot is in full contact with the floor (as mentioned above). This time is affected by the foot alignment changes and the effect will be discussed later along with the moments of the ankle joint (section 6.7.4.1).

On the sound leg, a trend of change was found in the FPY of most subjects, in that the first/second peak decreased/increased when the prosthetic foot was plantarflexed from the normal position. The first/second peak also showed a trend of increase/decrease when the foot was dorsiflexed from the normal position. However, these changes were not discernible with all subjects and not always consistent. The reason for this change of trend can be explained as discussed above regarding the changes in the braking and pushing off forces of the sound leg.

The Effect on the Medio-Lateral Force (FPZ):

The medio-lateral force (FPZ) of the prosthetic leg showed noticeable and consistent changes during the first half of stance phase (the transition time), and inconsistent variations were noted during the rest of the stance phase with the foot alignment changes. When the foot was dorsiflexed/plantarflexed from the normal position, the magnitude of the prosthetic FPZ decreased/increased in most subjects. These changes varied among subjects and their maximum value was 100% (subject ELCQAF) of that when the foot was in normal alignment. These changes are attributed to the decrease/increase in the horizontal force acting along the foot, which has components in the X (FPX) and Z (FPZ) directions, since the foot has a toe out angle with the line of progression. On the sound leg, no changes were found in the FPZ with the foot alignment changes, and the variations that exist can be attributed to step to step variations.

It should be mentioned that no differences were found in the effect of the foot alignment changes on the vertical and medio-lateral force that could

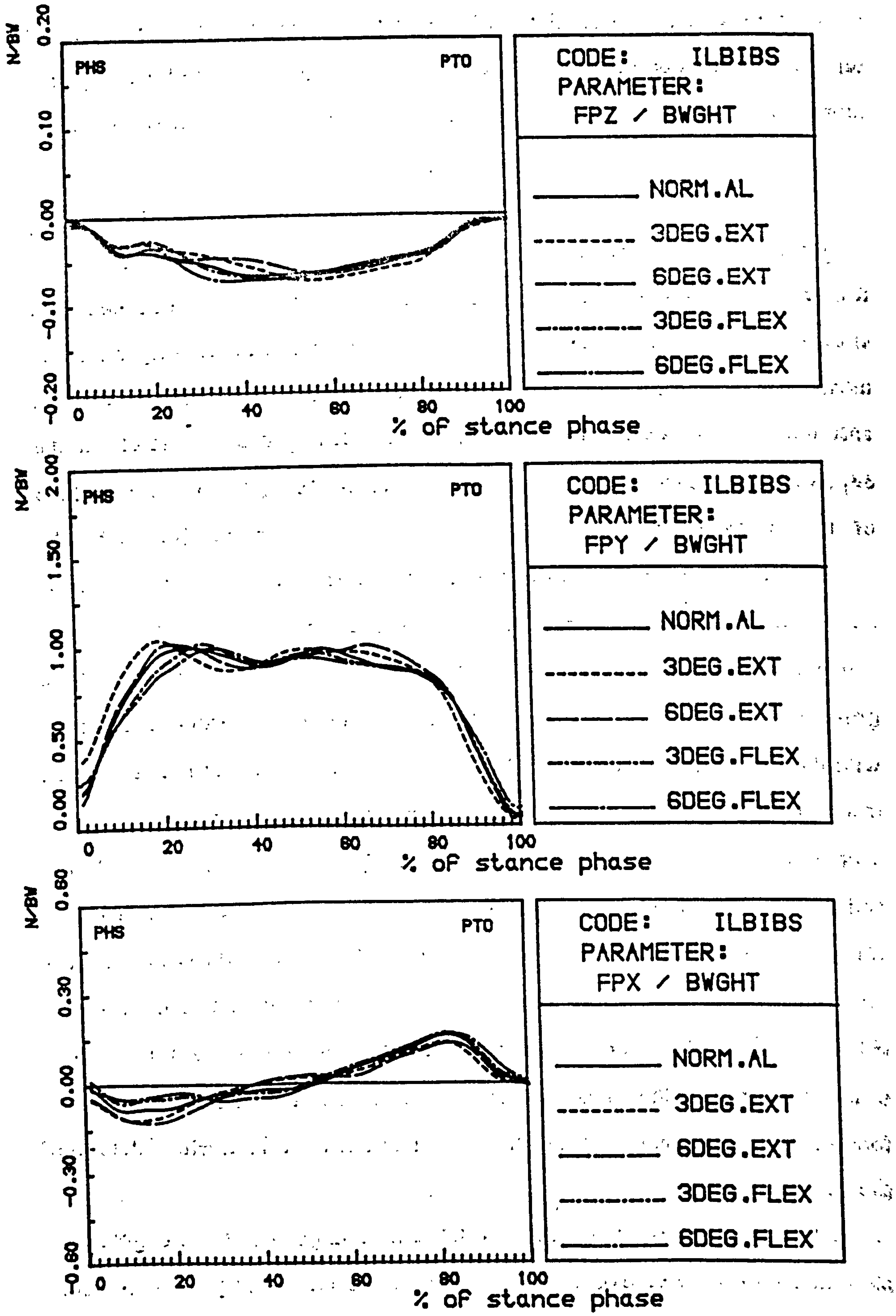


Figure 6.54 Effect of socket alignment changes on the variation with time of the ground reaction forces of the prosthetic leg for an AK amputee.

be attributed to the type of socket which was either quadrilateral or IC.

6.7.3.2 Effect of the Socket Changes

Figures 6.54 and 6.55 show the effect of the socket alignment changes on the ground reaction forces of the prosthetic and sound leg respectively, and tables 6.17, 6.18 and 6.19 show a summary of the results of all subjects.

The Effect on the fore-and-aft Force (FPX):

On the prosthetic side, the fore-and-aft ground reaction force showed noticeable and consistent changes with the socket alignment changes. Flexing the socket from its normal position by 6 degrees (in increments of 3 degrees), decreased the braking force (FXB) of the prosthetic leg in ten subjects by an average of 43.9% (ranging from 14% to 66% and proportional changes can be assumed) of that when the socket was in normal alignment. The remaining subject (ILBQBS) showed an increase in the FXB of the prosthetic leg by 57% of the normal value when the socket was flexed by 6 degrees (in increments of 3 degrees). These changes can be explained using the acquired kinematic and kinetic data and referring to figure 6.39 as follows:

Flexing the socket from its normal position will change the vector of the ground reaction force from R to R1 (corresponding to the position of the body CG) which passes on or behind the KJC causing instability to the prosthesis. Therefore, the subject will flex the trunk forwards by an angle θF (which was indeed found in section 6.7.1.2 and table 6.15) so that the vector of the ground reaction force, in its final position (RF), will pass ahead of the KJC and secure the stability of the prosthesis. Comparing vector R with RF, it is evident that the fore-and-aft component (FXB) of vector RF is smaller than that of vector R in condition that the absolute values of the two vectors are equal ($|R| = |RF|$). This indicates that the subject encountered less resistance by flexing the socket than when the socket was in normal alignment. Therefore, the transition time (TT) would be decreased when flexing the socket. However, as the socket was flexed from the normal position by 6 degrees (in increments of 3 degrees), the TT value decreased in only four subjects

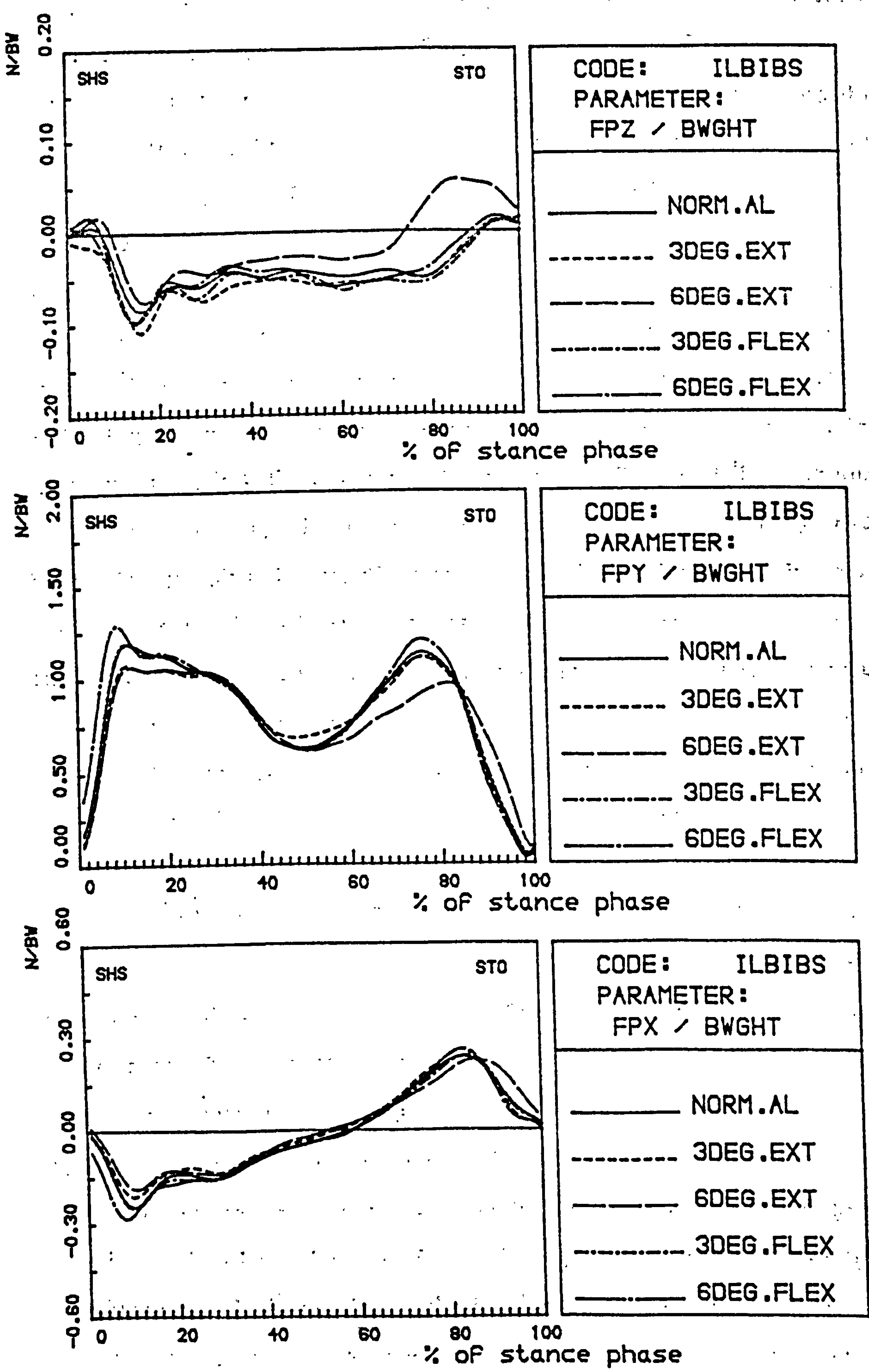


Figure 6.55 Effect of socket alignment changes on the variation with time of the ground reaction forces of the sound leg for an AK amputee.

(ILBQBS, PLAIBS, DLAQBS, ELCQBS) by an average of 16.3% (ranging from 6% to 30%) of that of the normal alignment, and the seven remaining subjects showed no change in the magnitude of TT. The reason that seven subjects showed no change in the value of TT is related to the unchanged ($P>0.1$) velocity of the subjects which would affect the TT value.

When the socket was flexed by 6 degrees (in increments of 3 degrees) from the normal alignment, the push off force (FXP) did not change in one subject (JLAQBS), but did increase by an average of 42% (ranging from 10% to 130%, and proportional changes can be assumed) of that of the normal alignment in the ten remaining subjects. This increase in the FXP can be explained by the same method used above to explain the decrease in the FXB when the socket was flexed. Referring to figure 6.40, it is clear that the fore-and-aft component of vector RF which represents the ground reaction force after flexing the socket, is larger than that of vector R which represents the ground reaction force when the socket was in normal alignment. Another reason for the increase in the FXP with increased socket flexion is that the effectiveness of the hip extensors muscles may increase and therefore, the moment generated by the prosthetic thigh will be increased, thus, the push off force will also be increased.

Extending the socket by 6 degrees (in increments of 3 degrees) from the normal alignment increased the braking force (FXB) of the prosthetic leg in all subjects, one subject (ILBQBS) showed a large increase (350%) while the ten remaining subjects showed an increase by an average of 34.5% (ranging from 15% to 100%, and proportional change can be assumed) of that when the socket was in normal alignment. This can be explained following a similar procedure to that followed above for studying the changes resulting from flexing the socket. Referring to figure 6.39, it is clear that the fore-and-aft component of vector R which represents the ground reaction force with the socket in normal alignment, is smaller than that of vector RE which represents the ground reaction force after extending the socket. It should be mentioned that the increase in the FXB would delay the transition time (TT), however, it was found that the TT value was delayed by 8% in subject ILBQBS only, advanced by 29% of that of the normal alignment in subject ILBIBS, and no changes were found in the TT value of the prosthetic side of the nine remaining subjects. This is related to the fact that all subjects were active and overcame the increase in the resistance without delay in the TT, and this in turn, is supported by the fact that no significant ($P>0.1$) changes were found in the velocity of the subjects with the socket alignment changes. Referring to figure 6.40, it can be seen that the push off force (FXP) should be decreased when extending the socket from its normal alignment (compare vector R with RE). This was found in eight subjects out of eleven, i.e. extending the socket by 6 degrees (in increments of 3 degrees) from the normal alignment, decreased the FXP of the eight subjects by an average of 21.8% (ranging from 8% to 35%, and proportional changes can be assumed) of that of the normal alignment, and no changes were found in the FXP of the three remaining subjects (ILBQBS, LRBQBS, MRCQBS).

Flexing/extending the socket from its normal position resulted in a trend to increase/decrease the braking force (FXB) and decrease/increase the push off force (FXP) on the sound leg. This trend of change was found in most subjects and is related to the changes in the prosthetic fore-and-aft force with the socket alignment changes which were discussed above, as such, an increase in the FXP of the prosthetic leg would result in an increase in the FXB of the sound leg, and a decrease in the FXB of the prosthetic leg would result from a decrease in the FXP of the sound leg.

The Effect on the Vertical Force (FPY):

The vertical ground reaction force of the prosthetic leg was more affected by flexing, than extending, the socket from the normal alignment. However, the noticeable changes were only at the instant at which the first vertical peak occurred. When the socket was flexed from the normal position by 6 degrees (in increments of 3 degrees), two subjects (DLAQBS and

PLAQBS) showed no change, one subject only (JLAQBS) showed an advance by 17%, and the eight remaining patients showed a delay in the occurrence of the first vertical peak of FPY, by an average of 38.4% (ranging from 9% to 64%, and proportional changes can be assumed) of that when the socket was in normal alignment. With the socket flexion changes, the magnitude of the first vertical peak of FPY of the prosthetic leg slightly increased in three subjects (JLAIBS, LRBQBS, JLAQBS), slightly decreased in subject ILBQBS, changed inconsistently in subject MRCQBS, and no changes were found in the six remaining subjects. These changes in the magnitude of the first vertical peak of FPY can be explained in a similar manner to that used to explain the changes in the FXB with increased socket flexion. Referring to figure 6.39, and comparing the vector R which represents the ground reaction force when the socket was in normal alignment with the vector RF which represents the ground reaction force after flexing the socket, it is clear that the vertical component of vector R is smaller than that of RF, if the magnitude of the resultant vector did not change, i. e. $|R| = |RF|$. This explanation is applicable to the three subjects who showed an increase in the first vertical peak of FPY when the socket was flexed. The reason that the other subjects did not follow the foregoing explanation may be related to the fact that they could not maintain the condition $|R| = |RF|$, because of a feeling of instability which was felt by the subjects when the socket was flexed. Thus, the subjects had to reduce the load which was supported by the prosthesis i.e. $|RF| < |R|$. The instability feeling at heel strike is also responsible for the delay in the instant of occurrence of the first vertical peak of FPY which was found in eight subjects, because the patients spent more time on the heel in order to ensure stability before fully loading the prosthesis.

No noticeable changes were found in the magnitude and time of occurrence of the trough and the second vertical peak of FPY of the prosthetic leg when the socket was flexed from its normal position. This shows that the delay in the occurrence of the first vertical peak did not affect the occurrence time of the trough and the second peak, and this in turn is in agreement with the finding that the transition time (TT) was reduced (or did not change) when the socket is flexed from its normal position. The unchanged magnitude of the trough and second vertical peak of FPY is related to the unchanged velocity of the subjects with the socket alignment changes.

When the socket was extended from the normal position, no noticeable changes in the pattern and magnitude of the FPY of the prosthetic leg were found. Although the changes in the FPX should be combined with changes in the FPY, this was not clearly shown. The reason for this is related to the small magnitude of the changes in FPY when the socket was extended relative to its original magnitude when the socket was in normal alignment. The changes in FPY can be explained by considering FPY and FPX simultaneously during stance phase. At heel strike, FXB was increased when the socket was extended thus, the FPY should have decreased (if the magnitude of the GRFV did not change), however, this was not the case because all subjects were active and could increase the speed and therefore, the magnitude of the ground reaction force vector would also increase. At push off the FXP decreased when the socket was extended thus, FPY should have increased, however, this was not the case either because extending the socket may reduce the effectiveness of the hip extensors and may create pain under the posterior brim of the socket during the push off phase.

On the sound side, FPY was not affected by the socket alignment changes, and the variations which are shown can only be related to step to step variations in the patients' gait.

The Effect on the Medio-Lateral Force (FPZ):

The medio-lateral ground reaction force was not changed on the sound side with the socket alignment changes. On the prosthetic side, flexing the socket from the original position resulted in a slight decrease in the FPZ of six subjects (ILBQBS, JLAIBS, ELCQBS, LRBQBS, PLAQBS, JLAQBS) during the first half of the stance phase, and extending the socket increased the FPZ

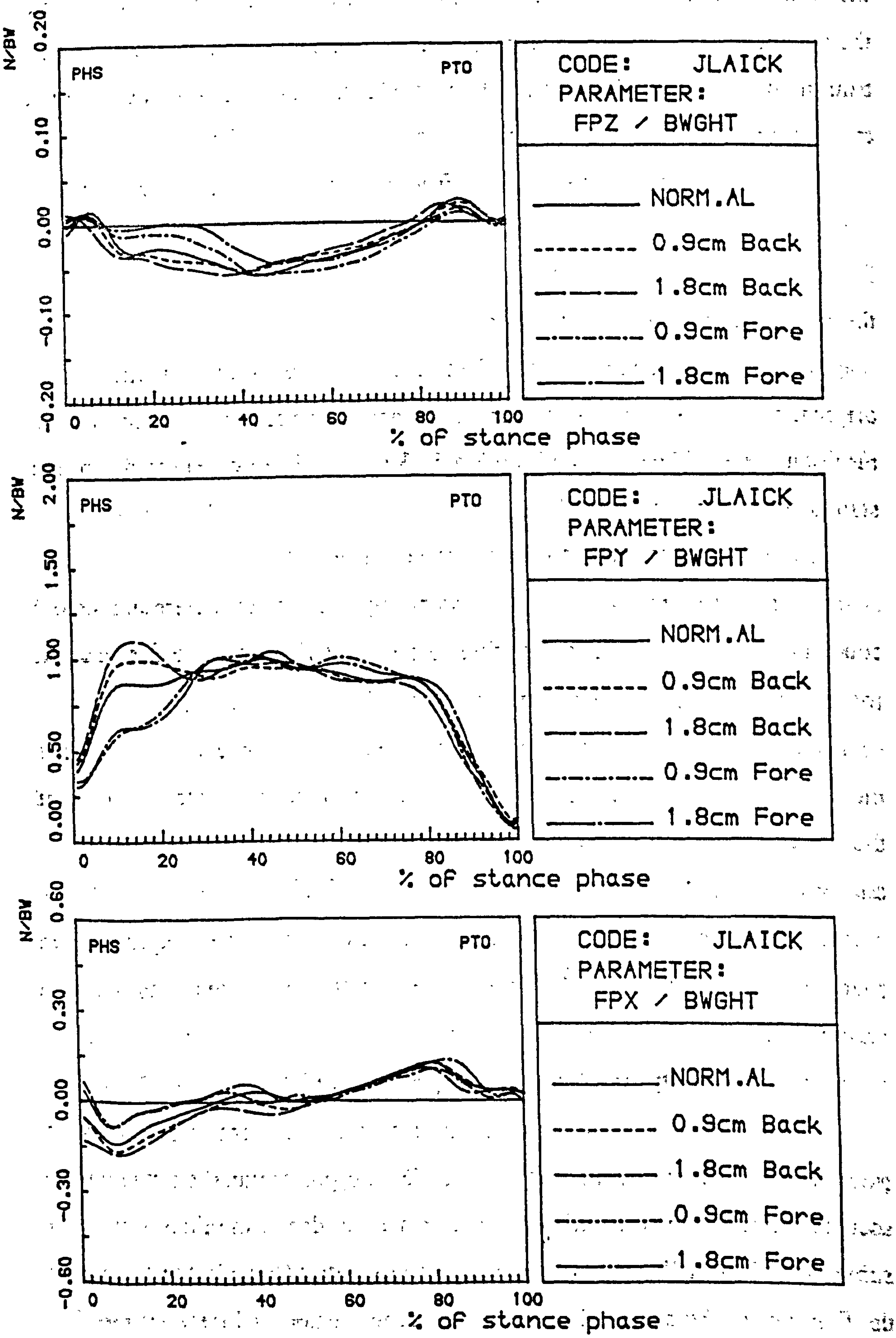


Figure 6.56 Effect of knee shifts on the variation with time of the ground reaction forces of the prosthetic leg for an AK amputee.

slightly during the first half of stance phase in five subjects (JLAIBS, ELCQBS, LRBQBS, PLAQBS, PLAIBS). The decrease/increase in the FPZ may be related to the fact that the alignment changes of the prosthesis, may change the horizontal force acting along the foot. Since the foot has a toe out angle with the line of progression, the force acting along the foot would have components on the X (FPX) and Z (FPZ) directions, and any change in the force acting along the foot will be reflected on FPX as discussed above and on FPZ as shown in this section.

6.7.3.3 Effect of the Knee Shifts

Figures 6.56 and 6.57 show the effect of the knee alignment changes on the ground reaction forces of the prosthetic and sound leg respectively for one representative subject, and tables 6.17, 6.18 and 6.19 show a summary of the results of all subjects.

The Effect on the Fore-and-Aft Force (FPX):

When the knee was shifted forwards by 1.8 cm (in increments of 0.9 cm) from its normal position relative to the hip-ankle line, the fore-and-aft force (FPX) of the prosthetic leg showed the following changes:

The decelerating force (FXB) decreased in ten subjects by an average of 46.1% (ranging from 33% to 157%, and proportional changes can be assumed) of that when the knee was in normal alignment, and the remaining subject (ILBQCK) showed an increase in the FXB by 143% of that of the normal alignment. The transition time (TT) advanced in six subjects by an average of 45.8% (ranging from 26% to 59%, and proportional changes can be assumed) of that when the knee was in normal alignment, and no change was found in the five remaining subjects (ILBQCK, PLAICK, JLAQCK, DLAQCK, LRBQCK). The push off force (FXP) increased in eight subjects by an average of 18.6% (ranging from 6% to 31%, and proportional change can be assumed) of that when the knee was in normal alignment, the three remaining subjects (ILBQCK, JLAQCK, TRAQCK) showed no change in the magnitude of FXP when shifting the KJC forwards. Shifting the KJC backward by 1.8

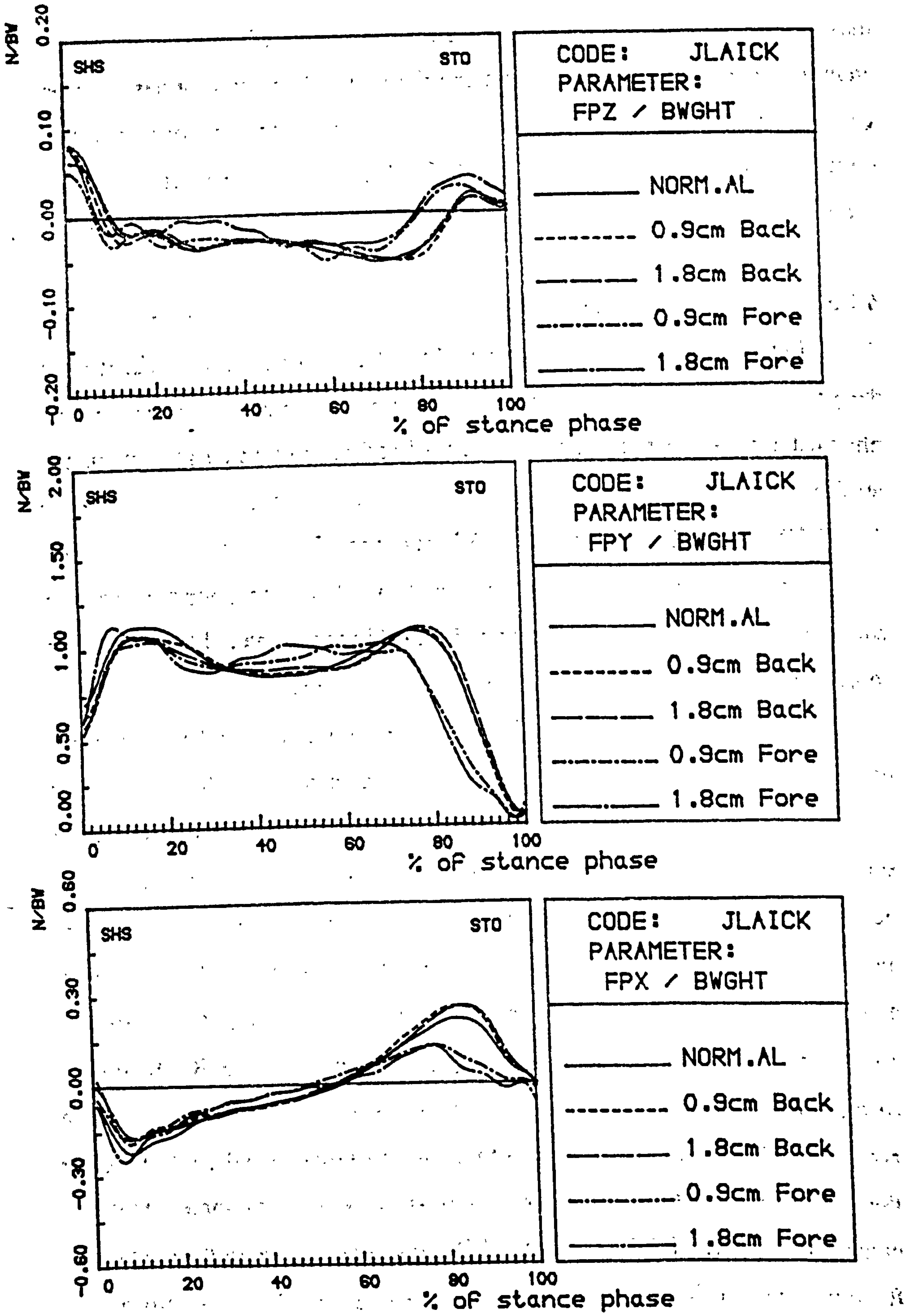


Figure 6.57 Effect of knee shifts on the variation with time of the ground reaction forces of the sound leg for an AK amputee.

cm (in increments of 0.9 cm) relative to the hip-ankle line resulted in the following changes in the fore-and-aft force of the prosthetic leg:

The braking force (FXB) increased in ten subjects by an average of 59.5% (ranging from 10% to 198%, and proportional changes can be assumed) of that when the knee was in normal alignment, and the remaining subject (LRBQCK) showed a decrease in the FXB by 22% of that of the normal alignment. The transition time (TT) was delayed in four subjects (ILBQCK, TRAQCK, LRBQCK, JLAICK) by an average of 28.5% (ranging from 8% to 73%) of that when the knee was in normal alignment, and no change was found in the seven remaining patients. The push off force (FXP) decreased in nine subjects by an average of 20% (ranging from 9% to 29%, and proportional changes can be assumed) of that when the knee was in normal alignment, one subject (ILBQCK) showed inconsistent changes and the remaining subject (JLAQCK) showed no change in the FXP when the KJC was shifted backward.

The above results were expected and can all be explained by using the acquired kinematic and kinetic data and referring to figures 6.43 and 6.44. In these figures, it can be seen that shifting the KJC forward from K to KF will cause instability of the knee joint and therefore, the patient will flex his trunk forwards by an angle θF (as indeed was found in section 6.7.1.3 and table 6.15) changing the vector of the ground reaction force from R to RF which passes ahead of the KJC (KF) and maintain the stability of the prosthesis. It can also be seen that shifting the KJC backwards from K to KB would force the subject to extend his trunk by an angle θB (see section 6.7.1.3 and table 6.15) in order to reduce the hip flexing moment at heel strike and ease the knee flexion prior to swing phase. This will change the position of the ground reaction force vector from R to RB. Thus, the ground reaction force vector is represented by R when the knee is in normal alignment, and by RF/RB when the knee is shifted forwards/backwards from the normal position. A simple comparison between vectors R, RF, and RB shows that shifting the KJC forwards/backwards relative to the hip-ankle line will decrease/increase the braking force (FXB) and increase/decrease the push off force (FXP) with the condition that $|R| = |RF| = |RB|$. The advance/delay in the TT which was obtained with shifting the knee forwards/backwards is attributed to the decrease/increase in the resistance encountered by the subjects with shifting the KJC forwards/backwards as listed above. The reason that some subjects did not show changes in the TT with the knee changes, is related to the fact that the velocity of the subjects had no significant ($P>0.1$) change with the knee alignment change (see section 6.4.3.3). It should be mentioned that the decrease/increase in the braking force can be influenced by the feeling of stability the patient has as the KJC was shifted forwards/backwards.

On the sound leg, shifting the KJC of the prosthetic leg forwards by 1.8 cm (in increments of 0.9 cm), increased the braking force (FXB) of the sound leg in nine subjects by an average of 24.9% (ranging from 5% to 43%) of that when the prosthetic KJC was in normal alignment, one of the two remaining subjects (LRBQCK) showed no change, and the other (TRAQCK) showed inconsistent changes. With the above forward shift of the prosthetic KJC, the push off force (FXP) of the sound leg decreased in eight subjects by an average of 33.4% (ranging from 20% to 60%), did not change in two subjects (DLAQCK, LRBQCK), and increased in the remaining subject (MRCQCK) by 85% of that when the prosthetic KJC was in normal alignment.

When the prosthetic KJC was shifted backwards by 1.8 cm (in increments of 0.9 cm) relative to the hip-ankle line, the braking force (FXB) of the sound leg decreased in nine subjects by an average of 22.9% (ranging from 8% to 55%) of that of the normal alignment, and did not change in the two remaining subjects (PLAICK, LRBQCK). The push off force (FXP) of the sound leg increased in seven subjects by an average of 26.7% (ranging from 11% to 48%), did not change in three subjects (JLAQCK, TRAQCK, LRBQCK), and decreased in the remaining subject (ELCQCK) by 22% of that when the prosthetic KJC was in normal alignment. The increase/decrease in the FXB, and the decrease/increase in the FXP of the sound leg when the prosthetic KJC was shifted forwards/backwards from the normal position, are influenced by the changes of the fore-and-aft force of the prosthetic leg which were found with the knee shifts. The increase/decrease in the prosthetic push off force resulted in an increase/decrease in the braking force of the sound leg, and an increase/decrease in sound side push off resulted in an increase/decrease in prosthetic braking force.

The Effect on the Vertical Force (FPY):

It was found that the only change in the FPY of the prosthetic leg was in the time of occurrence of its first vertical peak. When the KJC was shifted forwards by 1.8 cm (in increments of 0.9 cm), the time of occurrence of the first vertical peak of FPY of the prosthetic leg was delayed in eight subjects by an average of 39% (ranging from 9% to 117%) of that when the knee was in the normal position, and the three remaining subjects (ILBQCK, LRBQCK, MRCQCK) showed no change. The delay in the occurrence of the first vertical peak of FPY, was associated with a slight delay in the occurrence time of the trough of the FPY of some subjects (PLAQCK, TRAQCK, ILBQCK, ELCQCK). Shifting the KJC backwards by 1.8 cm (in increments of 0.9 cm), advanced the occurrence of the first vertical peak of the prosthetic FPY in five subjects (ILBQCK, PLAQCK, TRAQCK, ELCQCK, DLAQCK) by an average of 21.2% (ranging from 13% to 29%) of that of the normal alignment, and the six remaining subjects showed no change. The advance in the occurrence of the first vertical peak resulted in a slight advance in the time of occurrence of the trough of FPY, however, this was only found in three subjects (PLAQCK, TRAQCK, DLAQCK). These changes in the time of occurrence of the first vertical peak of the prosthetic FPY, are related to the instability which was felt by the patient when the KJC was located anteriorly, because the patient would be dwelling on the heel for a longer time in order to ensure the stability of the prosthesis before transferring full load onto it.

On the sound leg, the vertical ground reaction force (FPY) showed similar changes to those shown by the fore-and-aft force of the same leg with

Table 6.20 Effect of alignment changes on the ankle joint moments.

Ankle Joint	A/P Moment of the Ankle Joint (MAZ)		Transverse Moment (MAY)	Coronal Moment (MAX)
	Transition Time (TP)	Dorsiflexing Moment at Push Off		
Effect of the Foot Changes				
Prosthetic	+ in all amp. by 9% to 31%* as the foot was dorsiflexed by 12 degrees.	+ Slightly in 5 amputees when plantarflexing the foot.	+ Slightly in all amputees when plantarflexing the foot.	Slight increase in the everting moment of most amp. when plant.
Sound	The reported change is a slight increase (in 6 amp) in the plantarflexing moment which exists after HS as the foot changes were toward dorsiflexion.			
Effect of Socket Changes				
Prosthetic	+ in all amp. by 11% to 20% when extending the socket by 12 degrees.	No noticeable changes	+ Slightly the internal rotating MAY in 5 amp with socket flexion	No noticeable changes
Sound	The reported change is a slight increase (in most amp) in the plantarflexing moment which exists after HS, with flexing the socket.			
Effect of the Knee Shift				
Prosthetic	+ in all amp by 4% to 13% when shifting the KJC 3.6 cm forwards.	No noticeable changes	+ Slightly the internal rotating MAY in all amp when shifting the KJC anteriorly.	Slight +/- in the eve/inv MAX of 4 and inconsis. change in 7 amp with locating the KJC anteriorly.
Sound	+ in 4 amp by an average of 9% as the KJC shifted anteriorly.	A trend of +/- in the plant/dors MAZ with the forward shifts.	No noticeable changes	No noticeable changes

+ Increased   - Decreased   amp: amputee.

* All changes are in percentage of stance phase.

the knee changes, so that the first vertical peak increased/decreased and the second vertical peak decreased/increased when shifting the prosthetic KJC forwards/backwards. Although the changes in the FPY were not pronounced in most subjects, this trend of change was observed and was consistent in all subjects.

The Effect on the Medio-Lateral Force (FPZ):

In all subjects, the medio-lateral force (FPZ) of the prosthetic leg showed a tendency to decrease/increase during the first half of stance phase as the KJC was shifted forwards/backwards from the normal position. This can be related to the changes in the horizontal force acting along the foot, and can be explained as seen in section 6.7.3.2 for the changes in the FPZ which resulted from the socket alignment changes.

On the sound leg, no trend of changes was found in the FPZ with the knee alignment changes, and the observed variations can be attributed to step to step variations.

6.7.4 Effect of Alignment Changes on the Ankle Joint Moments

6.7.4.1 Effect of Foot Alignment Changes

The effect of foot alignment changes on the prosthetic ankle joint moments is shown in figures 6.58 and 6.59 for two representative subjects, and the effect on the sound ankle joint moments is shown in figure 6.60 for one of the two subjects. A summary of the results for all subjects is shown in table 6.20.

On the prosthetic leg, the pattern and magnitude of the AP ankle joint moment (MAZ) did not noticeably change with the foot alignment changes. However, the magnitude of the ankle dorsiflexing moment during the push off phase, was slightly increased in some subjects (JLAQAF, ILBIAF, DLAQAF, LRBQAF, ILBQAF) as the alignment changes were towards plantarflexing the foot. This is related to the fact that plantarflexing the foot will make the prosthetic toes more supportive to the subject than that when the foot was in normal alignment at the push off phase (see section 6.7.2.1). Thus, the

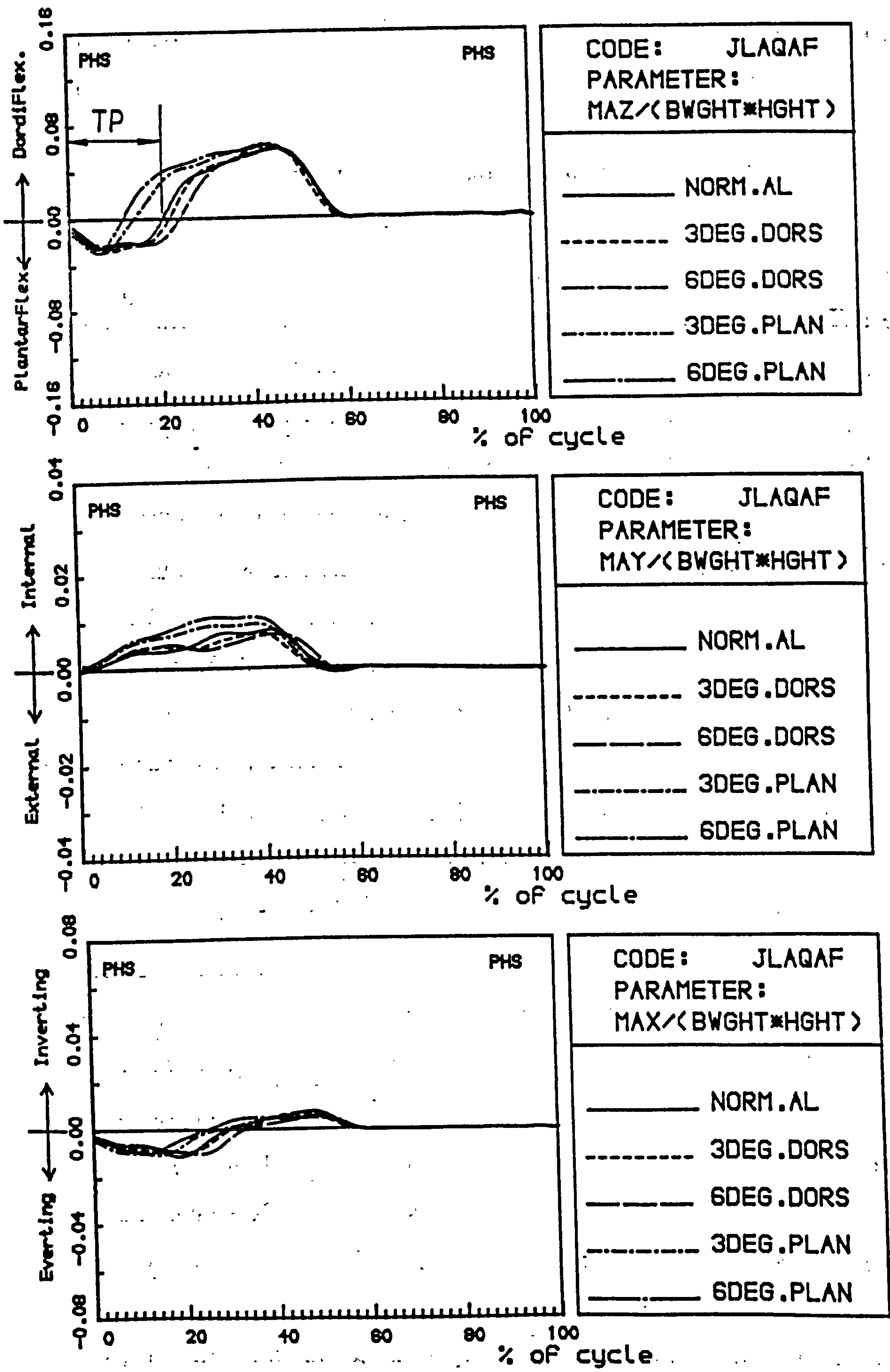


Figure 6.58 Effect of foot alignment changes on the variation with time of the ankle joint moments of the prosthetic leg for an AK amputee (subject JLAQAF).

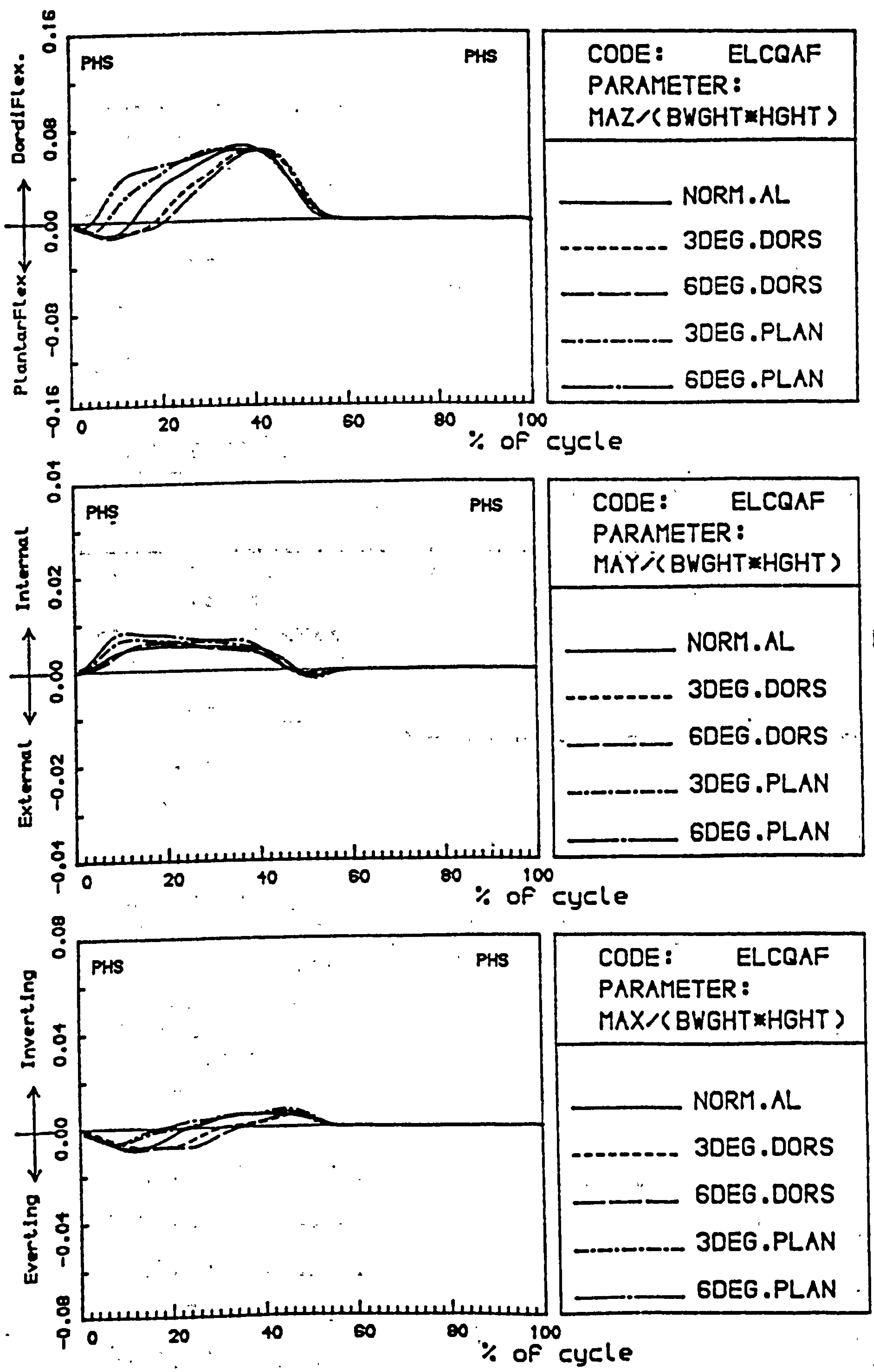


Figure 6.59 Effect of foot alignment changes on the variation with time of the ankle joint moments of the prosthetic leg for an AK amputee (subject ELCQAF).

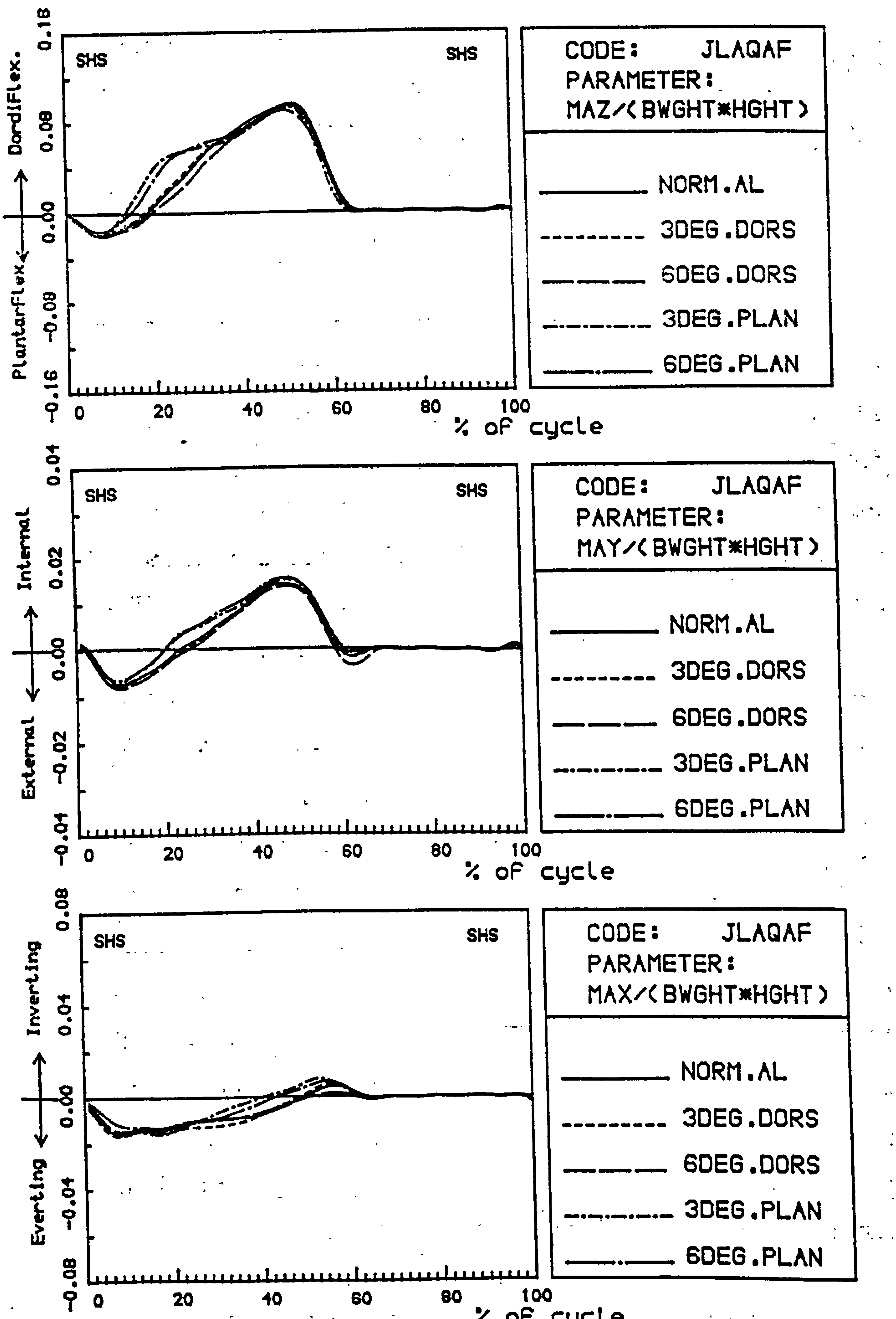


Figure 6.60 Effect of foot alignment changes on the variation with time of the ankle joint moments of the sound leg for an AK amputee (subject JLAQAF).

perpendicular distance between the AJC and the vector of the ground reaction force will increase with plantarflexing the foot.

It was also found that the magnitude of the plantarflexing moment of the prosthetic ankle at heel strike did not change with the foot alignment changes. However, the transition time (TP) (see fig. 6.58) during which the ankle moment changes from plantarflexing to dorsiflexing moment was noticeably affected by the foot alignment changes. Dorsiflexing the foot by 12 degrees (in increments of 3 degrees) delayed the TP of the prosthetic ankle moment in all subjects by an average of 20% (ranging from 9% to 31%, and proportional delay can be assumed of 5% per change) of stance phase. The delay in the TP shows that the subject was on the prosthetic heel for a longer time when the foot was dorsiflexed, than that when the foot was in its original position. This is related to the fact that dorsiflexing the foot will increase the foot-floor angle at heel strike; thus, the time consumed from heel strike until the foot is in full contact with the floor will be increased. Furthermore, because dorsiflexing the foot reduces the stability of the prosthetic knee, the patient will spend a long time on the heel to feel and ensure the knee stability before fully loading the prosthesis (delay in the occurrence of the first peak of FPY. See section 6.7.3.1). Thus, the TP will be delayed with dorsiflexing the foot.

In the transverse plane, no changes in the pattern of the prosthetic ankle joint moment (MAY) of any subject were found with the foot alignment changes. However, the magnitude of the internally rotating moment which dominated the MAY, slightly increased in all subjects (noticeably in subjects JLAQAF, TRAQAF and LRBQAF) during stance phase when the foot alignment changes were in a direction of foot plantarflexion. This may be attributed to the increase in the medio-lateral ground reaction force (FPZ) when plantarflexing the foot (see section 6.7.3.1). The increase in the FPZ is also responsible for the slight decrease in the everting moment (MAX) which was found at the prosthetic ankle of most subjects when plantarflexing the foot. As the increase in the FPZ was not noticeable during the second half of stance

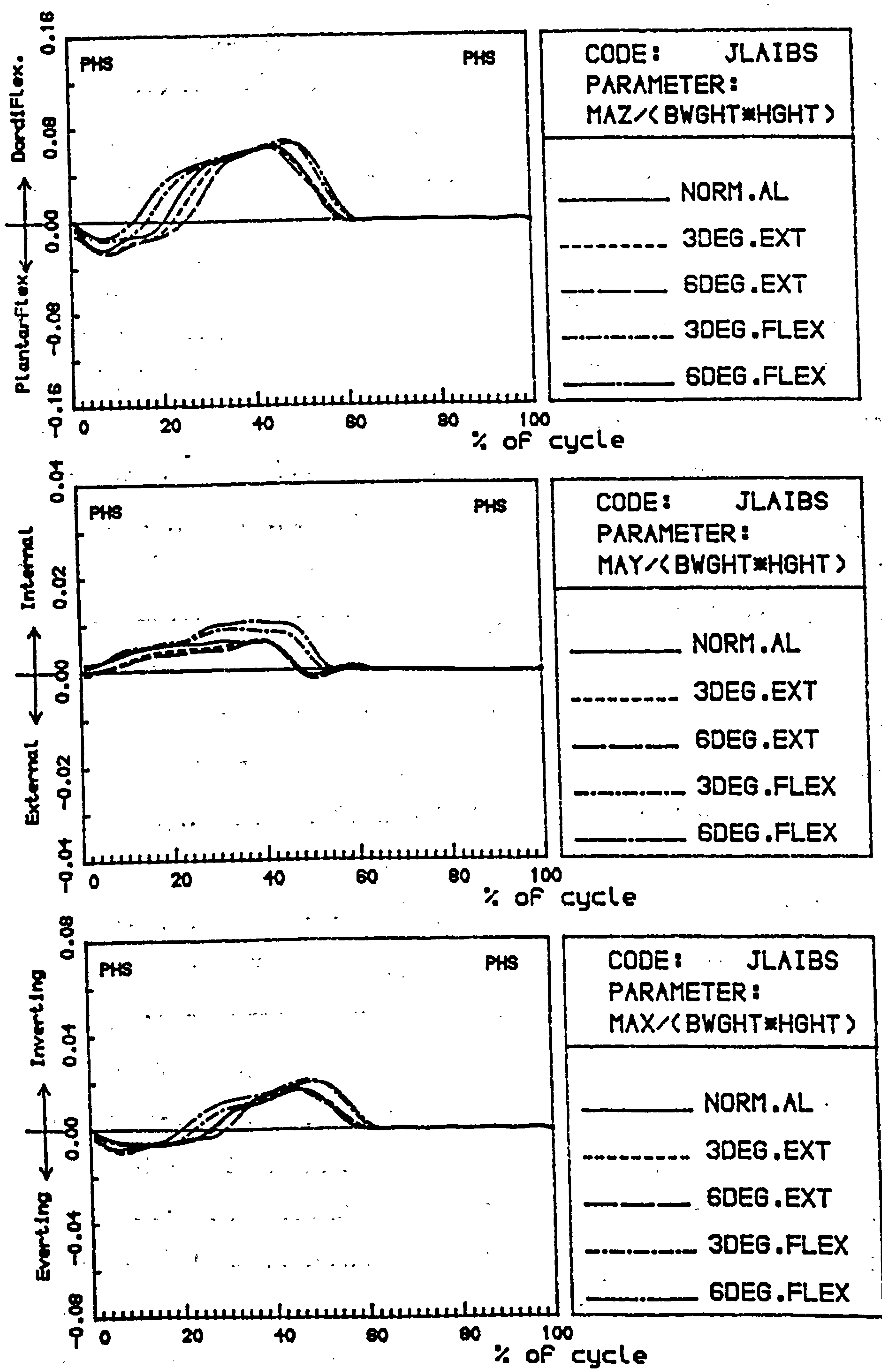


Figure 6.61 Effect of socket alignment changes on the variation with time of the ankle joint moments of the prosthetic leg for an AK amputee (subject JLAIBS).

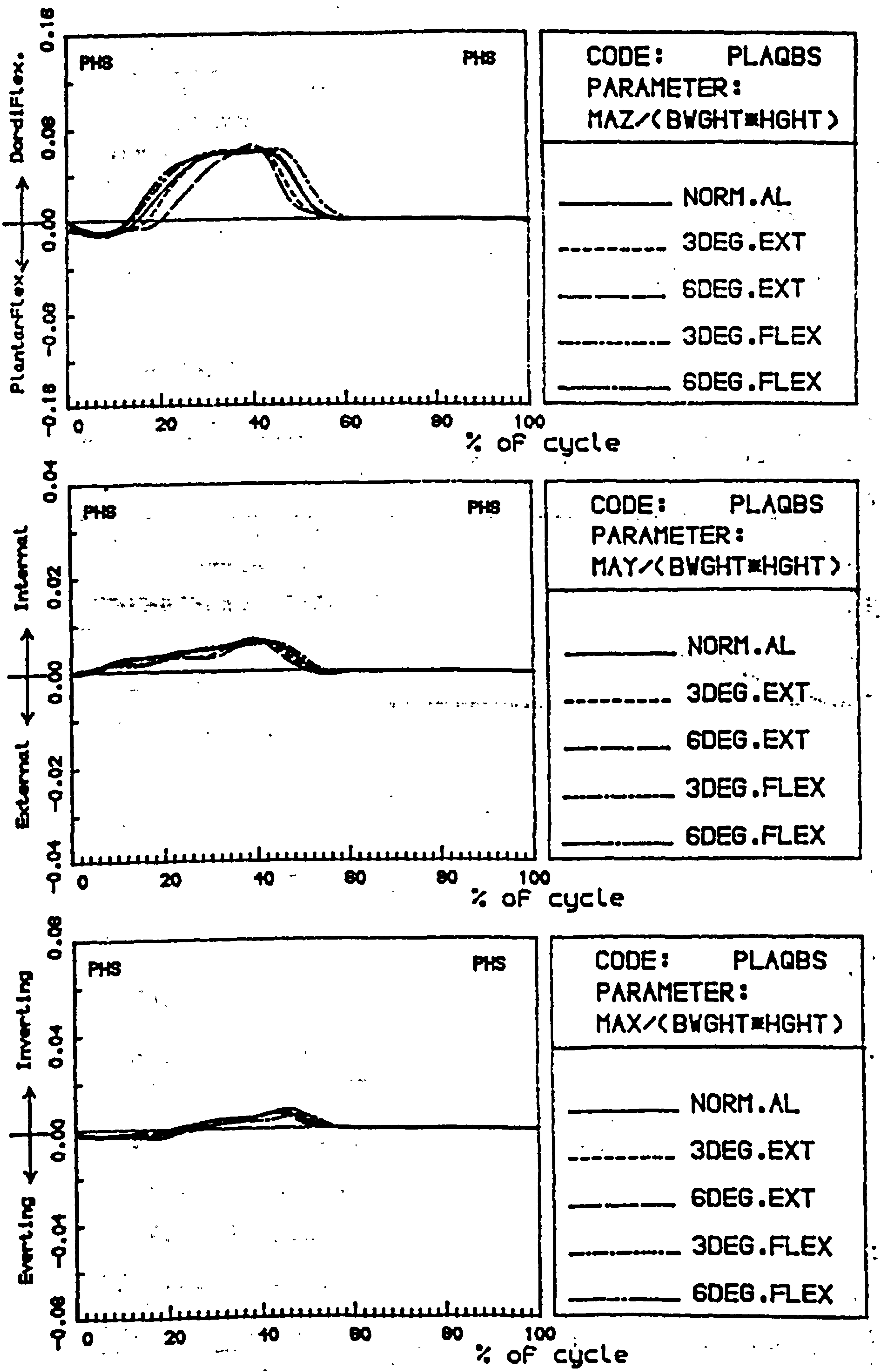


Figure 6.62 Effect of socket alignment changes on the variation with time of the ankle joint moments of the prosthetic leg for an AK amputee (subject PLAQBS).

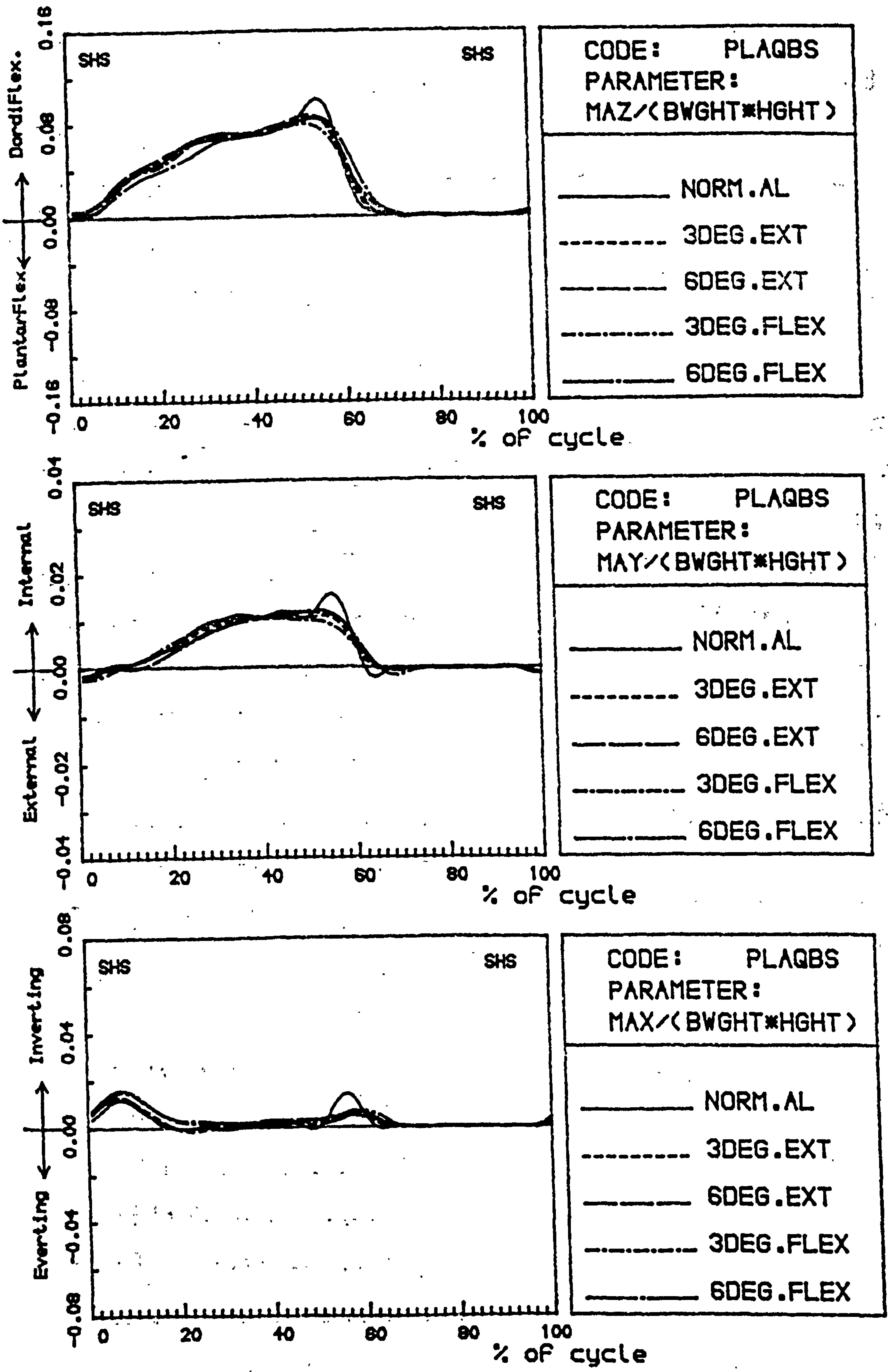


Figure 6.63 Effect of socket alignment changes on the variation with time of the ankle joint moments of the sound leg for an AK amputee (subject PLAQBS).

phase, its effect was not apparent on the inverting moment of MAX which existed during the second half of the stance phase.

On the sound side, the ankle joint moment showed no change in the transverse and coronal planes with foot alignment changes. In the AP plane, the plantarflexing moment of six subjects (JLAQAF, ILBQAF, LRBQAF, JLAIAF, PLAIAF, MRCQAF) increased slightly when dorsiflexing the foot. This is related to the increase in the braking force of the sound leg which resulted when the prosthetic foot was dorsiflexed (see section 6.7.3.1).

6.7.4.2 Effect of the Socket Alignment Changes

Figures 6.61 and 6.62 show the effect of socket alignment changes on the prosthetic ankle joint (AJ) moments of two subjects, and figure 6.63 shows the effect on the sound AJ moments of one subject, a summary of the results for all subjects is shown in table 6.20.

On the prosthetic side, no noticeable changes were found in the pattern and magnitude of the MAZ with the socket alignment changes, and the only detected change was in the transition time (TP). It was found that extending the socket by 12 degrees (in increments of 3 degrees) delayed the TP in all subjects by an average of 15% (ranging from 11% to 20%, and proportional changes can be assumed of 3.8% per 3 degrees change) of stance phase. This is related to the fact that extending the socket will increase the foot-floor angle at the heel strike, thus, a longer time will be needed for the period from heel strike until the foot is in full contact with the floor. This time is accounted as the main part of TP.

In the transverse plane (MAY), the internal rotating moment which dominated the MAY of the prosthetic leg showed a tendency to increase (subjects JLAIBS, MRCQBS, ELCQBS, LRBQBS, JLAQBS) with flexing the socket as seen in figure 6.61. This may be related to the fact that flexing the socket may increase the foot-floor angle at push off phase (similar to the foot plantarflexion case). Therefore, the force component which is parallel to the X axis of the shank, and acting at the foot centre of pressure in a manner that

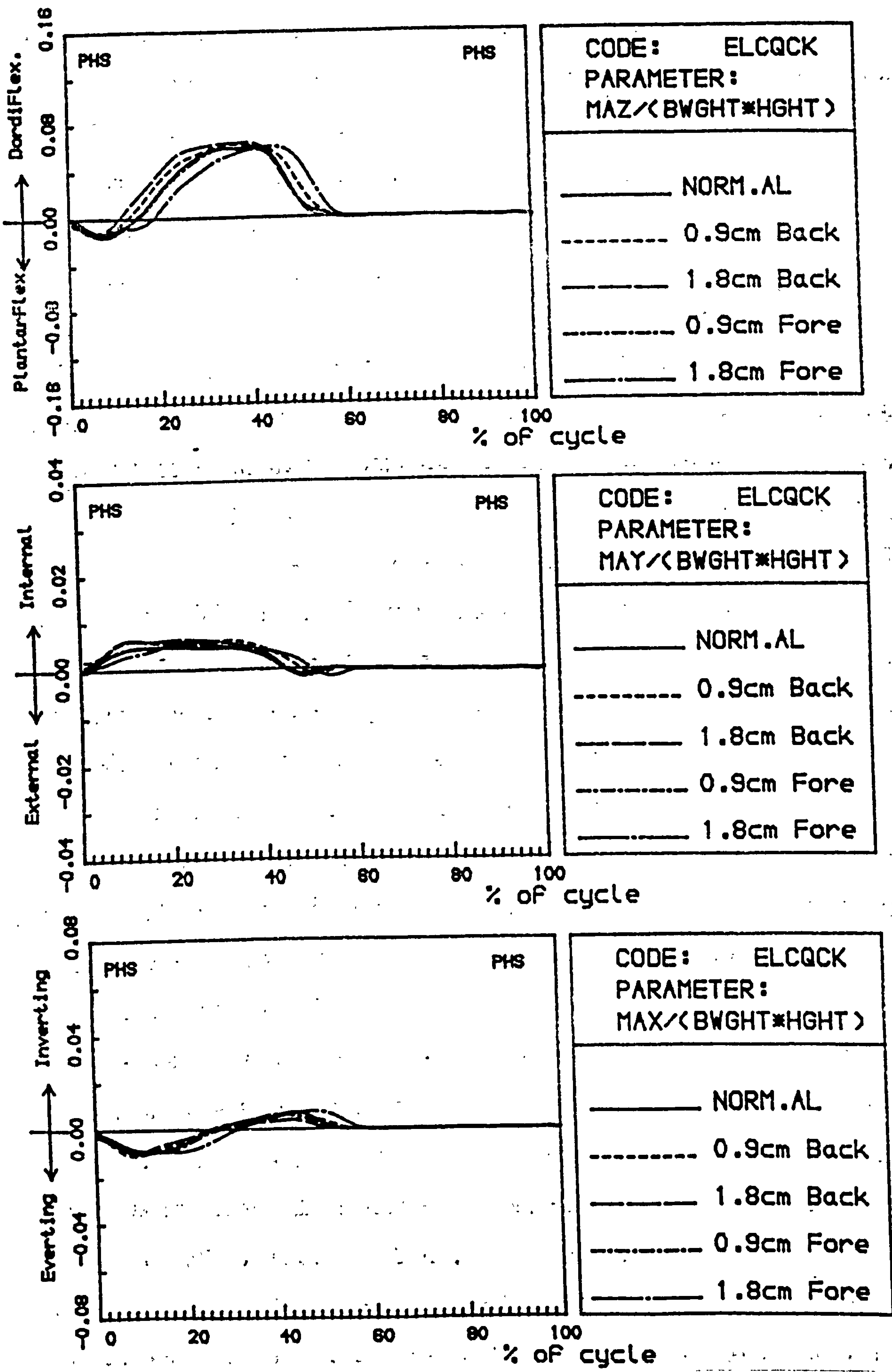


Figure 6.64 Effect of knee shifts on the variation with time of the ankle joint moments of the prosthetic leg for an AK amputee (subject ELCQCK).

increases the internally rotating moment of the ankle joint at push off phase, will be increased. This is supported by the fact that Zahedi et al (1988) found that the FX of the shank increased when the foot was plantarflexed (see fig. 3.37)

On the coronal plane (MAX), some variations were found in the prosthetic MAX when the socket alignment was changed, however, these variations were small in magnitude, inconsistent, and were not found in all subjects. Therefore, no conclusion could be reported on the MAX with socket alignment changes.

On the sound leg, the only detected change was in the ankle plantarflexing moment. Most subjects showed a slight increase in that moment when the socket was flexed. This is related to the increase in the braking force of the sound leg which was found (see section 6.7.3.2) when flexing the socket.

6.7.4.3 Effect of the Knee shifts

The effect of the knee changes on the prosthetic ankle joint moments is shown in figures 6.64 and 6.65 for two representative subjects. The effect on the sound ankle joint moments is shown in figure 6.66 for one of the two subjects, a summary of the results for all subjects is shown in table 6.20.

On the prosthetic side, although the braking and push off forces of the prosthetic leg changed with the knee shifts, the magnitudes of the prosthetic ankle plantar/dorsiflexing moments did not change with the knee shifts. This can be explained by the fact that FPY is the main factor for the MAZ, and small changes in the FPX may not affect the MAZ. The only change detected in the MAZ on the prosthetic leg is in the transition time (TP). It was found that shifting the KJC forwards relative to the hip-ankle line by 3.6 cm (in increments of 0.9 cm) delayed the transition time in all subjects by an average of 11.4% (ranging from 4% to 13%, and proportional changes can be assumed) of stance phase. This is related to the fact that shifting the KJC forwards relative to the hip-ankle line will increase the height of the prosthesis (and the height of the HJC) therefore, the subject will have to put his foot further

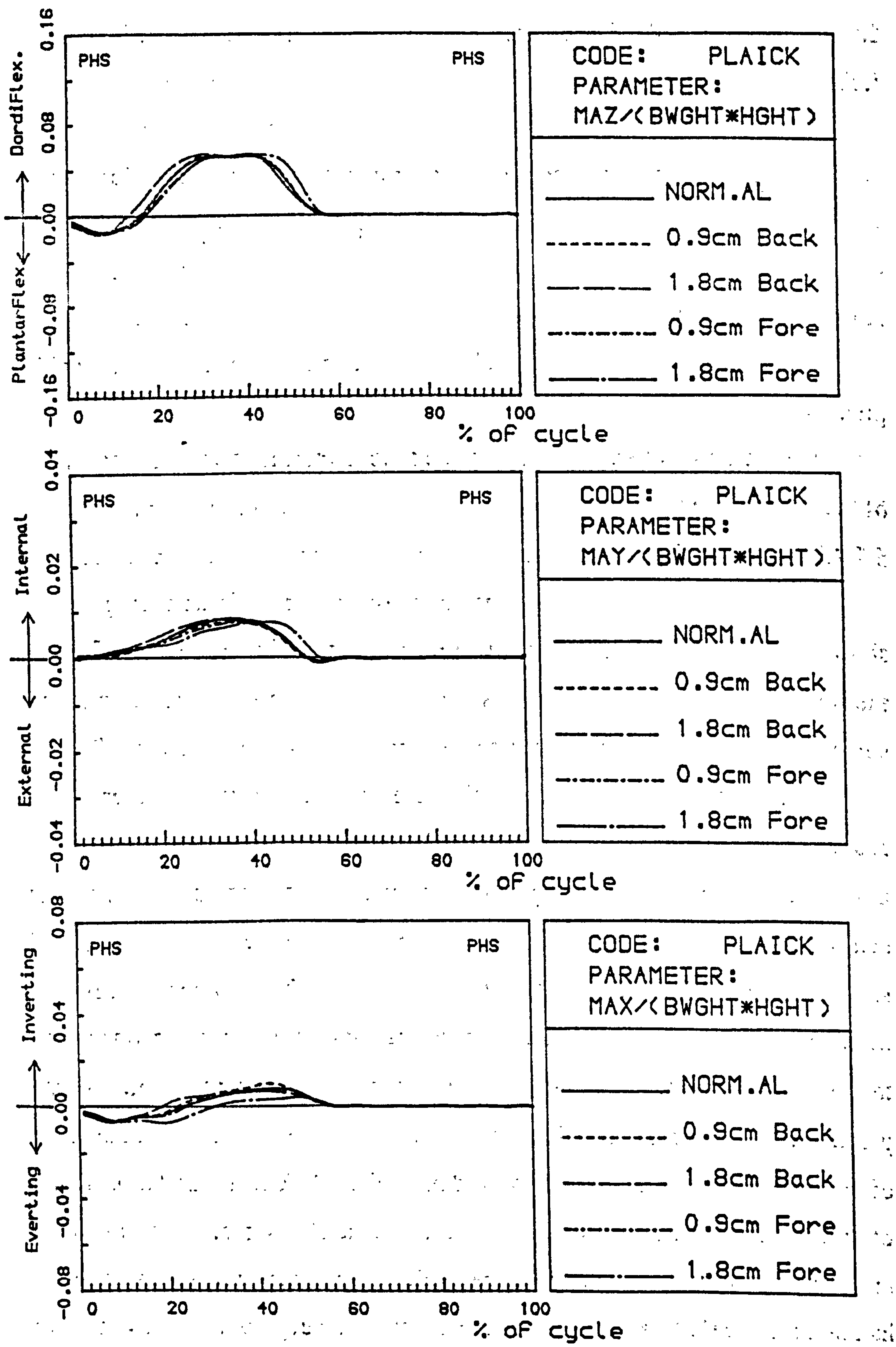


Figure 6.65 Effect of knee shifts on the variation with time of the ankle joint moments of the prosthetic leg for an AK amputee (subject PLAICK).

forwards in order to maintain the original height of the HJC at heel strike (see section 6.7.2.3), so that the push off force of the sound leg will be maintained. This (putting the foot further forwards) will reduce the stability of the prosthesis and increase the foot-floor angle at heel strike and therefore, the transition time will be delayed.

In the transverse plane, the internal rotating moment which dominated the MAY of the prosthetic ankle joint, exhibited a tendency to decrease in all subjects as the KJC was located anteriorly. This is related to the decrease in the medio-lateral ground reaction force (FPZ) which was found in all subjects (see section 6.7.3.3) as the KJC was located anteriorly relative to the hip-ankle line.

In the coronal plane (MAX), inconsistent variations were found in the MAX of the prosthetic leg with the knee shifts. However, some patients (PLAICK, LRBQCK, PLAQCK, JLAICK) showed a trend of increase/decrease in the eversion/inversion moment of the prosthetic ankle joint during stance phase as the KJC was located anteriorly relative to the hip-ankle line. This is also related to the decrease in the FPZ of the prosthetic leg which was found when locating the KJC anteriorly relative to the hip-ankle line, because the FPZ is working against/with FPY in generating the eversion/inversion moment at the ankle joint during stance phase.

On the sound side, shifting the prosthetic KJC backwards by 1.8 cm (in increments of 0.9 cm) showed no noticeable change in the MAZ of the sound leg. However, the plantarflexing moments of only three subjects (JLAICK, ELCQCK, TRAQCK) decreased by an average of 27.7% (ranging from 12% to 43%) of that when the knee was in normal alignment. Shifting the prosthetic KJC forwards by 1.8 cm (in increments of 0.9 cm) affected the MAZ of the sound leg in three subjects only (JLAICK, JLAQCK, ELCQCK), and no changes were found in the MAZ of the eight remaining subjects. It was found that the plantarflexing/dorsiflexing moments of the sound ankle joint increased/decreased in the three above mentioned subjects by an average of

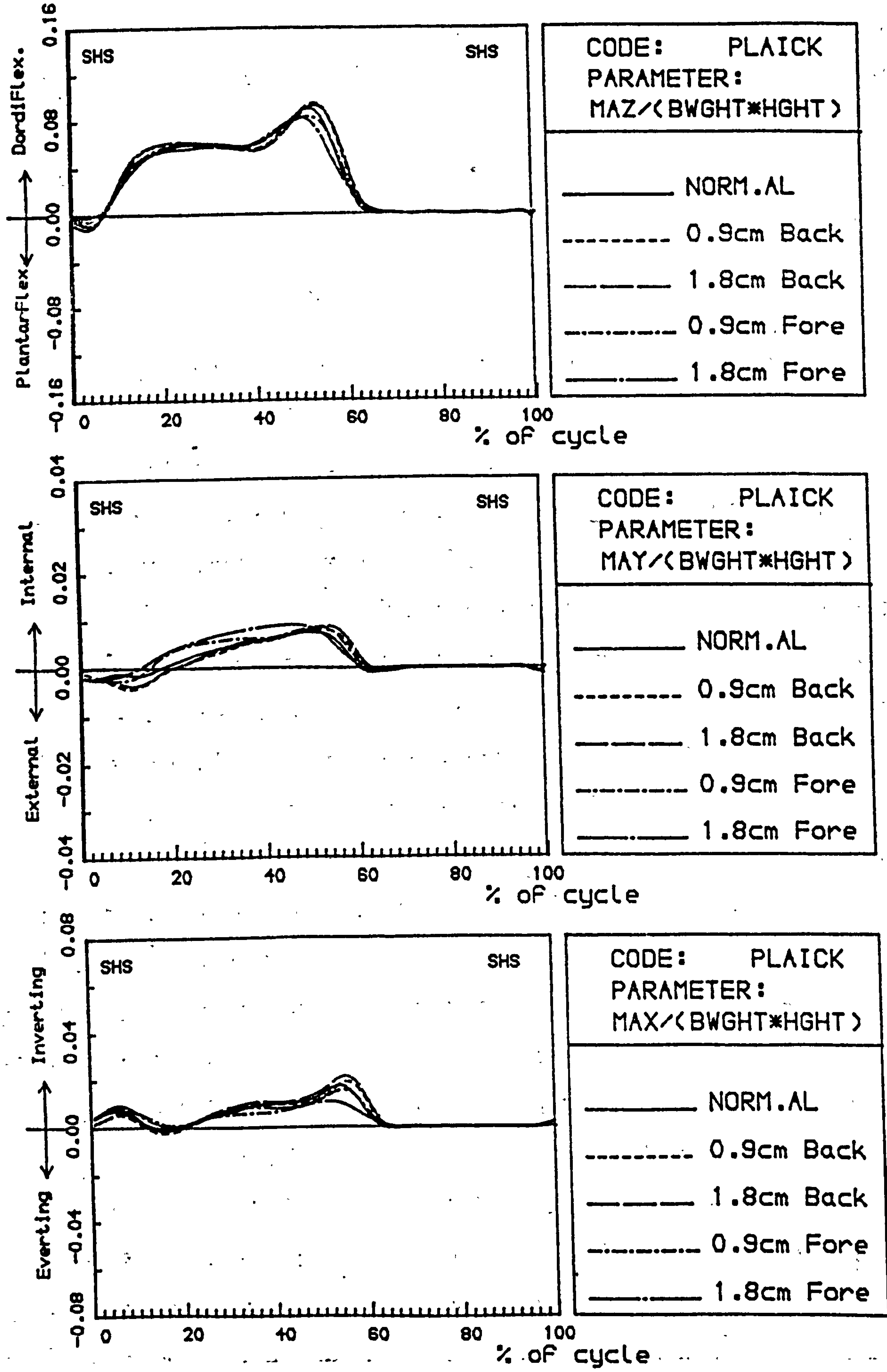


Figure 6.66 Effect of knee shifts on the variation with time of the ankle joint moments of the sound leg for an AK amputee (subject PLAICK).

Table 6.21 Effect of alignment changes on the knee joint moments.

Knee Joint	A/P Moment[1] of the Knee Joint (MKZ)		Coronal Moment[1] (MKX)
	Transition Time (TE)	Extending Moment at Stance Phase	
Effect of the Foot Changes[2]			
Prosthetic	+ in all amp. by approximately[4] 10% to 30%*.	No noticeable changes	+ slightly in all amp with plantarflexing the foot.
Sound	Slight increase/decrease in the knee flexing moment after HS, when dorsiflexing/plantarflexing the foot.		
Effect of Socket Changes[2]			
Prosthetic	Trend of slight decrease/increase when flexing/extending the socket.	+ in 9 amp. (26% -90%)* with 6 deg. socket ext. - in 7 amp. (10% -35%)* with 6 deg. socket flex.	No noticeable changes
Sound	Slight increase/decrease in the knee flexing moment after HS, when flexing/extending the socket.		
Effect of the Knee Shift[2]			
Prosthetic	- in 10 amp. (13%-72%)* with the back shift[3]. + in 5 amp. (9%-40%)* with the forward shift[3].	+ in 10 amp. (10%-105%)* with the back shift[3]. - in 10 amp. (14%-50%)* with the forward shift[3].	Slight increase in the MKX of most amputees when locating the KJC posteriorly.
Sound	Slight increase/decrease in the knee flexing moment which exists after HS, when shifting the KJC forwards/backwards.		

1 The knee moment in the transverse plane (MKY), showed similar change to MAY as they expressed in the frame reference of the same segment.
2 No changes were found on the knee moments of the prosthetic and sound leg during swing phase.
3 The shift was by 1.8 cm (in increments of 0.9 cm) relative to the hip-ankle line. 4 The foot was dorsiflexed by 12 degrees, in increments of 3 degrees.
TE is the transition time at which the knee extending moment starts to build up. + Increased - Decreased amp. amputee.
* Changes are expressed as a percentage of the normal value.

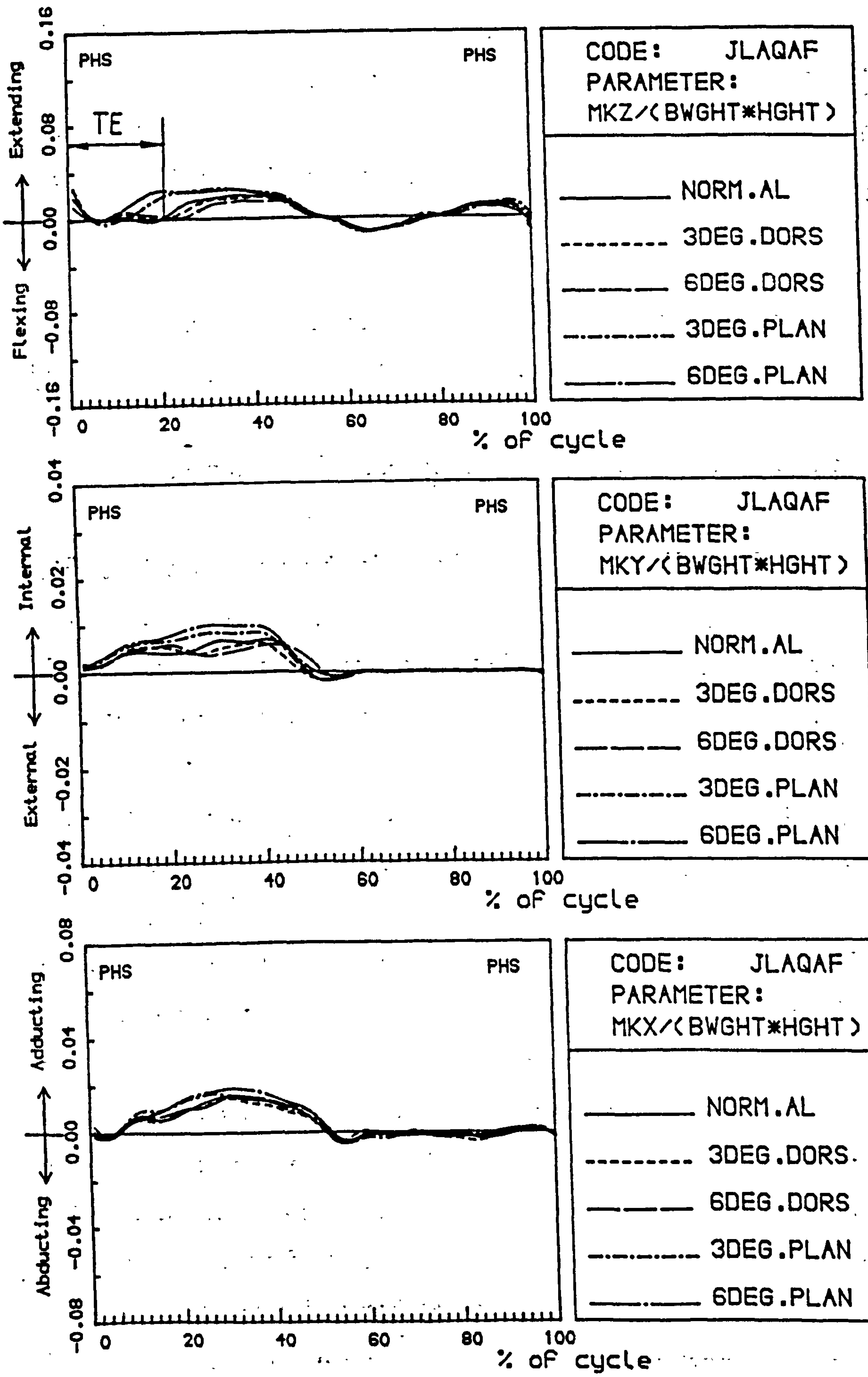


Figure 6.67 Effect of foot alignment changes on the variation with time of the knee joint moments of the prosthetic leg for an AK amputee (subject JLAQAF).

43%/17.7% of that when the knee was in normal alignment. These changes in the MAZ of the sound leg are attributed to changes in the fore-and-aft force which were found at the sound leg with the knee alignment changes. An unexpected change was found in the TP of the sound side of four subjects (ILAQCK, JLAQCK, ELCQCK, TRAQCK), namely the TP of these subjects delayed by an average of 9% of stance phase as the prosthetic KJC was located anteriorly by 3.6 cm (in increments of 0.9 cm).

In the coronal (MAX) and transverse (MAY) planes, inconsistent changes were found in the ankle joint moments of the sound leg, this is related to the inconsistent variations which were found in the FPZ of the sound leg with the prosthetic KJC shifts.

6.7.5 Effect of Alignment Changes on the Knee Joint Moments

6.7.5.1 Effect of the Foot Alignment Changes

The effect of the foot alignment changes on the knee joint moments of the prosthetic leg is shown in figures 6.67 and 6.68 for two representative subjects, and the effect on the sound knee joint moments is shown in figure 6.69. A summary of the results of all subjects is shown in table 6.21.

On the prosthetic side, no noticeable changes were found in the three moments at the knee joint during swing phase. During stance phase, the AP knee moment (MKZ) showed no change in the pattern and magnitude of the extending moment which dominated the stance phase. However, the transition time TE (see fig. 6.67) at which the knee extending moment starts to build up, was delayed in all subjects by a range from 10% to 30% of that when the foot was in normal alignment, when the foot was dorsiflexed by 12 degrees (in increments of 3 degrees). The changes in the TE of the prosthetic knee moment coincided with and are approximately equal to those found in the TP of the prosthetic ankle joint moment, and the reason for the TE changes is similar to that of the TP changes which was discussed in section 6.7.4.1.

In the transverse plane, the prosthetic knee moment (MKY) showed

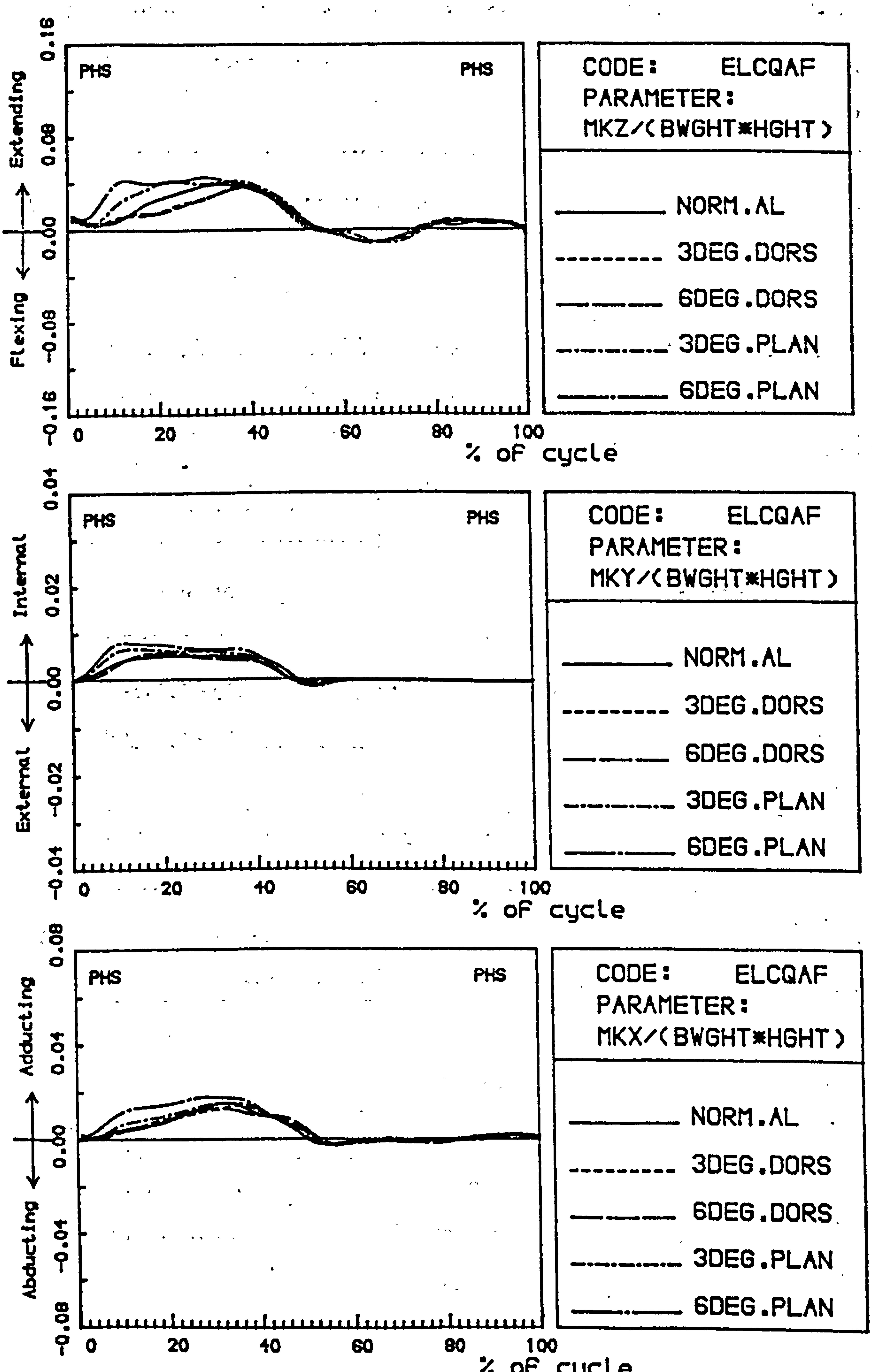


Figure 6.68 Effect of foot alignment changes on the variation with time of the knee joint moments of the prosthetic leg for an AK amputee (subject ELCQAF).

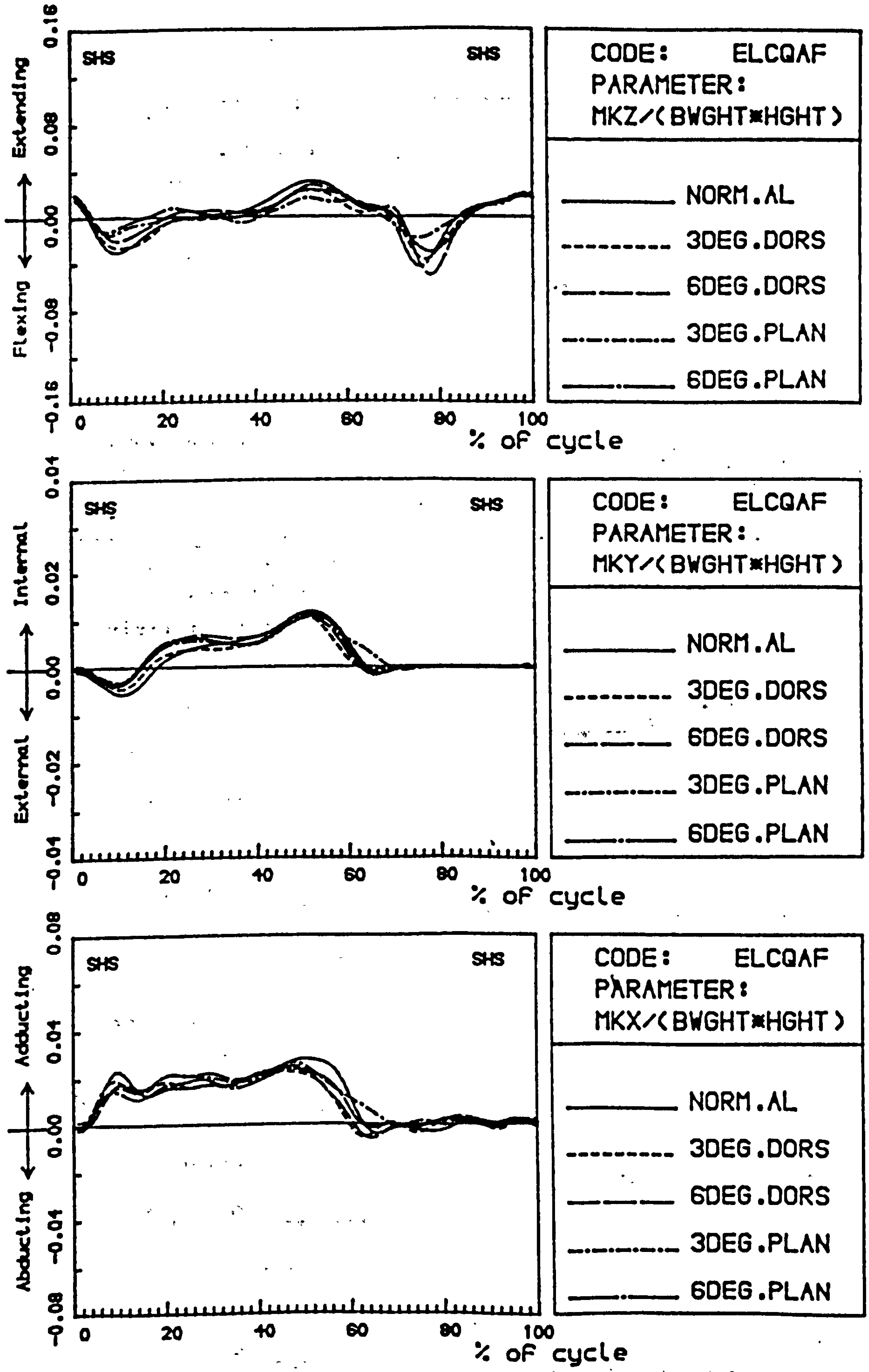


Figure 6.69 Effect of foot alignment changes on the variation with time of the knee joint moments of the sound leg for an AK amputee (subject ELCQAF).

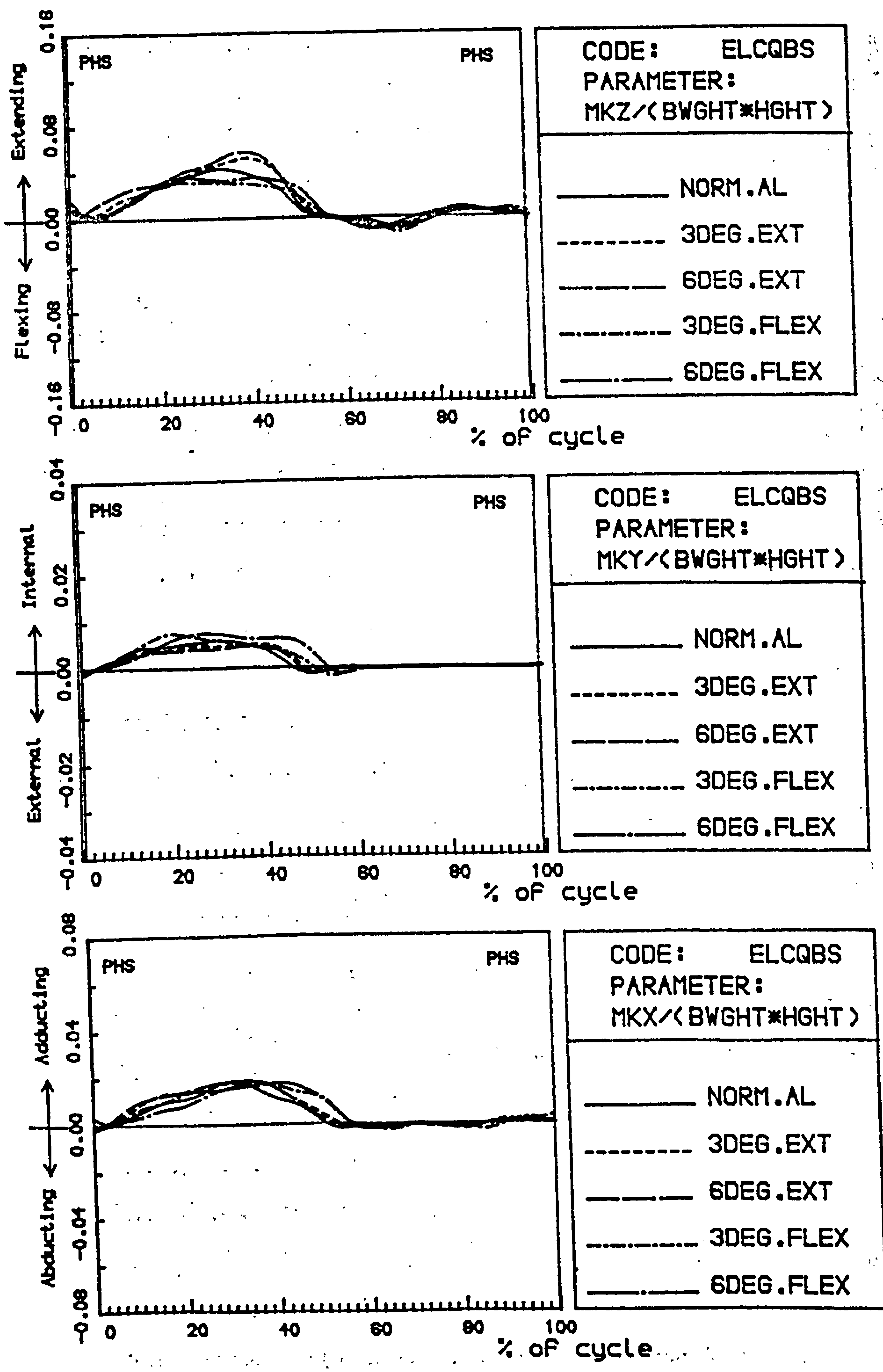


Figure 6.70 Effect of socket alignment changes on the variation with time of the knee joint moments of the prosthetic leg of an AK amputee (subject ELCQBS).

similar changes to that exhibited by the ankle moment (MAY) with the foot alignment changes. This was expected because the ankle and knee moments were calculated relative to the shank frame of reference and the mass moment of inertia about the Y axis (I_{yy}) was considered negligible.

On the coronal plane, the knee moment (MKX) of the prosthetic leg exhibited a slight increase in all subjects when plantarflexing the foot. This is attributed to the increase in the medio-lateral ground reaction force which was found at the prosthetic leg when the foot was plantarflexed.

On the sound side, the foot alignment changes had no noticeable effect on the three moments of the knee joint. However, most subjects exhibited a slight increase/decrease in the knee flexing moment after heel strike when the prosthetic foot was dorsiflexed/plantarflexed from its normal position. This is related to the increase/decrease in the braking force which was found on the sound leg as the prosthetic foot was dorsiflexed/plantarflexed from the normal position.

6.7.5.2 Effect of the Socket Alignment Changes

The effect of socket alignment changes on the prosthetic knee joint moments is shown in figures 6.70 and 6.71 for two representative subjects, and figure 6.72 shows the effect on the sound leg for one of the two subjects. A summary of the results of all subjects is shown in table 6.21.

On the prosthetic leg, during swing phase, no noticeable changes were found on the three knee moments. In the coronal plane (MKX), during stance phase, the knee moment was not noticeably changed and only slight variations were obtained, these variations can be attributed to the slight change in the prosthetic FPZ which was obtained with the socket alignment changes. In the transverse plane, the changes in the prosthetic knee joint moment (MKY) were similar to those found in the prosthetic ankle joint moment (MAY) which were discussed in section 6.7.4.2. This is because MAY and MKY were expressed in the shank frame of reference and the inertia effects were neglected.

In the AP plane, the knee joint moment (MKZ) showed noticeable

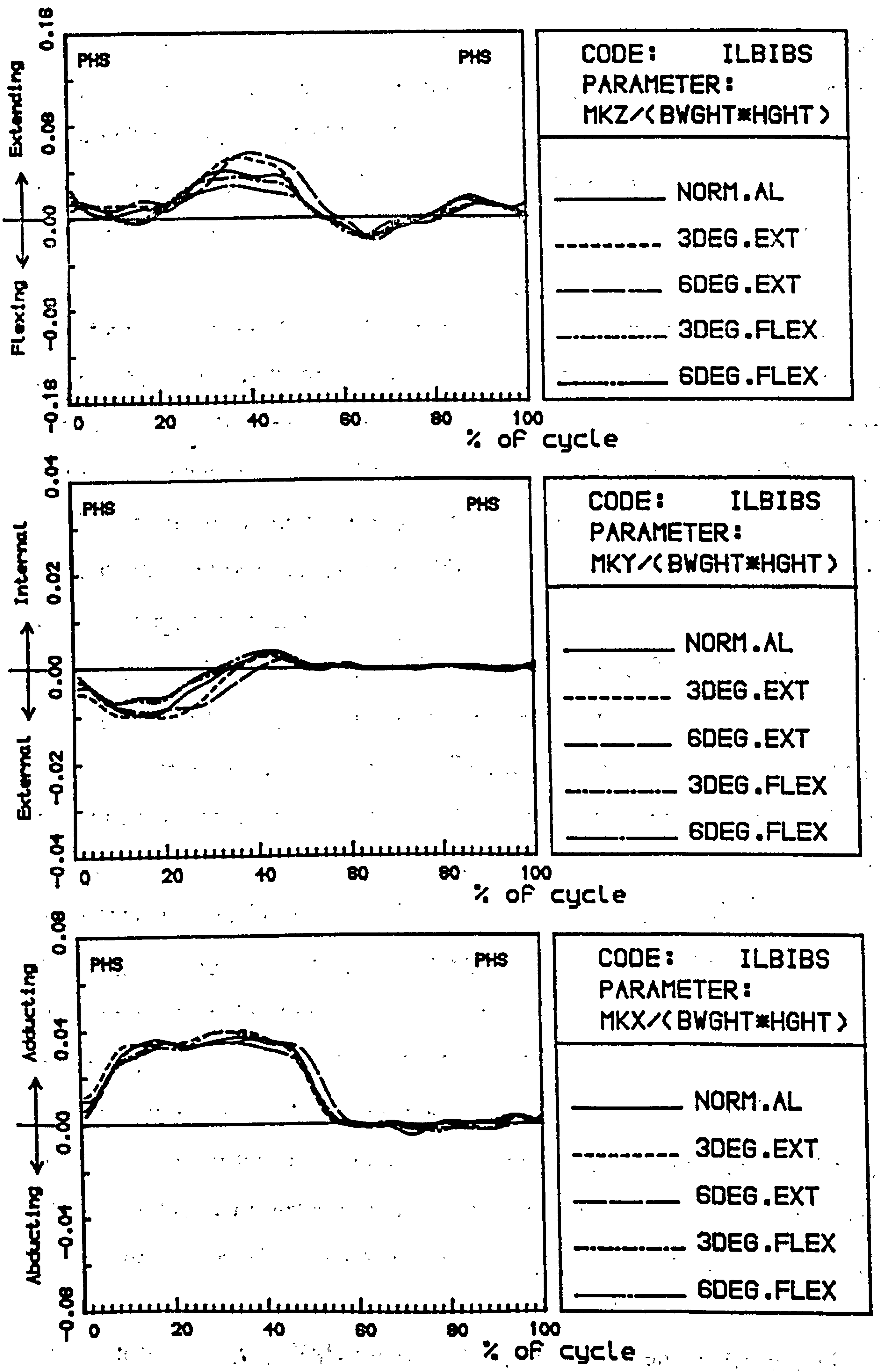


Figure 6.71 Effect of socket alignment changes on the variation with time of the knee joint moments of the prosthetic leg for an AK amputee (subject ILBIBS).

changes during stance phase with socket alignment changes, and these changes were mostly noticed on the pattern of the knee moment, and on the magnitude of the peak of the knee extending moment, which occurred just before the push off phase. The pattern of the knee extending moment was changed during stance phase, in that its shape changed from triangular to rectangular like when the socket was flexed from the normal position. When the socket was extended by 6 degrees (in increments of 3 degrees) from the normal position, the peak of the extending moment during the stance phase increased in nine subjects by an average of 49.9% (ranging from 26% to 90%, and proportional changes can be assumed) of the moment when the socket was in normal alignment. The two remaining subjects showed no change. Flexing the socket by 6 degrees (in increments of 3 degrees) from the normal alignment, decreased the knee extending moment at the push off phase in seven subjects by an average of 24.3% (ranging from 10% to 35%, and proportional changes can be assumed) of the moment when the socket was in normal alignment. No changes were found in three subjects (JLAIBS, ILBQBS, JLAQBS). The remaining subject (MRCQBS) showed an increase in the knee extending moment by 70% of that when the socket was in normal alignment.

The decrease/increase of the knee extending moment during stance phase is related to the decrease/increase in the perpendicular distance between the KJC and the vector of the ground reaction force when the socket is flexed/extended. The change in the perpendicular distance between the KJC and the force vector is influenced by the change in the trunk angular position θF/θE (see fig. 6.40) which was achieved by the subject in order to compensate for the flexion/extension of the socket. The change in the knee moment which resulted by extending the socket by 6 degrees (49.9%) is approximately double the change which resulted by flexing the socket by 6 degrees (24.3%). This is related to the fact that flexing the socket produces knee instability and therefore, the subject will have to flex his trunk in order to bring the vector of the ground reaction force ahead of the KJC (see section 6.7.1.2). In the case

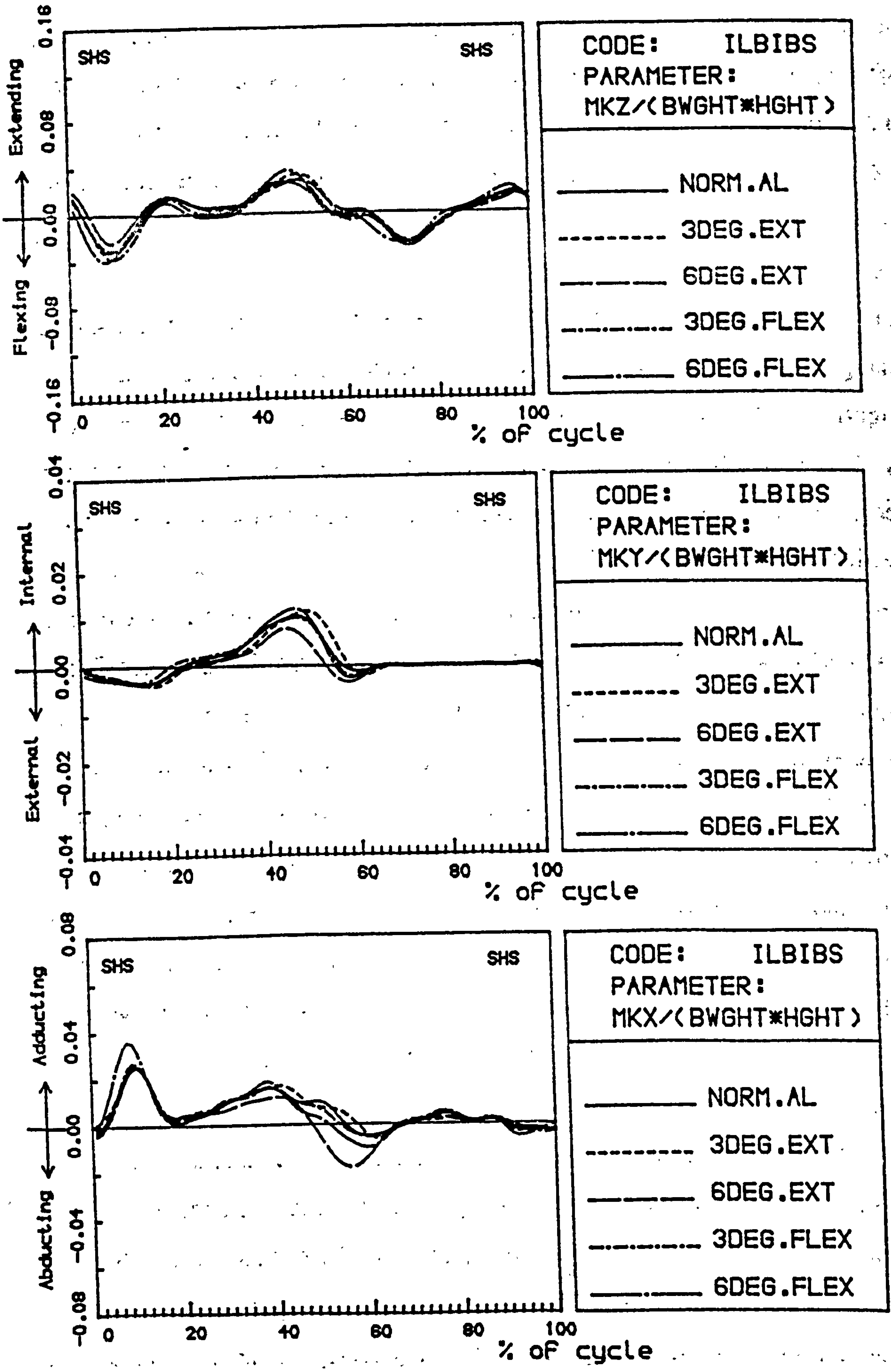


Figure 6.72 Effect of socket alignment changes on the variation with time of the knee joint moments of the sound leg for an AK amputee (subject ILBIBS).

of extending the socket, the subject dose not have to change the trunk position because the knee stability is maintained, and therefore, the change in the perpendicular distance between the KJC and the force vector will be larger when extending the socket than that when flexing the socket. This explanation is supported by the fact that five subjects only (section 6.7.1.2) showed a slight extension in the trunk position when extending the socket, and subject ILBQBS who showed the largest trunk extension showed no change in the knee moment when extending the socket.

The transition time (TE) of the prosthetic knee joint moment, showed a slight trend of advance/delay when the socket was extended/flexed. This is related to the fact that flexing the socket would decrease the stability of the prosthesis. Thus, the patient will spend a longer time dwelling on the heel than when the socket was in normal alignment, to ensure stability before applying full loading on the prosthesis.

On the sound leg, no noticeable changes were found in the three moments of the knee joint. However, it can be reported that the knee flexing moment which exists after heel strike was slightly increased/decreased as the socket was flexed/extended from the normal position. This is related to the increase/decrease in the braking force of the sound side which was found (section 6.7.3.2) when flexing/extending the socket from the normal position.

6.7.5.3 Effect of the Knee Shifts

Figures 6.73 and 6.74 show the effect of the knee shifts on the prosthetic knee joint moments of two subjects. Figure 6.75 shows the effect on the sound knee joint moments for one subject, A summary of the results of all subjects is shown in table 6.21.

On the prosthetic side, during the swing phase, the three knee moments showed no change when the knee was shifted. In the AP plane, shifting the knee backward by 1.8 cm (in increments of 0.9 cm) from the normal position relative to the hip-ankle line, increased the peak value of the knee extending moment during the stance phase in ten subjects by an average of 52.5%

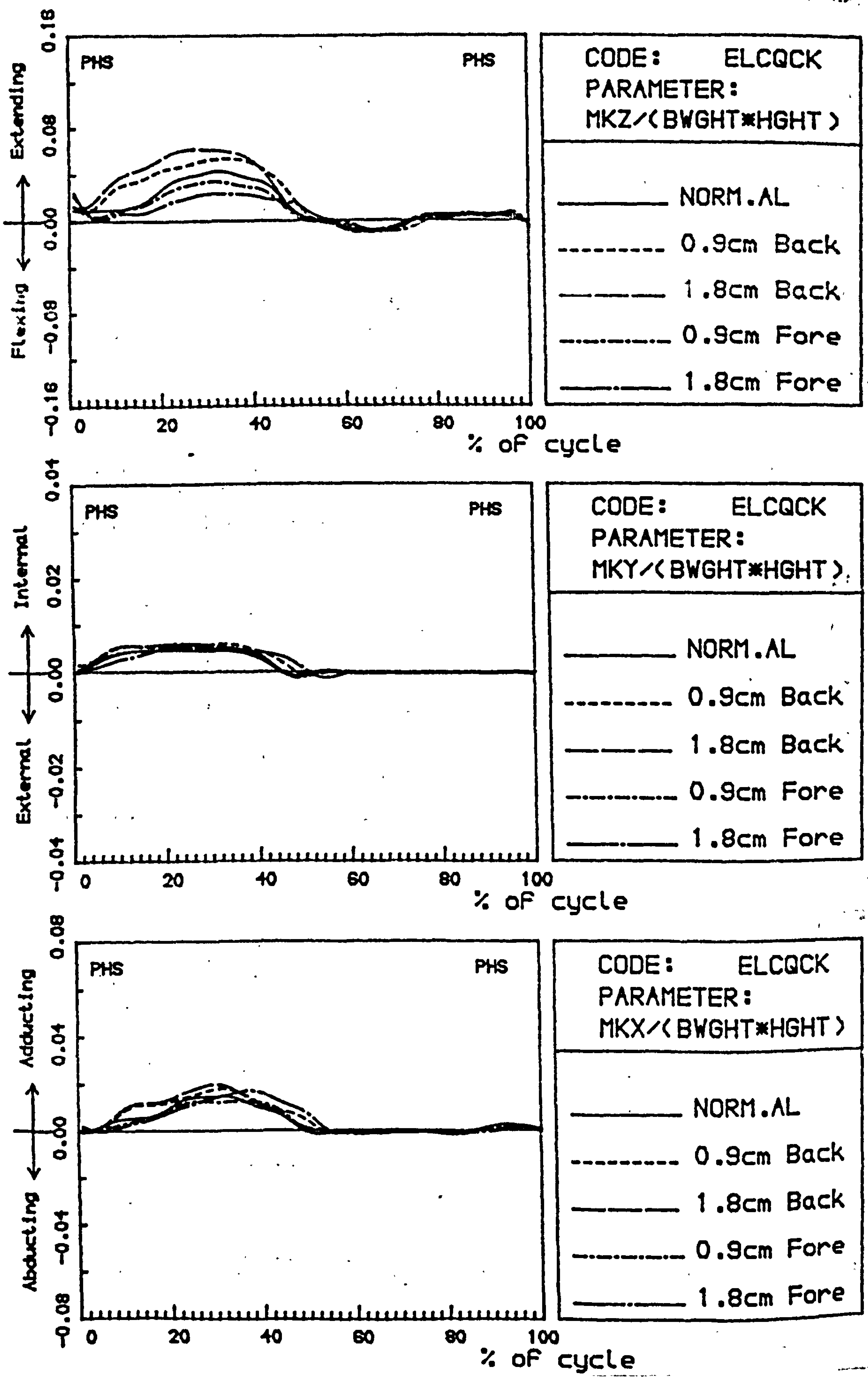


Figure 6.73 Effect of knee shifts on the variation with time of the knee joint moments of the prosthetic leg for an AK amputee (subject ELCQCK).

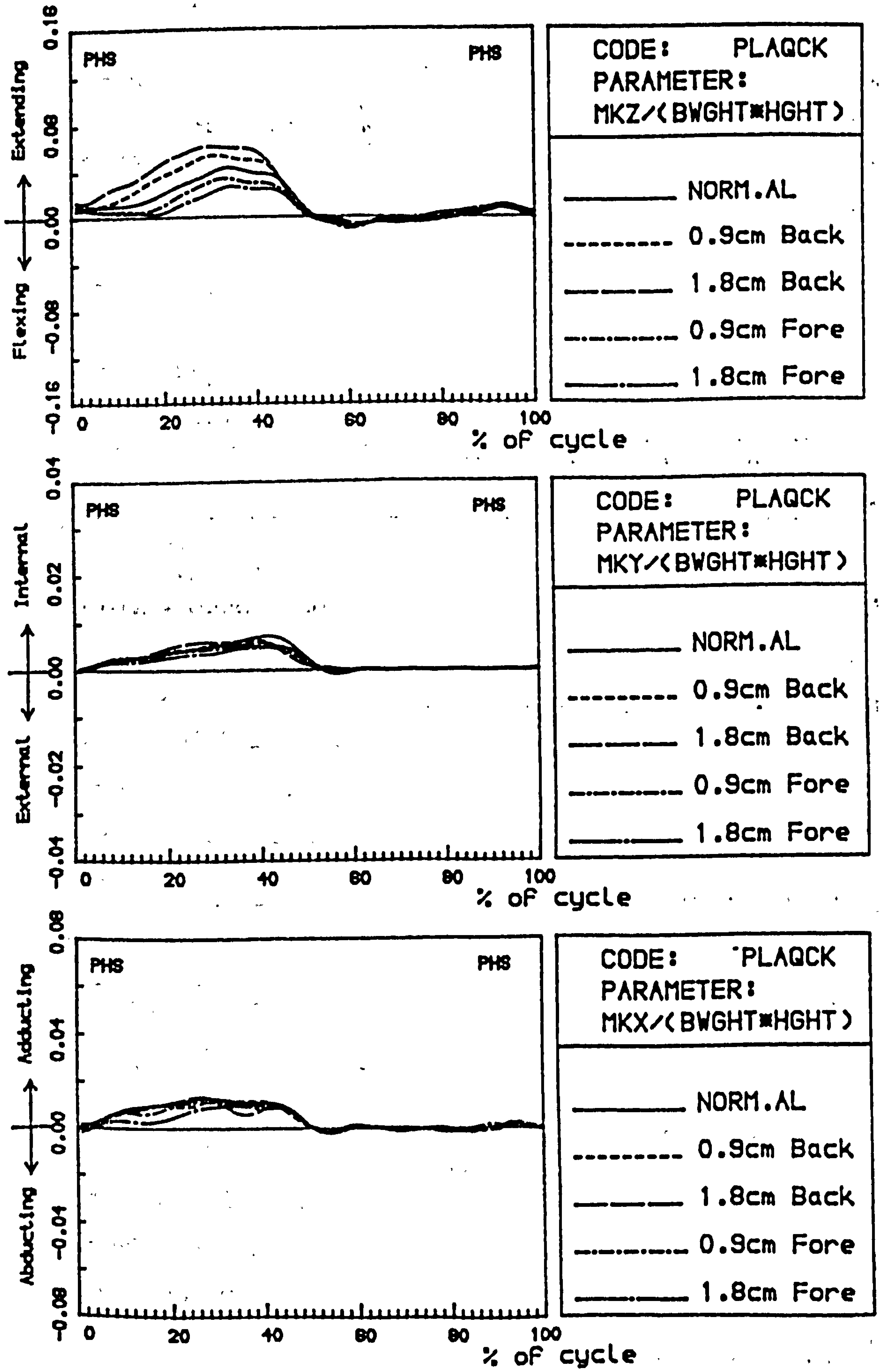


Figure 6.74 Effect of knee shifts on the variation with time of the knee joint moments of the prosthetic leg for an AK amputee (subject PLAQCK).

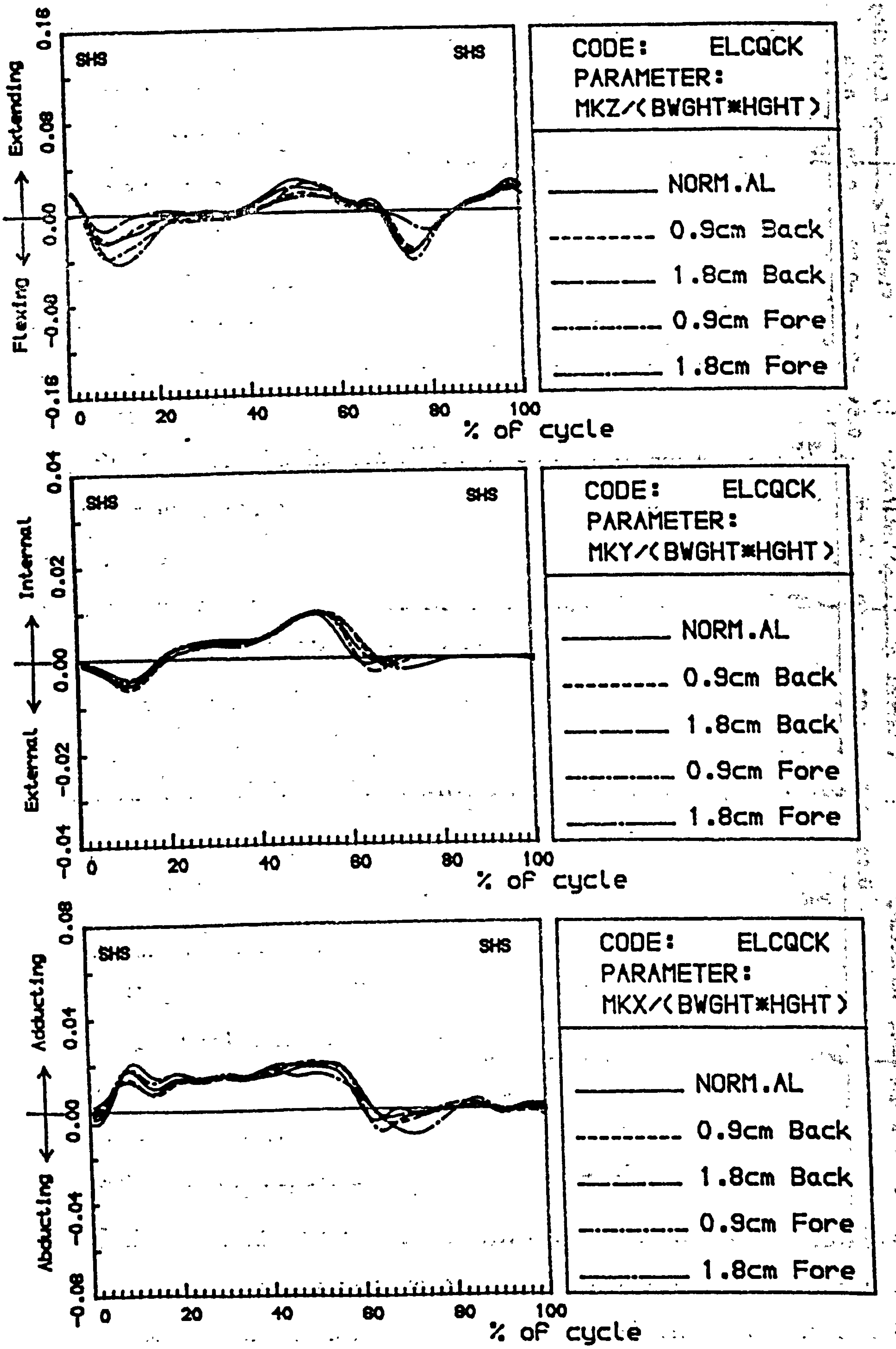


Figure 6.75 Effect of knee shifts on the variation with time of the knee joint moments of the sound leg for an AK amputee (subject ELCQCK).

(ranging from 10% to 105%, and proportional changes can be assumed) of that when the knee was in normal alignment, and no change was found in the knee moment of the remaining subject (ILBQCK). Shifting the KJC forwards by 1.8 cm (in increments of 0.9 cm) from the normal position, decreased the prosthetic knee extending moment during the stance phase in ten subjects by an average of 34.5% (ranging from 14% to 50%) of that when the KJC was in normal alignment, and no change was found in the knee moment of the remaining subject (JLAQCK). Thus, the AP knee moment of the prosthetic leg decreased/increased during the stance phase when the KJC was shifted forwards/backwards from the normal position. This is attributed to the decrease/increase in the perpendicular distance between the KJC and the vector of the ground reaction force when the KJC was shifted forwards/backwards relative to the hip-ankle line. This change in the perpendicular distance between the KJC and the force vector, is influenced by the degree of change in the trunk position which was achieved by the subject in order to restore/reduce the knee stability as the KJC was shifted forwards/backwards (see fig. 6.44). The variations in the knee moment changes among subjects, and the differences between the changes in the knee moment resulting by shifting the knee forwards and those resulting by shifting the knee backwards can be attributed to the differences among subjects in the trunk angular changes with the knee shifts.

It was also found that shifting the KJC backwards by 1.8 cm (in increments of 0.9 cm) relative to the hip-ankle line, advanced the TE of the prosthetic MKZ in ten subjects by an average of 44% (ranging from 13% to 72% and proportional changes can be assumed) of that when the KJC was in the normal position, and no change was found in the remaining subject (ILBQCK). Shifting the KJC forwards by 1.8 cm (in increments of 0.9 cm) delayed the TE of the prosthetic MKZ in five subjects (PLAICK, TRAQCK, JLAICK, ILBICK, PLAQCK) by an average of 16.6% (ranging from 9% to 40%) of that when the KJC was in the normal position, and no change was

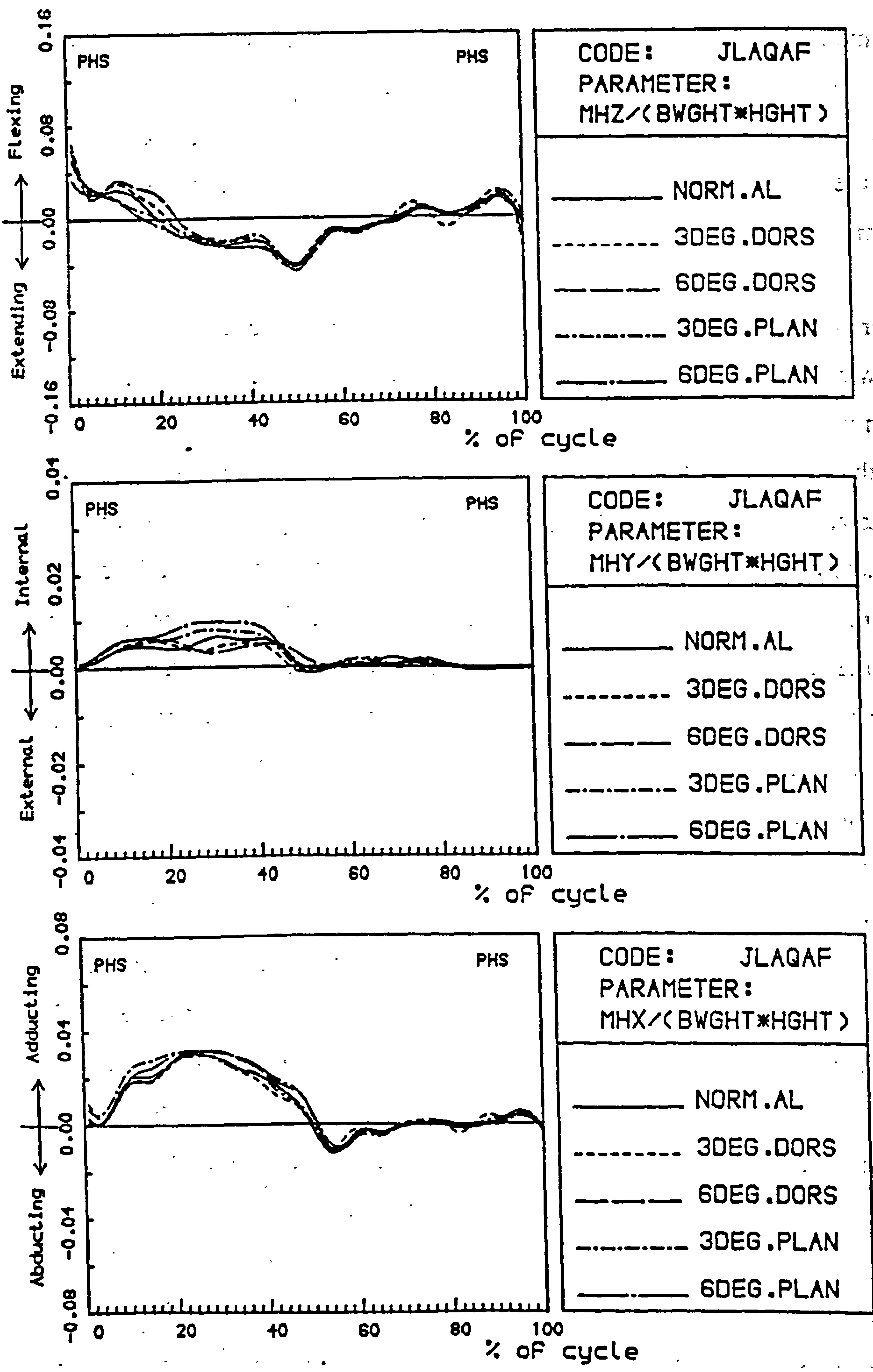


Figure 6.76 Effect of foot alignment changes on the variation with time of the hip joint moments of the prosthetic leg for an AK amputee.

found in the six remaining subjects. These changes are related to the increase/decrease in the stability of the prosthesis which were associated with the backwards/forwards shifts of the KJC, as the patient would delay loading on the prosthesis until he/she ensures its stability. The reason that six subjects showed no change in the TE when shifting the KJC forwards, can be attributed to the fact that when the forward of the knee shifts were carried out, the stability was reduced but the prosthesis was not unstable.

The above changes in the TE and in the peak value of the knee extending moment when the KJC was changed, caused a change in the pattern of the AP moment of the prosthetic leg, during the stance phase (see fig 6.73 and 6.74).

In the transverse plane (MKY), the change in the prosthetic knee joint moment was similar to that obtained on the ankle joint moment (the two joints belong to one segment which is subjected to a uniform torque). In the coronal plane (MKX), all subjects (except ILBQCK and JLAQCK) exhibited a trend of slight increase in the MKX as the KJC was located posteriorly especially during the first half of stance phase. This may be attributed to the slight increase in the medio-lateral ground reaction force (FPZ) as the KJC was located posteriorly (see section 6.7.3.3).

On the sound side, the only noted change in the knee joint moments with the knee shifts was in the flexing moment which exists after heel strike. The flexing moment showed a tendency to increase/decrease when the KJC of the prosthetic leg was shifted forwards/backwards, this is related to the increase/decrease in the braking force which was found on the sound leg as the prosthetic KJC was shifted forwards/backwards relative to the hip-ankle line.

6.7.6 Effect of Alignment Changes on the Hip Joint Moments

6.7.6.1 Effect of the Foot Alignment Changes

Figures 6.76 and 6.77 show the effect of the foot alignment changes on the hip joint moments on the prosthetic and sound leg respectively for a representative subject, table 6.22 shows a summary for the results of all

Table 6.22 Effect of alignment changes on the hip joint moments.

Hip Joint	A/P Moment of the Hip Joint (MHZ)	Transverse Moment (MHY)	Coronal Moment (MHX)
		Effect of the Foot Changes[1]	
Prosthetic	Slight increase in the HFM of 7 amputees when dorsiflexing[2] the foot.	Increased slightly in all amputees when plantarflexing the foot.	Slight increase in the hip adducting moment of some amputees when plantarflexing the foot.
Sound	Slight increase in the HFM of some amputees with dorsiflexing the foot.		
		Effect of Socket Changes[1]	
Prosthetic	No noticeable changes	No noticeable changes	No noticeable changes
Sound	No noticeable changes		
		Effect of the Knee Shift	
Prosthetic	A slight increase in the hip flexing moment of 7 amp when the KJC located anteriorly.	Slight increase in the MHY of all amputees when the KJC was located posteriorly.	Inconsistent variations.
Sound	Slight increase in the HFM (in some amputees) which exists after heel strike, with locating the prosthetic KJC anteriorly relative to the hip-ankle line.		

1 No noticeable changes were found in the hip joint moments during swing phase of the sound and prosthetic legs.

2 The foot dorsiflexion changes were by 12 degrees, in increments of 3 degrees.

HFM Hip Flexing Moment which exists at heel strike. amp: amputee.

subjects.

On the hip joint of the prosthetic leg, the moments in the AP (MHZ), transverse (MHY) and coronal (MHX) planes showed no trend of change with the foot alignment changes during the swing phase. During stance phase, dorsiflexing the foot resulted in a trend of slight increase in the magnitude of the flexing moment which exists after heel strike. This trend was found in seven subjects (JLAQAF, ILBQAF, MRCQAF, ELCQAF, TRAQAF, ILBIAF, LRBQAF), and it is related to the fact that dorsiflexing the foot decreased the braking force (FXB) of the prosthetic leg (see section 6.7.3.1 and fig. 6.35). FXB is the component of the ground reaction force which is responsible for the reduction of the hip flexing moment. Thus, a reduction in the FXB results in an increase in the hip flexing moment. Furthermore, dorsiflexing the foot increased the trunk flexion angle in most subjects (see section 6.7.1.1); this may increase the perpendicular distance between the HJC and the vector of the ground reaction force (see fig. 6.35) depending on the amount of change in the trunk flexion angle (θd).

In the transverse plane, the MHY of all subjects slightly increased (the increase is pronounced in subjects JLAQAF, TRAQAF, and LRBQAF) with plantarflexing the foot during the stance phase of the prosthetic leg. This is attributed to the increase in the FPZ which was found on the prosthetic leg with plantarflexing the foot (see section 6.7.1.1), because FPZ is the main contributor to MHY. The changes in the FPZ are also responsible for increasing the hip adducting moment which was found when the foot changes were toward plantarflexion. However, this trend of increasing the hip adducting moment was not found in all subjects and was not always consistent.

On the sound side, no noticeable changes were found in the hip joint moments with the foot alignment changes. However, it can be stated that some patients (ILBQAF, ILBIAF, ELCQAF, JLAIAF, PLAQAF) showed a trend of slight increase in the hip flexing moment which exists after heel strike as the foot changes were toward dorsiflexion. This is attributed to the increase

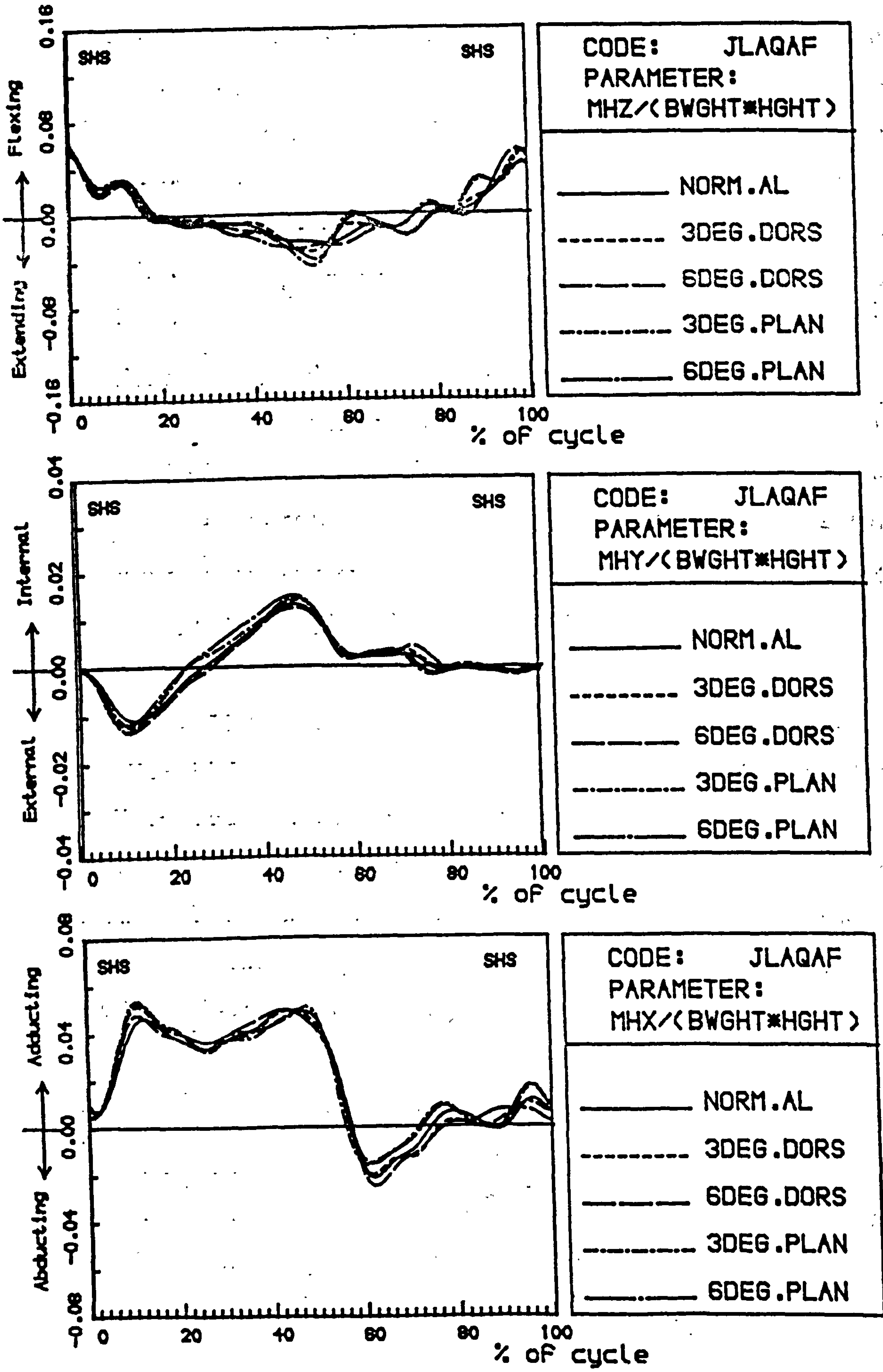


Figure 6.77 Effect of foot alignment changes on the variation with time of the hip joint moments of the sound leg for an AK amputee.

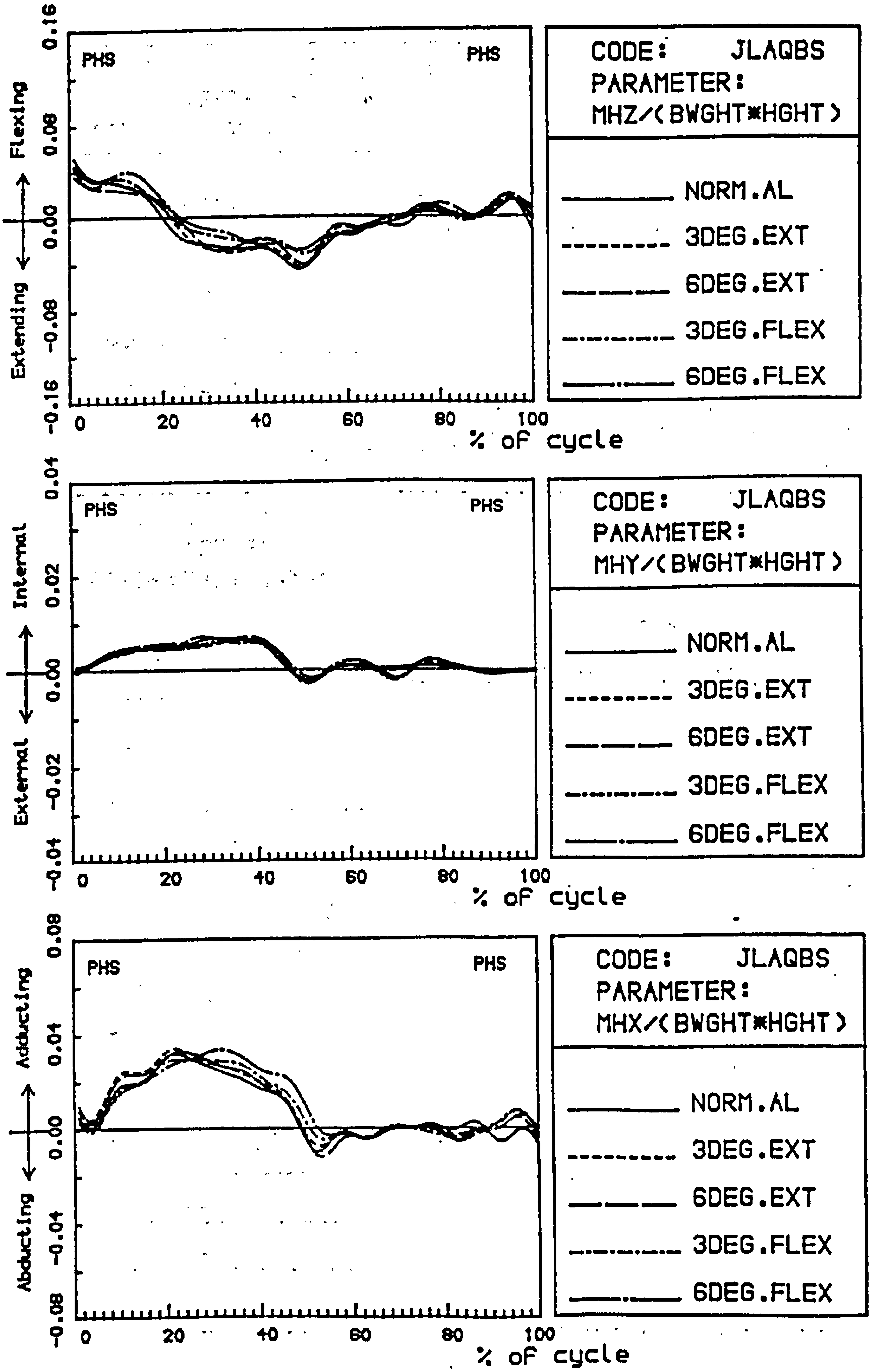


Figure 6.78 Effect of socket alignment changes on the variation with time of the hip joint moments of the prosthetic leg for an AK amputee.

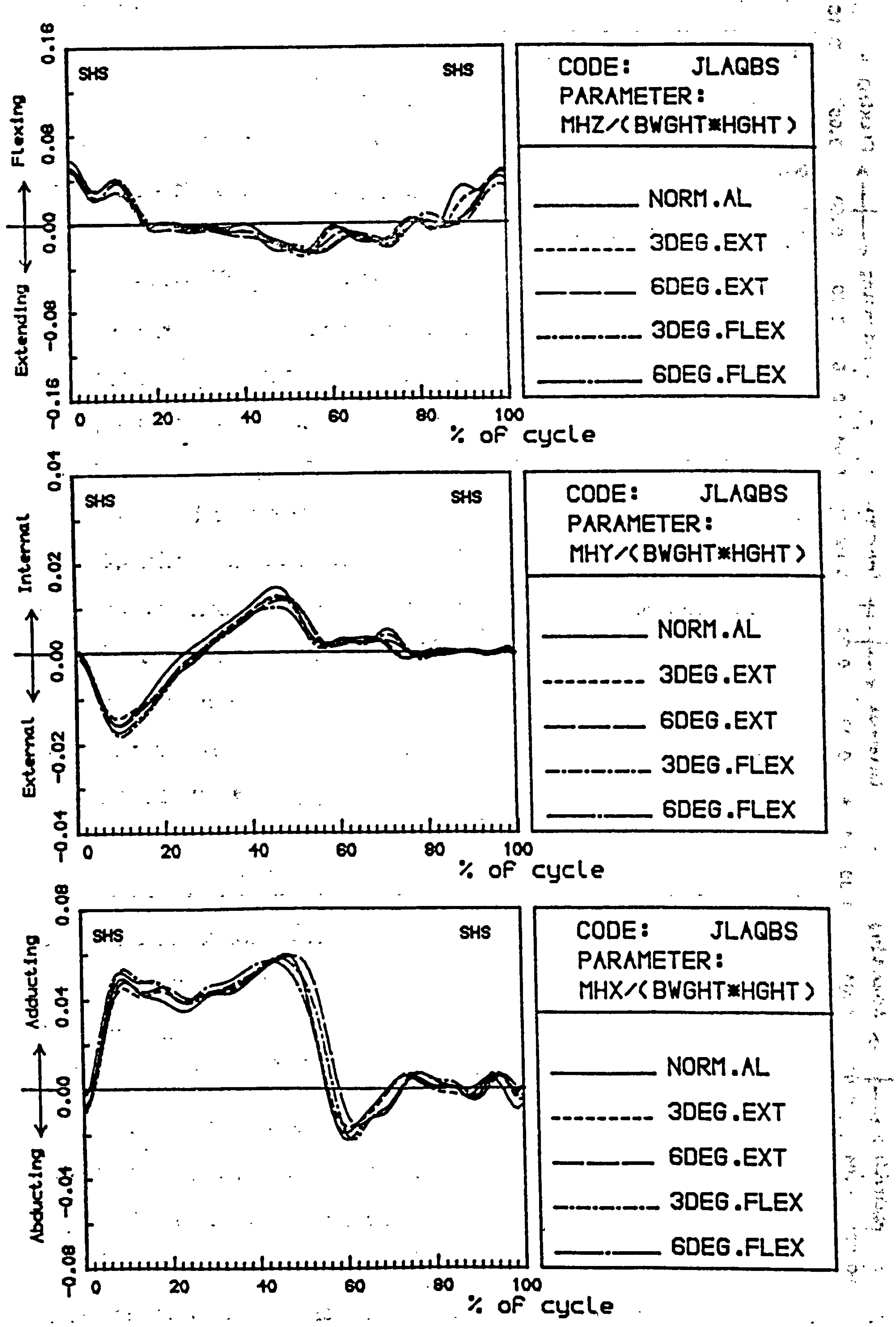


Figure 6.79 Effect of socket alignment changes on the variation with time of the hip joint moments of the sound leg for an AK amputee.

in the perpendicular distance between the HJC and the vector of the ground reaction force, as the subject had to flex his trunk with flexing the foot (see section 6.7.1.1 and fig. 6.35).

6.7.6.2 Effect of the Socket Alignment Changes

The effect of the socket alignment changes on the hip joint moments is shown in figures 6.78 and 6.79 for the prosthetic and sound sides respectively, table 6.22 shows a summary for the results of all subjects.

It was found that the socket alignment changes had no noticeable effect on the hip joint moments of the prosthetic and sound leg. This is related to the small trunk angular change which was found with the socket alignment changes (2 to 4.7 degrees as the socket was flexed by 6 degrees) in comparison to that obtained with the foot alignment changes (1.5 to 10 degrees as the foot was dorsiflexed by 6 degrees). Thus, the change in the perpendicular distance between the HJC and the ground reaction force vector was not noticeable with the socket changes, and therefore, the changes in the AP hip moment were also not noticeable. Furthermore, the changes in the FPZ which are responsible for changes in the MHY and MHX were also found to be small and not in all subjects (see section 6.7.3.2) with the socket alignment changes. Therefore, no changes were found in the MHY and MHX with the socket alignment changes.

6.7.6.3 Effect of the Knee Shifts

Figures 6.80 and 6.81 show the effect of the knee shifts on the hip moments of the prosthetic and sound leg respectively for a representative subject, table 6.22 shows a summary for the results of all subjects.

On the prosthetic side, the AP hip flexing moment (MHZ) increased slightly in seven subjects (JLAQCK, PLAQCK, MRCQCK, ILBICK, ELCQCK, TRAQCK, JLAICK) when the knee was located anteriorly. This case is similar to that which resulted on the prosthetic MHZ as the foot was dorsiflexed, and it can be explained by the same method by studying the trunk angular changes (see section 6.7.6.1). In the transverse plane, the MHY of all subjects showed a trend of slight increase during stance phase when the KJC

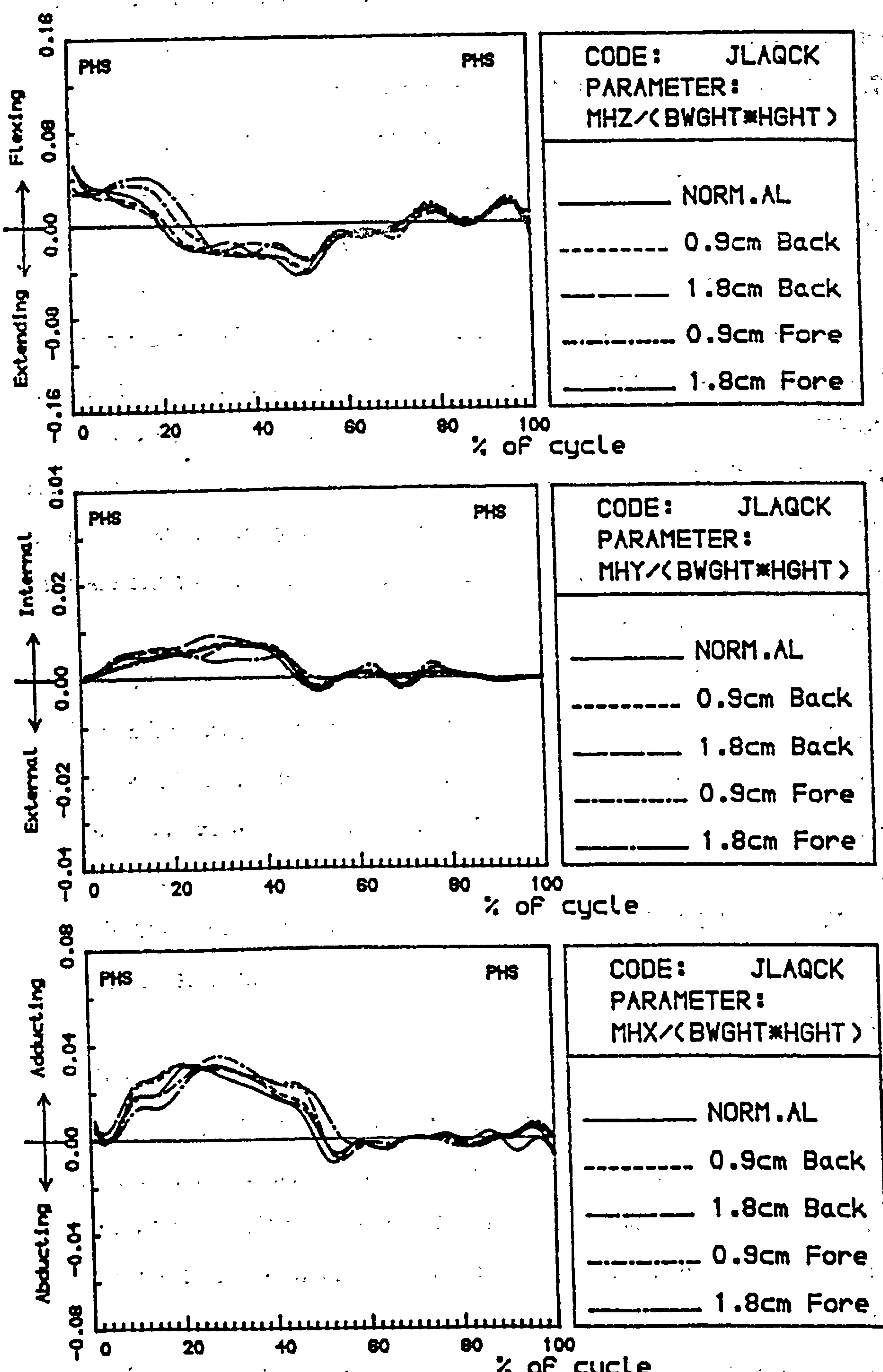


Figure 6.80 Effect of knee shifts on the variation with time of the hip joint moments of the prosthetic leg for an AK amputee.

was located posteriorly. This is related to the increase in the prosthetic FPZ which was found when the KJC was located posteriorly, and can be explained by a similar method to that used in section 6.7.6.1 for the increase in the MHY when the foot was plantarflexed. The hip adducting moment (MHX) of the prosthetic leg, was also increased during the first half of stance phase as a result of the increase in the FPZ when shifting the KJC backwards relative to the hip-ankle line. However, the increase in the MHX was not pronounced, not in all subjects and was not always consistent.

On the sound leg, no noticeable changes were found on the hip joint moments with the knee shifts. However, a slight increase in the hip flexing moment which exists after heel strike was shown in some subjects when the prosthetic KJC was located anteriorly. This is related to the changes in the trunk flexion angle which were found when locating the prosthetic KJC anteriorly, and can be explained as seen in section 6.7.6.1 for the changes resulted from dorsiflexing the foot.

In summary, the AP hip joint moment of the prosthetic leg increased/decreased during the first half of the stance phase, when the stability of the prosthetic knee was decreased/increased as a result of the alignment changes. This is related to the fact that the amputee was compensating for the lost stability by flexing the trunk forwards as seen above, therefore, the perpendicular distance between the HJC and the vector of the ground reaction force will be increased, and thus, the AP moment of the hip joint will also be increased.

It should be mentioned, that the above finding is in agreement with that found by Lowe (1969). Lowe studied the variation of the AP hip moment of an AK amputee fitted with six different knee mechanisms. The stabilised knee mechanism was found to be associated with a smaller hip flexing moment at heel strike than that of the non stabilised knee type, and when comparing the Blatchford Stabilised Knee (BSK) with the Single-Axis (SA) knee Lowe stated "BSK had a significantly lower maximum hip extensor moment (flexing

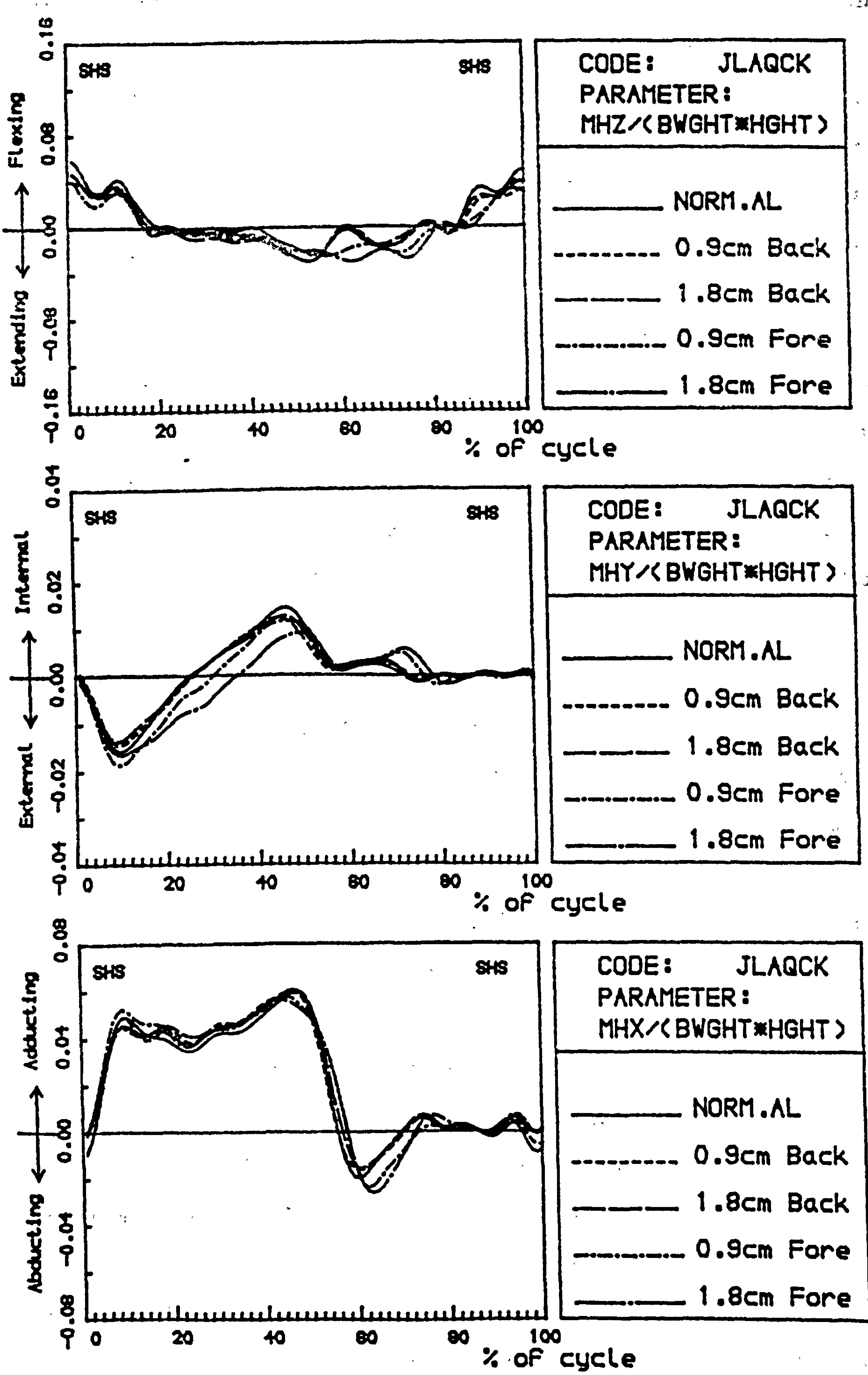


Figure 6.81 Effect of knee shifts on the variation with time of the hip joint moments of the sound leg for an AK amputee.

moment in the convention of this study) than SA. Although there was an increase in the maximum hip flexor moment produced (extending moment in the convention of this study), this was not significant in the experiments carried out".

6.8 Summary

Alignment Parameters:

The values obtained for the alignment parameters measured in this study can be used as a guide in setting the bench alignment of above knee prostheses. Thus, the dynamic alignment may be achieved by introducing the minimum amount of adjustment.

The knee set out alignment parameter, was more pronounced in prostheses which were fitted with IC sockets than that in prostheses fitted with quad sockets.

The socket flexion was reduced and socket abduction decreased (or became adduction) when the quad socket was replaced by an IC socket. However, since only three subjects were tested with IC sockets, definite conclusions cannot be drawn.

Temporal-Distance Parameters:

The temporal-distance parameters (in fact all the gait parameters), measured in this study for normal subjects and above knee amputees, are comparable to those obtained by other researchers, and can be used in clinical assessment of above knee amputees' gait, and in determining the degree of pathology in the non-amputees' gait.

For the normal subjects which were tested, the left step length was longer than the right step length. This was associated with slight differences between some of the other parameters of the right and left legs, such that, the ankle plantarflexion angle at push off was greater for the right leg than that for the left leg (fig. 6.5 and 6.6). Therefore, symmetry between the right and left legs cannot always be assumed, and it is recommended that in any gait analysis test, both the right and left legs should be treated separately and measured simultaneously.

Speed and style of walk were found to affect the gait parameters for the normal subjects and amputees. Subjects fitted with IC sockets walked slightly faster than those fitted with quad sockets as the former achieved a longer stride length. The step length of the prosthetic leg was significantly ($P<0.05$) longer than that of the sound leg for subjects who wore quad sockets. This difference was not significant ($P>0.05$) for subjects who wore IC sockets.

Stance phase duration of the sound leg was significantly ($P<0.01$) longer than that of the prosthetic leg for subjects fitted with quad sockets. For subjects fitted with IC sockets, the stance phase duration of the sound leg was noticeably (but not significantly, $P=0.08$) longer than that of the prosthetic leg.

Double support duration of the sound leg was also significantly ($P<0.05$) longer than that of the prosthetic leg for subjects fitted with quad sockets. For subjects fitted with IC sockets, the double support duration of the sound leg was not different ($P>0.1$) from that of the prosthetic leg. Again, the above conclusions are not definite since only three subjects were tested with IC sockets.

Effect of Alignment Changes on the Temporal-Distance Parameters:

Foot alignment changes and socket alignment changes had no significant ($P>0.1$) effect on the measured temporal-distance parameters of above knee amputees.

Shifting the KJC forward or backward relative to the hip-ankle line, both caused slight decreases in the velocity of the amputees. Also, shifting the KJC forwards decreased the step length of the prosthetic side and increased the stance phase duration of the sound side. Shifting the KJC backward decreased the double support time of the sound leg.

Kinetics and Kinematics Parameters:

Foot plantarflexion angle at push off phase, was larger for the sound leg of the amputees than that of the normal subjects in order to facilitate the achievement of a long step length for the prosthetic leg.

The knee angle of the prosthetic leg was maintained in full extension, and subjected to an extending moment during the stance phase in order to ensure stability in the prosthesis. During swing phase, the knee angle of the prosthetic leg was smaller for some patients than that for others and was also smaller than that of the normal subjects, because the prosthetic push off force of these patients was relatively smaller than that of the others.

At heel strike, the femur flexion angle of the prosthetic leg (approximately 20 degrees) was smaller than that of the sound side (approximately 24.5 degrees), and this flexion angle did not increase immediately after heel strike because the prosthetic knee did not flex during the stance phase.

Trunk M/L tilt of the above knee amputees was larger than that of the normal subjects, because of the amputees' poor medio-lateral stability. It was also larger toward the prosthetic side than toward the sound side, in order to reduce the pressure on the lateral-distal end of the femur.

No differences were found in the trunk M/L tilt between subjects fitted with IC and quad sockets.

The amputees' rotation of the torso in the transverse plane was similar to that of the normal subjects, because they walked with a relatively similar velocity.

For the vertical ground reaction force, the occurrence of the first peak is very sensitive, and it occurred at 22%, 12%, 16% and 33% of stance phase duration for the normal subjects, sound leg, prosthetic leg with IC socket, and prosthetic leg with quad socket respectively.

The "vaulting action" which is commonly seen at the sound leg of the amputees, was larger for subjects fitted with quad sockets than that of subjects fitted with IC sockets.

Some subjects showed a third peak at the prosthetic leg, and it is believed that this peak may be caused by the hip extensors.

In the direction of progression, the ground reaction force of the prosthetic leg was smaller than that of the sound leg and of the normal subjects, because a large braking force may endanger the stability of the prosthesis at heel strike, and the prosthetic ankle joint does not help in generating a large push off force.

Subjects fitted with IC sockets were found to have a larger braking and smaller push off force at the prosthetic leg as a fraction of the sound leg force, than those fitted with quad sockets.

The shape of the ankle dorsiflexing moment curve with time for the sound leg was different from that of the normal subjects, viz the ankle moment increased sharply just after the point at which the moment changed from plantarflexing to dorsiflexing. The ankle plantarflexing moment of the prosthetic leg which occurs after heel strike, showed a longer duration than that of the normal subjects. The ankle dorsiflexing moment of the prosthetic leg, which occurs at the push off phase was 61% of that of the normal subjects.

A/P hip moment of the prosthetic side was approximately 50% of that of the sound leg throughout stance phase. This can be related to the small vertical ground reaction force, and small femur flexion/extension angle during stance phase of the prosthetic leg in comparison to that of the sound leg. During the swing phase, the small A/P hip moment of the prosthetic leg can be attributed to the small mass of the prosthetic leg in comparison to that of the sound leg.

Effect of Alignment Changes on the Kinetics and Kinematics Parameters:

The A/P trunk tilt was found to be the most sensitive parameter to the alignment changes which were performed to the prosthesis in the A/P plane. The trunk angular position changed so that the position of the body CG moved forwards/backwards in order to enhance/reduce the stability of the prosthetic knee as the latter was affected by the alignment changes.

Increasing the dorsiflexion of the prosthetic foot by 6 degrees, the flexion of the socket by 6 degrees from the normal position, and shifting the KJC forwards by 1.8 cm relative to the hip-ankle line, increased the trunk flexion angle by ranges from 1.5 to 10, 2 to 4.7 and 1.7 to 10.6 degrees respectively, in comparison to those corresponding to the normal alignment. These trunk angular changes maintained the stability of the prosthetic knee, throughout the stance phase.

Increasing the plantarflexion angle of the prosthetic foot by 6 degrees, the extension angle of the socket by 6 degrees from the normal position, and shifting the KJC backwards by 1.8 cm relative to the hip-ankle line, decreased the trunk flexion angle by a range from 1.5 to 3 degrees, slightly, and a range from 1 to 3.3 degrees respectively, in comparison to those corresponding to the normal alignment. These changes in the trunk attitude reduced the hip flexing moment at heel strike and facilitated knee flexion prior to swing phase of the prosthetic leg.

The M/L trunk tilt showed a trend of slight increase toward the sound leg during the prosthetic swing phase, when the foot changes were toward plantarflexion and when the knee shifts were backwards. This is related to the fact that plantarflexing the foot and shifting the KJC backward may increase the length of the prosthesis, therefore, the patient would lean over the sound leg in order to provide clearance for the swinging leg and prevent stumbling.

The alignment changes had no noticeable effect on the angular displacements of the lower limb joints of the sound leg.

During swing phase, on the prosthetic side, the alignment changes had no noticeable effect on the thigh angle. However, shifting the KJC forward/backward relative to the hip-ankle line, showed a slight trend of increase/decrease respectively in the knee flexion angle during the swing phase. This is related to the increase/decrease in the push off force of the prosthetic leg which resulted when the KJC was shifted forward/backward.

During stance phase, the alignment changes had no noticeable effect on the knee flexion angle of the prosthetic side. Dorsiflexing/plantarflexing the foot and extending/flexing the socket from the normal position, decreased/increased the ankle dorsiflexion angle. This angle showed inconsistent variations with the knee shifts.

When the foot alignment changes were toward dorsiflexion, the socket changes were toward flexion and the knee shifts were forward, the thigh flexion angle increased at heel strike and the thigh extension angle at push off was decreased on the prosthetic leg.

On the prosthetic leg, when alignment changes were toward foot dorsiflexion, socket flexion, and when the knee was located anteriorly relative to the hip-ankle line, the following changes were found in the fore-and-aft ground reaction force for above-knee amputees: 1- Braking force (FXB) decreased relative to the normal value, in order to enhance the stability of the prosthetic knee. 2- Push off force (FXP) increased relative to the normal value, since the hip extensors were put in an advantageous position (socket and knee changes only) and the trunk was flexed forwards relative to its normal position. 3- The time (TT) up to the transition point in early stance, during which the fore-and-aft force is negative, advanced with the foot alignment changes, but not with the socket or knee alignment changes.

On the sound leg, fore-and-aft force was affected by the knee alignment changes only and not by the foot or socket changes. Shifting the KJC of the prosthetic leg forwards/backwards relative to the hip-ankle line, increased/decreased the braking force and decreased/increased the push off force on the sound leg in comparison to that when the prosthetic knee was in normal alignment. It should be noted, that the effect of the alignment changes on the braking and push off forces of the sound leg, was opposite to that found on the prosthetic leg.

Alignment changes had no effect on the vertical ground reaction force (FPY) of the sound leg. However, the magnitude of the first vertical peak of FPY acting on the prosthetic leg, decreased relative to the normal value, when the foot changes were toward dorsiflexion, socket changes toward flexion, and when the KJC was located anteriorly relative to the hip-ankle line.

On the prosthetic side, the time (TP) up to the transition point in early stance phase, during which the ankle joint is subjected to a plantarflexing moment, delayed relative to the normal value, as the foot alignment changes were toward dorsiflexion, socket changes toward extension, and when the KJC was located anteriorly relative to the hip-ankle line. The ankle joint moments of the sound leg, were not affected by the alignment changes.

On the prosthetic side, the time (TE) up to the point at which the knee extending moment starts to build up, delayed as the foot was progressively dorsiflexed, and the KJC was shifted forwards relative to the hip-ankle line. The knee extending moment during stance phase, increased as the socket extension was increased and as the KJC was shifted backwards relative to the hip-ankle line.

On the hip joint of the prosthetic leg, a slight increase was found in the hip flexing moment after heel strike, when the foot was dorsiflexed and when the KJC was located anteriorly relative to the hip-ankle line.

It can be concluded that the stability of the knee on the prosthetic leg is an important factor affecting the pattern of walk and the gait parameters, and in many cases appears to dictate the patient's response to alignment changes.

No noticeable differences were found in the effect of alignment changes on the gait parameters of subjects wearing quadrilateral and ischial containment sockets. However, since only three patients were tested using the IC socket, this can not be a definite conclusion.

Chapter Seven

Conclusions and Recommendations for Future Work

7.1 Conclusions

Biomechanics of Above-Knee Amputee:

(a) Sound Side vs Prosthetic Side:

The fore-and-aft ground reaction force in AK amputees (both braking and push off) was smaller on the prosthetic leg than that on the sound leg.

The A/P hip moment on the prosthetic side was approximately 50% of that of the sound leg throughout stance phase.

At heel strike, the femur flexion angle of the prosthetic leg was smaller than that of the sound side.

(b) Normal Subject vs Above-Knee Amputee:

The trunk M/L tilt in above knee amputees was larger than that in normal subjects, and it was also larger toward the prosthetic side than toward the sound side.

The ankle dorsiflexing moment of the prosthetic leg, which occurs at the push off phase was 61% of that of the normal subjects.

(c) Quad vs IC socket:

Subjects fitted with IC sockets walked slightly faster than those fitted with quad sockets. The step length of the prosthetic leg was significantly ($P<0.05$) longer than that of the sound leg for subjects who wore quad sockets. This difference was not significant ($P>0.05$) for subjects who wore IC sockets.

Stance phase and double support durations of the sound leg were longer than those of the prosthetic leg for subjects fitted with quad sockets. This was not the case for subjects fitted with IC sockets.

The "vaulting action" was larger for subjects fitted with quad sockets than that of subjects fitted with IC sockets.

The first peak of FPY occurred at 22%, 12%, 16% and 33% of stance phase duration for the normal subjects, sound leg, prosthetic leg with IC socket, and prosthetic leg with quad socket respectively.

When the quad socket was replaced by an IC socket, dynamic alignment resulted in reduced socket flexion and socket abduction, and increased knee set out.

Effect of Alignment Changes on the Gait Parameters of AK Amputees:

No noticeable effect was found for the alignment changes on the measured temporal-distance parameters of above knee amputees.

Increasing dorsiflexion of the prosthetic foot by 6 degrees, or the flexion of the socket by 6 degrees, or shifting the KJC forwards by 1.8 cm relative to the hip-ankle line from the normal alignment position, increased the trunk flexion angle by ranges from 1.5 to 10, 2 to 4.7 and 1.7 to 10.6 degrees respectively.

Increasing the plantarflexion angle of the prosthetic foot by 6 degrees, or the extension angle of the socket by 6 degrees, or shifting the KJC backwards by 1.8 cm relative to the hip-ankle line from the normal alignment position, decreased the trunk flexion angle by a range from 1.5 to 3 degrees, slightly, and a range from 1 to 3.3 degrees respectively.

When the foot alignment changes were toward dorsiflexion, or the socket changes were toward flexion, or the knee shifts were forward, the thigh flexion angle increased at heel strike and the thigh extension angle at push off was decreased on the prosthetic leg.

On the prosthetic leg, when alignment changes were toward foot dorsiflexion, or socket flexion, or when the knee was located anteriorly relative to the hip-ankle line, the following changes were found in the fore-and-aft ground reaction force for above-knee amputees: 1- Braking force (FXB) decreased relative to the normal value. 2- Push off force (FXP) increased relative to the normal value. 3- The time (TT) up to the transition point, advanced with the foot alignment changes, but not with the socket or knee alignment changes.

Shifting the KJC of the prosthetic leg forwards/backwards relative to the hip-ankle line, increased/decreased the braking force and decreased/increased the push off force on the sound leg in comparison to that when the prosthetic knee was in normal alignment.

On the <u>prosthetic side</u>, the time (TP) up to the transition point delayed relative to the normal value, as the foot alignment changes were toward dorsiflexion, or socket changes toward extension, or when the KJC was located anteriorly relative to the hip-ankle line.

On the <u>prosthetic side</u>, the time (TE) up to the point at which the knee extending moment starts to build up, delayed as the foot was progressively dorsiflexed, or the KJC was shifted forwards relative to the hip-ankle line. The knee extending moment during stance phase, increased as the socket extension was increased or as the KJC was shifted backwards relative to the hip-ankle line.

On the hip joint of the <u>prosthetic leg</u>, a slight increase was found in the hip flexing moment after heel strike, when the foot was dorsiflexed or when the KJC was located anteriorly relative to the hip-ankle line.

7.2 Recommendations for Further Works

1- More tests are recommended on patients wearing IC sockets in order to assess the function of the IC socket, and to draw definite conclusions regarding the claimed superiority of the IC over the quad socket, and regarding the effect of alignment changes on the gait parameters of patients wearing IC sockets.

2- A greater number of force plates and cameras incorporated in the system would allow monitoring of the subject and analysis of the gait for more than one stride. However, since the installation of many force plates would be expensive, and adjusting the patient's gait to strike properly many force plates may not be possible, a radio telemetry or data recording pylon is recommended to record the loads on the prosthetic leg.

3- An investigation for the muscle activities of the amputated leg of above-knee amputees during the gait cycle, would be useful in the analysis of the above-knee amputee's gait. Electromyography is suggested.

4- In order to assist the prosthetist in achieving the optimum alignment, a simple technique can be suggested. The technique should make use of the relative position between the vector of the ground reaction force and the knee joint centre of the prosthetic leg, and also consider the angular position of the trunk in space. A computer program can be used to display the position of the KJC, the ground reaction force vector and the angular position of the trunk simultaneously during walking, thus, the prosthetist would be able to judge the knee stability, the function of the prosthesis and the appearance of the subject.

5- A device is also suggested to help the prosthetist in setting the bench alignment, and in achieving the dynamic alignment. The device can consist of a series of gauges temporarily fixed on the components of the prosthesis, the output signal of these gauges can be fed to a computer which would quickly calculate and display the eleven alignment parameters of the prosthesis. This device can also help in reproducing a certain alignment if a prosthesis needed to be aligned and realigned.

Appendix A

Calculation of the knee shift referred to in section 6.3, and Tables Representing Some of Temporal-Distance Parameters Obtained for Above Knee Amputees, when the Alignment of the Prosthesis was Changed.

Calculating the Amount of Knee Shift:

Referring to figure 6.1, and knowing that the position of points H and K are determined relative to point A (see section 6.3), the angle θ can be found as:

$$\theta = \arccos \frac{\overline{AK}^2 + \overline{AH}^2 - \overline{KH}^2}{2 \times \overline{AK} \times \overline{AH}}$$

θ is positive when the HA line passes behind the KJC, i.e. the knee is set forward, and the KN distance is positive.

The distance KN and the coordinates x_{K1} , y_{K1} of point K_1 were calculated in section 6.3 as:

$$\overline{KN} = \overline{AK} \sin \theta$$

$$x_{K_1} = \overline{AK_1} \sin (\beta + \alpha)$$

$$y_{K_1} = \overline{AK_1} \cos (\beta + \alpha)$$

Point H_1 is the position of the HJC which resulted from changing the foot angular position, and its coordinates relative to the AJC can be found as:

$$x_{H_1} = \overline{AH} \sin (\beta + \psi)$$

$$y_{H_1} = \overline{AH} \cos (\beta + \psi)$$

Where: β is the foot change angle and it is positive when the change dorsiflexes the foot. Ψ is the original angle between the y_a axis and the HA line, and it is positive clockwise.

The coordinates (x_S , y_S) of the SRC (point S in fig. 6.1) can be calculated relative to the AJC as:

$$x_S = x_{K_1} + \overline{KS} \sin \phi$$

$$y_S = y_{K_1} + \overline{KS} \cos \phi$$

Where KS was measured on the coordinate measuring system (section 4.2.2). ϕ is the original angle between the SH_1 line and the vertical line which passes from S, and it is positive clockwise.

In the knee shift procedure, the foot angular change was coupled with a change in the socket angular orientation by the same angle β (β is toward extension in fig. 6.1). This will shift the HJC from point H_1 to point H_2, and the coordinates of point H_2 can be found relative to the AJC as:

$$x_{H_2} = x_{H_1} + \overline{SH_1} \; (\sin (\beta + \phi) - \sin \phi)$$
$$y_{H_2} = y_{H_1} + \overline{SH_1} \; (\cos (\beta + \phi) - \cos \phi)$$

Where β is positive when the change is extending the socket.

Thus, the angle θ_1 (positive when H_2A passes behind K_1) can be calculated from triangle AK_1H_2 and the distance K_1N_1 can be found as:

$$\overline{K_1N_1} = \overline{AK_1} \sin \theta_1$$

So, the shift " a " in the KJC relative to the line which connects the HJC to the AJC resulted from a simultaneous change in the foot and socket angular position by an angle β is:

$$a = \overline{K_1N_1} - \overline{KN}$$

The knee shift is forwards when the value of " a " is positive, and backwards when the value of " a " is negative.

Example:

Subject TRAQ is taken as an example. The coordinates (in cm) of the KJC and HJC relative to the AJC are: K(-0.4 , 41.3) and H(2.1 , 85.2). Thus, AK = 41.3 cm, KH = 43.97 cm and AH = 85.23 cm. Also, K_1S = 2.4 cm $\Rightarrow$ SH_1 = 41.57 cm.

$$\alpha = \arctan \frac{-0.4}{41.3} = -0.55 \; degree$$

$$\psi = \arctan \frac{2.1}{85.2} = 1.41 \; degree$$

$$\theta = \arccos \frac{(41.3)^2 + (85.22)^2 - (43.97)^2}{2 \times 41.3 \times 85.22} = 2.02 \; degree$$

It can be seen from angles α and Ψ that the HA line passes ahead of the K point $\Rightarrow$ θ is negative (θ = - 2.02). Thus,

$$\overline{KN} = 41.3 \sin (-2.02) = -1.46 cm$$

The following calculations are for the case of plantarflexing the foot and extending the socket by 3 degrees.

The coordinates of H_2 which resulted from extending the socket by 3 degrees (It should be noted here that socket extension is positive) are:

$$x_{K_1} = 41.3 \sin (-3 - 0.555) = - 2.56\ cm$$

$$y_{K_1} = 41.3 \cos (-3.555) = 41.22\ cm$$

$$x_{H_1} = 85.23 \sin (-3 + 1.41) = -2.26\ cm$$

$$y_{H_1} = 85.23 \cos (-3 + 1.41) = 85.2\ cm$$

$$\phi = \arctan \frac{x_{H_1} - x_{K_1}}{y_{H_1} - y_{K_1}} = \arctan \frac{(-2.36) - (-2.56)}{85.2 - 41.22} = 0.26\ degree$$

$$x_{H_2} = -2.36 + 41.57 (\sin (3 + 0.26) - \sin 0.26) = - 0.18\ cm$$

$$y_{H_2} = 85.2 + 41.57 (\cos 3.26 - \cos 0.26) = 85.13\ cm$$

Thus, $AH_2 = 85.13$ cm , $K_1H_2 = 43.97$ cm and $AK_1 = 41.3$ cm.

$$\theta_1 = \arccos \frac{\overline{AH_2}^2 + \overline{AK_1}^2 - \overline{K_1H_2}^2}{2 \times \overline{AH_1} \times \overline{AK_1}}$$

$$\theta_1 = \arccos \frac{(85.13)^2 + (41.3)^2 - (43.97)}{2 \times 85.13 \times 41.3} = 3.39\ degrees$$

It is clear from the coordinates of K_1 and H_2 relative to the AJC that the H_2A line is passing ahead of K_1 (this can be confirmed by calculating the angles of H_2A and K_1A with the y_a axis), therefore, the angle θ_1 is negative ($\theta_1 = -3.39$).

$$\overline{K_1N_1} = 41.3 \sin(- 3.39) = - 2.44\ cm$$

Therefore, the knee shift a is:

a = - 2.44 - (-1.46) = - 0.98 cm and the KJC is shifted backwards from the line which connects the HJC to the AJC.

Table A1 Temporal-Distance parameters of 11 AK amputees. (3 Degrees Foot Dorsiflexion change, from the normal position)

Sub. Code	Velocity [m/s]	Cycle Duration [s]	Stride Length [m]	Step Width [m]	Sound Step Length [m]	Prosth step Length [m]	Prosthetic Stance *	Sound Stance *	Prosthetic D Support *	Sound D Support *
JLAQ	1.034 (0.017)	1.300 (0.053)	1.335 (0.026)	0.066 (0.004)	0.635 (0.020)	0.700 (0.010)	0.561 (0.006)	0.642 (0.009)	0.106 (0.004)	0.111 (0.008)
PLAQ	0.754 (0.036)	1.593 (0.092)	1.240 (0.056)	0.197 (0.022)	0.550 (0.026)	0.690 (0.032)	0.542 (0.009)	0.662 (0.016)	0.111 (0.008)	0.104 (0.020)
ILBQ	1.310 (0.015)	1.213 (0.012)	1.567 (0.037)	0.212 (0.044)	0.754 (0.012)	0.813 (0.028)	0.573 (0.008)	0.622 (0.008)	0.097 (0.001)	0.114 (0.001)
MRCQ	1.186 (0.056)	1.180 (0.028)	1.381 (0.009)	0.165 (0.009)	0.783 (0.005)	0.598 (0.014)	0.525 (0.001)	0.708 (0.019)	0.117 (0.003)	0.133 (0.020)
TRAQ[1]	0.987 (0.003)	1.450 (0.014)	1.409 (0.035)	0.126 (0.017)	0.663 (0.009)	0.746 (0.027)	0.572 (0.025)	0.660 (0.023)	0.095 (0.001)	0.150 (0.001)
LRBQ	0.713 (0.033)	1.640 (0.053)	1.167 (0.029)	0.112 (0.019)	0.500 (0.020)	0.667 (0.012)	0.522 (0.012)	0.683 (0.025)	0.084 (0.003)	0.133 (0.012)
DLAQ[1]	0.854 (0.040)	1.480 (0.085)	1.230 (0.025)	0.176 (0.026)	0.500 (0.001)	0.730 (0.026)	0.527 (0.020)	0.687 (0.011)	0.120 (0.007)	0.107 (0.025)
ELCQ	0.759 (0.032)	1.480 (0.053)	1.114 (0.021)	0.196 (0.039)	0.533 (0.015)	0.581 (0.020)	0.569 (0.013)	0.662 (0.009)	0.124 (0.003)	0.120 (0.015)
Aver.	0.950 (0.218)	1.417 (0.169)	1.305 (0.147)	0.156 (0.050)	0.615 (0.112)	0.683 (0.091)	0.549 (0.022)	0.671 (0.021)	0.107 (0.014)	0.122 (0.016)
JLAI	1.029 (0.019)	1.360 (0.053)	1.396 (0.017)	0.129 (0.019)	0.675 (0.012)	0.722 (0.012)	0.570 (0.006)	0.628 (0.008)	0.102 (0.004)	0.111 (0.012)
PLAI[1]	0.953 (0.034)	1.290 (0.014)	1.229 (0.064)	0.216 (0.007)	0.555 (0.009)	0.674 (0.056)	0.565 (0.006)	0.664 (0.004)	0.122 (0.001)	0.122 (0.001)
ILBI	1.167 (0.041)	1.327 (0.050)	1.564 (0.018)	0.138 (0.051)	0.722 (0.037)	0.842 (0.023)	0.564 (0.015)	0.624 (0.009)	0.099 (0.018)	0.104 (0.004)
Aver.	1.050 (0.108)	1.326 (0.035)	1.396 (0.168)	0.161 (0.048)	0.651 (0.086)	0.746 (0.087)	0.566 (0.003)	0.639 (0.022)	0.108 (0.016)	0.112 (0.009)

Each reading in this table is the average for 3 different runs carried out within one day. Figures between brackets are one standard deviation.
* The unit is: proportion of the cycle duration.
1 Data of this subject were averaged for 2 different runs carried out within one day.

Table A2 Temporal-Distance parameters of 11 AK amputees. (3 Degrees foot plantarflexion change, from the normal position).

Sub. Code	Velocity [m/s]	Cycle Duration [s]	Stride Length [m]	Step Width [m]	Sound Step Length [m]	Prosth step Length [m]	Prosthetic Stance *	Sound Stance *	Prosthetic D Support *	Sound D Support *
JLAQ	1.110 (0.049)	1.213 (0.031)	1.350 (0.032)	0.103 (0.039)	0.631 (0.010)	0.719 (0.022)	0.578 (0.006)	0.632 (0.003)	0.114 (0.003)	0.114 (0.003)
PLAQ	0.842 (0.032)	1.407 (0.031)	1.213 (0.027)	0.206 (0.039)	0.561 (0.039)	0.653 (0.024)	0.556 (0.006)	0.673 (0.027)	0.131 (0.014)	0.112 (0.016)
ILBQ	1.252 (0.025)	1.220 (0.020)	1.520 (0.031)	0.201 (0.024)	0.696 (0.026)	0.824 (0.006)	0.581 (0.007)	0.618 (0.014)	0.102 (0.009)	0.113 (0.002)
MRCQ[1]	1.198 (0.035)	1.180 (0.028)	1.407 (0.074)	0.211 (0.023)	0.723 (0.018)	0.685 (0.056)	0.550 (0.013)	0.683 (0.007)	0.117 (0.003)	0.133 (0.003)
TRAQ[1]	0.949 (0.014)	1.550 (0.042)	1.418 (0.028)	0.118 (0.029)	0.649 (0.012)	0.769 (0.016)	0.567 (0.012)	0.656 (0.009)	0.089 (0.002)	0.146 (0.005)
LRBQ	0.692 (0.059)	1.687 (0.145)	1.158 (0.052)	0.122 (0.011)	0.503 (0.048)	0.655 (0.017)	0.515 (0.011)	0.676 (0.021)	0.082 (0.007)	0.121 (0.011)
DLAQ	0.911 (0.063)	1.393 (0.046)	1.267 (0.034)	0.195 (0.018)	0.515 (0.041)	0.742 (0.010)	0.533 (0.016)	0.679 (0.002)	0.118 (0.013)	0.108 (0.005)
ELCQ	0.710 (0.036)	1.433 (0.058)	1.048 (0.044)	0.243 (0.020)	0.476 (0.019)	0.572 (0.028)	0.587 (0.008)	0.694 (0.072)	0.142 (0.012)	0.153 (0.060)
Aver.	0.958 (0.212)	1.385 (0,177)	1.298 (0.155)	0.175 (0.052)	0.594 (0.093)	0.702 (0.078)	0.558 (0.025)	0.664 (0.026)	0.112 (0.020)	0.125 (0.017)
JLAI	1.089 (0.038)	1.333 (0.042)	1.454 (0.036)	0.108 (0.021)	0.707 (0.022)	0.747 (0.021)	0.586 (0.004)	0.611 (0.004)	0.108 (0.005)	0.104 (0.003)
PLAI[1]	0.968 (0.028)	1.260 (0.028)	1.228 (0.074)	0.240 (0.008)	0.522 (0.043)	0.705 (0.031)	0.555 (0.001)	0.680 (0.004)	0.125 (0.003)	0.125 (0.003)
ILBI	1.008 (0.034)	1.460 (0.072)	1.484 (0.033)	0.134 (0.023)	0.668 (0.032)	0.816 (0.010)	0.577 (0.008)	0.613 (0.012)	0.104 (0.010)	0.099 (0.005)
Aver.	1.022 (0.062)	1.351 (0.101)	1.389 (0.140)	0.161 (0.069)	0.632 (0.098)	0.756 (0.056)	0.573 (0.016)	0.635 (0.039)	0.112 (0.011)	0.108 (0.011)

Each reading in this table is the average of 3 different runs carried out within one day. Figures between brackets are one standard deviation.
* The unit is: proportion of the cycle duration.
1 Data of this subject were averaged for 2 different runs carried out within one day.

Table A3 Temporal-Distance parameters of 11 AK amputees. (3 Degrees Socket Flexion change,from the normal position).

Sub. Code	Velocity [m/s]	Cycle Duration [s]	Stride Length [m]	Step Width [m]	Sound Step Length [m]	Prosth step Length [m]	Prosthetic Stance *	Sound Stance *	Prosthetic D Support *	Sound D Support *
JLAQ	1.019 (0.019)	1.340 (0.020)	1.371 (0.011)	0.084 (0.012)	0.652 (0.010)	0.720 (0.011)	0.568 (0.014)	0.637 (0.005)	0.103 (0.002)	0.118 (0.013)
PLAQ	0.807 (0.024)	1.393 (0.042)	1.135 (0.056)	0.255 (0.017)	0.553 (0.020)	0.582 (0.039)	0.590 (0.017)	0.698 (0.041)	0.123 (0.008)	0.180 (0.046)
ILBQ[1]	1.317 (0.031)	1.210 (0.014)	1.592 (0.042)	0.263 (0.023)	0.732 (0.007)	0.860 (0.037)	0.569 (0.007)	0.626 (0.004)	0.098 (0.001)	0.114 (0.001)
MRCQ[1]	1.270 (0.003)	1.120 (0.000)	1.393 (0.031)	0.142 (0.038)	0.687 (0.022)	0.706 (0.008)	0.553 (0.012)	0.667 (0.000)	0.123 (0.000)	0.114 (0.012)
TRAQ	1.042 (0.022)	1.420 (0.000)	1.494 (0.045)	0.163 (0.035)	0.751 (0.029)	0.743 (0.038)	0.579 (0.008)	0.648 (0.008)	0.097 (0.000)	0.144 (0.008)
LRBQ	0.623 (0.029)	1.760 (0.001)	1.097 (0.049)	0.149 (0.056)	0.494 (0.018)	0.602 (0.042)	0.536 (0.013)	0.678 (0.017)	0.079 (0.000)	0.146 (0.011)
DLAQ	0.893 (0.011)	1.467 (0.023)	1.303 (0.021)	0.161 (0.016)	0.628 (0.024)	0.675 (0.040)	0.547 (0.014)	0.664 (0.018)	0.103 (0.006)	0.121 (0.038)
ELCQ	0.697 (0.033)	1.460 (0.072)	1.049 (0.025)	0.229 (0.037)	0.474 (0.014)	0.575 (0.021)	0.617 (0.018)	0.757 (0.021)	0.135 (0.007)	0.252 (0.031)
Aver.	0.959 (0.252)	1.396 (0.191)	1.307 (0.197)	0.182 (0.060)	0.619 (0.104)	0.687 (0.099)	0.570 (0.026)	0.672 (0.041)	0.109 (0.018)	0.148 (0.048)
JLAI	1.025 (0.016)	1.287 (0.012)	1.333 (0.020)	0.090 (0.004)	0.663 (0.009)	0.670 (0.011)	0.592 (0.012)	0.663 (0.016)	0.112 (0.008)	0.158 (0.017)
PLAI[1]	0.981 (0.001)	1.260 (0.000)	1.235 (0.012)	0.198 (0.037)	0.544 (0.008)	0.691 (0.004)	0.570 (0.011)	0.664 (0.011)	0.125 (0.000)	0.125 (0.000)
ILBI	1.048 (0.044)	1.393 (0.046)	1.456 (0.023)	0.199 (0.038)	0.673 (0.011)	0.782 (0.025)	0.575 (0.007)	0.623 (0.011)	0.113 (0.004)	0.099 (0.003)
Aver.	1.018 (0.034)	1.313 (0.070)	1.341 (0.111)	0.162 (0.063)	0.627 (0.072)	0.714 (0.059)	0.579 (0.012)	0.650 (0.023)	0.117 (0.007)	0.127 (0.029)

Each reading in this table is the average of 3 different runs carried out within one day. Figures between brackets are one standard deviation.

* The unit is: proportion of the cycle duration.

1 Data were averaged for 2 different runs carried out within one day.

Table A4 Temporal-Distance parameters of 11 AK amputees. (3 Degrees socket extention change, from the normal posision).

Sub. Code	Velocity [m/s]	Cycle Duration [s]	Stride Length [m]	Step Width [m]	Sound Step Length [m]	Prosth step Length [m]	Prosthetic Stance *	Sound Stance *	Prosthetic D Support *	Sound D Support *
JLAQ	1.057 (0.011)	1.307 (0.012)	1.382 (0.011)	0.090 (0.023)	0.641 (0.018)	0.741 (0.012)	0.578 (0.004)	0.623 (0.012)	0.111 (0.009)	0.106 (0.001)
PLAQ	0.785 (0.024)	1.493 (0.031)	1.193 (0.022)	0.246 (0.009)	0.531 (0.010)	0.662 (0.024)	0.551 (0.015)	0.678 (0.021)	0.119 (0.002)	0.123 (0.019)
ILBQ[1]	1.283 (0.055)	1.210 (0.071)	1.558 (0.041)	0.178 (0.019)	0.696 (0.003)	0.862 (0.038)	0.569 (0.010)	0.626 (0.002)	0.098 (0.006)	0.114 (0.007)
MRCQ[1]	1.243 (0.033)	1.080 (0.028)	1.366 (0.001)	0.141 (0.002)	0.750 (0.012)	0.616 (0.013)	0.555 (0.001)	0.682 (0.005)	0.127 (0.003)	0.127 (0.003)
TRAQ	0.949 (0.014)	1.553 (0.064)	1.463 (0.056)	0.170 (0.020)	0.698 (0.044)	0.765 (0.068)	0.564 (0.009)	0.665 (0.002)	0.089 (0.004)	0.153 (0.006)
LRBQ	0.623 (0.008)	1.760 (0.035)	1.103 (0.010)	0.128 (0.051)	0.460 (0.023)	0.643 (0.014)	0.524 (0.012)	0.674 (0.007)	0.079 (0.002)	0.131 (0.006)
DLAQ[1]	0.903 (0.037)	1.470 (0.042)	1.316 (0.010)	0.171 (0.010)	0.632 (0.017)	0.684 (0.008)	0.550 (0.003)	0.651 (0.010)	0.107 (0.003)	0.107 (0.016)
ELCQ	0.662 (0.035)	1.593 (0.099)	1.080 (0.047)	0.214 (0.066)	0.419 (0.025)	0.660 (0.022)	0.603 (0.004)	0.716 (0.053)	0.157 (0.011)	0.175 (0.059)
Aver.	0.938 (0.247)	1.433 (0.221)	1.308 (0.170)	0.167 (0.049)	0.603 (0.120)	0.704 (0.081)	0.562 (0.023)	0.664 (0.031)	0.111 (0.024)	0.130 (0.024)
JLAI	1.143 (0.046)	1.233 (0.058)	1.420 (0.049)	0.096 (0.043)	0.655 (0.027)	0.766 (0.040)	0.574 (0.017)	0.639 (0.029)	0.117 (0.013)	0.112 (0.005)
PLAI[1]	0.893 (0.002)	1.350 (0.014)	1.205 (0.039)	0.232 (0.024)	0.521 (0.021)	0.684 (0.018)	0.548 (0.016)	0.686 (0.014)	0.124 (0.012)	0.124 (0.009)
ILBI[1]	0.986 (0.001)	1.460 (0.028)	1.449 (0.032)	0.217 (0.023)	0.612 (0.051)	0.837 (0.019)	0.561 (0.001)	0.642 (0.003)	0.122 (0.002)	0.095 (0.002)
Aver.	1.007 (0.126)	1.348 (0.114)	1.358 (0.133)	0.182 (0.075)	0.596 (0.068)	0.762 (0.077)	0.561 (0.013)	0.656 (0.026)	0.121 (0.004)	0.110 (0.015)

Each reading in this table is the average for 3 different runs carried out within one day. Figures between brackets are one standard deviation.
* The unit is: proportion of the cycle duration.
1 Data were averaged for 2 different runs carried out within one day.

Table A5 Temporal-Distance parameters of 11 AK amputees. (0.8 cm Knee Forward Shift, from the normal position)

Sub. Code	Velocity [m/s]	Cycle Duration [s]	Stride Length [m]	Step Width [m]	Sound Step Length [m]	Prosth step Length [m]	Prosthetic Stance *	Sound Stance *	Prosthetic D Support *	Sound D Support *
JLAQ	1.035 (0.017)	1.313 (0.012)	1.362 (0.010)	0.095 (0.012)	0.679 (0.012)	0.683 (0.003)	0.570 (0.011)	0.650 (0.008)	0.100 (0.008)	0.135 (0.015)
PLAQ	0.766 (0.048)	1.513 (0.110)	1.163 (0.037)	0.243 (0.043)	0.584 (0.007)	0.579 (0.031)	0.562 (0.017)	0.683 (0.019)	0.118 (0.008)	0.140 (0.029)
ILBQ[1]	1.297 (0.016)	1.220 (0.028)	1.591 (0.014)	0.141 (0.057)	0.746 (0.009)	0.845 (0.022)	0.589 (0.002)	0.613 (0.009)	0.105 (0.009)	0.113 (0.003)
MRCQ[1]	1.127 (0.034)	1.180 (0.000)	1.290 (0.060)	0.206 (0.043)	0.656 (0.065)	0.633 (0.005)	0.500 (0.047)	0.717 (0.024)	0.117 (0.000)	0.117 (0.024)
TRAQ	1.089 (0.023)	1.360 (0.020)	1.476 (0.054)	0.121 (0.022)	0.741 (0.037)	0.734 (0.026)	0.570 (0.011)	0.671 (0.013)	0.101 (0.001)	0.154 (0.020)
LRBQ	0.629 (0.040)	1.747 (0.133)	1.110 (0.080)	0.134 (0.044)	0.483 (0.055)	0.627 (0.027)	0.535 (0.025)	0.681 (0.041)	0.080 (0.006)	0.148 (0.014)
DLAQ	0.894 (0.020)	1.433 (0.050)	1.274 (0.014)	0.175 (0.032)	0.553 (0.025)	0.721 (0.011)	0.523 (0.015)	0.698 (0.027)	0.101 (0.008)	0.134 (0.034)
ELCQ	0.882 (0.019)	1.353 (0.050)	1.218 (0.037)	0.230 (0.004)	0.567 (0.022)	0.651 (0.019)	0.539 (0.006)	0.689 (0.008)	0.112 (0.010)	0.131 (0.005)
Aver.	0.965 (0.214)	1.390 (0.180)	1.311 (0.161)	0.168 (0.054)	0.626 (0.094)	0.684 (0.083)	0.549 (0.029)	0.675 (0.032)	0.184 (0.012)	0.134 (0.014)
JLAI	0.801 (0.118)	1.487 (0.070)	1.205 (0.126)	0.117 (0.021)	0.598 (0.078)	0.607 (0.048)	0.598 (0.016)	0.704 (0.026)	0.110 (0.012)	0.204 (0.038)
PLAI[1]	0.965 (0.003)	1.300 (0.028)	1.242 (0.010)	0.216 (0.021)	0.573 (0.029)	0.669 (0.019)	0.553 (0.001)	0.659 (0.003)	0.114 (0.013)	0.114 (0.008)
ILBI[1]	1.082 (0.042)	1.380 (0.057)	1.480 (0.004)	0.153 (0.011)	0.685 (0.010)	0.705 (0.014)	0.571 (0.003)	0.636 (0.005)	0.107 (0.006)	0.114 (0.005)
Aver.	0.949 (0.141)	1.389 (0.094)	1.309 (0.149)	0.162 (0.050)	0.619 (0.059)	0.660 (0.050)	0.574 (0.023)	0.666 (0.035)	0.110 (0.004)	0.144 (0.052)

Each reading in this table is the average for 3 different runs carried out within one day. Figures between brackets are one standard deviation.
* The unit is: proportion of the cycle duration.
1 Data were averaged for 2 different runs carried out within one day.

Table A 6 Temporal-Distance parameters of 11 AK amputees. (0.8 cm Knee Backward Shift, from the normal position)

Sub. Code	Velocity [m/s]	Cycle Duration [s]	Stride Length [m]	Step Width [m]	Sound Step Length [m]	Prosth step Length [m]	Prosthetic Stance *	Sound Stance *	Prosthetic D Support *	Sound D Support *
JLAQ	1.036 (0.041)	1.347 (0.058)	1.391 (0.023)	0.093 (0.022)	0.643 (0.016)	0.748 (0.007)	0.581 (0.008)	0.615 (0.001)	0.107 (0.007)	0.103 (0.004)
PLAQ	0.708 (0.035)	1.600 (0.069)	1.169 (0.015)	0.280 (0.005)	0.535 (0.014)	0.634 (0.015)	0.564 (0.005)	0.695 (0.006)	0.140 (0.008)	0.132 (0.002)
ILBQ	1.311 (0.023)	1.213 (0.012)	1.614 (0.041)	0.209 (0.025)	0.727 (0.011)	0.887 (0.029)	0.578 (0.013)	0.632 (0.011)	0.114 (0.001)	0.114 (0.001)
MRCQ	1.215 (0.025)	1.133 (0.042)	1.361 (0.007)	0.168 (0.039)	0.686 (0.043)	0.675 (0.038)	0.549 (0.005)	0.671 (0.012)	0.121 (0.004)	0.116 (0.008)
TRAQ	0.953 (0.062)	1.527 (0.083)	1.414 (0.042)	0.159 (0.014)	0.676 (0.033)	0.738 (0.037)	0.564 (0.021)	0.651 (0.010)	0.091 (0.005)	0.138 (0.014)
LRBQ	0.624 (0.032)	1.793 (0.042)	1.107 (0.040)	0.135 (0.047)	0.439 (0.009)	0.668 (0.032)	0.522 (0.025)	0.673 (0.034)	0.077 (0.002)	0.129 (0.016)
DLAQ	0.899 (0.033)	1.407 (0.050)	1.261 (0.028)	0.217 (0.015)	0.501 (0.014)	0.760 (0.014)	0.509 (0.011)	0.720 (0.015)	0.121 (0.005)	0.122 (0.024)
ELCQ	0.773 (0.055)	1.480 (0.106)	1.175 (0.014)	0.211 (0.052)	0.516 (0.020)	0.659 (0.014)	0.572 (0.019)	0.703 (0.029)	0.138 (0.006)	0.152 (0.022)
Aver.	0.940 (0.241)	1.438 (0.212)	1.312 (0.167)	0.184 (0.058)	0.590 (0.105)	0.721 (0.081)	0.555 (0.026)	0.670 (0.036)	0.114 (0.022)	0.126 (0.015)
JLAI	1.089 (0.021)	1.320 (0.020)	1.458 (0.040)	0.110 (0.037)	0.671 (0.014)	0.787 (0.029)	0.587 (0.012)	0.612 (0.006)	0.109 (0.009)	0.104 (0.002)
PLAI[1]	0.968 (0.051)	1.310 (0.042)	1.275 (0.005)	0.228 (0.011)	0.550 (0.003)	0.725 (0.009)	0.557 (0.018)	0.662 (0.000)	0.128 (0.015)	0.105 (0.003)
ILBI[1]	1.116 (0.047)	1.320 (0.028)	1.478 (0.094)	0,161 (0.054)	0.617 (0.045)	0.860 (0.049)	0.582 (0.033)	0.634 (0.003)	0.127 (0.034)	0.105 (0.002)
Aver.	1.058 (0.079)	1.317 (0.006)	1.404 (0.112)	0.166 (0.059)	0.613 (0.061)	0.791 (0.068)	0.575 (0.016)	0.636 (0.025)	0.121 (0.011)	0.105 (0.000)

Each reading in this table is the average of 3 different runs carried out within one day. Figures between brackets are one standard deviation.
* The unit is: proportion of the cycle duration.
1 Data were averaged for 2 different runs carried out within one day.

Appendix B

Kinetic and Kinematic Results for Normal Subjects

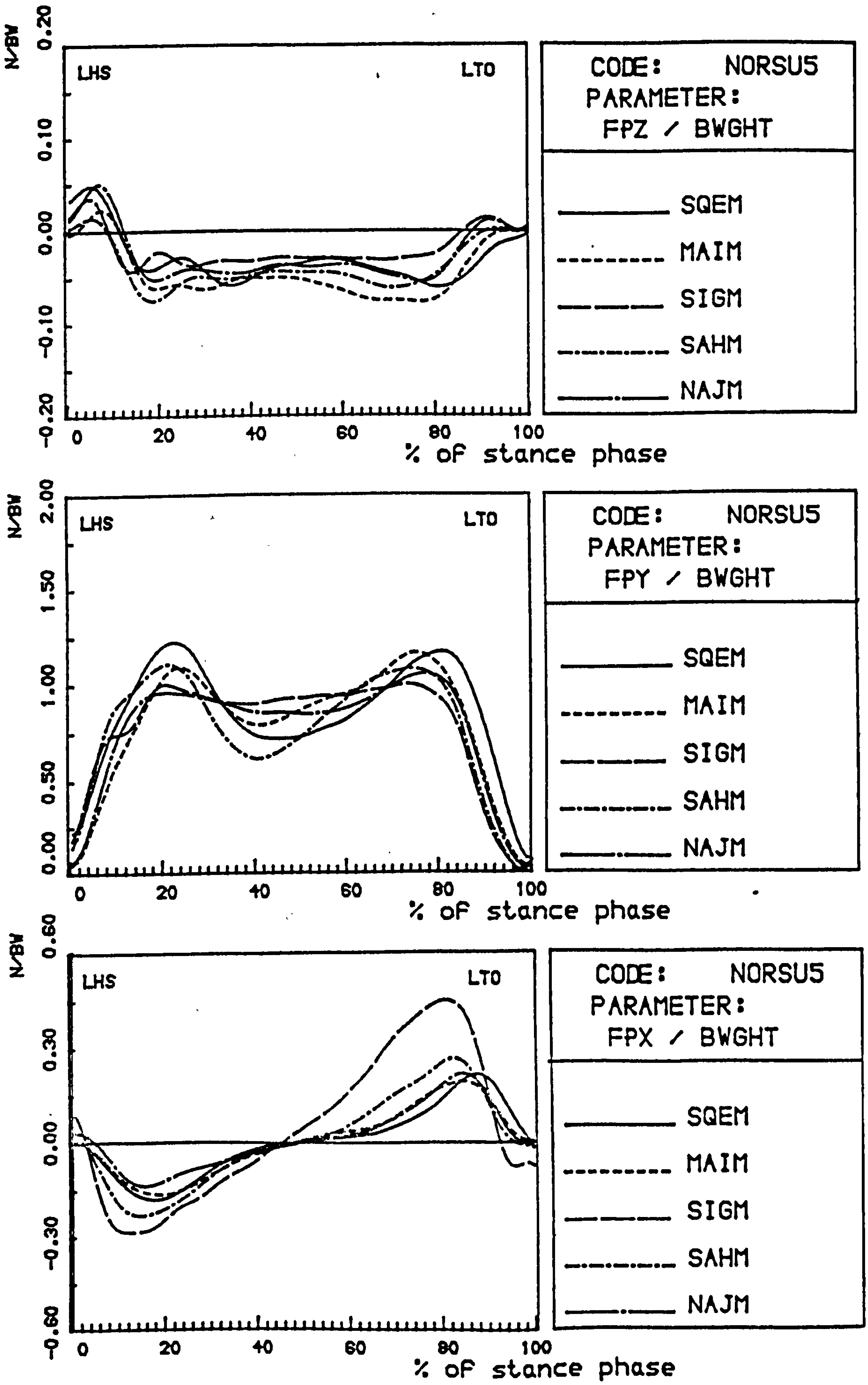


Figure B.1 Ground reaction forces with time for five normal subjects. (Left leg).

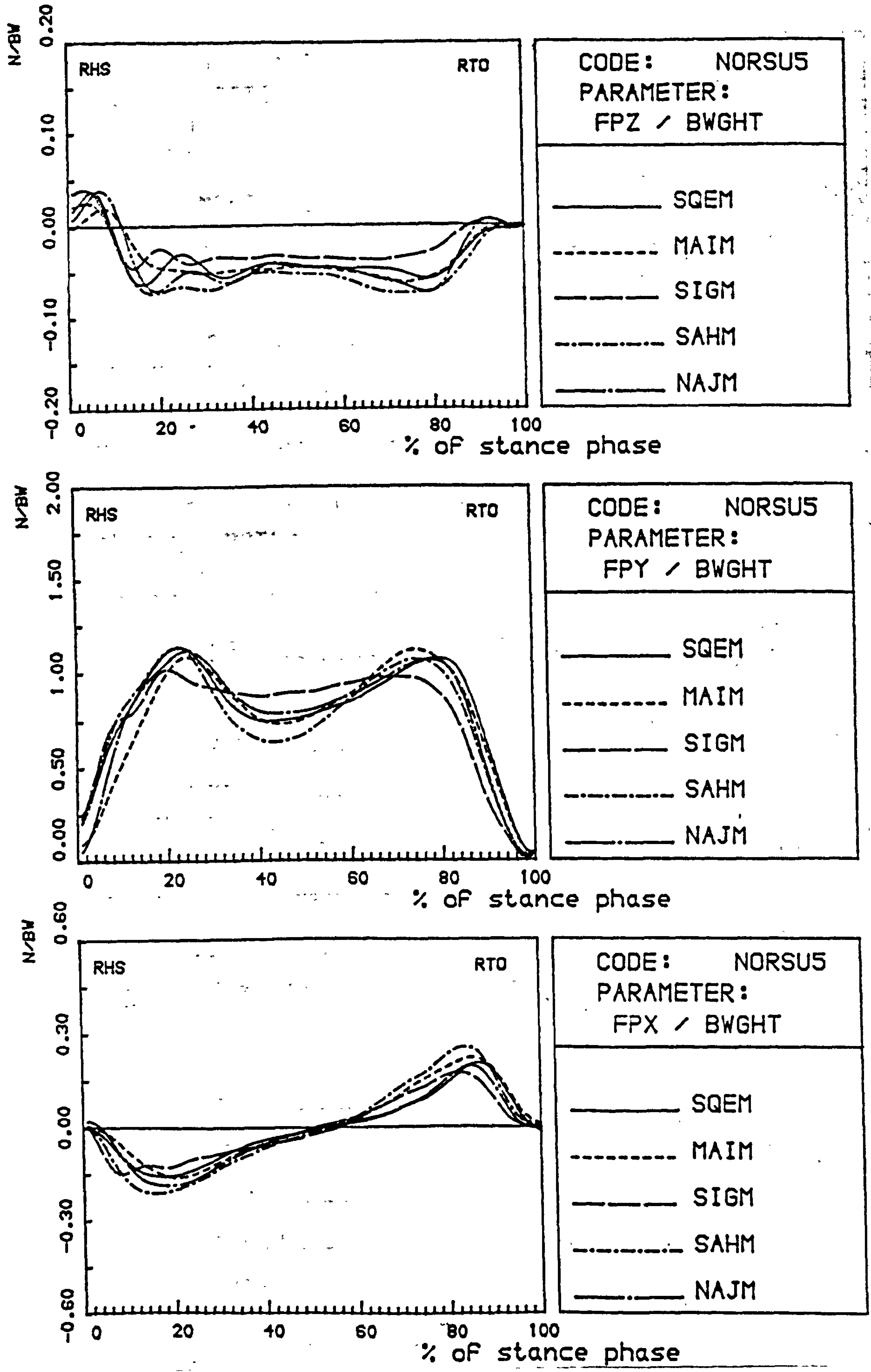


Figure B.2 Ground reaction forces with time for five normal subjects. (Right leg).

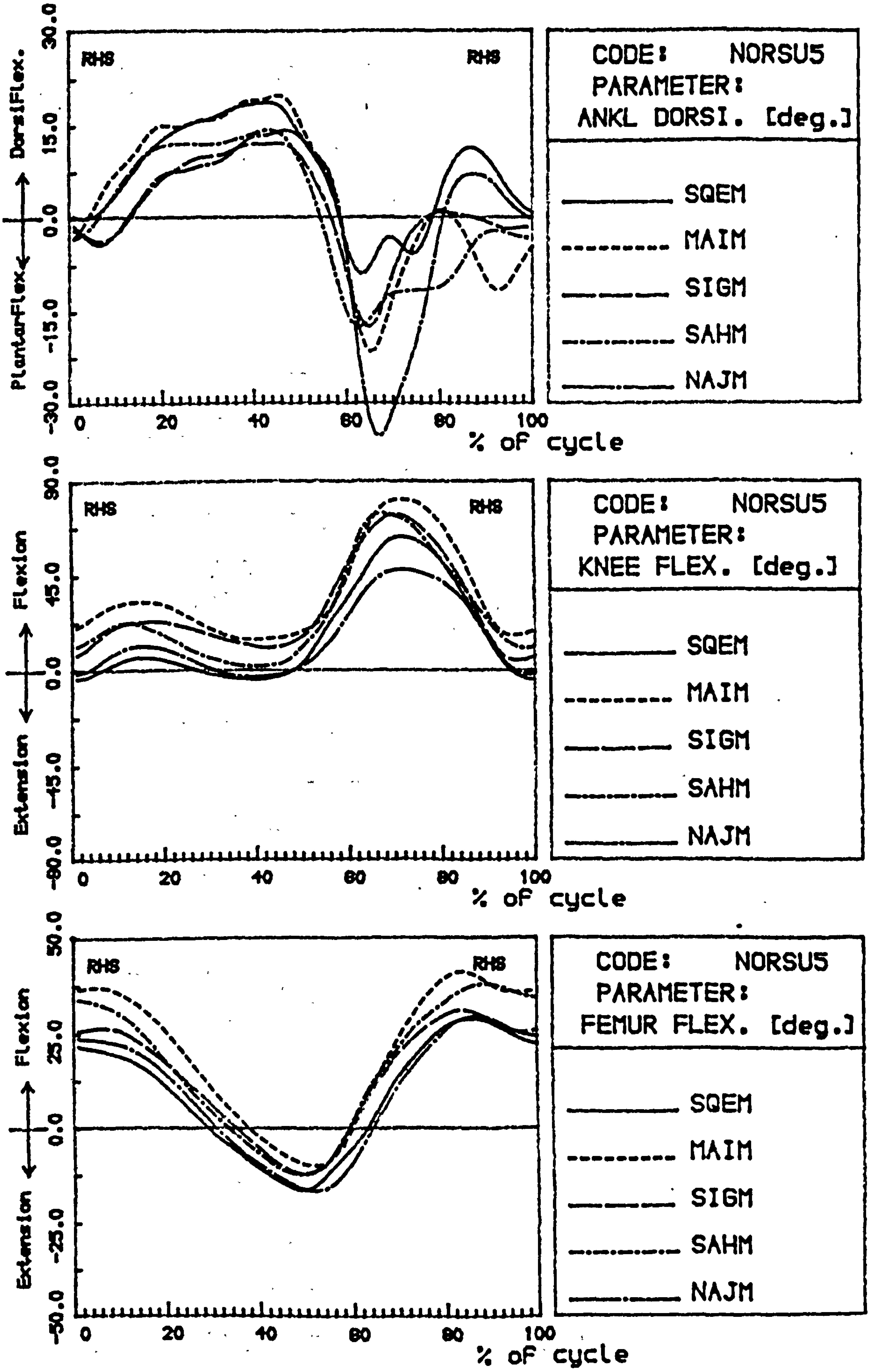


Figure B.3 A/P angular displacement of the lower limb joints with time for five normal subjects. (Right leg).

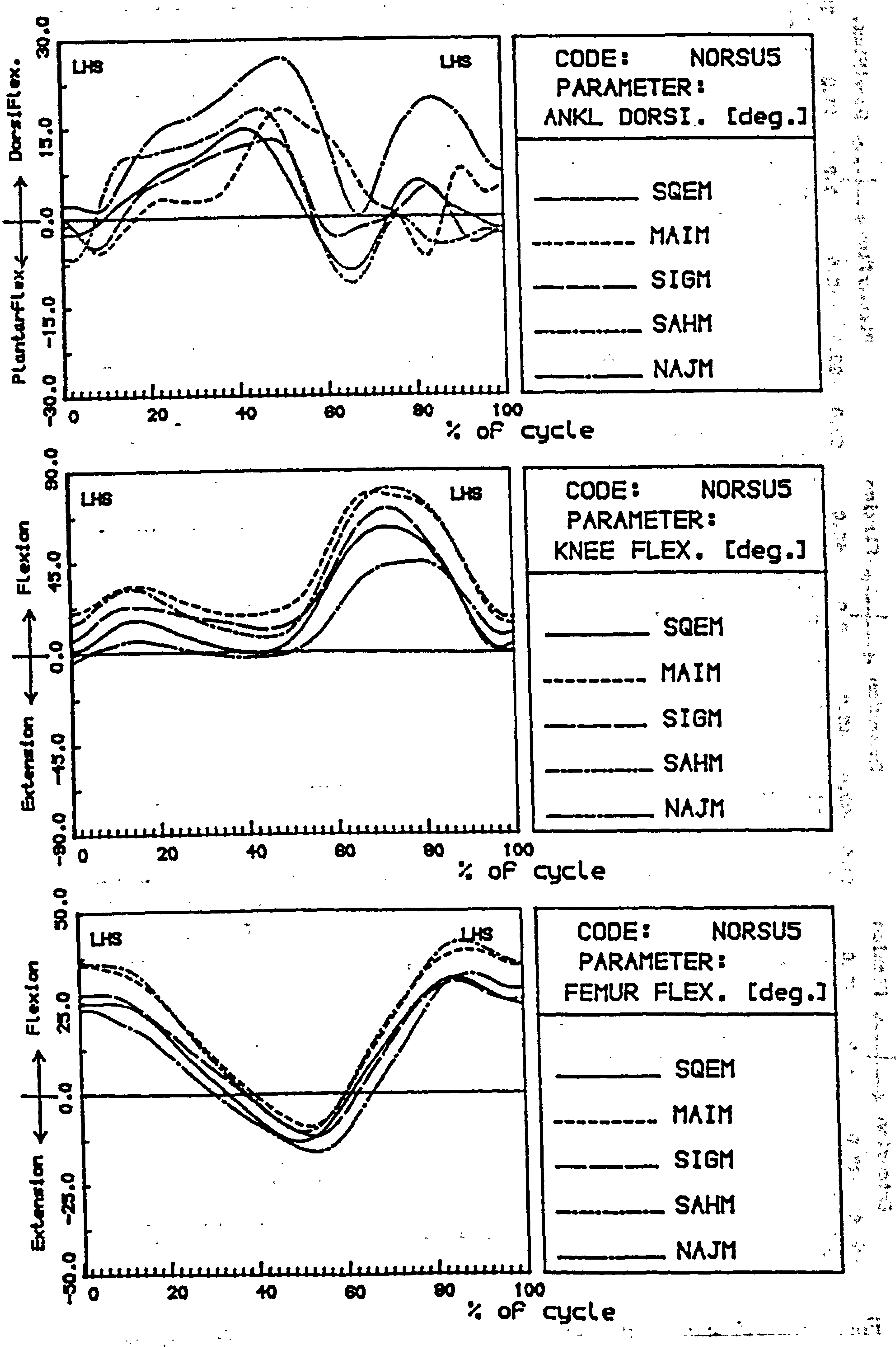


Figure B.4 A/P angular displacement of the lower limb joints with time for five normal subjects. (Left leg).

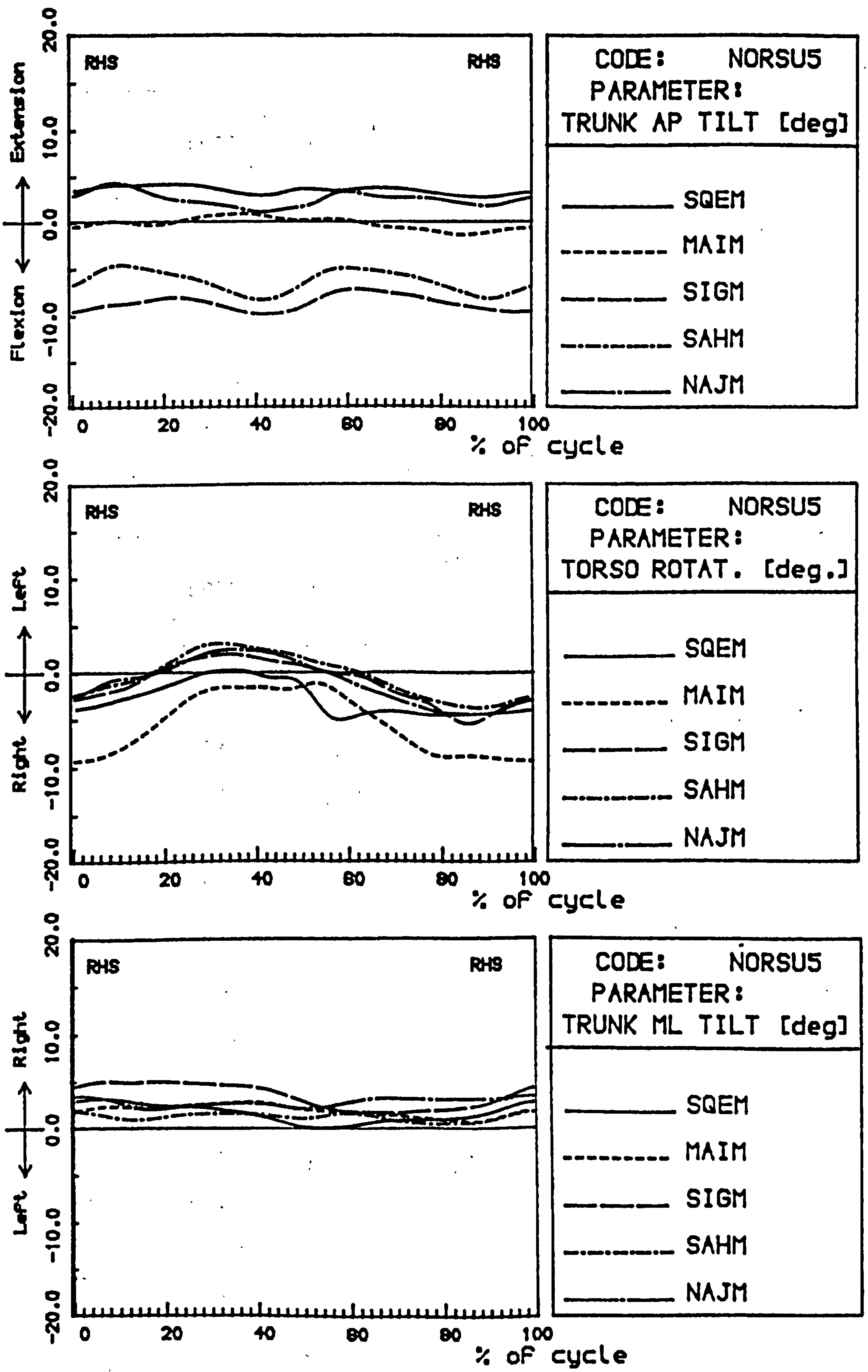


Figure B.5 A/P and M/L angular displacements of the trunk and transverse rotational displacement of the torso with time for five normal subjects.

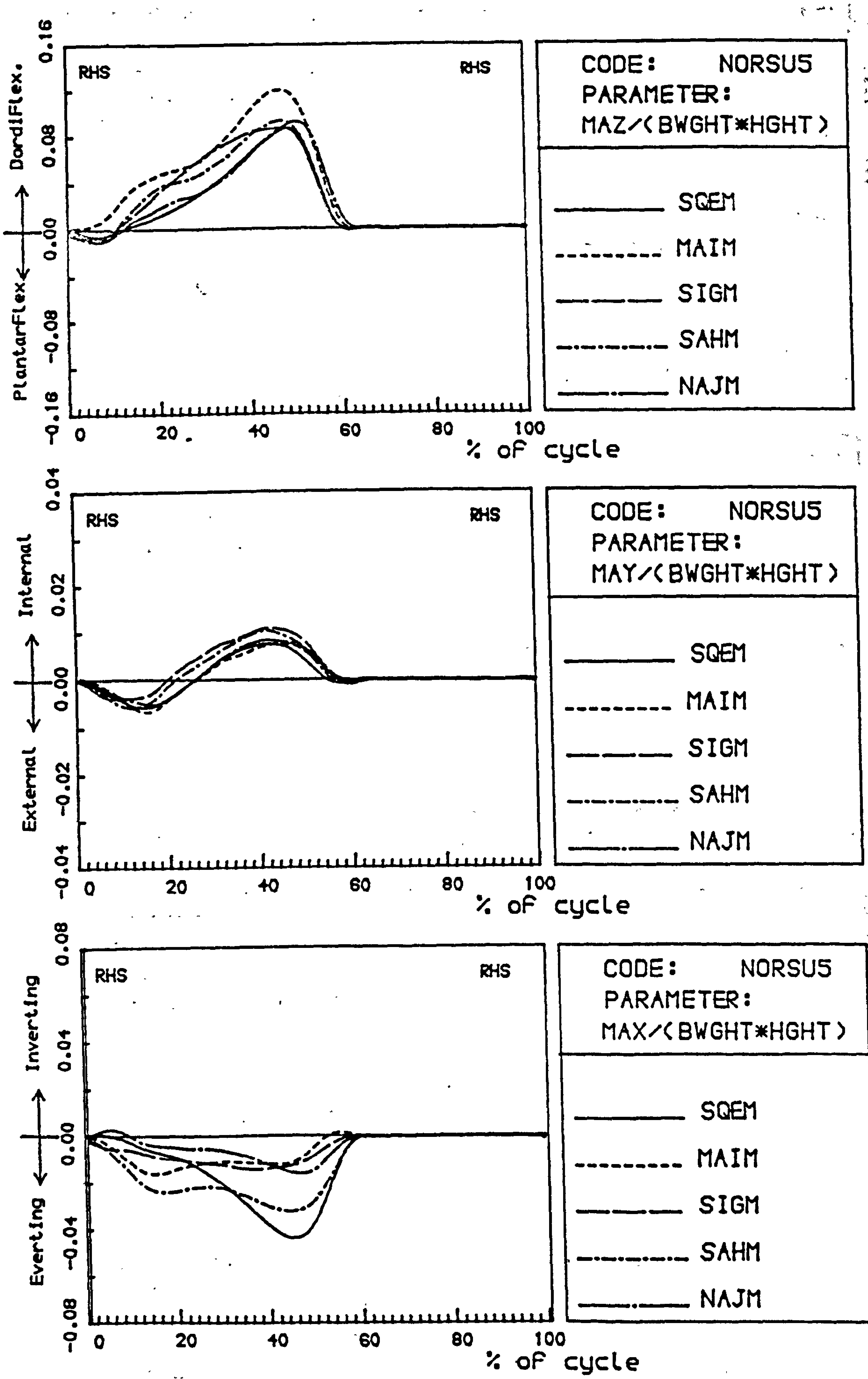


Figure B.6 Ankle joint moments with time for five normal subjects. (Right leg).

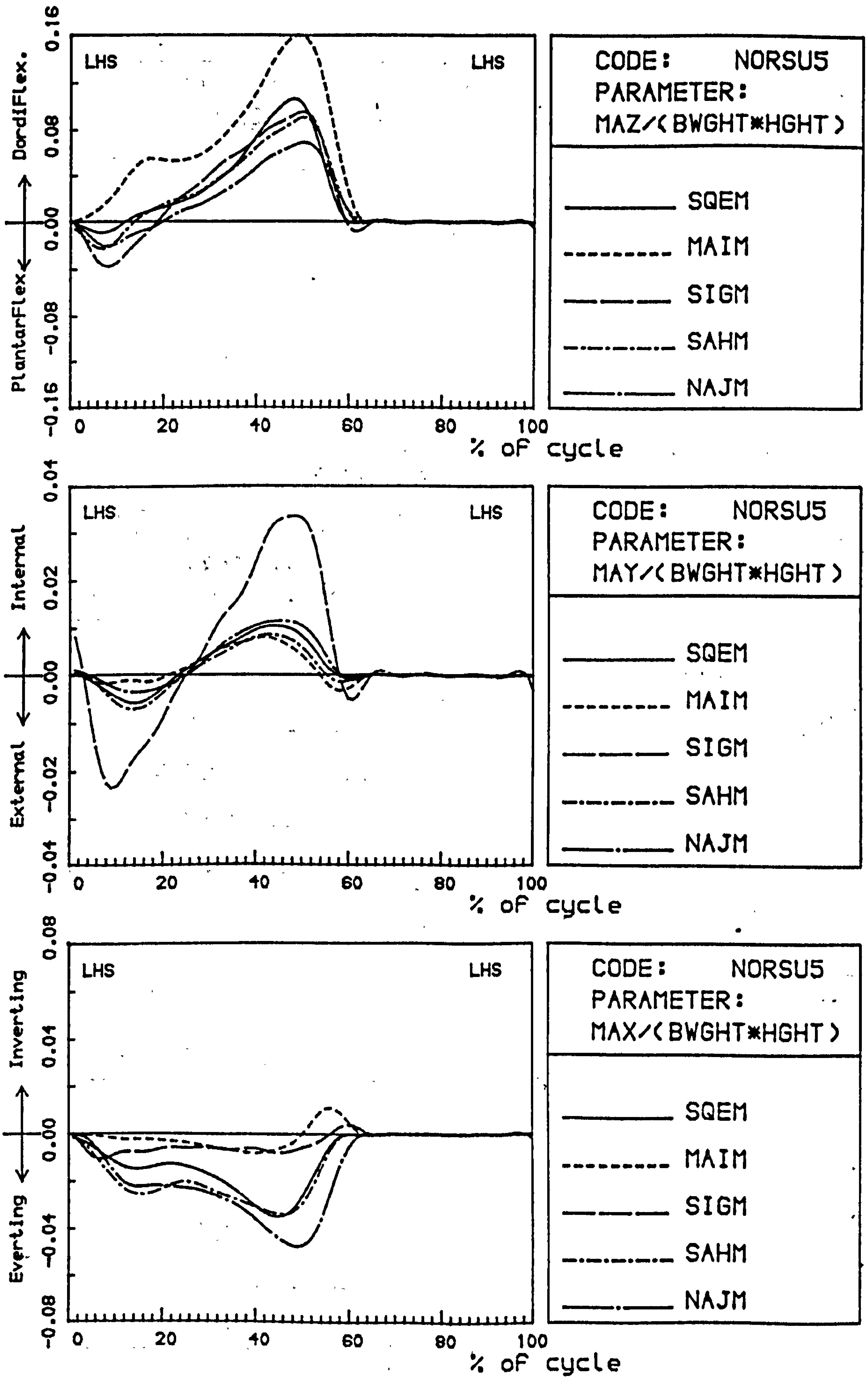


Figure B.7 Ankle joint moments with time for five normal subjects. (Left leg).

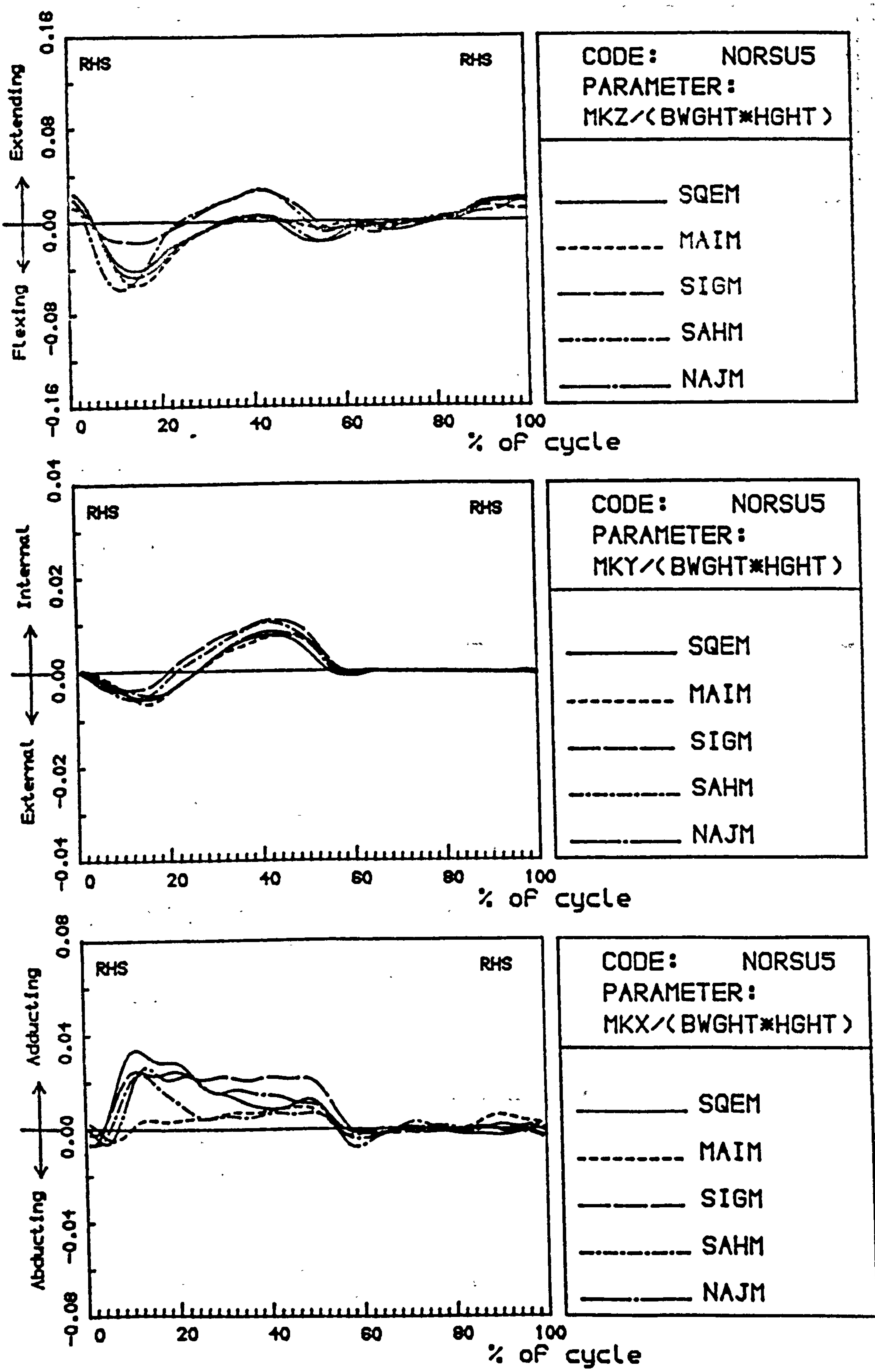


Figure B.8 Knee joint moments with time for five normal subjects. (Right leg).

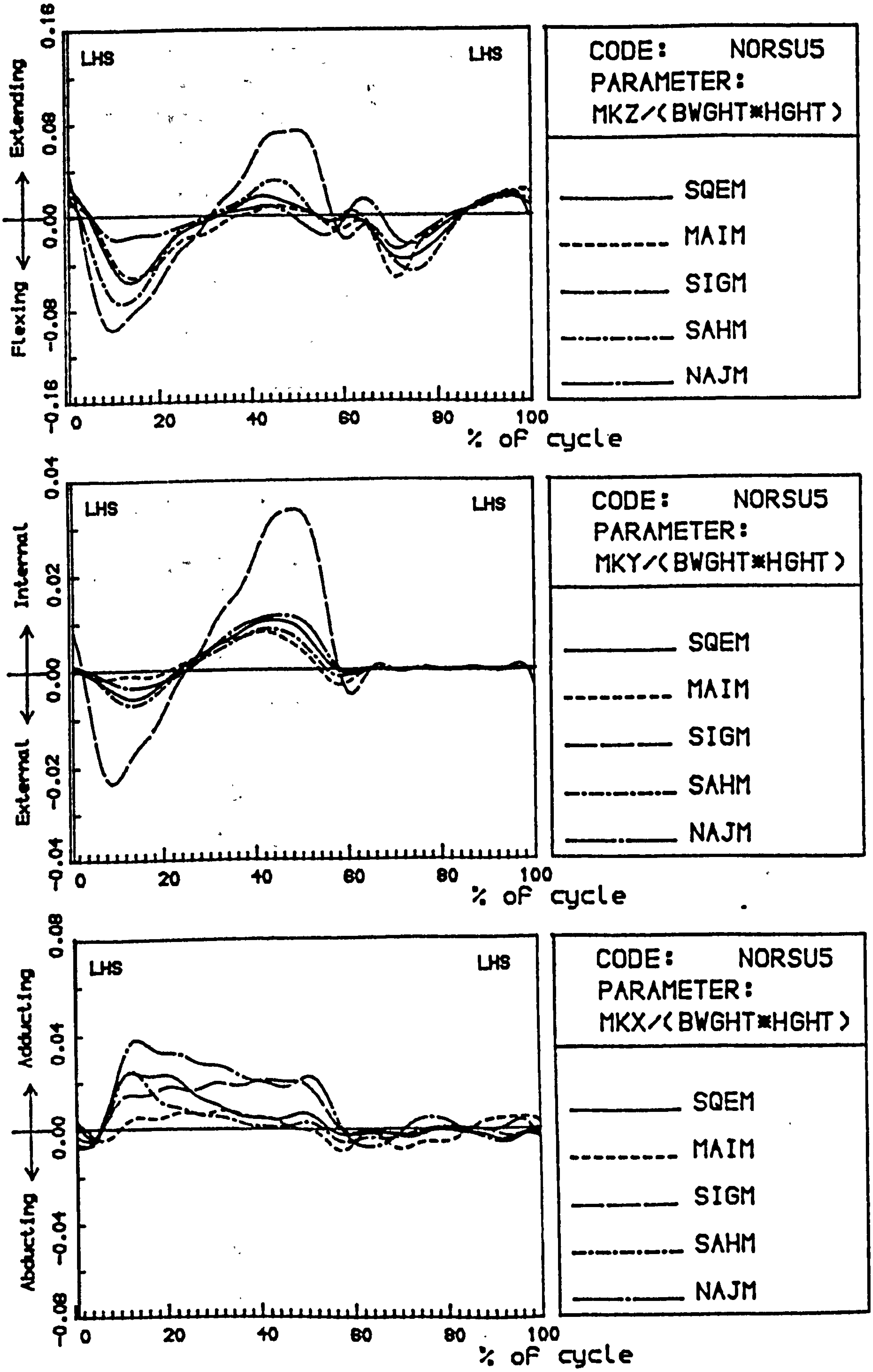


Figure B.9 Knee joint moments with time for five normal subjects. (Left leg).

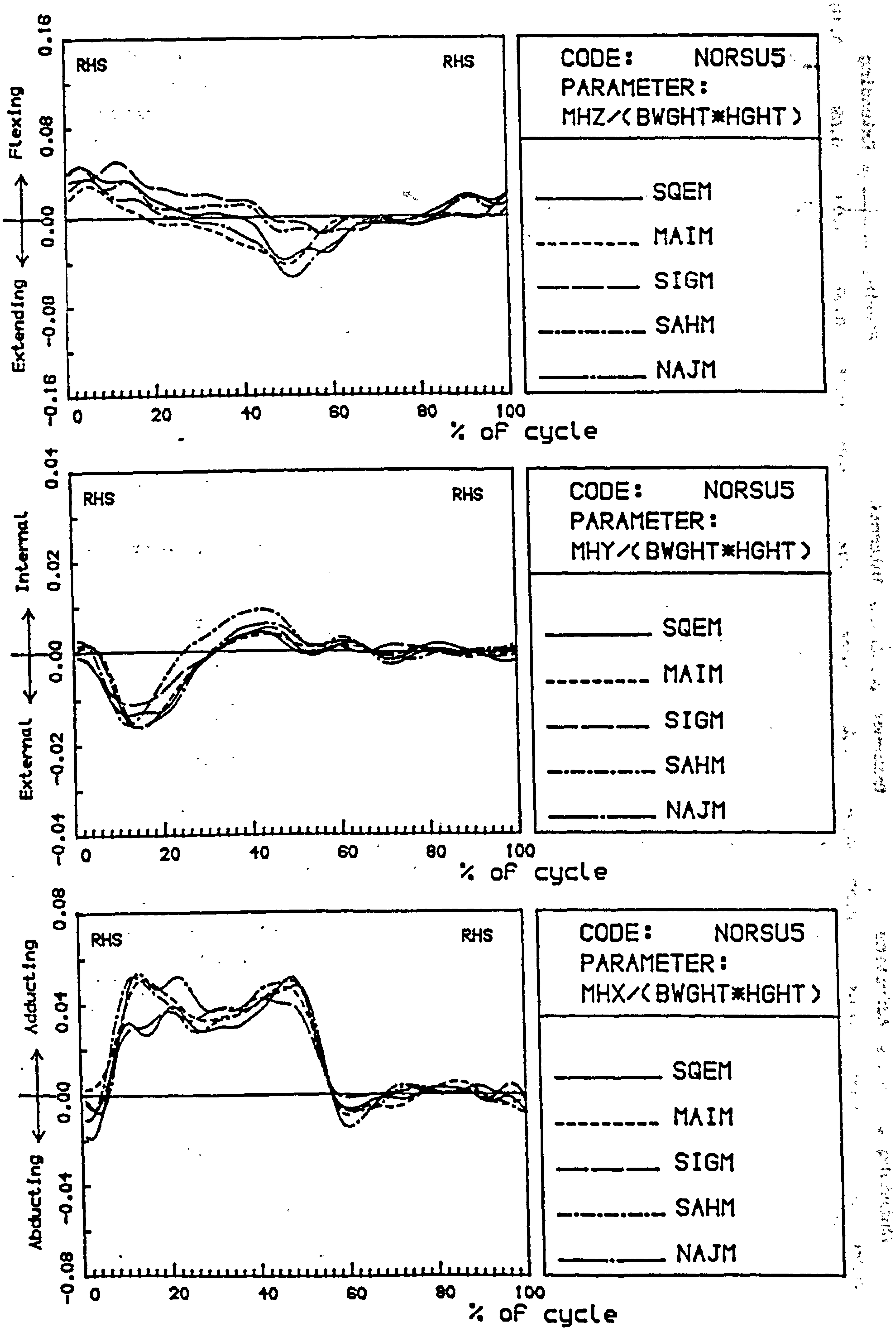


Figure B.10 Hip joint moments with time for five normal subjects. (Right leg).

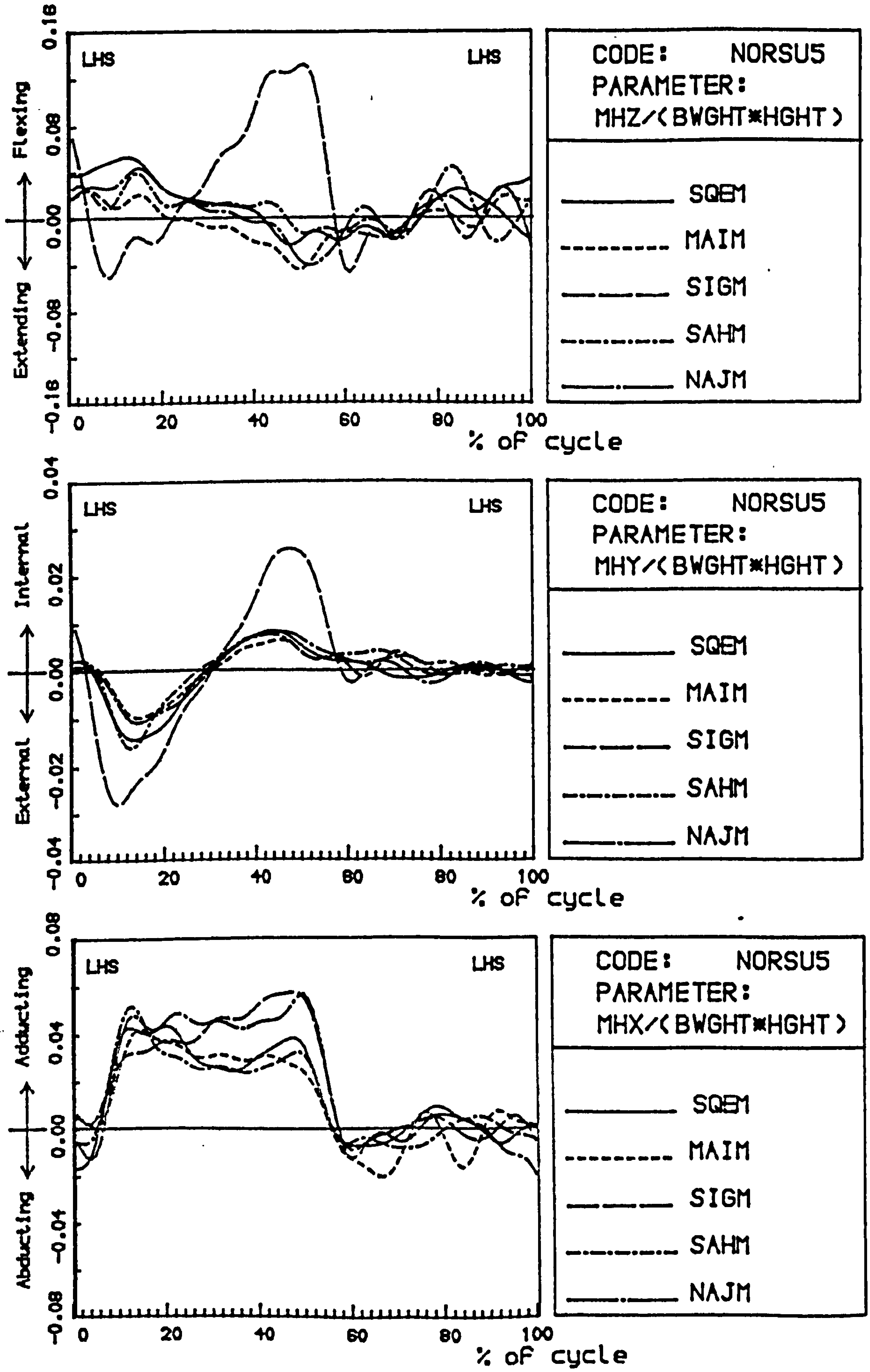


Figure B.11 Hip joint moments with time for five normal subjects. (Left leg).

Appendix C

Kinetic and Kinematic Results for Above Knee Amputees

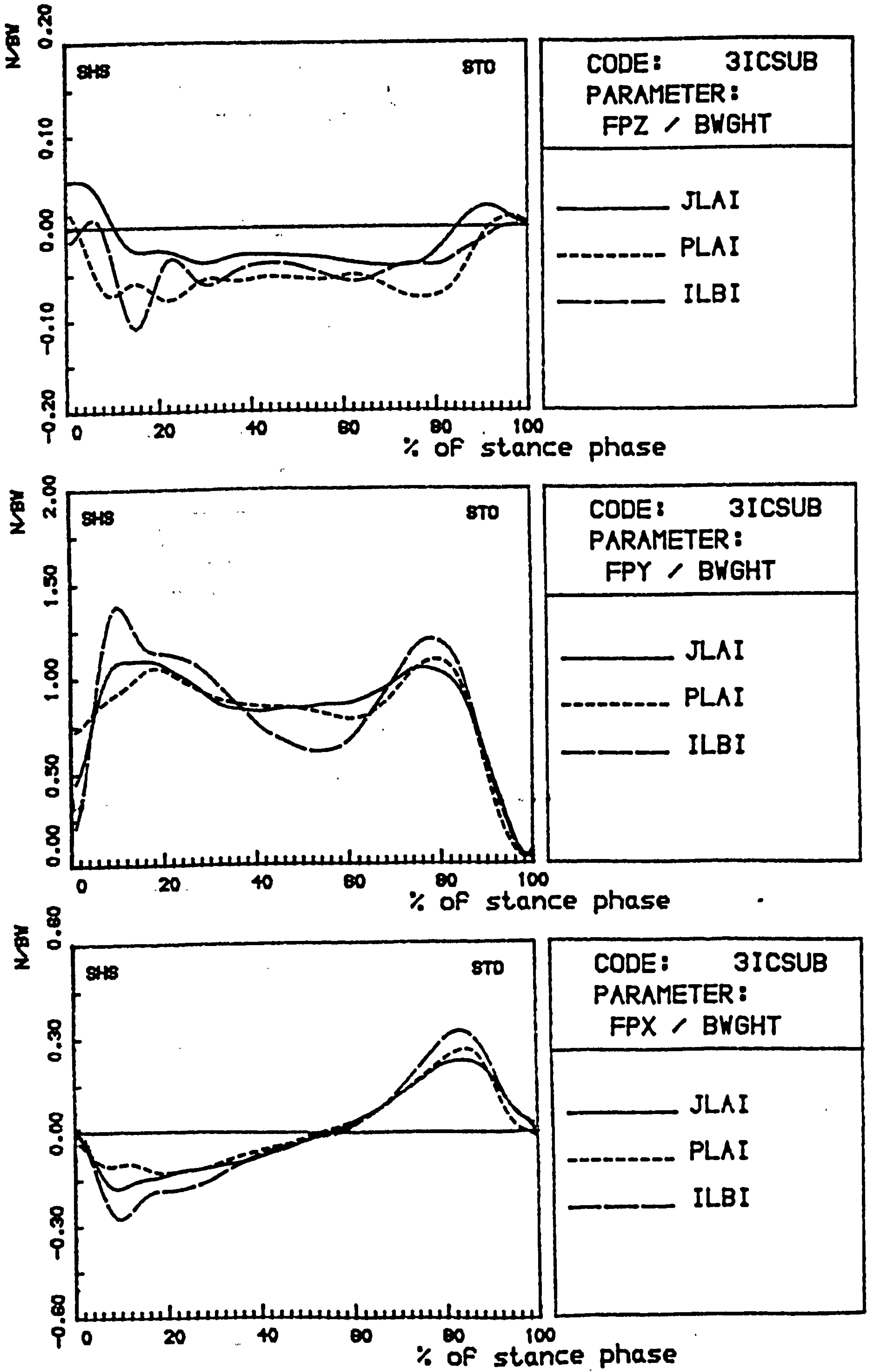


Figure C.1 Ground reaction forces on the sound leg with time for three AK amputees fitted with IC sockets. (Normal alignment).

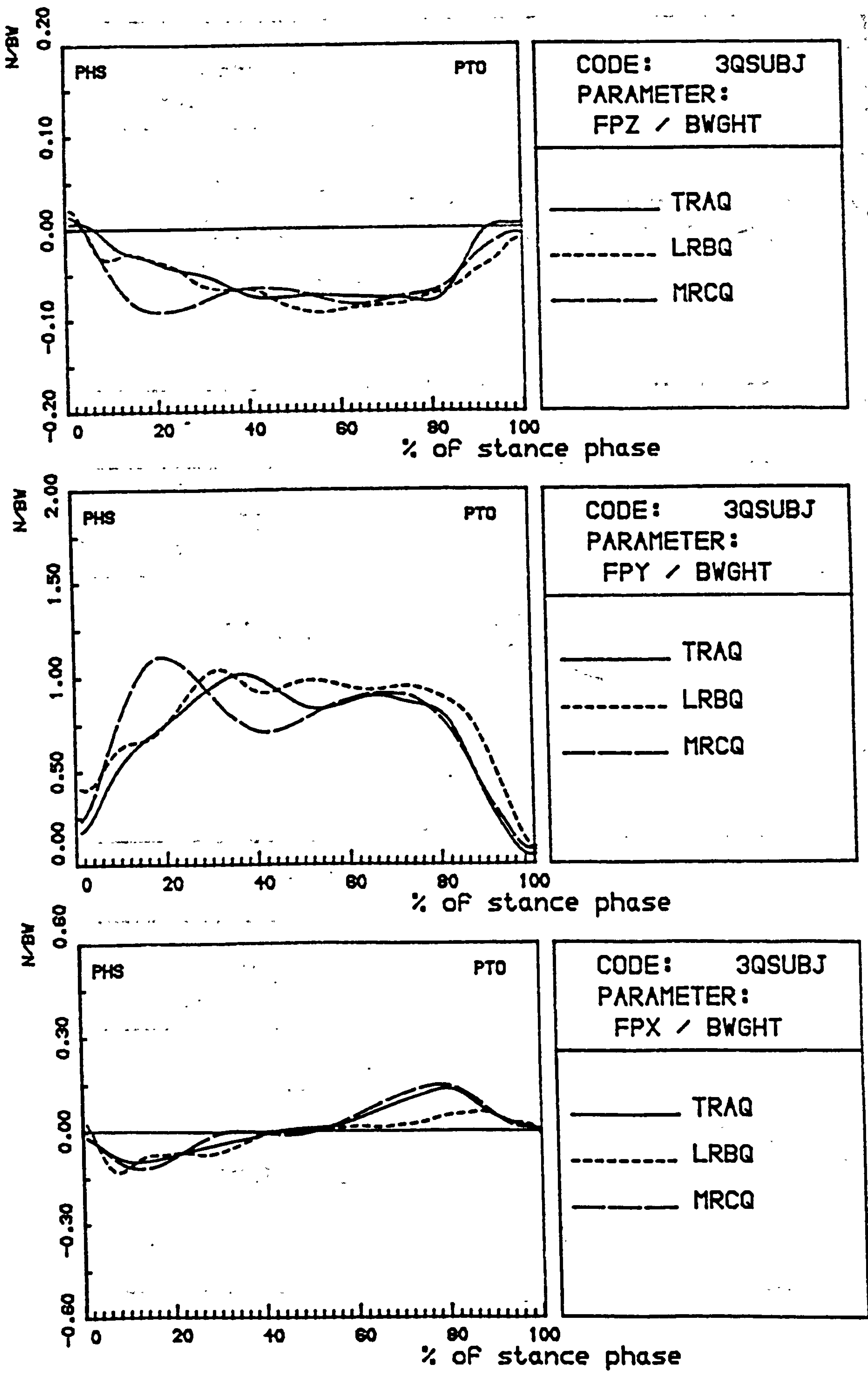


Figure C.2 Ground reaction forces on the prosthetic leg with time for three AK amputees fitted with quadrilateral sockets. (Normal alignment).

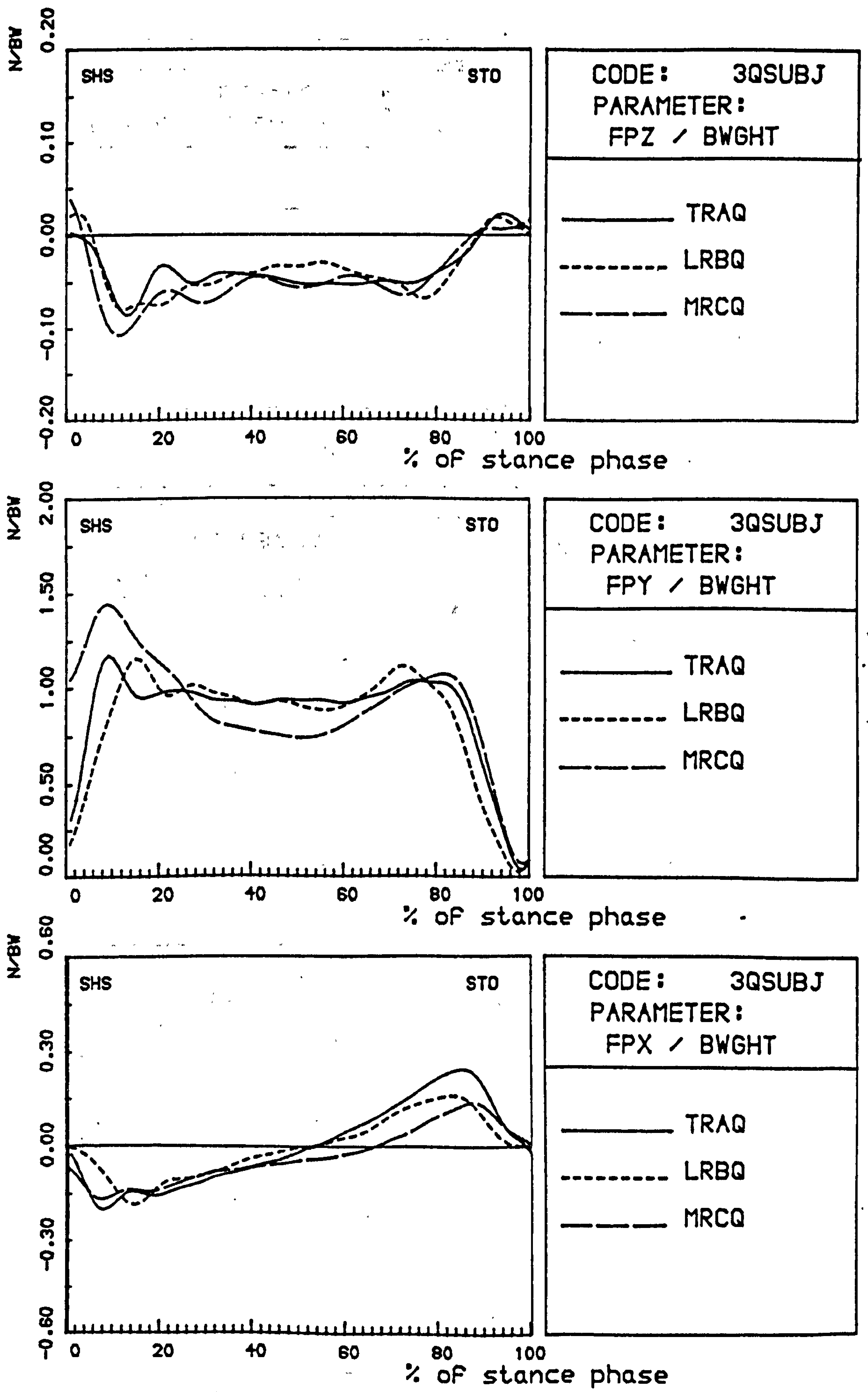


Figure C.3 Ground reaction forces on the sound leg with time for three AK amputees fitted with quadrilateral sockets. (Normal alignment).

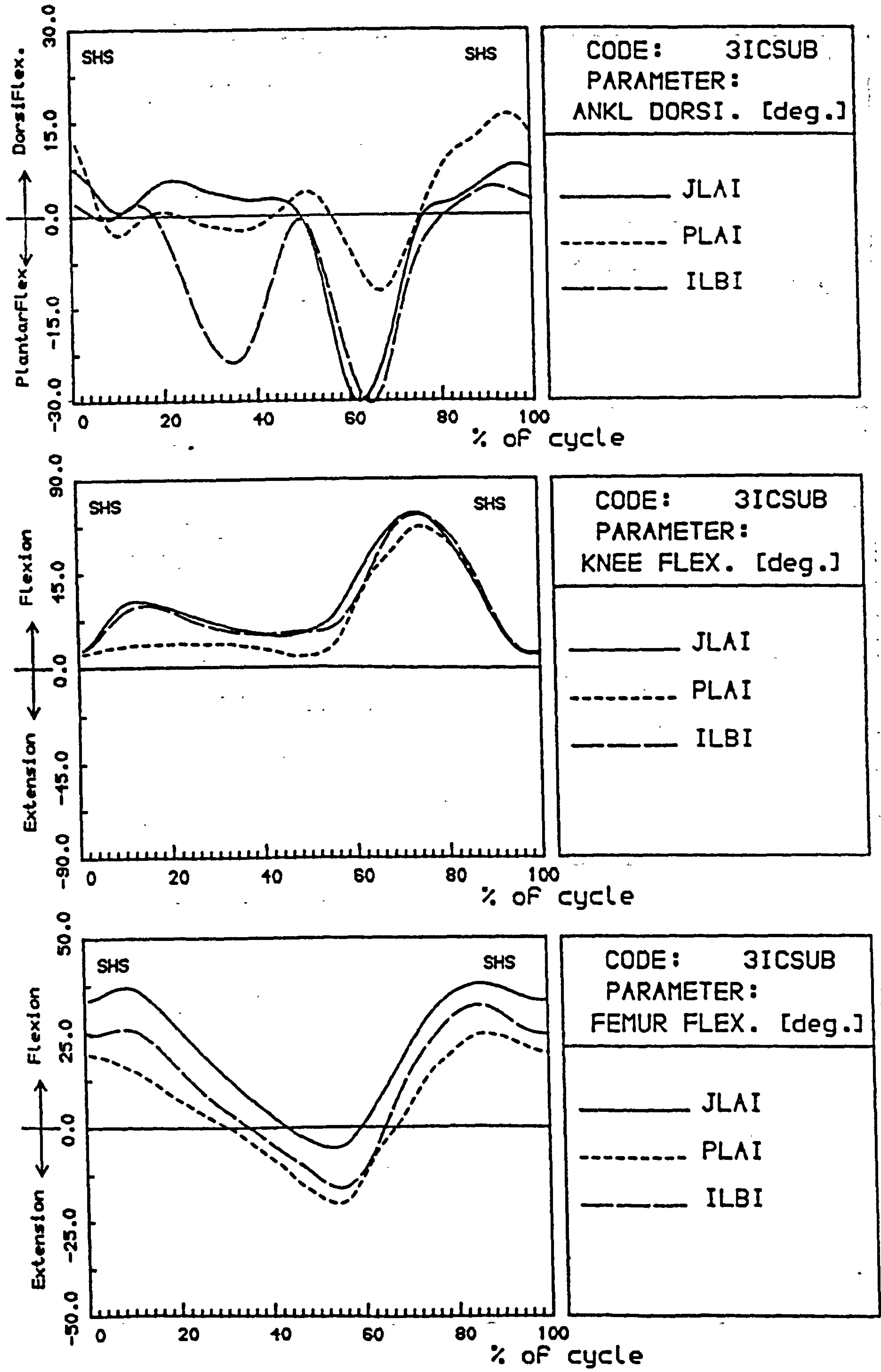


Figure C.4 A/P angular displacement of the lower limb joints with time for the sound leg for three AK amputees fitted with IC socket. (Normal alignment).

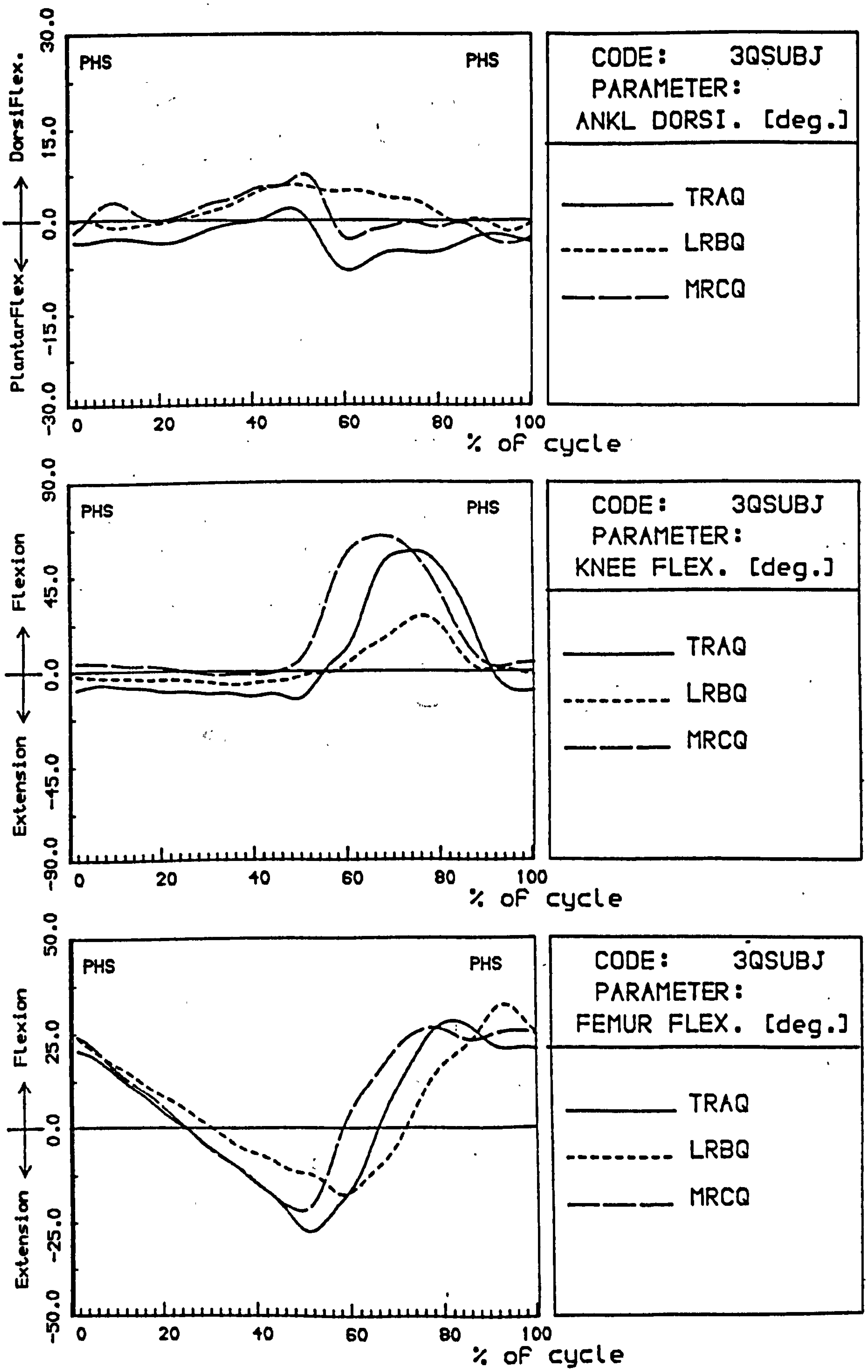


Figure C.5 A/P angular displacement of the lower limb joints with time for the prosthetic leg for three AK amputees fitted with quad sockets. (Normal alignment).

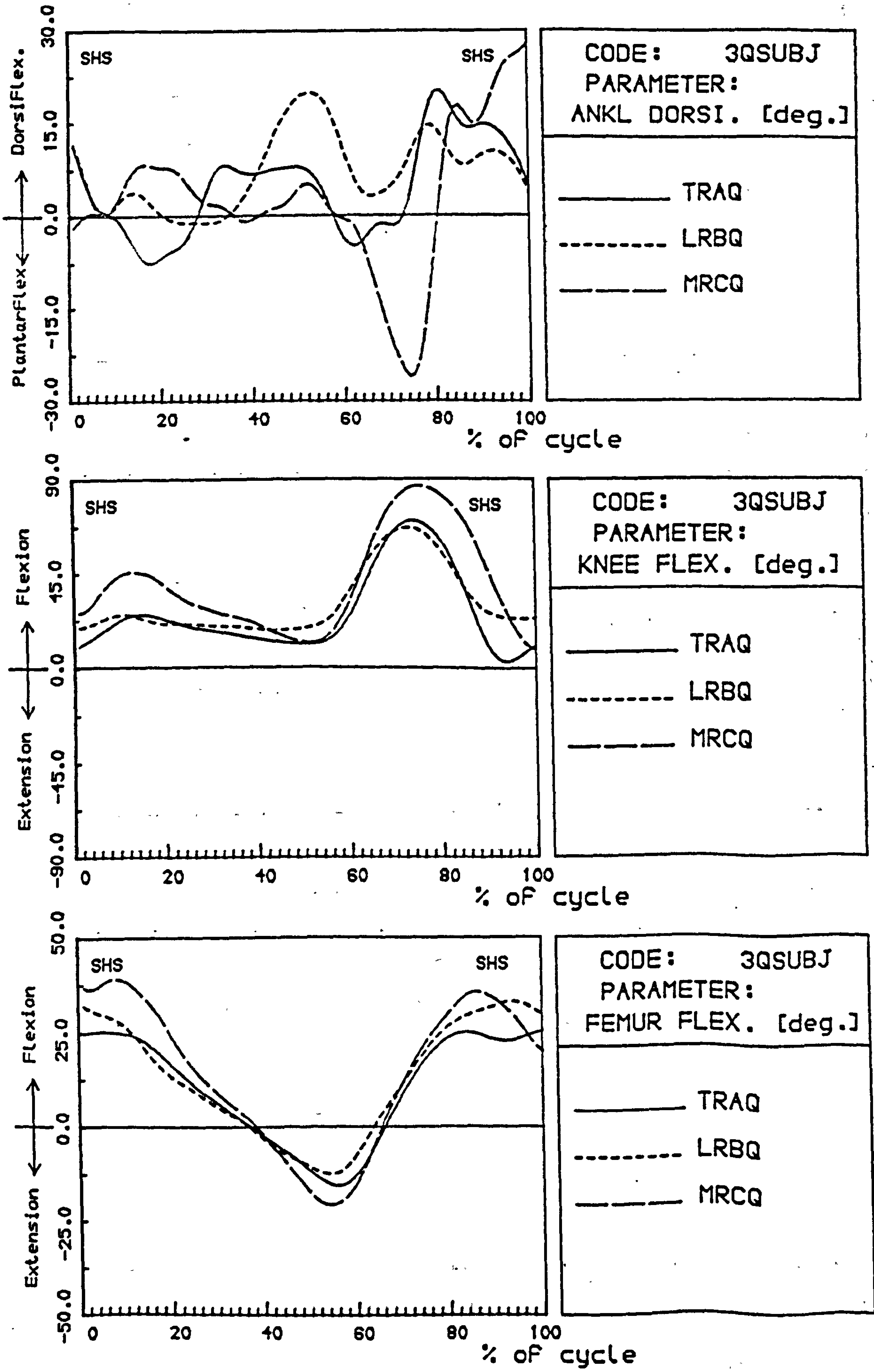


Figure C.6 A/P angular displacement of the lower limb joints for the sound leg with time for three AK amputees fitted with quad sockets.
(Normal alignment).

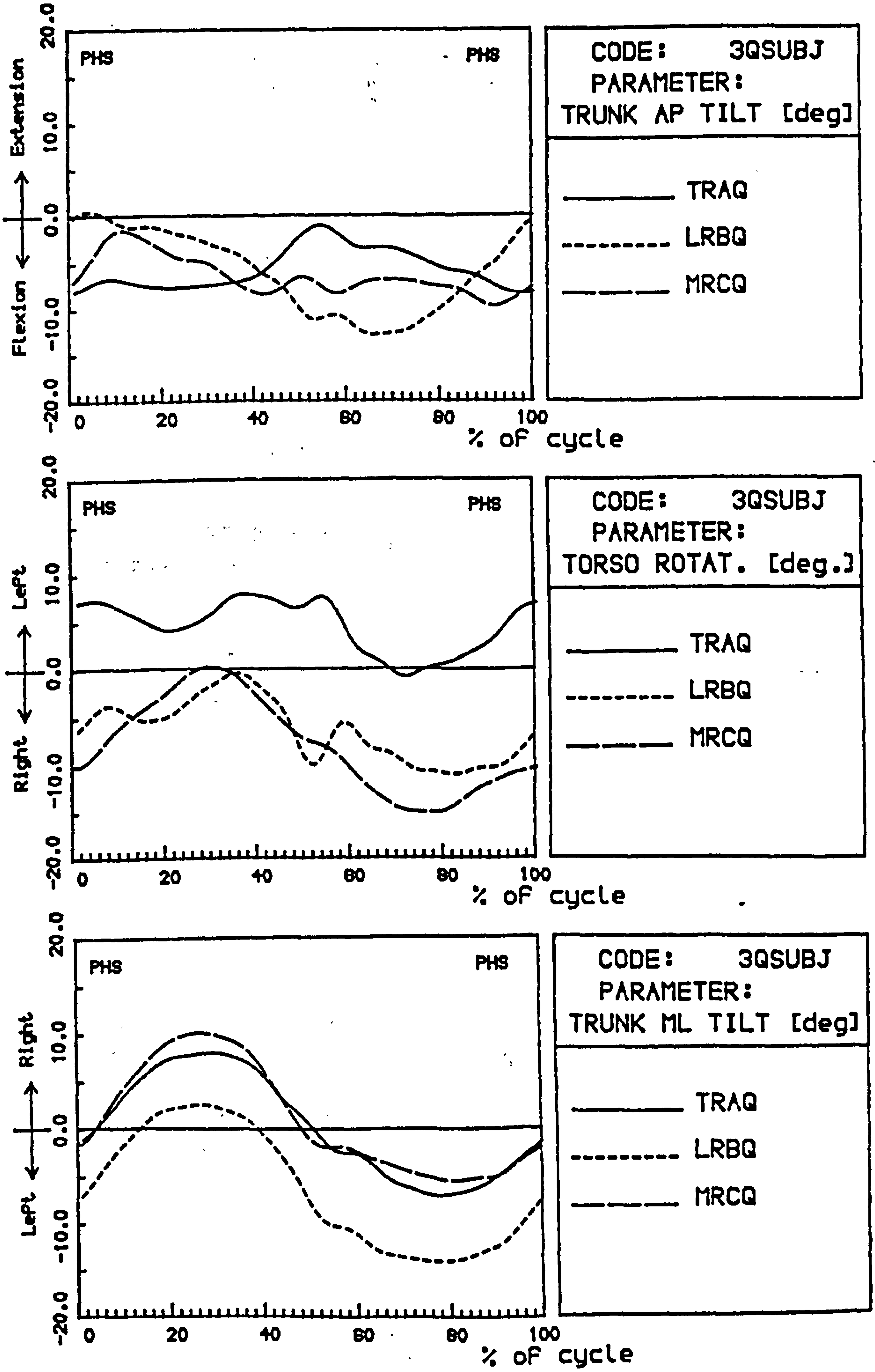


Figure C.7 A/P and M/L angular displacements of the trunk and transverse rotational displacement of the torso with time for three AK amputees fitted with quadrilateral sockets. (Normal alignment).

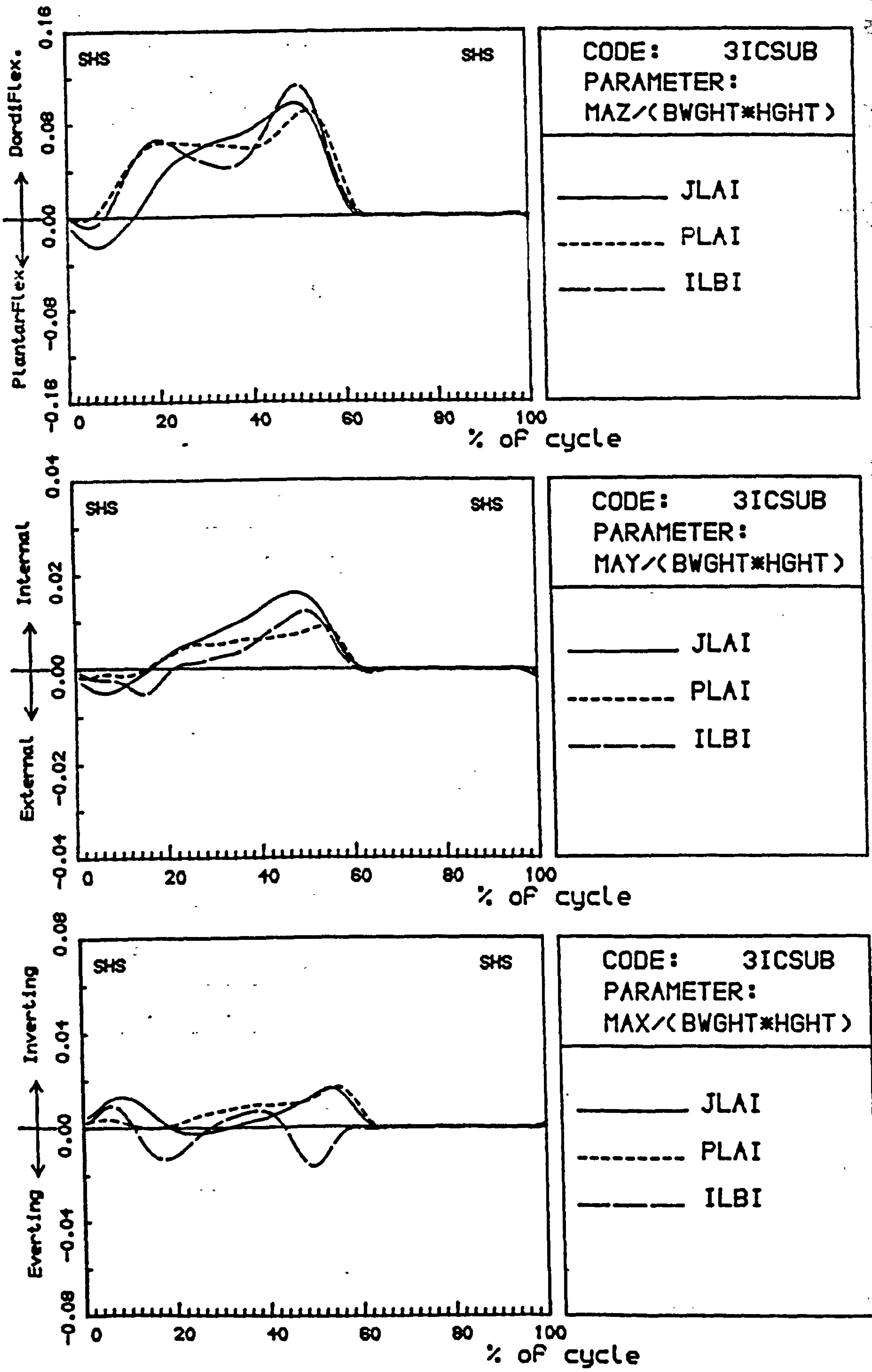


Figure C.8 Sound ankle joint moments with time for three AK amputees fitted with IC sockets. (Normal alignment)

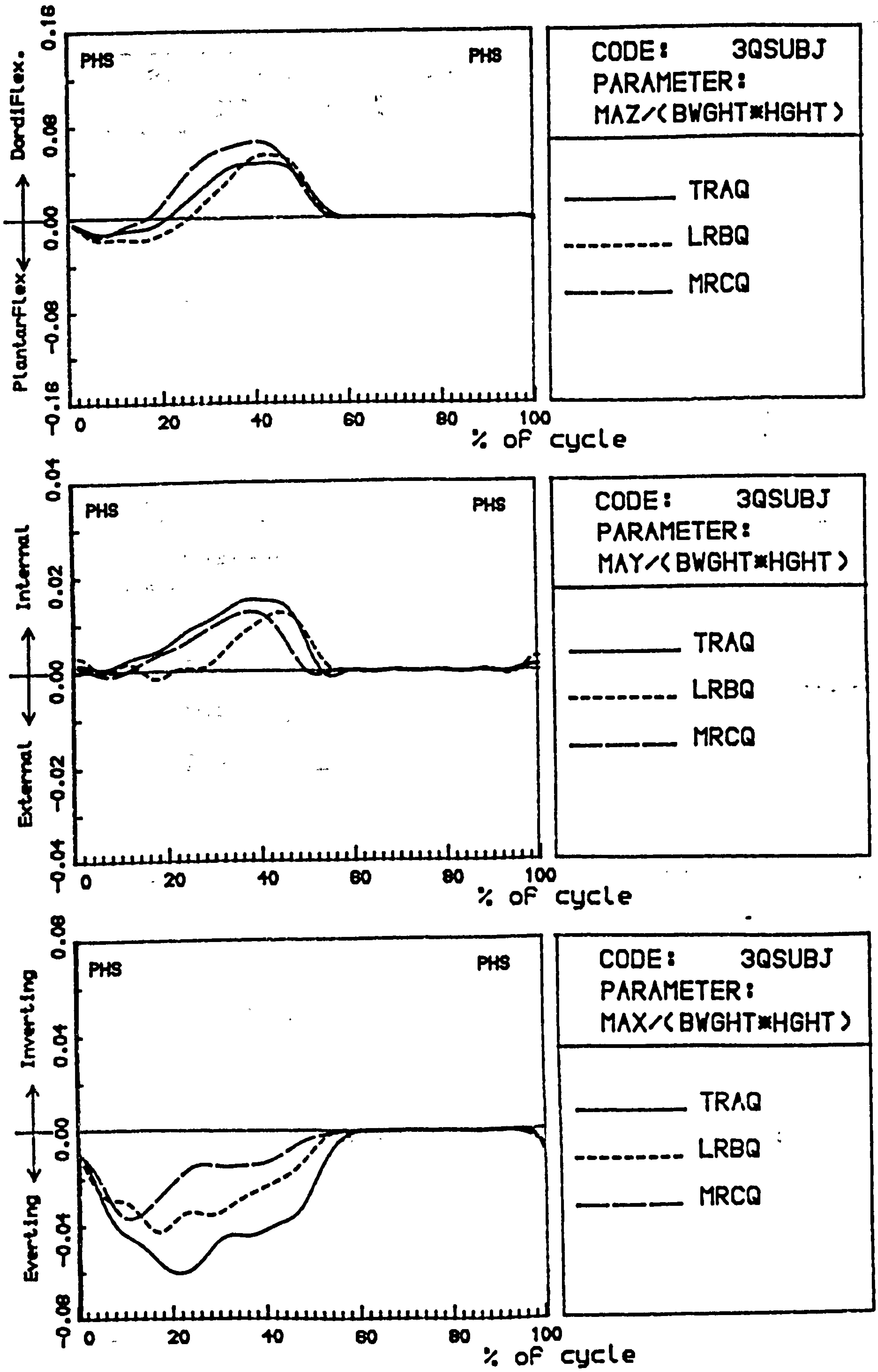


Figure C.9 Prosthetic ankle joint moments with time for three AK amputees fitted with quadrilateral sockets. (Normal alignment)

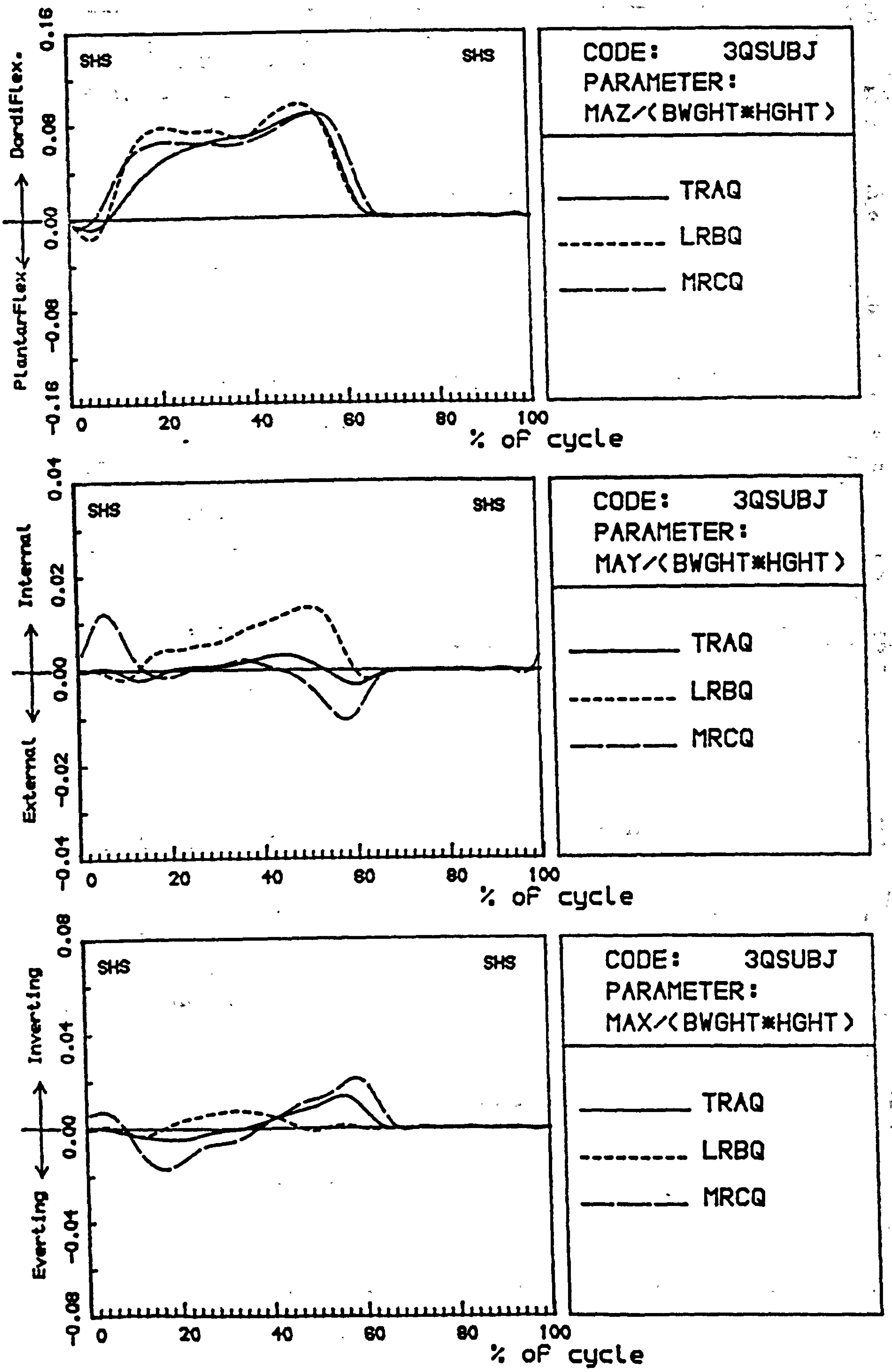


Figure C.10 Ankle joint moments on the sound leg with time for three AK amputees fitted with quadrilateral sockets. (Normal alignment)

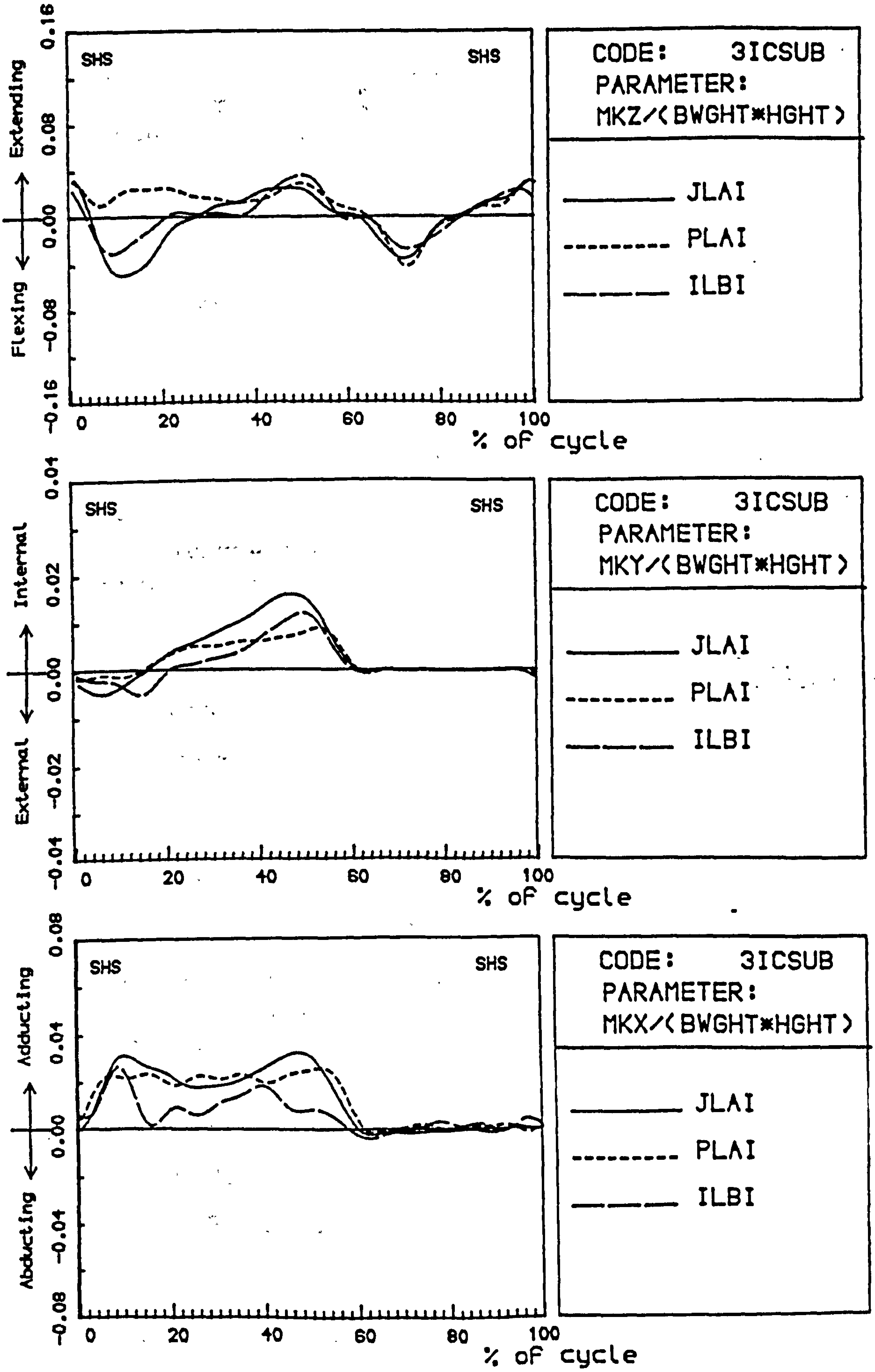
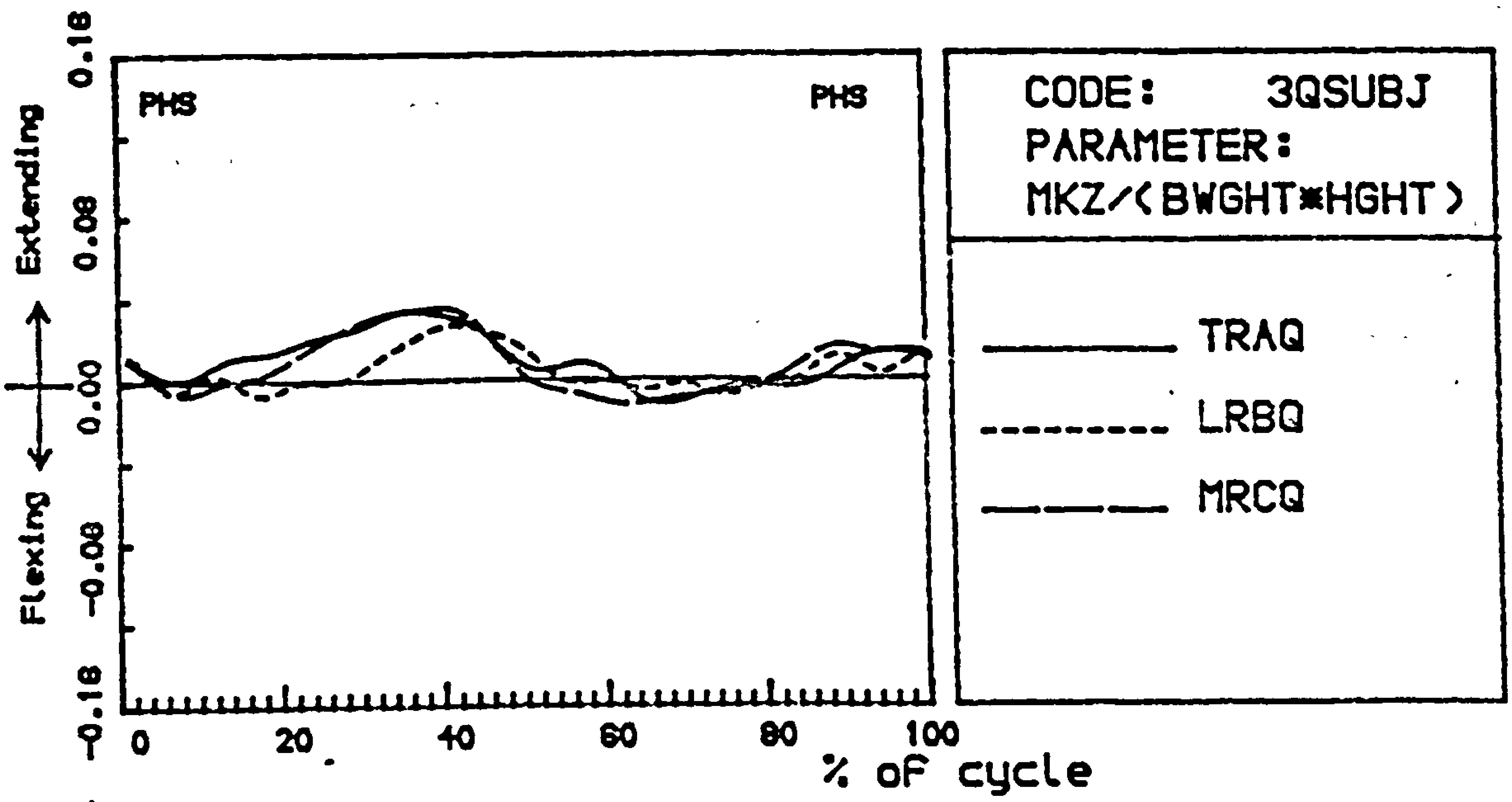
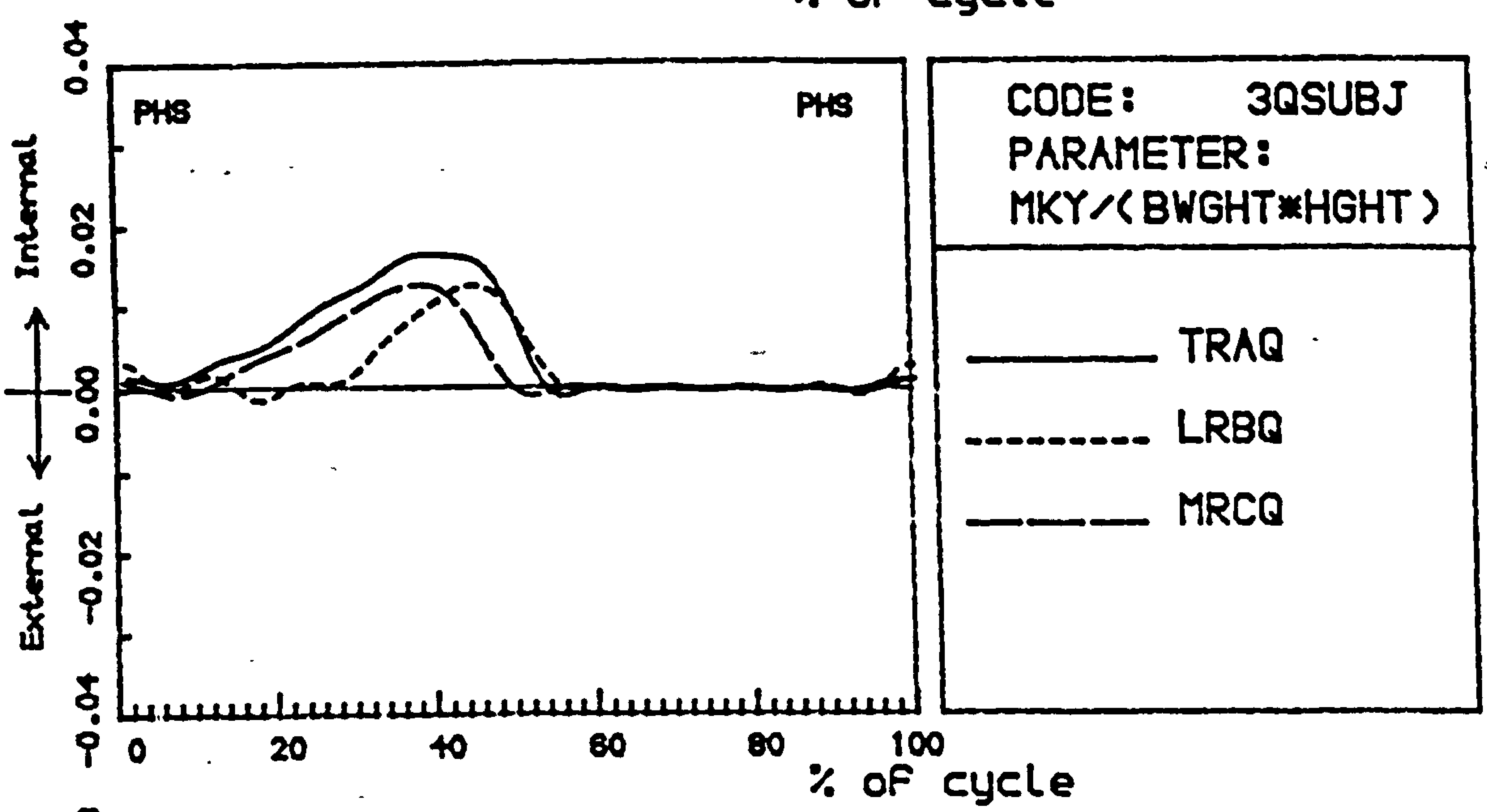


Figure C.11 Sound knee joint moments with time for three AK amputees fitted with IC sockets. (Normal alignment)





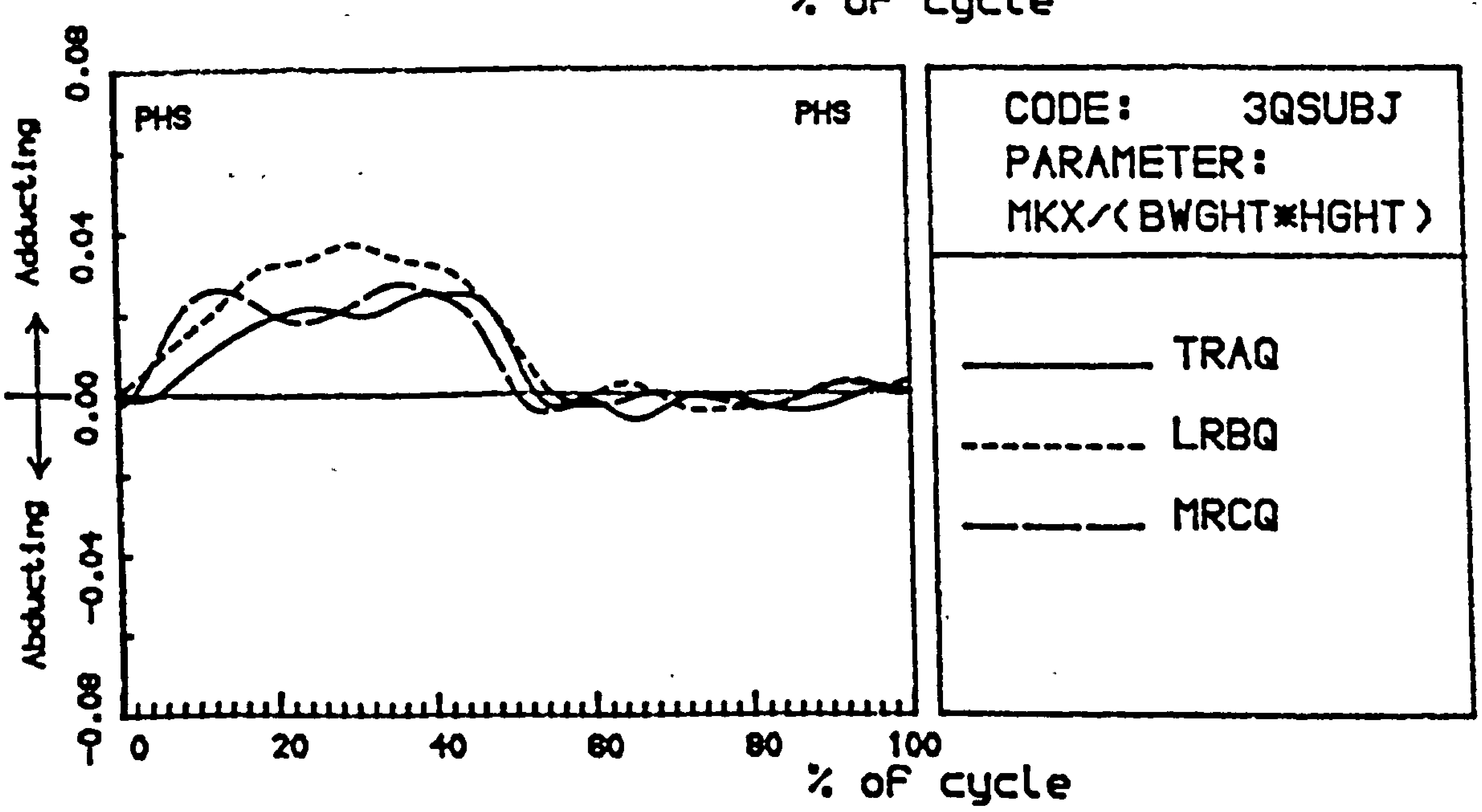


Figure C.12 Prosthetic knee joint moments with time for three AK amputees fitted with quadrilateral sockets. (Normal alignment)

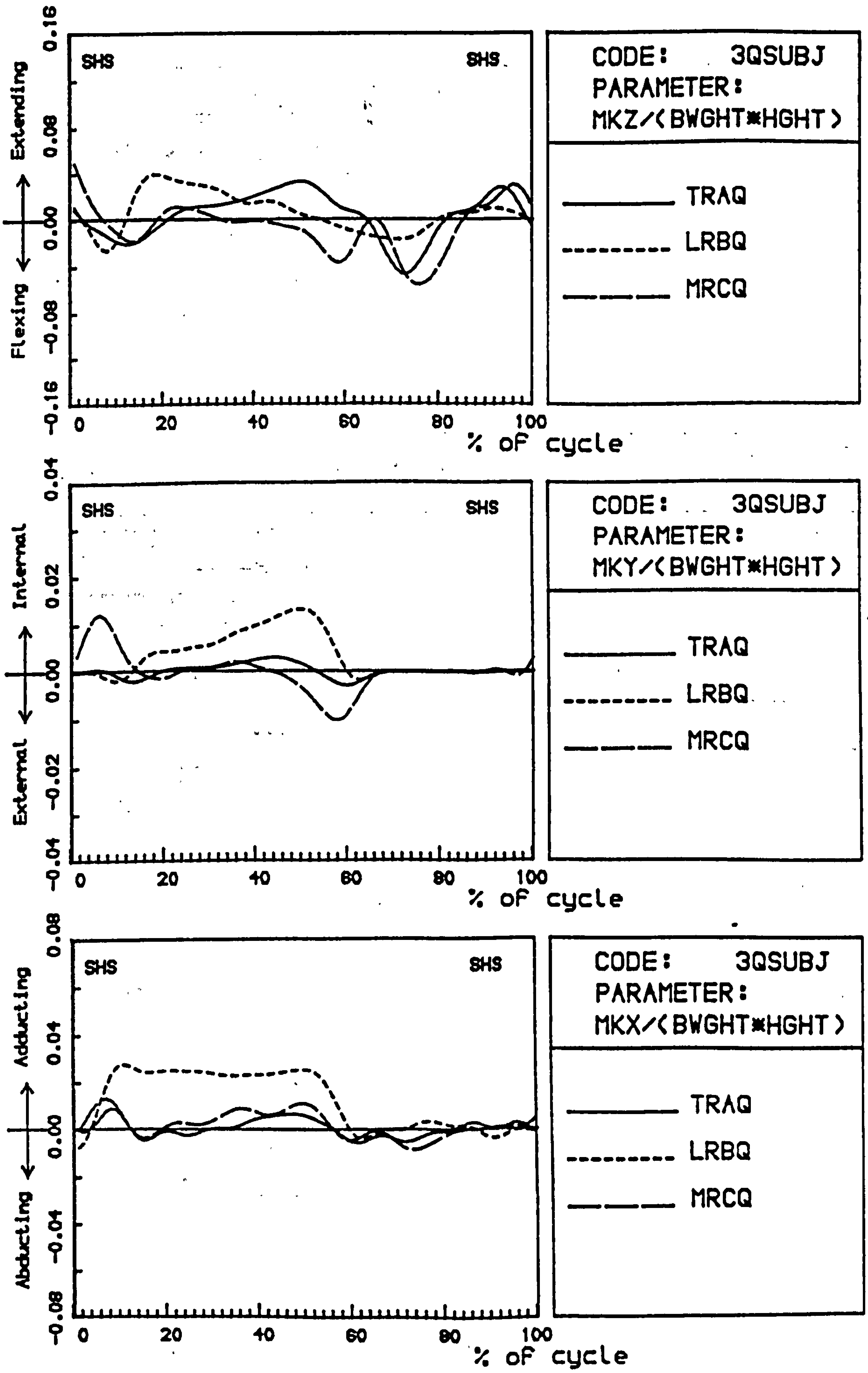


Figure C.13 Knee joint moments on the sound leg with time for three AK amputees fitted with quadrilateral sockets. (Normal alignment)

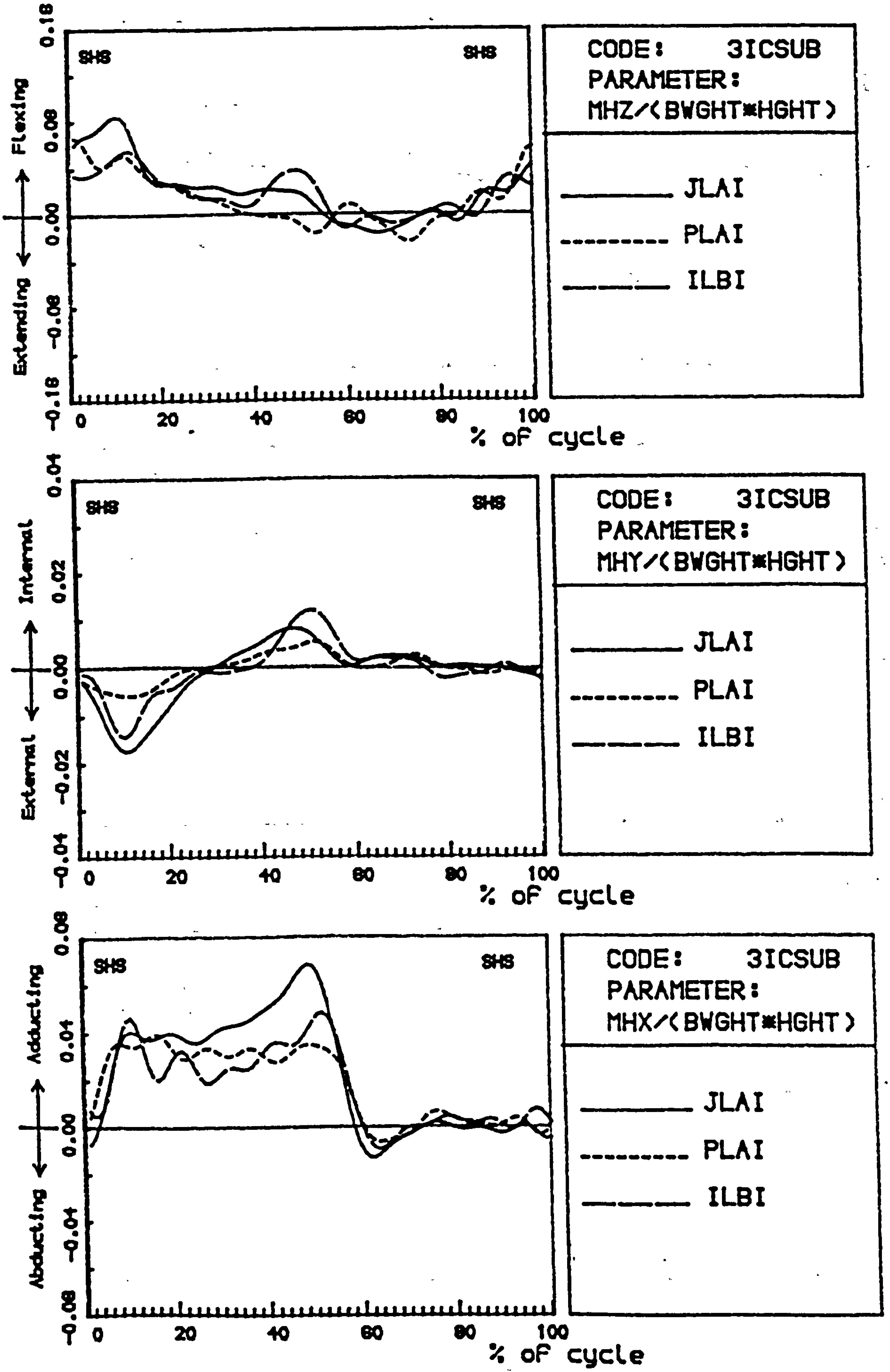


Figure C.14 Hip joint moments on the sound leg with time for three AK amputees fitted with IC sockets. (Normal alignment).

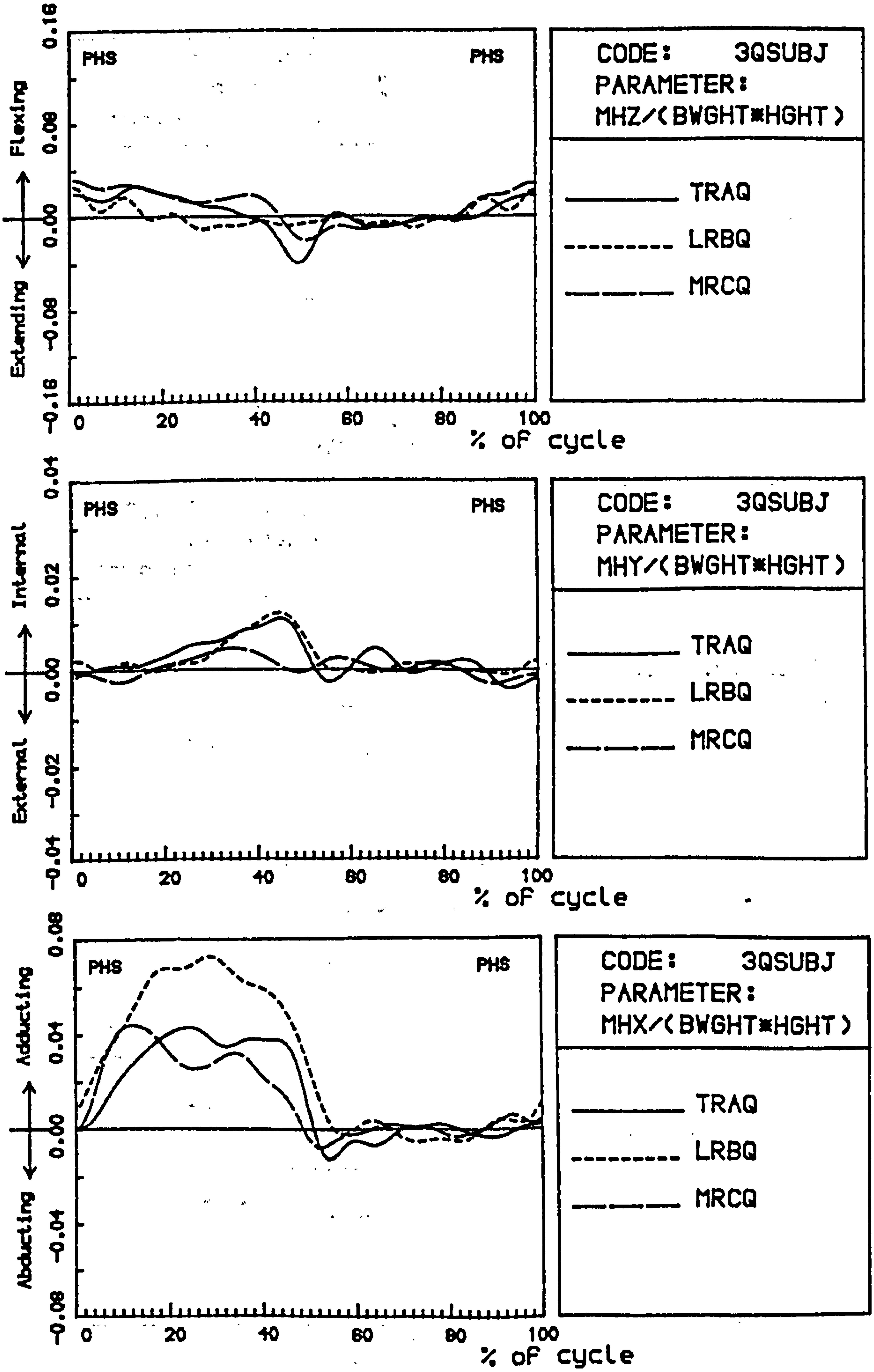


Figure C.15 Hip joint moments on the prosthetic leg with time for three AK amputees fitted with quadrilateral sockets. (Normal alignment).

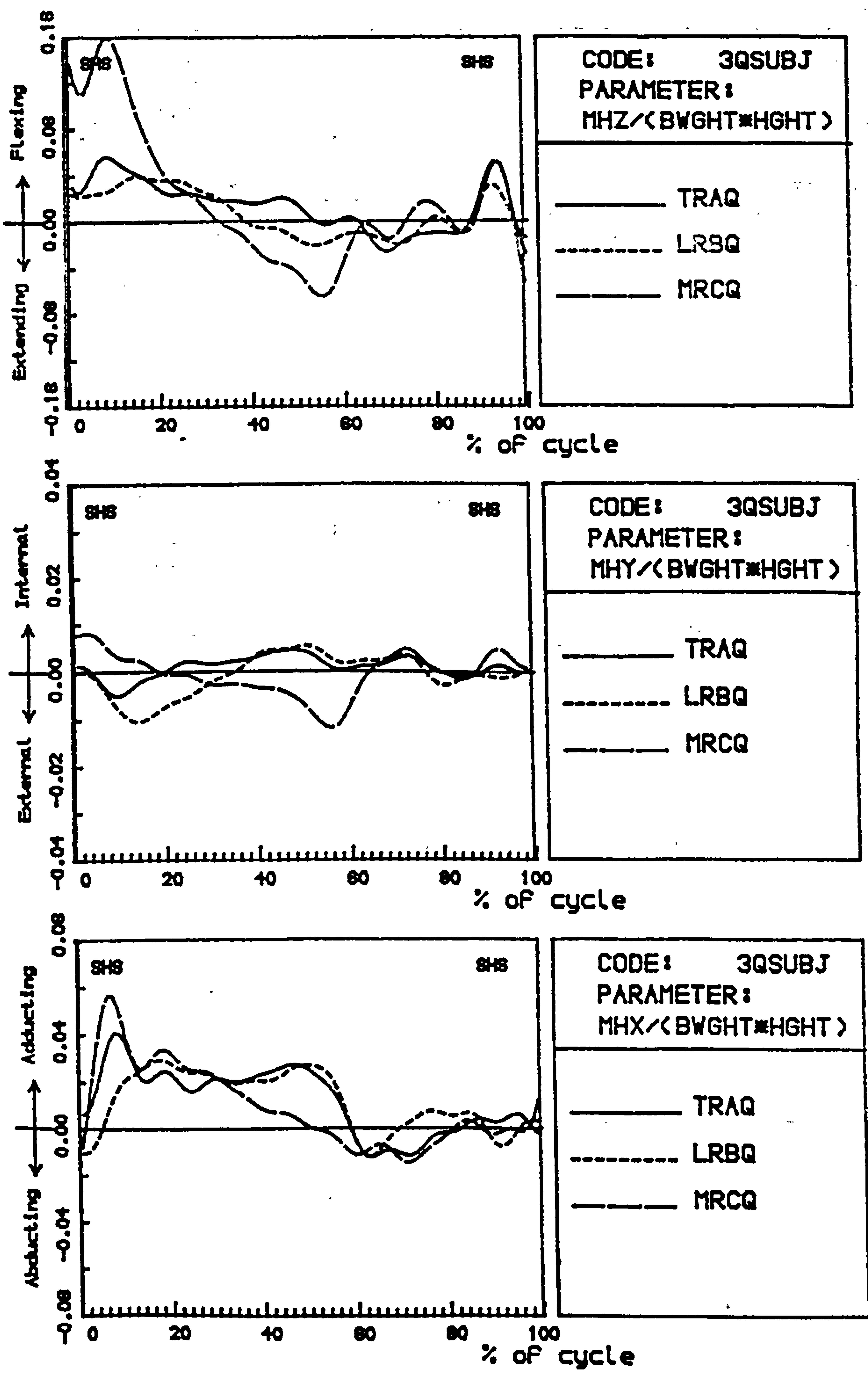


Figure C.16 Hip joint moments on the sound leg with time for three AK amputees fitted with quadrilateral sockets. (Normal alignment).

Appendix D

Evaluation of the effect of the TV errors and of the location error on the results

It was found (sections 4.3.2.2) that the TV system has a maximum absolute static accuracy of 5.2 mm, 3.9 mm and 2.4 mm in the X, Y and Z directions respectively (these errors are the average of the errors obtained for left/front and right/front camera sets), and a minimum absolute static accuracy of -3.4 mm, -2.8 mm and -5 mm in the X, Y and Z directions respectively. It was also found (section 4.3.2.3) that the TV system has an overall relative dynamic accuracy of 1:318. In order to evaluate the effect of these errors on the results calculated, the hip joint moments were chosen. The hip joint was chosen because the effect of the relative dynamic error on this joint is larger than that on any other joint, since the hip joint is located on the largest link (shank and thigh) to be subjected to that error.

It was also reported (section 5.3) that the method of locating the HJC has an error of ± 1.07 cm.

Because the effect of the above listed errors (static, dynamic and location) on the moments of the hip joint is additive, they were all combined and evaluated together for subject MFFM, and their effect is shown in figures D1 and D2 for the left and right hip joints respectively. In these figures the curve labelled MFFM was calculated without considering the effect of the errors. Curve labelled MFFM+E was calculated considering the errors in the following manner:

1- The maximum absolute static error of each direction was added to the respective coordinate of the static marker. i.e. 5.2 mm, 3.9 mm and 2.4 mm were added to X, Y and Z coordinates respectively of each static marker.

2- Each coordinate of the HJC was increased by 1.07 cm to evaluate the effect of the hip location error.

3- The length of each body segment (shank and thigh) which were used in calculating the hip moments was multiplied by 1.0031 which gives the effect of the relative dynamic error.

Curve MFFM-E was calculated using the same method as curve MFFM+E but considering the minimum errors(i.e. adding the negative errors).

Then, the hip moments were calculated during the gait cycle for curves MFFM, MFFM+E and MFFM-E. It was found that the above listed errors resulted in the following errors on the hip joint moments:

1- Maximum errors of ± 8% and ± 14% of the original moment (which was calculated without considering the errors, curve MFFM) were found on the A/P hip moment on the left and right legs respectively.

2- Maximum error of ± 14% of the original moment (curve MFFM) was found on the transverse hip moment of the left and right legs.

3- A maximum errors of ± 15% and ± 12% of the original moment

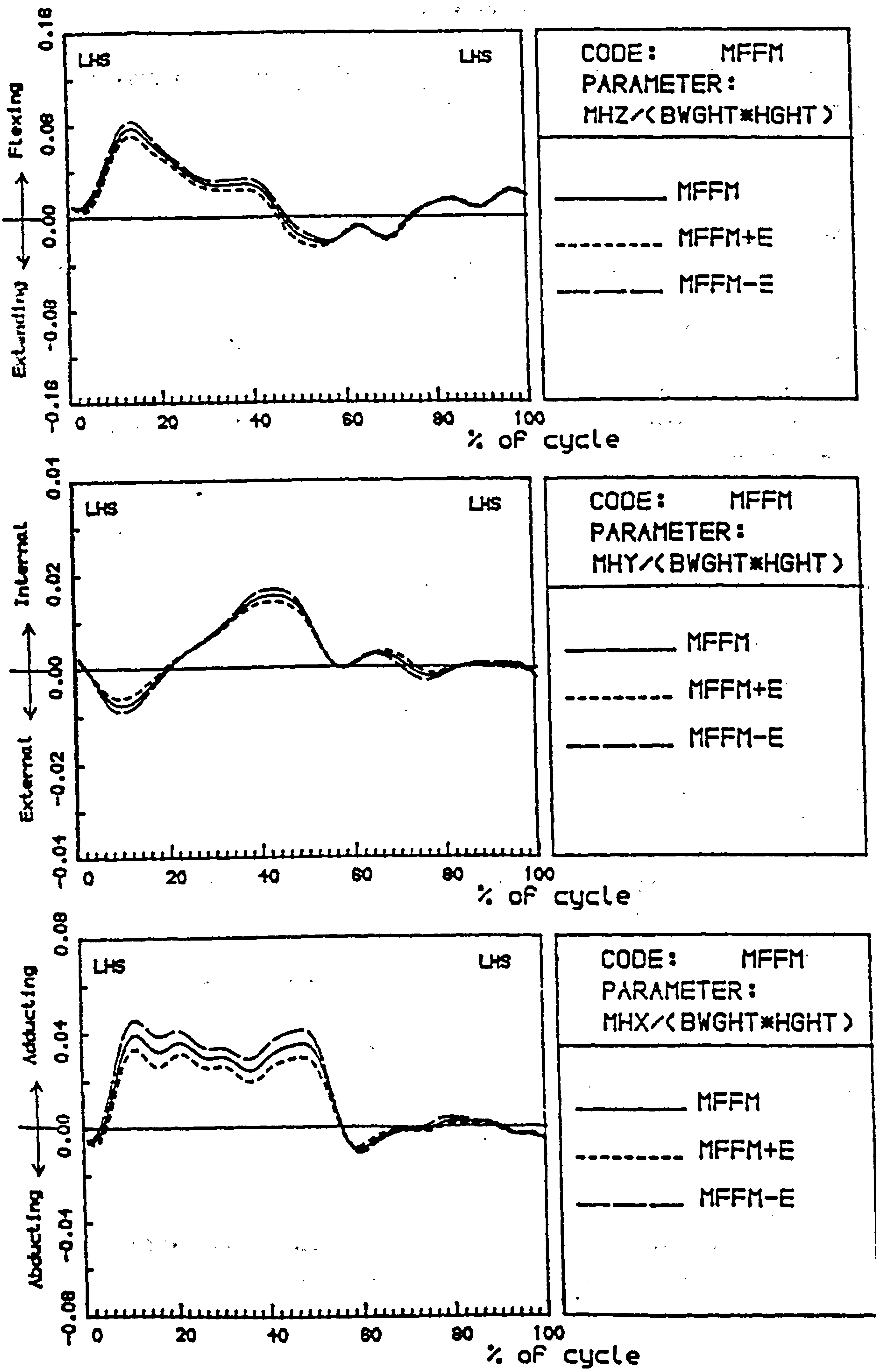


Figure D.1 Effect of the TV errors and of the location error on the moments of the hip joint. (Left Leg).

(curve MFFM) were found on the coronal moment of the hip joint of the left and right legs respectively.

The differences in the percentage errors between the left and right legs, are related to the differences in the magnitude of the original moments between the left and right legs, and the magnitudes of the errors on the left leg are approximately similar to those of the right leg.

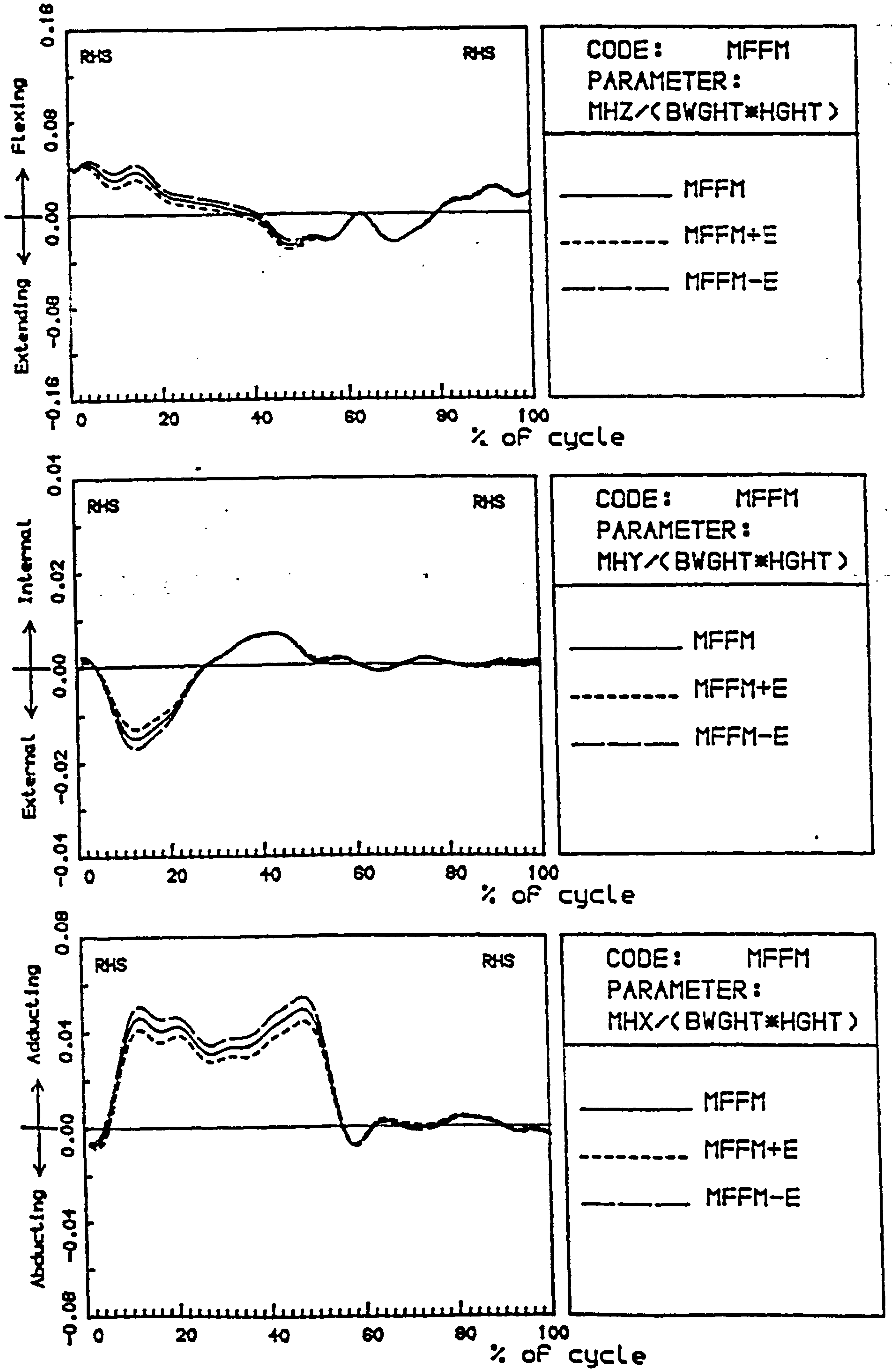


Figure D.2 Effect of the TV errors and of the location error on the moments of the hip joint. (Right Leg).

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