A COMPUTER ANALYSIS OF ENERGY TRANSFERS IN SWING-THROUGH CRUTCH GAIT WITH CHILDREN

by

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ABSTRACT

A Computer Analysis Of Energy Transfers And Power Flow In Swing-Through Crutch Gait With Children

The purpose of this study was to investigate swing-through crutch gait with children assisted by the development of a computer package to aid in data analysis. A recent technique of human motion analysis utilizes the mapping of energy transfers between body segments. Eight volunteer subjects ambulated with crutches as cinefilm data and axial and lateral bending moment crutch forces were recorded. The cinefilm was digitized, converted to absolute coordinates, digitally filtered and processed to derive instantaneous, segment energy levels. Although energy transfer patterns have been established for adults walking with crutches, child studies are absent in the literature. The study evaluated the adequecy of the 9-segment crutch gait model (Wells, 1979) for use with children. Energy transfer patterns were also established with children for use as a future reference

in the evaluation of pathological gait. Mechanical inefficiencies in individual gait patterns were identified, increasing understanding of this special locomotion problem. Inefficiencies were manifested in pathological, instantaneous, segmental power and energy curves, augmented by foot-contact and axial crutch force information. The 9-segment model was found to be adequate for analysis of normal healthy children. Although crutch length is a critical determinant in efficient gait, the formula for finding the correct length varies among individuals. 'Normalized' work output values, corrected for body weight and distance covered per stride, are much greater than those found in adults.

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Chapter I

THE PROBLEM

1.1 INTRODUCTION

Most of us are fortunate enough to avoid having to use crutches. However, there are some people whose only mode of ambulation is with the aid of axillary crutches. Some children face this prospect for lengthy periods of time due to injury and/or disease. There is little to be found in the literature on the biomechanical aspects of this rather unique form of ambulation. Scientific investigation into crutch gait did not begin until the sixties when Peacock (1966) undertook an electromyographic analysis. In the mid-seventies, the basic mechanics of crutch gait were analysed (Shoup et al, 1974). A study by Wells (1979) introduced the concept of energy transfers into the study of crutch gait. Efficiency, as a measure of human performance, involves some measure of mechanical energy output divided by a measure of metabolic input (Winter, 1979B). Energy transfers deal with the numerator of this equation. Further analysis of crutch gait, utilizing the technique of mapping energy transfers will give more insight into efficiency while walking with crutches. With greater understanding of the mechanical relationships, improvements in walking technique, crutch design and orthotic appliances can be achieved.

1.2 RATIONAL

People using crutches have a reduced capacity for physical movement due to their disability. It is physiologically more costly to ambulate with crutches as opposed to the conventional bipedal mode (Childs, 1964), consequently research is required to understand the energy processes of crutch gait. In the present study, a reduction in energy cost pertains to mechanical efficiency rather than 'internal' physiological efficiency. In other words, this investigation did not involve cardiovascular, biochemical or muscle physiological efficiency. The rational for this study, is to fulfill the need to gain further information on the biomechanical parameters of crutch gait to facilitate locating individual movement ineffecencies. To a handicapped person, this could mean the difference between being slightly tired and utterly exhausted at the end of his day. These factors emphasized the purpose of this study. In order to analyse crutch gait for mechanical efficiency, a model of the body, integrated with crutches, must be established and confirmed. In addition, gait evaluation centres (clinics and hospitals) have not had a standardized assesement procedure for crutch gait. Thus, this study described a technique that has potential to fill this void. However, before such a technique could be introduced and implimented on a broad scale, a bank of data would be required for comparative purposes. By establishing movement patterns through the mapping of energy transfers technique, comparisons of gait pathologies will show the unnecessary, energy consuming movements in handicapped individuals using crutches.

1.3 STATEMENT OF THE PROBLEM

The study of crutch gait started in the early 1960's but was quite crude in its methodology. From previous studies, the biomechanician has a creditable knowledge of which muscles are working (Peacock, 1966), and the angles that the body joints exhibit (Childs, 1964, Shoup et al, 1974). However, researchers did not devise a method to compute inefficiencies of energy expended during swing-through crutch gait. In other words, there is not a scientific, analysis procedure that will help the patient expend less energy. Such analysis could be enhanced with physiological methods in the future. Once a method is established for analysis, there must be data available of efficient crutch walkers to be able to distinguish between efficient and inefficient gait.

The problem, therefore, is to apply the method of analysing energy transfers (Quanbury, 1975, Winter and Robertson, 1978, Pierrynowski et al, 1980) and the crutch model of Wells (1979), to swing-through crutch gait. Secondly, energy transfer patterns that are efficient must be established for use, as a clinical proceedure in the future, as a reference to compare to those patterns of inefficient crutch gait. Once the method of analysis is employed, the secondary problem of establishing so called 'normal' data for future comparison and reference is in need of a solution.

1.4 HYPOTHESIS

The 9 segment model, postulated in this study, is hypothesized to be adequete in evaluating crutch gait for mechanical efficiency. Secondary to this primary postulate is that instantaneous energy levels and instataneous power curves are hypothesized to contain similar patterns for the 'normal' efficient crutch gait ambulators. Distinct, common features are hoped to be recognized and identified, thus establishing a data base for future comparative reference.

1.5 SCOPE OF THE STUDY

This study was an analysis of energy transfer patterns of eight children using axillary crutches. The subjects were artificialy handicapped with a leg brace to simulate an injury such that weight could be borne by only one leg.

1.6 LIMITATIONS OF THE STUDY

The study was limited to eight subjects due to the constraints of time as well as the availability of suitable subjects. The equipment (instrumented crutches, amplifiers, recorders) limited the location of the collection of data to the Biomechanics Laboratory (University of Ottawa).

1.7 DEFINITIONS OF TERMINOLOGY

Attenuation- the reduction in amplitude of a given signal (Miller and Nelson, 1973)

Bonded Strain Guage- strain sensitive elements arranged to facilitate bonding to a surface in order to meassure applied strains.

Error- the algebraic difference between the indicated value and the true value of the measurand expressed in percent of the full-scale output (Miller and Nelson, 1973).

Excitation- the external electrical voltage and/or current applied to a transducer for its proper operation.

Gait Cycle- the events that occur between successive ipsilateral foot contacts on the walking surface.

Heel-Strike (H-S)- The instant in time where the heel makes contact with the ground.

Hysteresis- the maximum difference in the signal output of a transducer at any given measurand value within the specified range when the value is approached first with increasing and then with decreasing measurand, in other words, a measure of non-linearity of the transducer signal output.

Linearity- the closeness of a calibration curve to a specified straight line (Miller and Nelson, 1973).

Noise- any unwanted electrical disturbance or spurious signal which modifies transmitting, displaying, or recording of desired data (Miller and Nelson, 1973).

Peak-to-Peak- the algebraic difference between maximum positive and negative values of a varying signal (Miller and Nelson, 1973).

Scale factor- the ratio of full scale output to the value of the measurand at full range (Miller and Nelson, 1973).

Signal- the output emanating from a device (Miller and Nelson, 1973).

Stance Time- elapsed time one foot is in contact with the walking surface, in seconds.

Stride Length- the distance between two consecutive ipsilateral foot contacts on the walking surface.

Swing Time- elapsed time one foot is not in contact with walking surface.

Toe-off (T-O) - The instant in time where the toe leaves the ground to initiate leg swing-through.

Chapter II

REVIEW OF THE LITERATURE

2.1 INTRODUCTION

Due to the diverse nature of the present study, the review of literature was subdivided into related areas. In 1974, Shoup et al, outlined the crutch literature demonstrating the lack of biomechanical analysis of crutch gait. Surprisingly, the article revealed that most of the literature, until then, dealt with crutch modifications, history and the art of fitting crutches.

Epstein(1937) described the history of crutch use and found that the axillary crutch has remained, more or less in the same general form for the past 4800 years. Attempts at modification of the basic design are outlined in the Atlas Of Orthotics(1975).

2.2 PROBLEMS ASSOCIATED WITH CRUTCH USE

When dealing with crutch gait, there are several parameters that must be considered to avoid complications. Although crutches are utilized to aid ambulation in those either temporarily or permenantly disabled, improper fitting and technique can cause further injury. Ettien(1980) reported cases of axillary artery aneurysms caused from the pressure of the axillary pad of the crutch. Of all reported cases, the patients were elderly, suffered weakness of lower extremities and had used axillary crutches for mobilization for years. Abbot and Darling(1973) and Brooks and Fowler(1964) noted that the axillary artery and radial nerve are prone to crutch induced injury and must be protected through proper gait technique and crutch fitting.

Use of crutches is not restricted to those with only lower extremity disabilities. Upper extremity amputees must overcome the great handicaps of lack of strength, balance and stabilization, but have in some cases, become successful crutch users(Tucker, 1979). Such studies illustrate the diversity of problems that must be anticipated by investigators of crutch gait.

Some incomplete quadriplegics (one who has injured the cervical cord but still has muscle innervation below the injury) with limited intrinsic hand grasp have become independent ambulators through crutch modification (Lamonda, 1979, Virga, 1979). Cerza(1978) described crutch modifications to aid in ambulation for patients with lower extremity monoparesis. This entailed the use of a strap and brace from the afflicted leg to the lower position of the crutch so the energy of the crutch would be transfered to the leg producing movement.

Thus one can see problems associated with crutch use are neither few, nor simple.

2.3 BASIC KINEMATICS

A brief overview of the development of investigation into human locomotion was described by Cappozzo et al(1975). Photography facilitated the first method of study which developed after the 1860's. As the art developed, biomechanists utilized photography to aquire a more complete knowledge of movements and forces involved in human locomotion. Muybridge(1955) filmed many locomotor activites of both men and women, in the late 1800's. His frame by frame analysis technique laid the cornerstone for much of the work in contemporary biomechanics. As much of todays research efforts are spent on pathological locomotor patterns, and many measurements must be taken, mathematical modeling becomes a necessity. Therefore, the body has been modeled as an articulated system of links in dynamic equilibrium. Such models employ a few assumptions for simplification purposes. Joints are usually considered to be frictionless and pinned. The displacement of blood and soft tissue is disregarded as segments are assumed homogenious rigid bodies. Hands and feet are described as single segments(Miller and Nelson, 1973). The model enables the collection of accurate data as well as conservation of man/hours during data analysis. The length of an anatomical link as defined by Dempster(1955), is an axial line connecting two adjacent joint centres or, a straight line between the joint centre and mass centre of the segment in the case of a terminal segment. The joint centres are not always easily identified by surface landmarks. Clauser et al(1969) suggested that segment lengths (links) be determined from using the available bony landmarks. The mass particles that comprise each segment are considered to act at the centre of gravity. These segmental mass centre locations were determined in a classical study by Braune and Fischer(1889). They concluded that the mass centres of extremities were approximately four-ninths of the total segment length from the proximal joint. The segmental (link) moments of inertia with respect to an axis through the mass centre were reported by Fischer(1906). A very comprehensive anthropometric investigation was conducted by Dempster(1955). He calculated moments of inertia by setting the segments into pendulum systems and recording the period of oscillation.

The body, when modelled as a system of links, moves as a result of external forces on the system. Such forces are ground reactions, inertial, gravitational, and the action of internal forces performed by muscular contractions(Cappozzo et al, 1975). Under the influence of such forces, the links undergo angular diplacements about the joint axes. The resultant angular displacements have a direct relationship with the moments provided by muscular forces at each joint.

The center of gravity of each link in the body interacts with applied forces to produce the movements characteristic in locomotion. Murray et al(1967) described the relationships between the postions of the centre of gravity of the body and the centres of applied force. For example, if the body is to be accelerated forward, the centre of pressure of the driving force must be applied behind the vertical line through the centre of gravity. For accelerating backwards, the driving force must be applied in front of the vertical centre of gravity line. However, Murray et al(1967) found that for stationary maintenance of the body, the centre of pressure fluctuated both in front of and behind the line of gravity. Thus, the accelerative and decelerative forces that maintain the stationary position must fall within the base of support. It is these forces that cause forward motion on crutches.

Since the link segment model outlined by Capozzo et al(1975) is accepted as an accurate representation, the body will be considered as a system of articulating links in this study(Winter, 1979B, Miller and Nelson, 1973). This concept will be greatly expanded in the subsection dealing with energy transfer.

The dynamics involved in the movement of the body links are well documented and can be found in several biomechanics textbooks (Miller and Nelson, 1973, Winter, 1979B, Plagenhoef, 1971)

2.4 CRUTCH GAIT PARAMETERS

In the prior subsection it was noted the literature is scarce on crutch gait analysis from an biomechanical viewpoint. For this reason, many of the methods for analysis must be borrowed from the better documented area of walking gait.

In addition to the previously cited work of Braune and Fischer(1889), and Fischer(1906), Bernstein(1930) made significant progress on human locomotion analysis. He devised a technique that determined X, Y, and Z coordinates of landmarked joints using only one camera in conjunction with a plane mirror. This was achieved by intersecting the optical axis of both the real and mirrored cameras. Consequently he was able to develop formulae using similar triangles to gain the joint coordinates(Miller & Nelson, 1973).

In the early 1950's, a comprehensive study on normal adult locomotion(Saunders et al, 1953) greatly contributed to the base of knowledge which was needed for analysing locomotor pathologies. Saunders et al(1953) outlined six basic determinants of normal gait which influence the vertical and horizontal displacement of the centre of gravity. The determinants were: pelvic tilt, pelvic rotation, knee flexion at mid-stance, the foot mechanism, the knee mechanism and lateral displacement of the pelvis. Once the segmental displacements of the centre of gravities were understood for walking, the same method could be applied to crutch gait analysis (Burnett and Johnson, 1971). When devising a mathematical model for crutch gait, the main difference between the Townsend and Seireg (1972) model of normal walking gait and a crutch model is the pelvic movement. Pelvic and thoracic motions are 180 degrees out of phase between sides when walking. This is not true for crutch gait as swing-through gait is essentially bilaterally symetrical (Shoup et al, 1974) (Bergmann et al, 1978).

Sankarankutty et al (1979) conducted a study to compare the metabolic and physiological energy costs between different types of crutches. Although higher heart rates were recorded for subjects using axillary crutches, the subjects did consider the axillary variety to be the easiest to use in addition to being the most comfortable. (The various types of crutches can be found in the ATLAS OF ORTHOTICS (1975)) Due to the subject preference for axillary crutches (Sankarankutty et al, 1979) and their universal popularity, this study shall utilize the standard, hospital issue, axillary crutch (ATLAS OF ORTHOTICS, 1975).

The study of crutch gait as a separate entity, really began in 1964 with the first analysis of swing-through crutch gait (Childs, 1964). It was an examination of paraplegic ambulation where the arms and head became sources of power to move the non-functioning mass (the lower limbs).

The study emphasized the concepts of dynamic equilibrium, balance, and power in crutch gait. Power, in this investigation, was defined as a transfer of energy from the arms and head to the legs through a whip-like action of the torso. This transfer of energy was noted, however, it was not measured in any way.

Peacock(1966) conducted an electromyographic study during crutch walking to demonstrate the pattern of activity in the upper limb and trunk muscles. In terms of crutch type, he found no consistant difference between the use of elbow and axillary crutches in EMG signals. Subsequently, Peacock(1966) made the observation that crutch walking involves the weight of the upper body being taken on the hands, resulting in the functional origins and insertions of muscles to be 'reversed' in the upper arm. Peacock's(1966) work was the first to note such a functional reversal of muscle attachments when the body mass is supported by the upper limbs.

Shoup et al(1974) reported on the mechanics of crutch gait by recording the displacements of body segments in different planes. They noted that for 'normal walking', the vertical fluctuations of the joints were smallest for those joints that were furthest from the floor. Thus the shoulder and hip have smaller vertical excursions than do the knee and ankle. For crutch motion, this condition is completely reversed. They also documented the lack of a shock absorbing system for the upper body from forces transmitted up through the crutch, which is present in the lower body when walking due to knee and back anatomical design. They suggested a shock absorbing system built into the crutch to absorb impact forces. Subsequently, a more complete study is required to examine the magnitude of such energy transfers which is in the scope of the present study. The following section describes the nature of such transfers of energy and a method of evaluatng the magnitude of the transfers. Shoup et al(1974) also suggested further investigation of crutch gait with other subject groups such as children, female adults and elderly people.

Ground reaction forces during swing-through crutch gait were examined by Stallard et al(1980) to see if the nature and magnitude of these forces would have repercussions for patients with injury or disease to the lower limb. They reported the average minimum vertical ground reaction to the leg was 132% body weight, 16% higher than reported for normal walking. The horizontal force on the lower limb in the direction of progress was 35% body weight while 15% body weight is reported for normal walking. Vertical ground reaction to crutches was on the average 54% body weight on the side of the landing leg and 51% on the contralateral side. Horizontal ground reaction on the crutches at right angles to the direction of ambulation was 09% body weight on the right and .08% body weight on the left. Further study to determine the effect of these forces on the total body is required.

The first kinematics and energy variations analysis of swing- through crutch gait was conducted by Wells(1979). He noted that Shoup et al(1974) utilized subjects who had full use of their legs. Therefore, they could push off, and lengthen their stride resulting in reduced trunk motions. The subjects in the Shoup et al(1974) study were also able to employ hip and knee flexions giving ground clearance during the swing phase. Wells(1979) pointed out such a mode of locomotion was not common for disabled crutch users. Consequently Wells(1979) conducted an investigation examining the work required to move the body as an absolute sum of positive and negative energy changes while restricting the mobility of the musculo-skeletal system to closer simulate the conditions of disabled crutch users. The nature of energy exchange is discussed in the next section.

The methodology of Wells(1979) appeared, for the most part, valid and complete. His data collection technique consisted of cinematographic and force platform records of crutch gait in the saggital plane. He used three male adult subjects ranging from 21 to 32 years of age. The 16mm cine camera was positioned 12 meters orthogonally to the walkway, with a frame rate of 30 to 35 frames per second. The crutch tip transducer and cine film were synchronized by means of a clock superimposed on the film which also generated a timing pulse on the chart recording of crutch tip forces. The body joints were landmarked and orthodic braces were placed on the knee and hip in various combinations. A computer program was used to analyse the film displacement data, rendering the energy levels of the 4 segment model. Data on the body segment parameters were obtained from Drillis and Contini(1966) and Dempster(1955). Data was filtered by a piecewise polynomial curve fitting technique. This setup served as a basis in the present study, however, the rational for this study necessitated variations outlined in chapter 3.

Fisher(1981) studied the energy cost of ambulating with crutches but used Vo2 as a measure of energy consumption. However he noted, as the average crutching speed increased, the rate at which step frequency increased was significantly greater than that of stride length. He hypothesized this result to be related to the pattern of potential and kinetic energy changes in the gait cycle. He suggested using the analysis methods of Wells(1979) and Quanbury et al(1975) to develop this relationship further.

Wells(1979) used male adults who were artificially incapacitated. This type of study has not been conducted using children as subjects.

As all previous crutch gait studies have used adults as subjects, one must consider the variable segment lengths if dealing with children. It has been suggested (Burnett and Johnson, 1971) that the characteristics of mature adult gait appear much sooner in children than previously thought. In crutch gait, the smaller segment lengths, compared to adults, may not agree with the conclusions of Burnett and Johnson (1971) which will warrant additional consideration. This is imperative when developing a rationale for the present study. It was the conclusion of Foley et al (1979) that adult data cannot be scaled down for use with children due to the adult segment proportions not equating with children. Foley et al (1979) data showed the average linear displacements, velocities and accelerations were larger for children. They concluded that the lower masses and inertias of the smaller limbs permitted more rapid accelerations with lower force levels. Therefore, energy exchanges during swing through crutch gait may be different in children as oppossed to adults.

2.5 ENERGY TRANSFER AND POWER FLOW

Wells (1979) has been the only author to analyse crutch gait from an energy transfer viewpoint, but he did not give an explanation as to its theory. An overview of the full development of energy transfer and power flow is required for complete understanding.

The concept of energy transfer in its infancy was referred to as 'external work' (Fenn, 1930, Elftman, 1939, Cavagna et al, 1963). This term was misleading as Zarrugh(1981) pointed out there was no exchange of energy between the body and external environments save for the effects of air resistance. Fenn(1930) determined the work done by the body during running by recording foot pressure measurements and obtaining body displacement data from film. Elftman(1939) and Fenn(1930) used data from a single walking cycle for a single subject to study energy changes and work done in the body. Fenn(1930) used film to record body displacements and measured foot pressure to determine the work done by the body. Elftman(1939) using great foresight, calculated potential energy(PE) and kinetic energy(KE) for the foot, shank, thigh, and trunk during the full gait cycle. Elftman(1939) also recorded the rate of doing work(power) in the above segments, noting that positive values indicated energy expended while negative values indicated energy received by the segment.

The average energy requirements of the body, during walking, for a group of 4 subjects on the basis of displacements was reported by Bresler and Barry(1951). This was achieved by the development of equations for energy, using Newtonian mechanics, in three planar dimensions. In a later study (Bresler et al, 1957) calculated the energy and power requirements of the legs in amputees and concluded that more fluid trunk motions may result in a decrease of the total energy need due to the large mass of this segment.

Cotes and Meade(1960) also attempted to measure the amount of energy expenditure in the body during walking. They used the vertical lift of the trunk as a measure, calculating lift as change in height of trunk multiplied by body weight. However, they did not take into account the forward and lateral accelerations of the body's centre of gravity.

Cavagna et al(1963), and Cavagna and Margaria(1966) computed the amount of 'external work' required to increase the translational mechanical energy of the body's centre of mass. This was achieved by accelerometry of the trunk(Cavagna et al, 1963) and by integration of force plate data(Cavagna, 1975). In the 1963 study, Cavagna et al noted the use of gravity to accelerate forward the centre of mass of the body. The resultant increased velocity gave sufficient momentum to raise the centre of mass again during leg implantation. This shift between potential and kinetic energy was called energy recovery and was used as a measure of efficiency by noting energy losses. Cavagna et al's(1976) equation for the calculation of recovery of mechanical energy was:

$$R = \frac{(|Wv| + |Wf| - Wext)}{(|Wv| + |Wf|)} \qquad (1)$$

(Cavagna et al, 1976, pp 643)

Wv= work to lift centre of mass of body

Wf= work to increase forward speed

Wext= total mechanical energy involved

Cavagna et al(1976) attempted to determine the most efficient cadence for walking. As the equation for recovery of mechanical energy suggests a difference in phase between potential and kinetic energy during walking, Cavagna et al(1976) compared the mechanism to a pendulum. In this attempt, they developed a model of the leg and shank as a pendulum over the foot employing the equation:

vertical displcement of trunk (swing period) = S S cos (@/2)

(Cavagna et al, 1976, pp 653)

S = length of leg

@ = amplitude of rotation

This assumes that @ is sufficiently small to assume the chord is equal to the arc(Cavagna et al, 1976). However, this model rapidly deteriorates with an increase in speed and the presence of any gait pathology. In addition, Cavagna et al(1976) made the assumption that the energy of the body's centre of mass represented the sum of all segment energies. This was later found inaccurate as it does not take into consideration between-segment energy exchanges(Quanbury et al, 1975).

Gersten et al(1969) computed 'external work' and energy requirements in a similar manner directly from the displacement of the body's common centre of mass using a triaxial accelerometer attached to the trunk, close to the estimated centre of gravity of the body. However, energy changes in the limbs were not accounted for in the calculations. Lukin et al(1967) also computed potential energy and kinetic energy from the change and lift, in the body's centre of mass, but fell short in the total calculations for the same reason as Gersten et al(1969). Despite the shortcoming, the method employed by Lukin et al(1967), is worth noting. They attached strings to the approximate locations of the mass centres in the segments to record their displacements which were used to compute energy levels. This was a direct method to record limb displacement rather than gaining such information from film.

Beckett and Chang(1968) found hip moments to be the most important factor in gait. They felt the quadriceps (acceleration phase) and the hamsrings (deceleration phase) were the most significant power units in walking. Any other muscle groups, such as those which produce knee movement, were largely for control and did not influence the flow of power in the body to any great extent. In conclusion Beckett and Chang(1968) believed that one could determine the most efficient gait cadence if given the parameters of the body. In other words, the most efficient gait cadence would be achieved by minimizing the energy used, per distance covered.

To this point in the development of energy transfer, the simplification of using the total body centre of gravity, or point mass model, introduced error into energy calculation. Lukin and Ralston(1969) took a substantial step towards the present-day, thought on energy transfer, by utilizing the link segment model. They calculated the mechanical energy level of the body segments rather than just considering the displacement of the total body centre of gravity as their predecessors had done. The body was divided into the following segments:head-arms-trunk, thigh, shank and foot. The theory involved was confirmed by Smith(1975). Lukin and Ralston(1969) determined limb volumes by water displacement and used Dempster's (1955) equations to approximate the segment centre of gravities. Despite their major advancement, rotational kinetic energy was not considered in the calculations.

A study of energy analysis measuring displacement, ground reaction forces and electromyographic patterns in the lower limb was conducted by Cappozzo et al(1976). They concluded that the constant level of energy of the HAT (head, arms and trunk) was the key to efficiency during walking gait.

Quanbury et al(1975) developed the current thought on power flow and energy transfer using a computer-TV system that provided information about the absolute trajectories of body segments during walking. They pointed out calculation of power flow can also be derived by considering the instantaneous rate of change of the total energy of the body segments as well as by the traditional method of Elftman(1939). Elftman's(1939) method of calculating power flow was based on the following principals. As the work performed by muscles is rarely constant with time, muscle power is calculated as a function of time(Winter & Robertson, 1978). Thus, the equation for muscle power across a joint is:

$$Pm = Mj * wj \quad W \qquad (2)$$

where: Pm is the muscle power in watts

Mj is the net muscle moment (Nm)

wj is the joint angular velocity (rad/sec)

(Winter, 1979B)

Pm can be either a positive or negative value depending on whether the muscle is positively or negatively accelerating the segment. When the power is a negative value, the muscles are absorbing mechanical energy while energy is generated when the power is a positive value(Winter, 1979B). Inasmuch as power is the rate of doing work, work done by a muscle is the integral of muscular power with respect to time. Elftman(1939) explained how power flow to and from segments were the result of two types of forces. The first type were intersegment forces acting at a joint. To calculate power flow due to acceleration of a limb segment, all external forces, the mass, and the acceleration of the centre of mass of the segment must be known. Newton's laws were then applied in equations to determine the forces in each limb segment as illustrated by Quanbury et al(1975). Power flow due to joint forces is expressed as Fj * Vj where Fj is the force acting at the joint and Vj is the absolute velocity of the joint(Robertson and Winter, 1980). This can be explained by the following: since power is the rate of doing work, or dW/dt, and work is force times displacement, the equation for power is also F * ds/dt. The ratio of the infinitesimal displacement over the infintesimal time can be substituted by velocity. Hence, P = F * V where F and V are vector quantities(Winter, 1979a).

The second type of force contributing to power flow is that due to tendon moments. This involves an equation that sums the moments created by muscular contractions, gravitational acceleration, and forces due to joint acceleration about the joint axis of rotation. Quanbury et al(1975) compared this method of determining intersegment power flow to the calculation of instantaneous power and found them to be in agreement. In addition, while the calculation of instantaneous power requires no assumptions due to the nature of the equations involved, power flow calculations required four assumptions. They were: (1) pure rotation about an ideal hinge, (2) constant moment of inertia about the joint, (3) no dissipative forces (frictional viscosity), (4) joint forces act through the centre of rotation of the joint (Quanbury et al, 1975). By making the preceeding assumptions, Quanbury et al (1975) demonstated that power flow was equivalent to the instantaneous segment power which was easily calculated from the slope of the instantaneous energy curve.

The calculation of the instantaneous energy of any body segment is the sum of PE(m*g*h), KE(0.5*m*(v**2)), and RE(0.5*I*(w**2)) resulting in the following equation:

$$Ei = mgh + 1/2mv**2 + 1/2Iw**2 \qquad (2)$$

Ei= the energy of the ith segment (j)

m= mass of segment (kg)

g= acceleration due to gravity (m/(s**2))

h= height of centre of mass of segment above datum (m)

v= absolute velocity of c of m. (m/s)

I= moment of inertia about the centre of mass (kg*m**2)

w= absolute rotational velocity of segment (rads/sec)

Winter et al (1976) furthered analysis using instantaneous energy where the importance of power flow to and from the segment is emphasized rather than the absolute energy level. Thus, the changes in energy of a segment rather than the absolute energy level provide more comprehensive information which will now be demonstrated.

Winter and Robertson (1978) established patterns of generation, absorption and transfer of mechanical energy during walking gait. They listed these power patterns in tables, which were compared with the results from this present study for major similarities and differences.

Quanbury et al (1975) explained the additional information that can be derived from further computations involving the instantaneous energy values. Winter (1979B) calculated internal work performed during gait through the equation:

$$Wwb = \sum_{j=1}^{N} \left| \sum_{i=1}^{S} (\Delta Ei,j) \right| \quad J \qquad (3)$$

Wwb is the total body energy change of all segments over all sample periods. This internal work (Wwb) calculation accounts for energy exchanges within each segment, energy exchanges between adjacent segments, all potential and kinetic energy components and both positive and negative work. Total energy is calculated from the Quanbury et al (1975) equation. Winter's work was replicated and verified by Zarrugh (1981).

Pierrynowski et al (1980) calculated segment energy changes and classified them into change due to muscular activity and change due to passive mechanisms. This paper presented the full equations and assumptions involving energy transfer and 'internal work' calculations. The present study will involve the equations from the Pierrynowski et al (1980) work, therefore it would be in order to review the theory and rational involved in the calculations.

Pierrynowski et al (1980) combined equations for the calculation of work done assuming no exchanges within or between segments (Wn) with Winter's (1979A) equation for internal work (Wwb). The value of (Wn) is questionable by itself as it renders excessively high work values (Pierrynowski et al, 1980). However, for the calculation of the magnitude of the energy transfers within and between segments (Twb), (Wn) is necessary as:

$$Twb = Wn - Wwb \quad J \qquad (4)$$

In order to calculate energy transfers between segments (Tb) one must first calculate work done assuming there had been energy transfers within segments but not between segments (Ww). Energy transfers between segments occur when there is a translational movement of the joints, hence, one segment is doing work on another (Winter, 1979B). Although the value of work assuming energy transfers within but not between segments Ww on its own is not of great value, it can be used to calculate the energy transfers between segments:

$$Tb = Ww - Wwb \quad J \qquad (5)$$

Body segments contain components of all three energies, and the various combinations that appear during movement are due to the exchanges of these energies within the segment. The calculation of energy transfers within all segments can be derived from:

$$Tw = Twb - Tb \quad J \qquad (6)$$

To understand the origin of the equation for calculating Wwb one must start with the Quanbury et al(1975) equation for calculation of instantaneous energy for each segment, at each instant in time (equation 2). The sum of the absolute changes of the total energy over time yields Wwb. Mathematically this results in:

$$Wwb = \sum_{j=1}^{N} \left| \sum_{i=1}^{S} (\Delta Ei,j) \right| \quad J \qquad (7)$$

Ei,j is the total energy change during the jth period of time for the ith segment and N is the number of samples while S is the number of segments.

To calculate work done by the body which accounts for energy transfers within but not between segments, the absolute changes of the curve of total energy of each segment at each frame were summed over one movement cycle, hence:

$$Ww = \sum_{i=1}^{S} \sum_{j=1}^{N} \left| (\Delta Ei,j) \right| \quad J \qquad (8)$$

To calculate the work done assuming no energy transfers between or within segments(Norman et al, 1976), the absolute changes of PE KE, and RE of each segment were summed across all frames and segments:

$$Wn = \sum_{i=1}^{S} \sum_{j=1}^{N} (|\Delta PEi,j| + |\Delta KEi,j| + |\Delta REi,j|) \quad J \qquad (9)$$

Therefore the key to understanding the work equations is in the placement of the absolute bars for they determine the extent of energy that is to be summed. Pierrynowski et al(1980) also added a correction factor into the calculations as total body energy rarely ends at the same level of energy as it had at the beginning of the stride. This correction was calculated by taking the difference between the beginning and ending energy levels assuming the floor was flat and the body maintained a constant forward velocity. Thus:

$$COFF = \sum_{i=1}^{S} Ei,j1 - \sum_{i=1}^{S} Ei,jN \qquad (10)$$

The correction value was subtracted from Wwb and Wn, and their corrected values rendered the correct Tw, Tb and Twb.

The preceeding equations compute values that are of no use unless one is able to interprete and assess them in a diagnostic setting. The total work values over specific intervals of time are a measure which can monitor improvements(Winter, 1979B). When the total mechanical energy values (equation #2) are plotted on a graph, positive and negative work done is illustated. Inefficiencies in movement are indentified by uncharacteristic changes in the total mechanical energy curve and by an increase in the value of total internal work (the sum of energy changes over all frames) (Winter, 1979B).

If an inefficient movement exists as diagnosed from the total internal work calculation, then the total energy curve for each segment can be examined to locate, both the location and temporal occurance of the inefficiency. Winter (1978) reported four major causes of inefficient movement. The first were muscular contractions whereby the agonist and antagonist contracted together without producing movement at a joint. In other words, the flexors performed positive work which was directly absorbed by the paired extensors. While co-contractions occured in many pathologies, they are most frequent in hemiplegia and spastic cerebral palsy (Winter, 1978). The second source of inefficient movement was isometric contractions of muscles against gravity. In normal locomotion, fluid limb motion allows a smooth interchange of energy. In pathological gaits, limb segments are held in near isometric contractions against gravity. Winter (1978) documented a palsied child walking with crutches holding her leg off the ground for excessive amounts of time resulting in prolonged hamstring activity. The third source of inefficiency was the presence of jerky movements. Winter (1978) used the example of the smooth, fluid movements of the ballet dancer who deftly converts positive work into other energy forms during efficient movements. Movements characterised by a succession of stops and starts increases positive and negative work values thus increasing the metabolic cost. The fourth source of inefficiency was caused from the generation of energy at one joint at the same time as absorption occurs at another joint. Winter (1978) illustrated this concept with an example from normal gait. The pushoff leg generates energy at the same time as the weight accepting leg absorbs energy. However, measurement of such energy exchanges cannot be simplified for routine clinical use as it requires a detailed power flow analysis(Winter, 1978).

Once problems and inefficiencies in movement are indentified a solution must be sought and implimented. In the case of ambulation, attemps at solving such difficulties generally take two directions. The subject either partakes in revised gait training or an orthotic device is prescribed. In cases where orthotics are already present, redesign is the usual course of action(Winter, 1978).

2.6 INSTRUMENTATION

The literature contained ideas for transducers to measure axial force in crutches and their displacement in space. Zarrugh and Radcliffe(1979) utilized a system of strings attatched to locations on the body to monitor total displacements. However this method confined the subject to a treadmill.

The axial force transducers outlined in Baxter et al(1969) Bergmann(1979), Cochran et al(1973), and Klenerman and Hutton(1973) are either unnecessarily expensive, heavy and bulky or are so compressive that they change the style of swing-through crutch gait. Sydenham(1980) explained the advantages and disadvantages of all types of force transducers. The design aims for the elastic member in a transducer is to achieve a linear movement of adequate magnitude that suits the available microdisplacement sensors(Sydenham, 1980). The elastic member must have low mechanical hysteresis, a long fatique life and resistance to environmental factors such as humidity and temperature. For the present study, a single beam type transducer was designed. The ATLAS OF ORTHOTICS(1975) detailed the resistance, elasticity, plasticity, ductility and mechanical fatique characteristics of materials used in transducers and orthodics. The beams of transducers are strain gauged to measure the amount of deformation which indicate the applied forces. The electrical resistance strain gauge consists of a conductor which is cemented to the test object. This allows the strains to be transmitted from the test object directly to the gauge. The small changes in the gauge length that are caused by the applied load induce small changes in resistance of the conductor which are detected by the measuring instrumentation (Vaughan, 1975). The gauge factor describes the relationship of change in gauge resistance and change in gauge length. The circuit measuring resistance changes can be of three configurations. The first, the quarter bridge measures the change in resistance of one strain gauge by comparing the voltage to a 'dummy', non-functional gauge. The second configuration, known as the half bridge, utilizes two strain gauges. The advantages of this setup are a greater resultant out of balance signal giving greater accuracy and provision for the cancelation of temperature influences. The full bridge, which is the third configuration, uses four strain gauges resulting in additional sensitivity. This is achieved by orthogonally cemented pairs of strain gauges which compensate for temperature change and extraneous strains by achieving balance through an additional, adjustable resistor (Vaughan, 1975).

2.7 FITTING OF CRUTCHES

The ATLAS OF ORTHOTICS (1975) described the fitting proceedure of axillary crutches and the correct body posture to be maintained throughout the gait cycle. To find the proper length of the crutch, the distance from the anterior axillary fold to the heel is measured with the patient in a supine position. Thus, when the patient stands, the axillary pad of the crutch should be two finger breadths below the axilla, with the crutch tips placed six inches laterally from the base of the fifth toe. The height of the handgrips should be sufficient to give 20 degrees of flexion at the elbow. Thus, proper fitting is designed to allow the body weight to be borne on the handgrips - not the axillary pads. The top third of the crutch is to improve balance when pressed against the rib cage by the upper arm (Atlas of Orthodics, 1975). The wrists are held firmly in extension. The hips are in a neutral to slightly hyperextended position. Maintenance of an overall erect posture is imperative for the swing-through gait.

2.8 DATA FILTERING

Due to the fact that displacement data, from human movement studies, is differentiated to produce velocities, any noise in the signal will be amplified. Any data reduction system inherantly adds noise to the pure signal (Winter, 1979B). The mechanics of noise reduction in data collection was outlined by Winter et al(1974) where the concept of both signal and noise are contained in the frequency spectrum is explained. Every signal has harmonic content. In other words, as Winter(1979B) explained, any waveform can be the sum of a number of sine waves. The process that breaks down the waveform into the various frequency components is called a Fourier transformation or Harmonic analysis. The harmonic analysis evaluates the amplitude of the individual frequencies. The lowest frequency, called the fundamental frequency(f), becomes a reference base as higher frecuncies are expressed as multiples of 'f', called harmonics. In repetitious movement, such as crutch gait, the fundamental frequency is the frequncy of one complete gait cycle (approx. .5 Hz). Winter(1979B) found that in walking gait, the first seven harmonics contained 99.7% of the signal power. Thus, any signal component above the seventh harmonic was considered noise. It must be stressed that a fourier analysis can only assess one signal at a time. Therefore, many signals must be processed in order to analyse a complex movement such as walking. Thus not all the signals that are used to evaluate one movement will have identical harmonic content. The investigator must choose a cutoff harmonic that contains a high signal to noise ratio that suits all the signals.

The sources of signal noise can be due to camera vibrations, sprocket misalignment and errors when deriving the coordinates of body landmarks from film(Winter, 1979B). Therefore, if the harmonic components that contain little signal but substantial noise can be eliminated, a clean, accurate signal can be produced. Plagenhoef(1968) used a Chebyshev least squares polynomial curve fitting technique but it was found to smooth data too much(Pezzack et al, 1977). Curve fitting techniques assume that the data is a certain order polynomial whereby the computer selects coefficients using such criteria as minimum mean square error, to give a 'best fit'. Pezzack et al,(1977) concluded that the second order, recursive Butterworth digital filter outlined by Winter(1974) was the most acceptable filter and should be used on the signal prior to any differentiation. Once again, Winter(1979B) detailed the theory of digital filtering. The purpose of digital filtering is to select the signal containing frequencies, while rejecting the high frequency noise. Thus, it is desirable to allow the low frequency component of the signal to pass through while attenuating high frequency components above the selected cutoff frequency. Frequency cutoff selection is determined from the previously explained harmonic analysis. Although the digital filter attenuates frequencies higher than the cutoff frequency, some high frequency signal remains. In addition, some of the useful signal is attenuated in the frequency range of the cutoff. Hence, cutoff selection must be considered with care(Winter, 1979B).

Fader and Gold(1967) covered the mathematical theory of digital filtering which is beyond the scope of this paper, however Winter(1979B) described the filtering process. The order of a filter decides the 'sharpness' of cutoff with the sharpness increasing as the order is increased. The format of the digital filter is:

$$X'(nT) = a0X(nT) + a1X(nT-T) + a2X(nT-2T) + b1X'(nT-T) + b2X'(nT-2T)$$

X'= filtered output coordinates

X= unfiltered coordiate data

nT= the nth sample frame

(nT-T)= the (n-1)th sample frame

(nT-2T)= the (n-2)th sample frame

a0, a1, a2, b1, b2= filter coefficients

Filter coeffiecients are constants that depend on filter order, sampling frequency and cutoff frequency. The ratio of the sampling frequency to cutoff frequency determine the coefficients.

Winter(1979B) documented the phase shift of the output signal relative to the input due to the filtering process. In second-order filters, the shift is 90 degrees which causes 'phase distortion'. To correct the phase lag, the data can be filtered again, only backwards. Therefore, the cutoff is sharper by a factor of two, which results in a fourth-order, zero phase shift filter. Hatze(1981) agreed that the Butterworth filter of Winter(1974) did indeed provide a computed second derivative that agreed well with a directly measured acceleration record. Still Hatze(1981) pointed out, the Butterworth filter required the user to decide upon a cutoff frequency to attenuate the signal at the most useful harmonic. However, as previously pointed out, from a harmonic analysis of the signal, the investigator determines the cutoff frequency by computing the signal power at each harmonic.

2.9 BODY SEGMENT PARAMETERS

The estimation of body segment parameters were briefly mentioned in an earlier section. Dempster(1955) used eight male cadavers in a study from which the results are widely cited(Miller and Nelson,1973). Since that time Clauser et al(1969) and Liu et al(1971) have provided additional information on adult males using cadavers. Drillis and Contini(1966) used the reaction change method to estimate segment weight but could not determine segment moments of inertia. Casper et al(1971) reported determination of body segment parameters on living subjects through a radiation technique. Gamma rays were focussed through the body segments which evaluated density and mass by the number of photons that were able to penetrate. The technique was furthered by Zatsiorsky and Seluyanov(1981), however, segment information on children was not reported.

The only report to date on inertial characteristics of children was the result of Jensen's(1978) work. He recorded the dimensions of the segments by digitizing photographic records of both front and side views of the subject as reflected in a 45 degree mirror. A mathematical model was then used to estimate inertial characteristics. The model is based on the assumptions that the body is composed of elliptical zones, two centimeters wide and segment densities are known(Jensen,1978). The masses and volumes of the elliptical zones are summed over the segment lengths from which segment masses and moments of inertia are computed. Jensen(1981) has followed 12 boys over three years (n=32) determining their inertial properties. It was interesting to note that inertial properties are a function of development as Jensen(1981) noted change with respect to age.

2.10 SUMMARY

The preceeding review has examined considerations for crutch usage and reviewed previous investigations into crutch gait analysis. The theory of kinematics and energy transfer in the body during locomotion established a base for the rationale and methodology of the present study. In addition, information on digital filtering and instrumentation familiarizes the reader with the various theories and methods that must be considered when undertaking kinetic investigations. It is hoped that this information provided a foundation upon which this study established energy transfer patterns in children walking with crutches. Thus, the following study attempted to analyse child crutch gait for efficiency, while relating the factors that were outlined in this chapter.

Chapter III

METHODOLOGY

3.1 INTRODUCTION

The swing-through crutch gait of 8 subjects was analysed. The subjects were 'normal' with no known gait pathologies. All subjects were children. The 'normal' subjects were given crutches of various lengths to ambulate.

The crutches provided were standard wood, axillary crutches, fitted to the patients in accordance to the Atlas of Orthotics(1975). The swing-through gait model of Cohen(1979) was followed and imitated, as closely as possible, by the subjects. Therefore, the placement of the tip of the crutch was 4-6 inches in front of the unaffected foot at the beginning of each gait cycle.

3.2 COLLECTION OF ORIGINAL DATA

The subjects were asked to walk (with crutches of varying lengths) along a walkway following a straight line. A demonstration was given to the group of subjects illustrating the correct method of crutching. Each subject was allowed to have a very short practice trial to correct any blatant faults. The lengths of crutches used in the trials were decided upon by the following process. First, crutch length, as predicted by Cohen(1979) was considered the fitted length, and then trials of 4cm longer and shorter than this length were performed. The length of 4cm was chosen as this was the size of two increment stages in the length adjustable crutch. Landmark data was collected from the following joints: R and L Gleno-humeral, R and L Ulnar styloid, midpoint between greater trochanters, R and L femoral condyle, R and L lateral malleolus, and R and L fifth metatarsal-phalangeal joint. These sites were chosen following the methodology of Wells(1979) and Winter(1979B). Not all landmarks were in full view of the camera during all phases of the gait cycle. In such cases the landmark locations were extrapolated from the body segments and markers that were in view. From this joint data, the centre of gravity of the following segments was determined: Head-neck-trunk, R arm-crutch, L arm-crutch, R thigh, L thigh, R shank, L shank, R foot, L foot. This is in accordance with Murray et al(1967) as landmarks directly on the soft tissue over the segment centre of gravity tends to move more than tissue over joints. The head was considered rigid on the trunk as the pilot study revealed that there is not distinguishable movement between the two during crutch gait. It was this 9-segment model that was evaluated(illustrated in Appendix 1). However, a 10-segment model was also constructed by considering the Head-Neck, and Trunk as two separate links. The ear canal landmark facilitated this task.

Cine camera (1) (Locam 16mm) was set 25-30 feet from, and orthogonal to the walkway in the sagital plane. Cine camera (2) (Bolex 16mm/telephoto lens) was positioned with its focal axis parallel to the walkway to record the lateral movements of the centre of gravity of the arm-crutch segment. The frame rate of both cameras was set at 20 frames per second thus giving about 35 frames per gait cycle. The cameras were orthogonally alligned and leveled, on a stationary reference frame. A scaling stick was filmed to later scale the raw data. The film used was KODAK tri-x 400 ASA, black and white. The walkway was illuminated with 4 Ianabeam tungsten lights. Additional subject data collected were height, weight, length of crutch and crutch weight. The axial force transducer on the right crutch was of a single beam type, constructed of aluminum for its resistance to change from humidity, temperature and resistance toward mechanical fatigue under loads experienced in this study (Vaughan, 1975). Lateral crutch torques were also measured and recorded from this transducer. The strain gauges (5 m.m. in length, gauge resistance-120 ohms, gauge factor-2.1%, gauge factor change with temperature-0.015%, and temperature compensated for aluminum) were bonded to the aluminum with Kodak adhesive(EC-10). Four strain gauges were used, two on each surface of the aluminum beam. Each pair was alligned orthogonally to cancel out any apparent strain due to temperature variations. The lead wires from the gauges were directed to a Honeywell Full Bridge(Accudata 218) amplifier which amplified and directed the signal to the Honeywell U-V Visicorder(model 1508 B). Triaxial force plate(Kistler) data were also recorded by the visicorder of the foot-ground interface during the support phase of the right foot for one cycle.

3.3 ANALYSIS

The co-ordinates of the 11 joint markers, for each frame, were derived from a Numonics digitizer(courtesy of the University of Waterloo, Dept. of Kinesiology). Absolute angles, relative to the horizontal were derived from the segment joints by computer. The source for the body segment parameters of children was from the work of Jensen(1981). This consisted of regression equations for determining the segment masses and radius of gyration information from which inertial properties were resolved. In order to compute the moment of inertia of the arm-crutch segment, the following method was used. The masses of the arm(Jensen, 1981) and crutch were established. A static setup was utilized to ascertain the combined centre of mass of the total arm-crutch segment. In order to aquire the moment of inertia of the arm-crutch segment, the inertias of the arm and crutch about their proximal axis' were computed individually.

Since the arm and crutch share the same axis of rotation (the gleno-humeral joint) in crutch gait, their separate inertia's about this common proximal joint can be summed. Once the combined inertia about the proximal joint and the distance of the combined centre of mass to the proximal end are known, the moment of inertia about the centre of gravity can be computed using the parallel axis theorem. The moment of inertia (proximal) of the crutch was determined from the period when the crutch was set into motion as a pendulum. The coordinates from the landmarked joints were processed by a computer analysis program.

The computer package required the input landmark coordinates, the proximal distance of the centre of gravity of the arm-crutch segment, the masses of the segments from regression equations (Jensen, 1981), the moments of inertia of segments about their centres of gravity (determined from Jensen (1981) data which can be converted using the parallel axis theorem), the number of frames of data, a scaling factor and a baseline (datum) y-coordinate. The filter sample rate and cutoff were fixed for the study at values of 20/sec and the 5th harmonic respectively. As the stride frequency is approximatly 1 gait cycle/2 seconds (1/2 hz.) the fifth harmonic is equivalent to 2.5 Hz. As the second order Butterworth filter becomes a fourth order, zero phase shift filter, the cutoff frequency becomes the original frequency of 2.5 Hz divided by .802. Therefore, the filter cutoff frequency is 3.12 Hz. Thus, the ratio of the sample rate over the cutoff frequency is greater than 4 which satisfies the sampling theorem. These values were chosen as the result of a pilot study which indicated that nearly all of the signal was contained in the first five harmonics for all the coordinate displacement traces. The whole computing process only required one execution run to render the complete output. This was achieved by executing the program through a WATFIV compiler with the filter as a WATFIV-S subroutine of the main program. Another subroutine in FORTRAN IV to determine the coordinates of the segment centres of gravity was also part of the main WATFIV-S program. The center of gravity subroutine utilized the equations for computing segment centres of gravity from Dempster(1955) (as cited in Winter, 1979B) but included the values for segment masses as a percent of total body mass from Jensen(1981). The segmental centre of mass coordinates, along with the segment angles, were then filtered by a second order, zero phase shift digital filter as described by Winter(1979B). The filtered centre of gravity coordinates and angles were read into the WATFIV-S main program. The angles were converted from degrees to radians while PE, KE, RE, and TE(total energy) for each segment over each frame was calculated. The next step in the program computed instantaneous power for each segment and over each frame. The following values for work were then calculated: work assuming energy transfers within and between segments, work assuming no energy transfers, work assuming energy transfers within segments only. The work values were required for the next loop in the computations to render: energy transferred within and between segments, energy transfered within segments only, energy transfered between segments only. The format of the output for segment energies and powers was designed in order that they could be easily plotted over the gait cycle by a calcomp plotting machine.

The plotted curves of instantaneous energies and powers for the 'normal' subjects were compared for similarities. This was to achieve the goal of establishing so called average curves for 'normal' subjects. The characteristics of the curves were noted, from which the 9 segment model of crutch gait was evaluated.

Chapter IV

RESULTS AND DISCUSSION

4.1 INTRODUCTION

Twenty-one trials were performed by eight subjects, ambulating with crutches of various lengths. Anthropometrics, crutch and foot impact forces, and displacement data recorded on film were collected. The following comparative analysis of the derived mechanical energies, powers, mechanical work, crutch forces, foot-floor forces, and segment model evaluation was detailed in chapter 2. Although all subjects were shown, and asked to perform swing-through crutch gait, some trials were of swing-to gait for various reasons that will be explained.

4.2 SUBJECT DATA

Data of the eight subjects is listed in Table 1. Inertial parameters of the body segments are listed in Table 2 which were derived from regression equations developed by Jensen(1981) as such reference data on children is extremely rare.

Moments of inertia of the various crutch lengths used by subjects in this study are found in Table 3. The codes for

TABLE 1

SUBJECT DATA

SUBJECT	SEX	AGE	HEIGHT (m)	weight (kgf)
B	m	10	1.38	35.4
C	m	10	1.32	35.4
D	m	9	1.45	44.3
E	m	9	1.46	37.7
F	m	11	1.46	35.7
G	f	11	1.34	33.3
H	f	9	1.32	27.1
I	f	9	1.30	29.3
means		9.75	1.38	34.77
S.D.		.886	.068	5.24

the trials starts with a letter which corresponds to the subject, the next two digits are the trial number while the last three digits were the crutch length used. Each trial was comprised of a few gait cycles. The most typical cycle from the trial was chosen for analysis unless otherwise noted.

TABLE 2

SEGMENT MASS AND INERTIA DATA

MASS/SEGMENT (kgf)

SEGMENT	SUBJECTS							
	B	C	D	E	F	G	H	I
H-N-T	19.3	19.3	24.8	20.7	18.5	16.1	15.4	17.7
H-N	4.4	4.4	6.2	4.9	3.6	3.8	4.1	3.7
Trunk	14.8	14.8	18.6	15.8	14.9	12.3	11.3	14.0
arm-cr	2.7	2.8	3.2	2.8	2.8	2.4	2.2	2.6
thigh	3.5	3.9	3.9	3.4	4.2	2.6	2.6	3.6
shank	1.8	1.7	2.2	1.9	1.8	1.5	1.3	1.6
foot	.7	.7	.8	.7	.7	.6	.5	.7
TOTAL	35.3	35.4	44.3	37.7	35.7	29.3	27.1	33.3

SEGMENT MOMENT OF INERTIA (KgfM**2)

H-N-T	.9	.9	1.01	.89	1.07	.46	.5	.9
H-N	.024	.016	.028	.018	.013	.007	.015	.013
Trunk	.44	.44	.524	.45	.49	.23	.27	.42
arm-cr	.23	.23	.22	.21	.21	.18	.17	.17
thigh	.06	.05	.08	.06	.07	.04	.04	.04
shank	.018	.023	.033	.028	.026	.015	.015	.023
foot	.0006	.0008	.000	.0003	.000	.000	.0002	.000

Note: Calculations were carried to three decimal points. Minor discrepancies between the sum and the total values were due to roundoff error.

TABLE 3

CRUTCH DATA

LENGTH (cm.)	CENTRE OF MASS PROX. (cm.)	MOM. INERTIA PROX. (Kgm**2)
91	49.5	.25
99	52	.32
102	52.5	.325
104	53	.34
106	54	.347
110	55.5	.38

4.3 MODELS

The instantaneous energy and power records in graphic form were produced and examined for all trials using both the 9 and 10 segment models. The criteria upon which the traces were evaluated are as follows: 1) The temporal occurances and general pattern of both postitive and negative peaks, 2) The interaction between the various component energies in energy traces, 3) The absolute magnitude of trace peaks 4) The number of peaks in one gait cycle regardless of magnitude, 5) Comparison of traces between adjacent segments.

The mechanics of crutch gait vary among subjects (Peacock, 1966, Wells, 1979). The 9-segment model, due to its property of the head-neck-trunk (H-N-T) as one rigid link, is more simplistic than the 10-segment model. However, if the 9-segment model is to be more practical than the

10-segment model, the supposition of the head moving in symmetry with the trunk must be true. The only advantage of the 10-segment model will be to describe motion where the head moved independantly as in paraplegic ambulation(Childs, 1964). The subjects in this study demonstrated the 9-segment model to be adequete in describing 'normal' crutch gait.

In general, the energy and power traces for the 10-segment trunk and the 9-segment H-N-T were of similar shape with the 10-segment trunk being of smaller magnitude. For example, a typical power curve for the 10-segment trunk and 9-segment H-N-T, of the same trial(C02102), demonstrate this phenomenon(Figure 1).
The instantaneous peaks of power are greater in the 9-segment H-N-T due to the additional mass contribution of the head and neck.

If the head, neck and trunk move as a unit, in other words, they can be accurately modelled as a rigid link, then one would expect the power curves of the head-neck to be of the same shape as of the trunk in each trial. This appears to be the case as can be seen in Figure 2, which typifies power curves of the trunk and head-neck segments of the 10-segment model. (trial D12110)
The magnitude of the power curve for the head-neck segment is surprisingly large (being of much smaller mass) due to

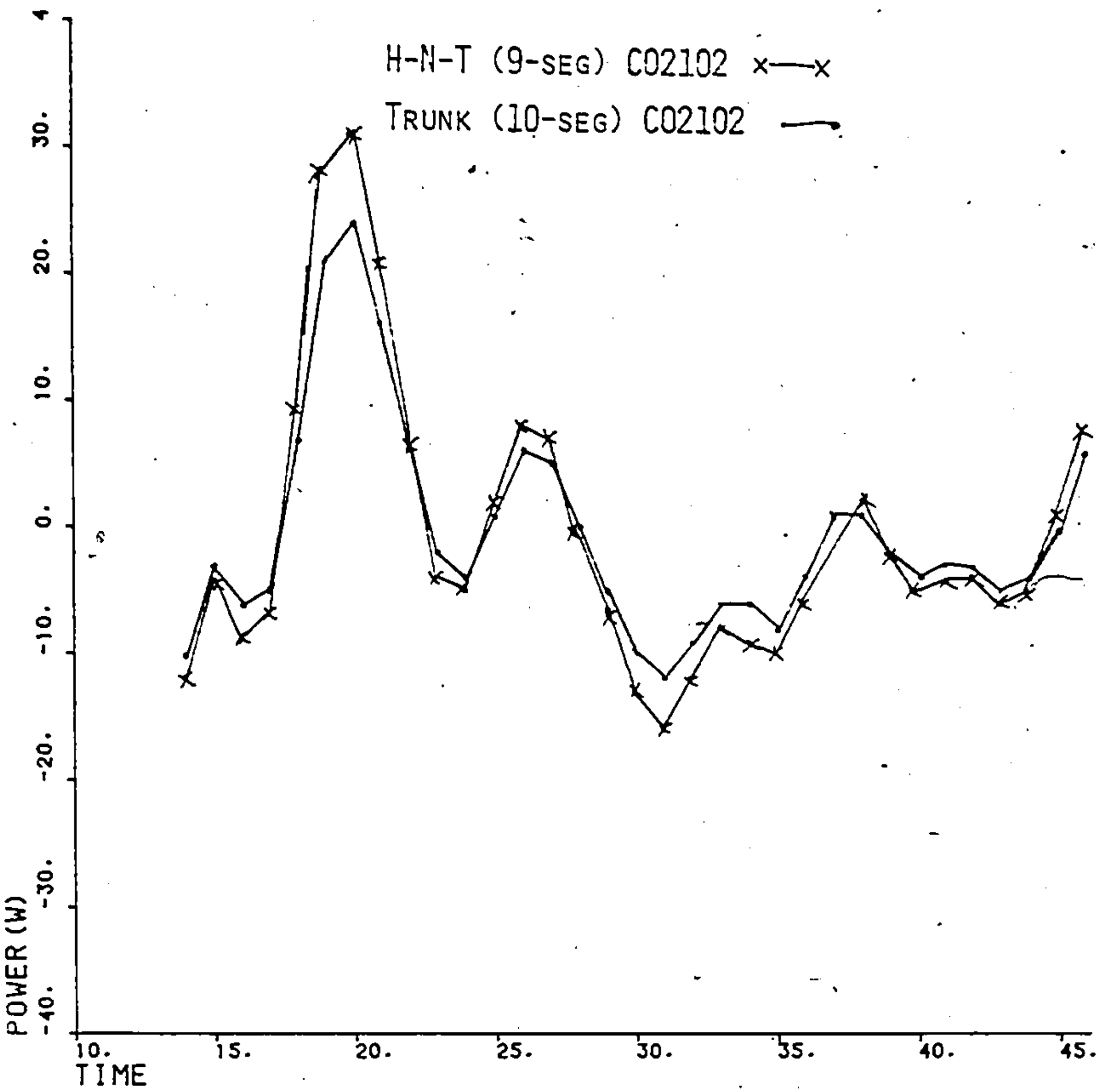


Figure 1: 9-segment H-N-T vs 10-segment trunk

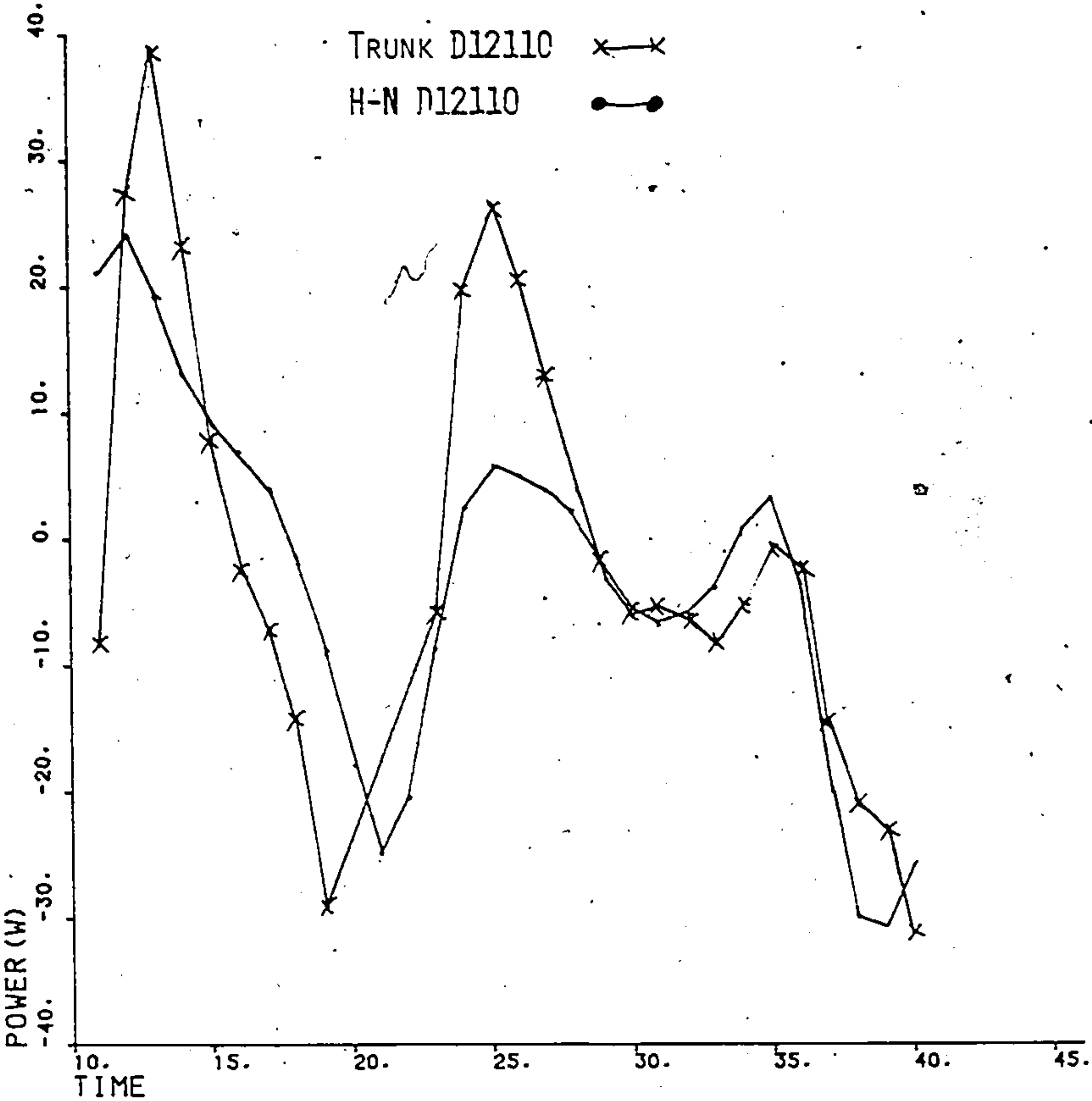


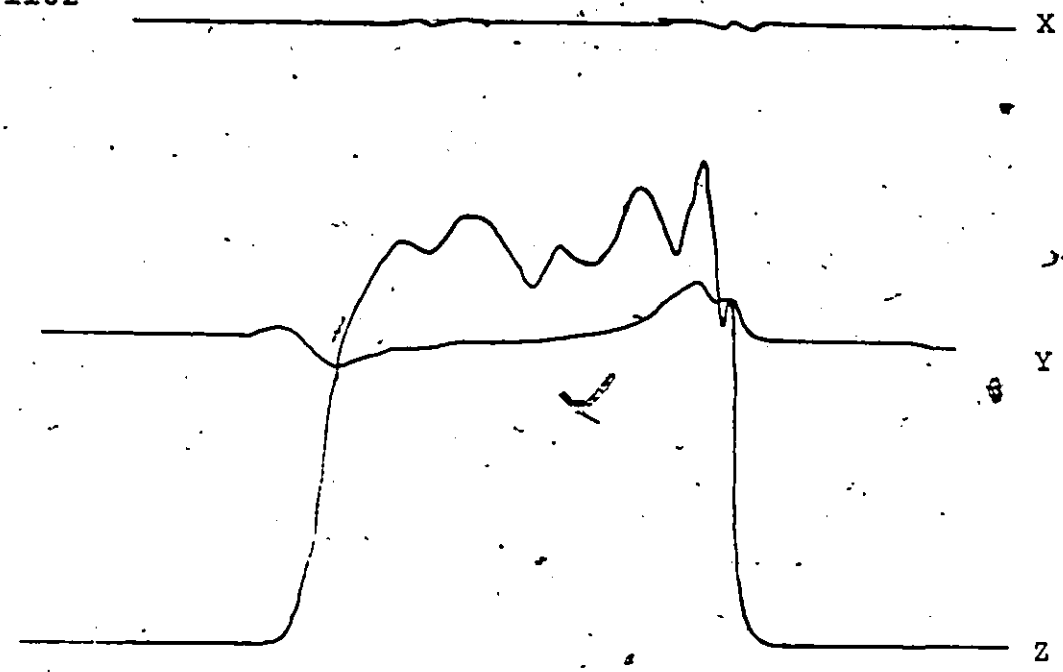
Figure 2: 10-Segment Model, Trial D12110

its higher orientation above the ground than the trunk. The fact of the head-neck and trunk power curves sharing a common shape concludes for normal crutch gait in children the 10-segment model does not provide additional information over the 9-segment model. Therefore in this case, the difference in peak mmagnitudes of the two traces is due to a difference in segment mass. Rather the temporal occurances of the peaks are similar which leeds to the conclusion that the separate link of the head-neck provides no additional information to gait analysis. However, the 10-segment model, or another variation of the 9-segment model, may be of greater use in the analysis of pathological crutch gait such as those exhibited by paraplegics and cerebral palsy victims. With this information in mind, a full, detailed analysis of all links in the 9-segment model will be undertaken utilizing the 'normal' children involved in this study.

4.4 ANALYSIS OF FOOT CONTACT FORCES

The triaxial force platform recorded the horizontal X (perpendicular to direction of travel) and Y (in the direction of travel); and the vertical Z directional forces. Traces of the foot contact forces are displayed below and on the following pages. Forces of all trials are not available as subjects did not always manage placement of the striking foot squarely onto the platform. Despite the lack of complete records from all trials, similarities in the existing traces suggested trends.

B01102



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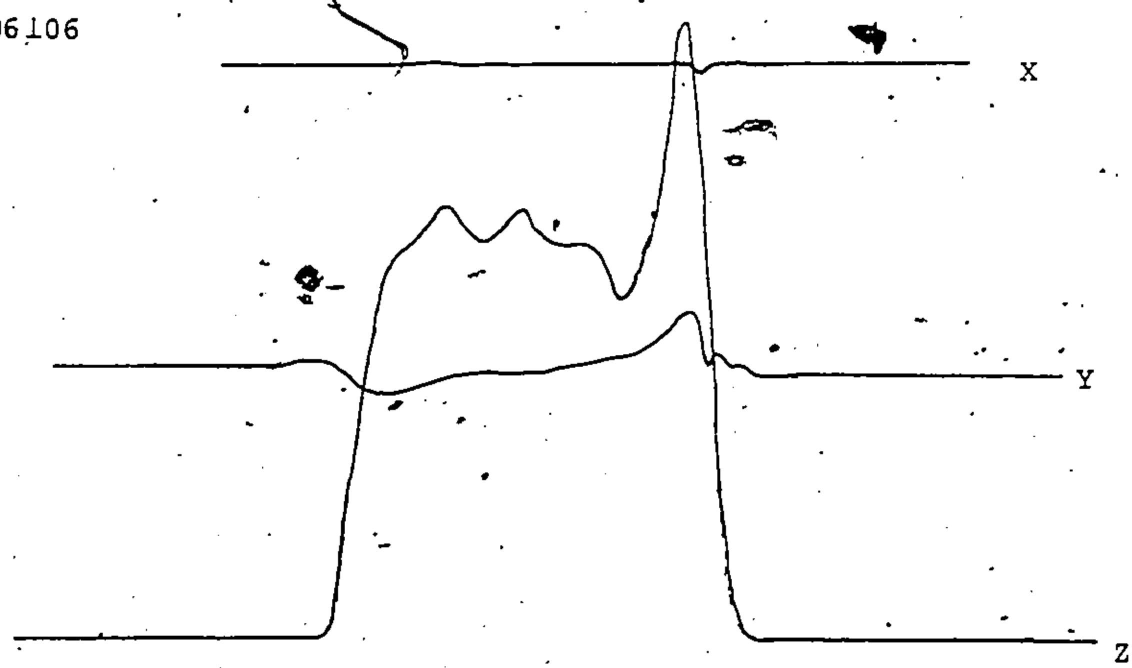

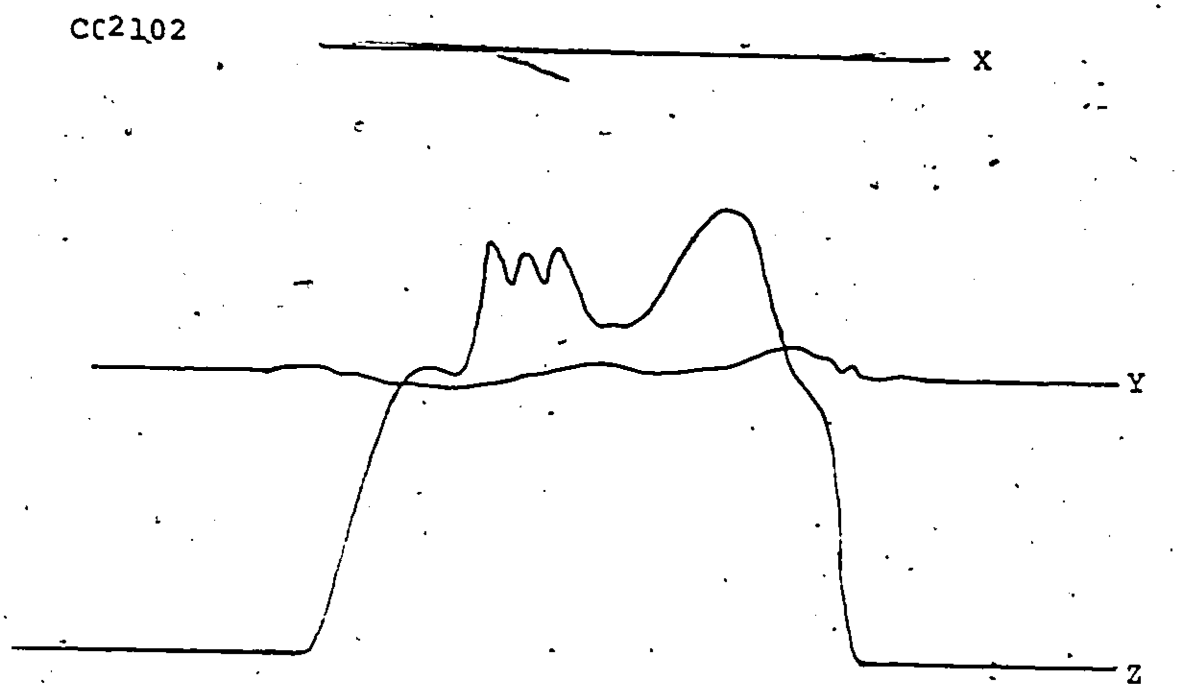
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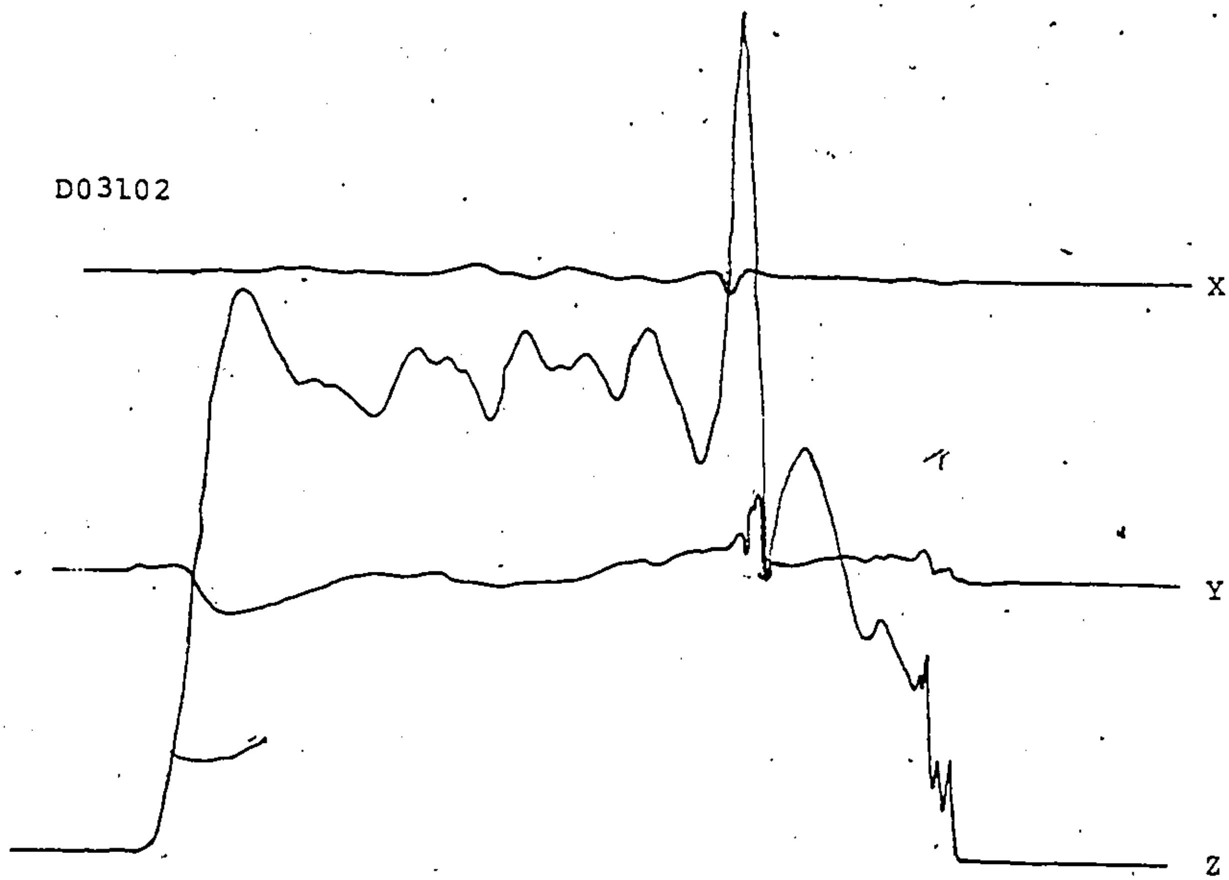
X
Y
Z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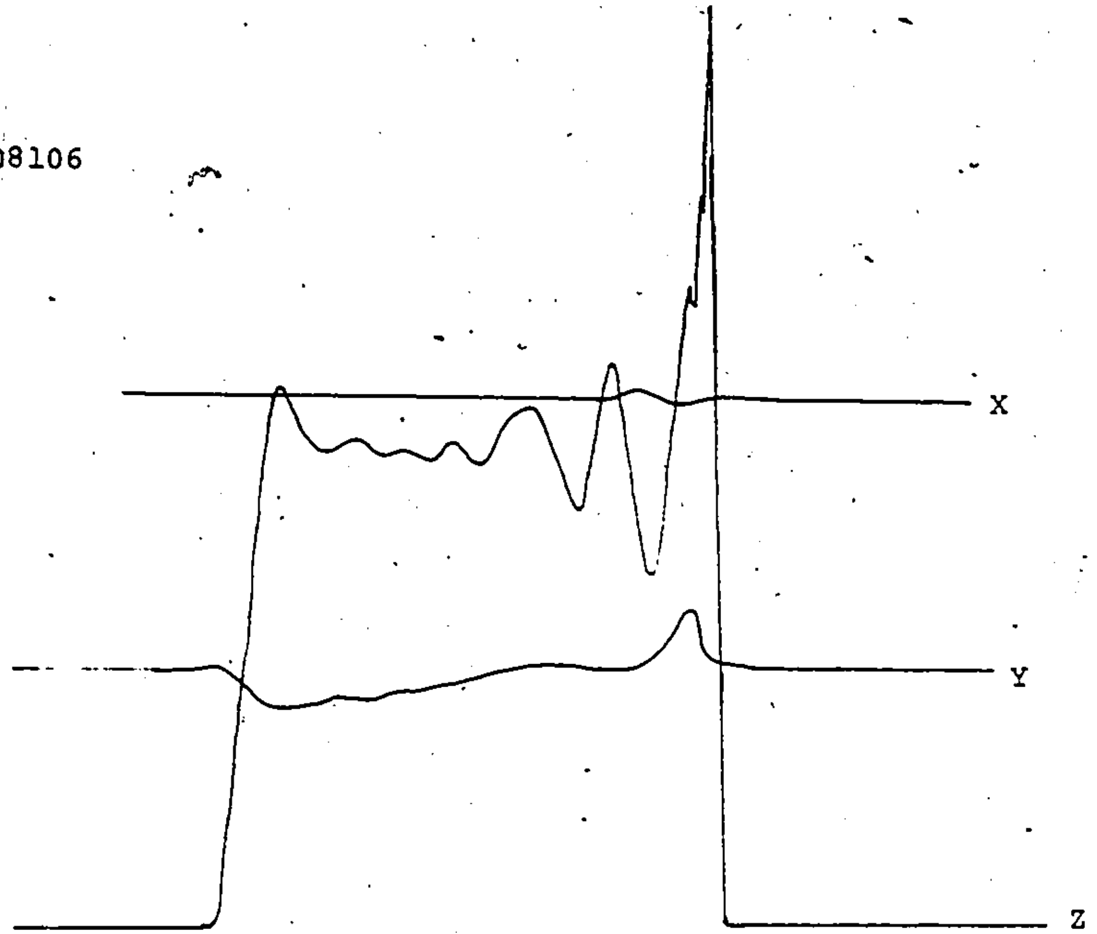
X
Y
Z

D08106



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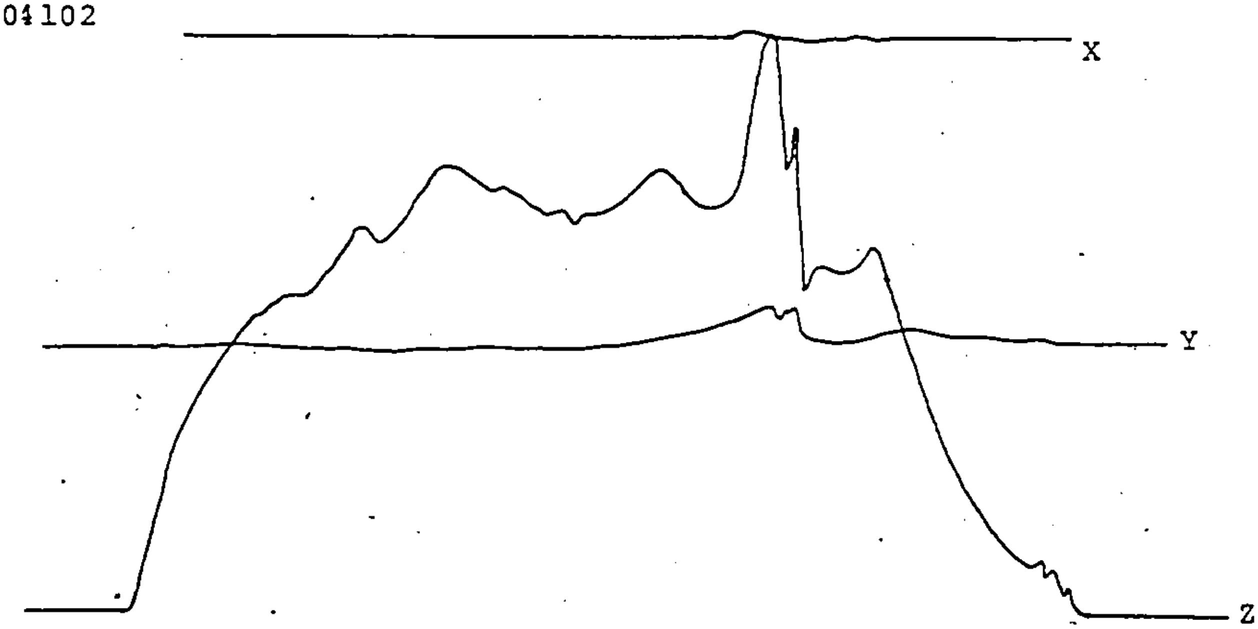

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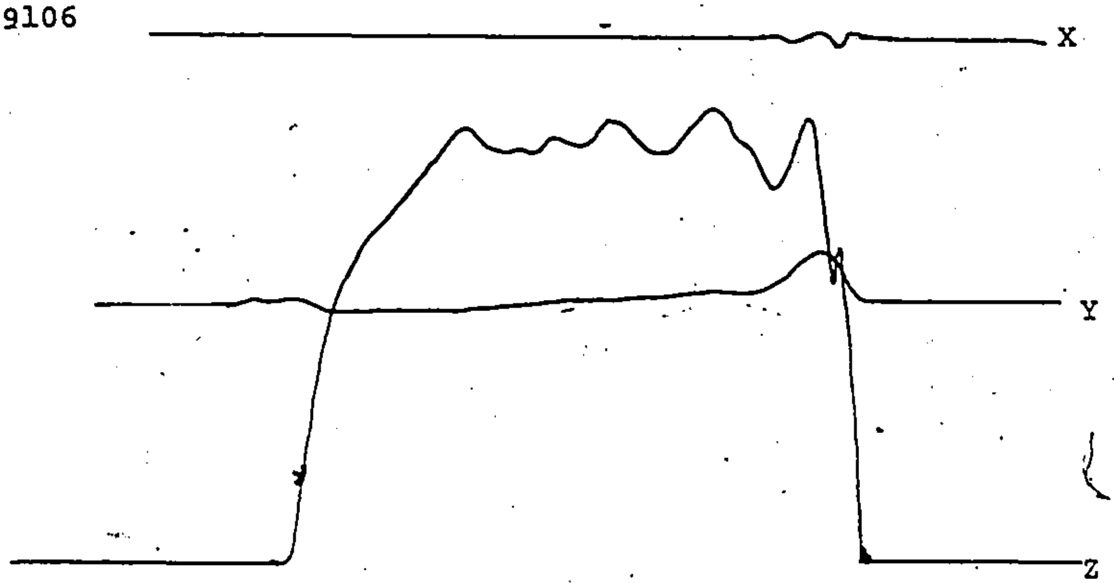

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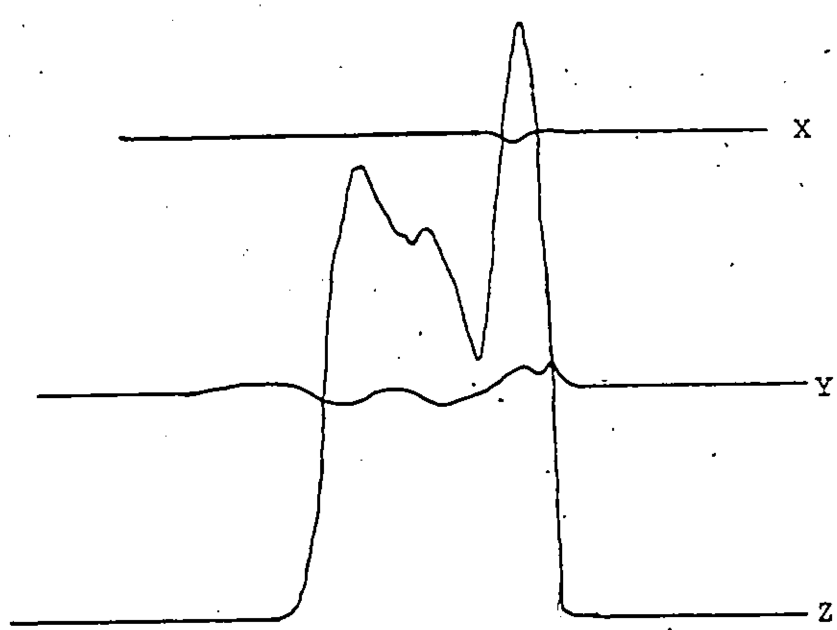

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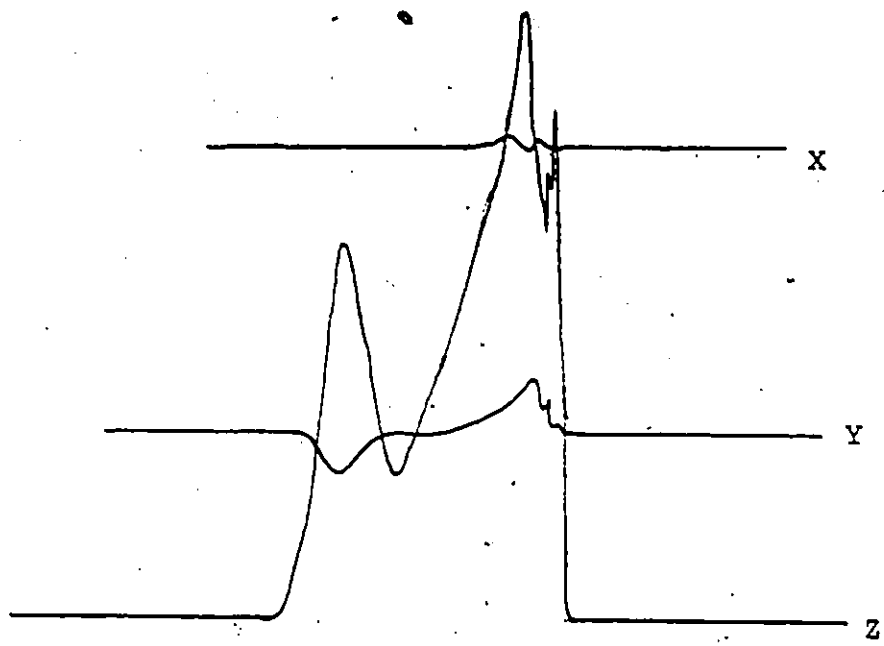

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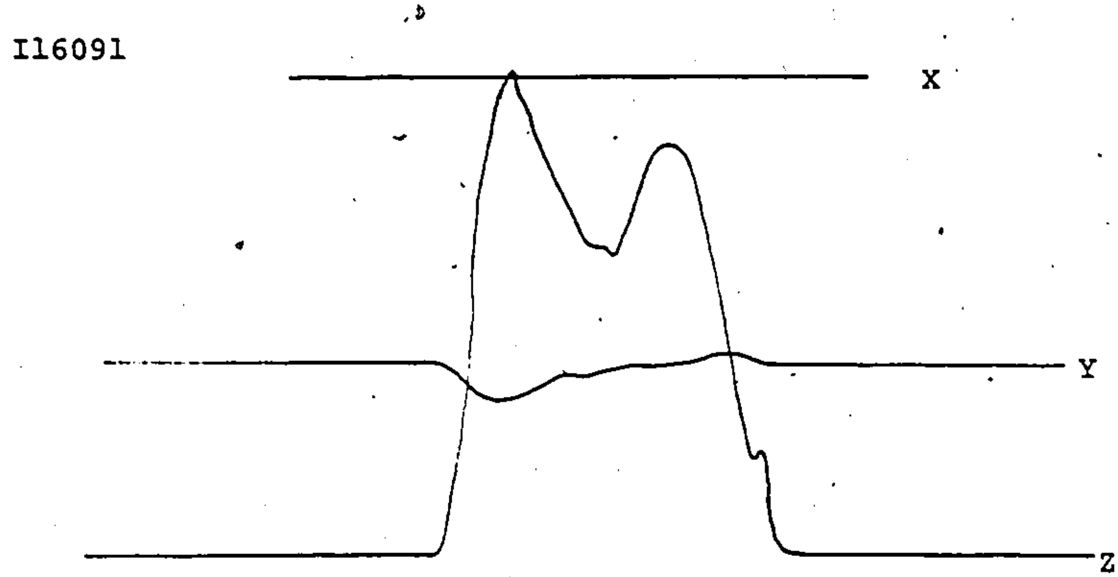
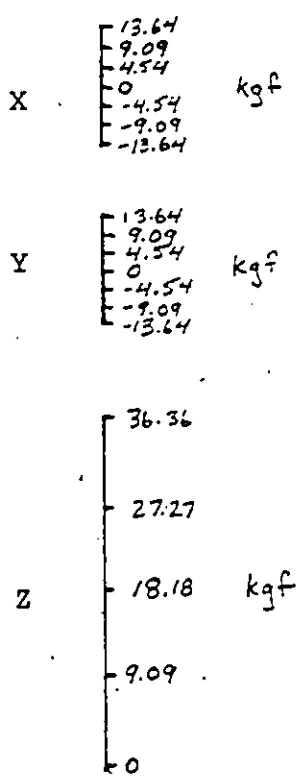


Calibration Of Force Plate



Subject B manifested lower peak Z forces with fitted crutches over long crutches(+4cm) by approximately 75%. The peak force in the long crutch trial occurred at heel-strike(H-S). The remaining portion of single leg support produced roughly the same forces as the fitted crutch trial. The magnitude of the peak Z force in the long crutch trial was approximately 140 lbs. or 175% that of body weight. The Y directional forces exhibit remarkable similarities between trials, thus indicating consistancy in energy absoption and generation patterns in the fore-aft plane. The long crutch resulted in higher peak Z force but no real change in Y direction foot contact forces in subject B.

Subject C, using crutches of 102cm which were of correct length, had a Z direction force application pattern that would almost suggest a reluctance to apply pressure upon heel strike in order to support the body mass. Qualitative film analysis revealed balance problems and slipping of the axillary pad resulting in the abnormal Z direction trace.

Subject D displayed the highest peak Z forces of the subject population, approximately 200 lbs. or 200% of body weight. The landing of the support leg appeared very abrupt on film with an actual oscillation or springing action occuring in the knee and ankle for a period of recovery after initial energy absoption. This oscillation appeared to be present in Z force traces after the occurance of the peak force when using both short (102cm) and fitted (106cm) crutches. The magnitude of the oscillations were positive and negative 20 lbs. from body weight. The X directional force trace during the short crutch trial, demonstrated unsteadiness in the sideways plane depicted in the deviations throughout the period of foot-ground contact. The Y direction trace exhibited a large peak at H-S indicating an abrupt change in motion in the forward direction. Film confirmed this observation as forward momentum was instantaneously halted at heel strike only to be resumed shortly afterwards as the cycle commenced with crutch-swing. The pattern of force application in the fitted crutch trial was for the most part, absent of the high frequency oscillations associated with balance problems. Slightly greater peak forces were recorded with fitted crutches, however the subject appeared to benefit as an improvement in balance was achieved.

The foot impact forces of subject E using fitted crutches, indicate very even force application in the Z direction. During this trial, the subject demonstrated a smooth flowing gait. There was an absence of any pushoff force indicating that the body mass was able to conserve linear momentum during the trial. However, there were X directional forces present at heel strike indicating force application that was not directly parallel to the direction of travel, hence detrimental to efficiency. However no evidence of balance problems were obtained during qualitative cine analysis. Comparing this trace with that of a trial where short crutches were used, balance problems were evident both in the Z direction trace and from film observation. The temporal duration of foot-platform contact was also greatly increased by the short crutches suggesting a much slower gait.

The force trace of subject F, using fitted crutches, resembled a typical trace of normal walking. There were two distinct, Z direction peaks, one occuring at H-S and the other at toe pushoff (T-O). The Y direction trace also indicated negative and positive body acceleration from H-S to pushoff. A small force was recorded during H-S in the X direction. During this trial the subject preferred to use a type of gait that resembled swing-to rather than swing-through style, where the planting leg does not break a line drawn between the planted crutches during the swing phase.

Subject G was equipped with fitted crutches during this trial. Large Z direction positive and negative peaks described the motion as manifesting large force changes. In fact, this subject took the longest strides, was the shortest in physical stature, and due to the short duration of force application was the fastest moving. The results were large peaks at H-S and toe pushoff in both the Z and Y directions.

Subject I was furnished with crutches 4cm too short during this intriguing trial. The Y and Z directional toe pushoff forces were of greater magnitude than the H-S peak force. This indicated more energy generated at toe-off than was absorbed at heel-strike. This subject had no observable problems but rather exhibited a smooth, flowing gait.

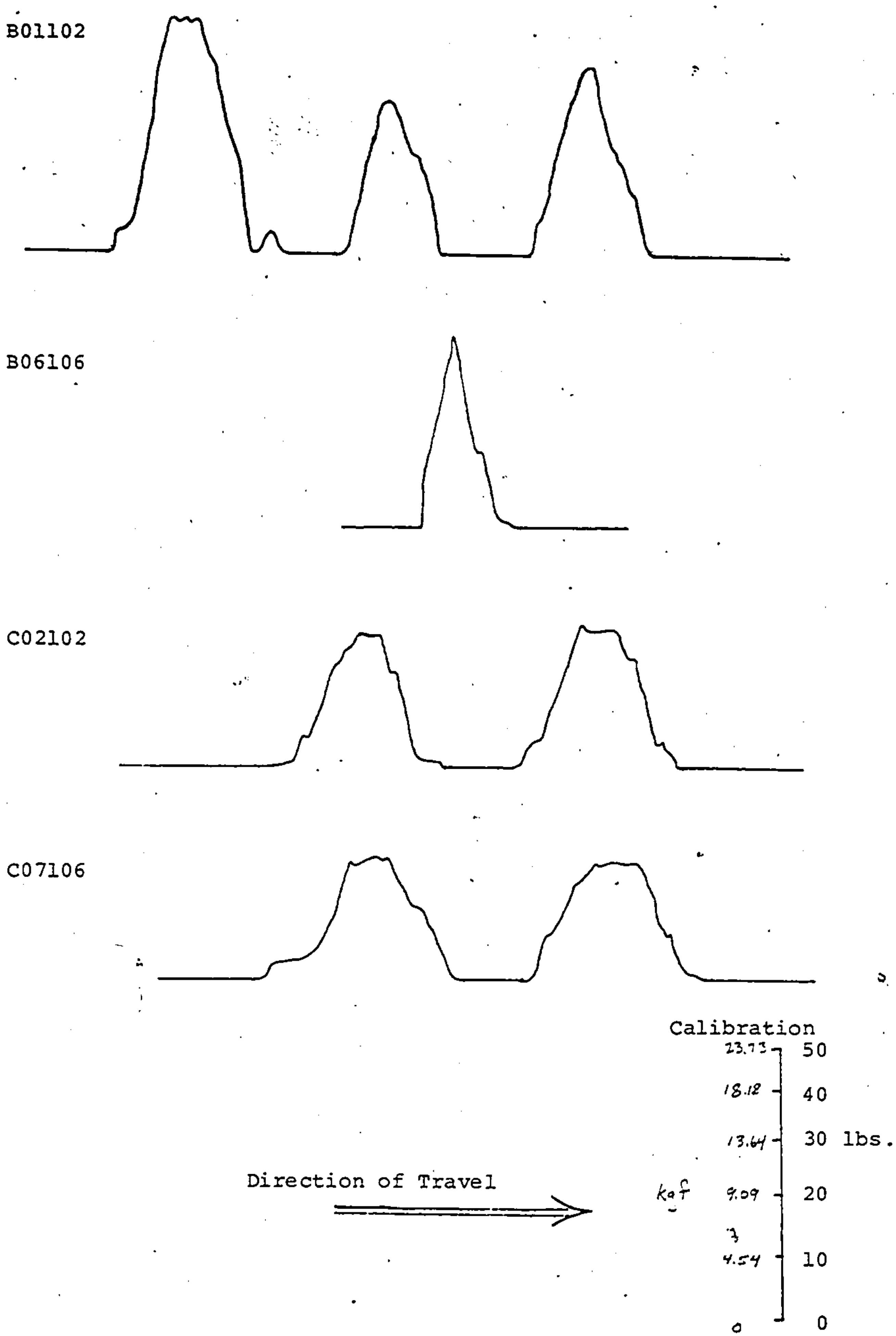
In general terms, foot contact forces provided insight into the total movement scheme. Forces in the X direction should be at a minimum as they do not contribute to forward progression, but do indicate problems of balance. The Y direction generally indicated those 'braking', energy absorbing forces during H-S that are due to a relatively rigid knee joint which is required for the support of the body weight. Force in the negative Y direction at toe pushoff is required for forward linear velocity providing the necessary momentum to carry the body through the double crutch support phase. The same energy absorbing and generating forces should be evident in the Z direction for similar reasons. The 'sharpness' of the Z-force slope at H-S indicates the severity of the foot planting mechanism. The presence of higher frequency patterns in the three force curves were, in most instances, indicators of balance problems. Thus, it was difficult to construct an ideal pattern of force application, but rather, features should be considered that may indicate the aformentioned sources of inefficiency in crutch gait.

4.5 AXIAL CRUTCH FORCE ANALYSIS

A quarter bridge transducer just above the rubber cup on the lower shaft of the crutch recorded axial loading (compressive force down the long axis of the crutch). Force traces of all trials were not recorded due to paper drive failure in the recorder. However, sixteen traces were obtained from the testing trials which contain additional information such as peak force loading, time spent in double crutch support and force loading patterns.

Upon examination of traces from two trials of subject B, no set pattern appeared, although peak forces were generally in the 13-16 kgf range which was close to 50% body weight signifying that crutch strike (C-S) was not severe.

The force traces of subject C had quite gentle loading and unloading slopes which was indicative of a non-injurous gait style.



B01102
B06106
C02102
C07106
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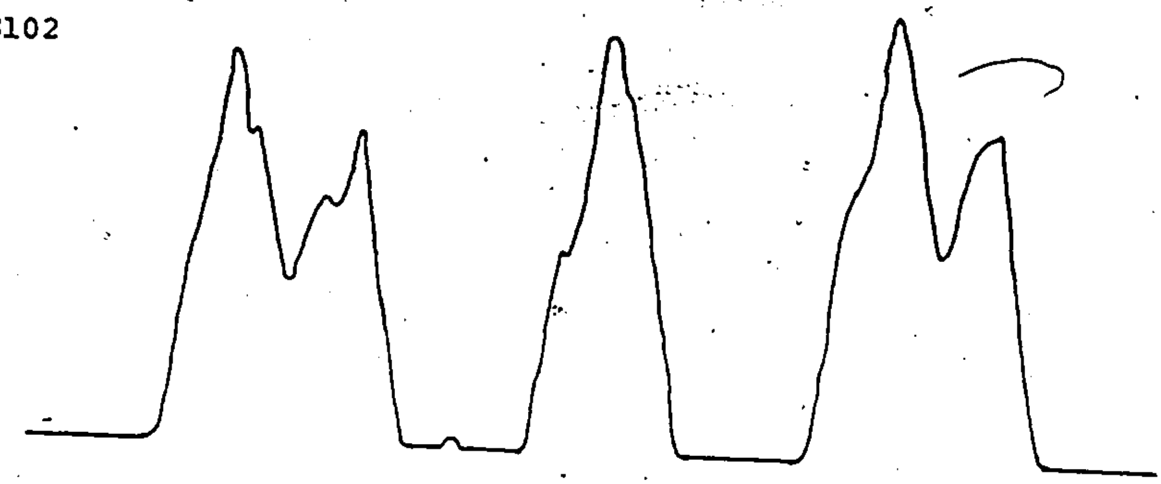
18.12 40
13.64 30 lbs.
kgf 9.09 20
4.54 10
0 0
Direction of Travel

D03102



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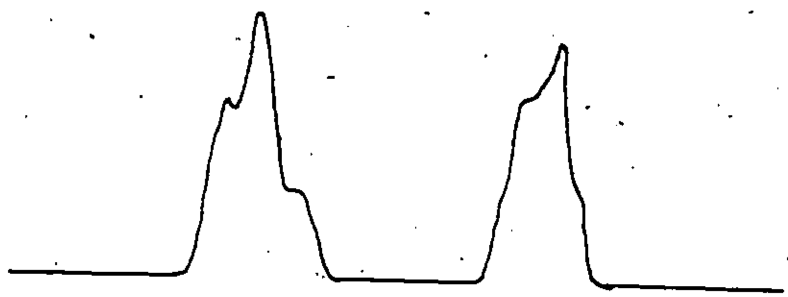

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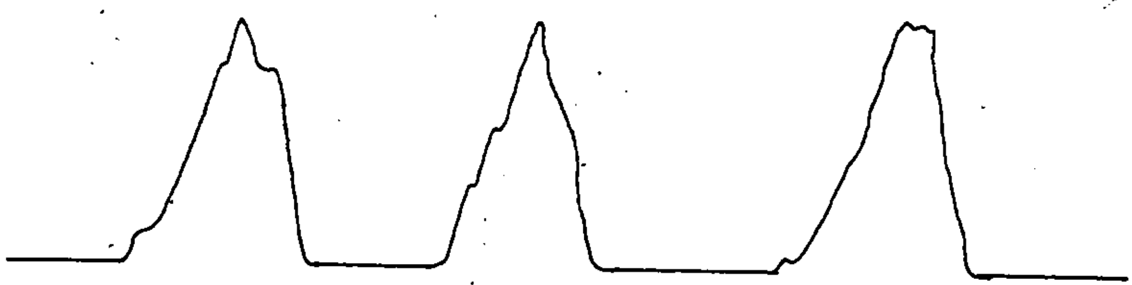

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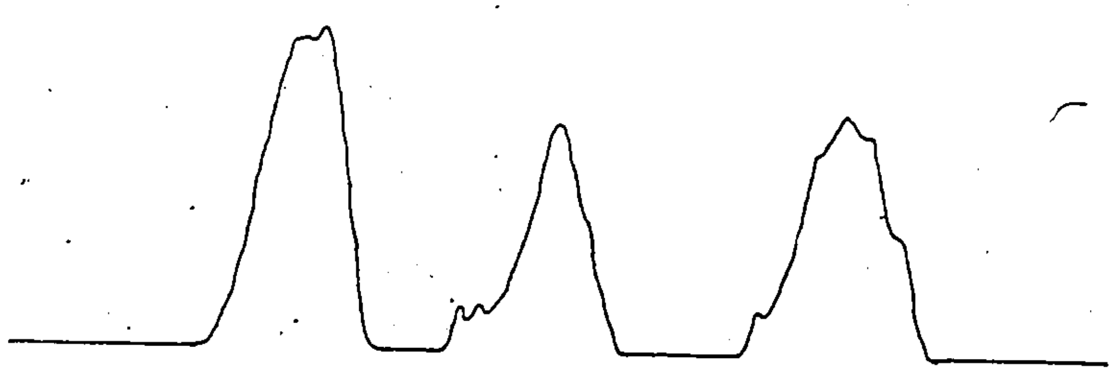

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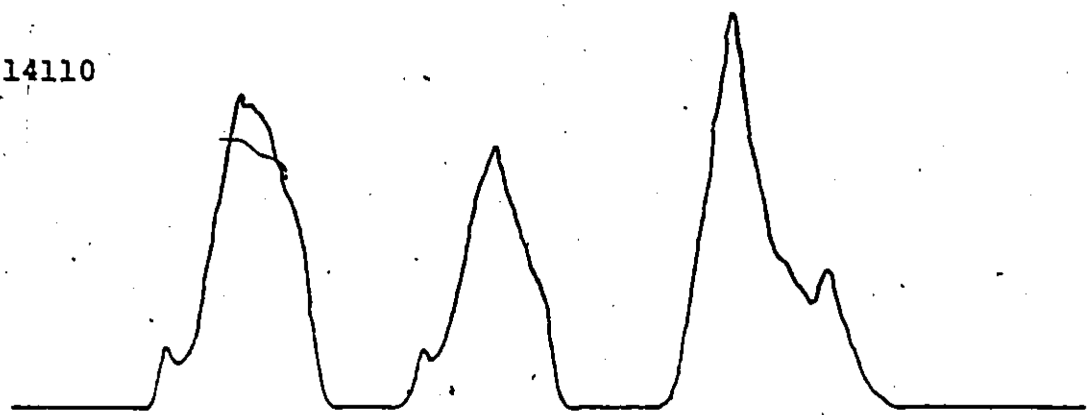

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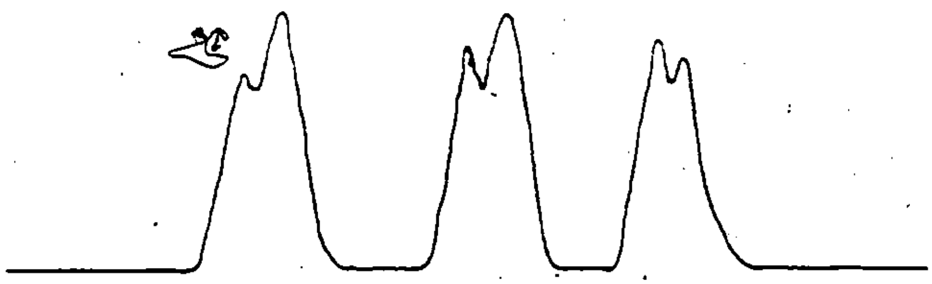

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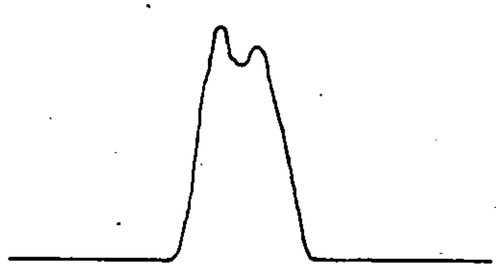

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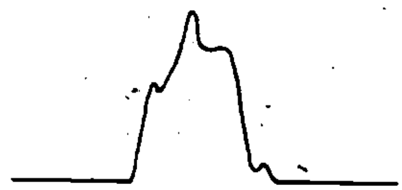
H18091



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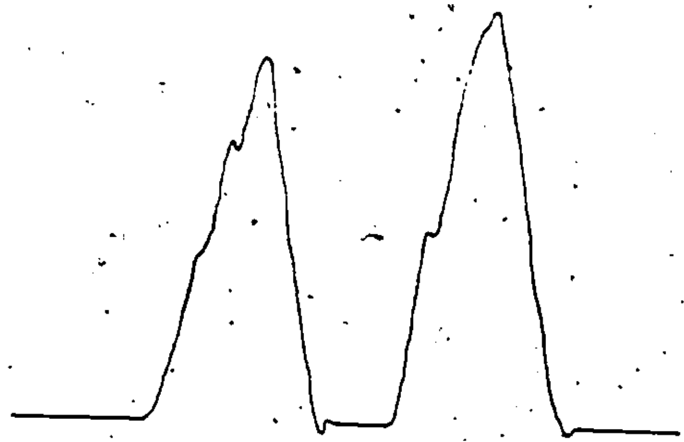

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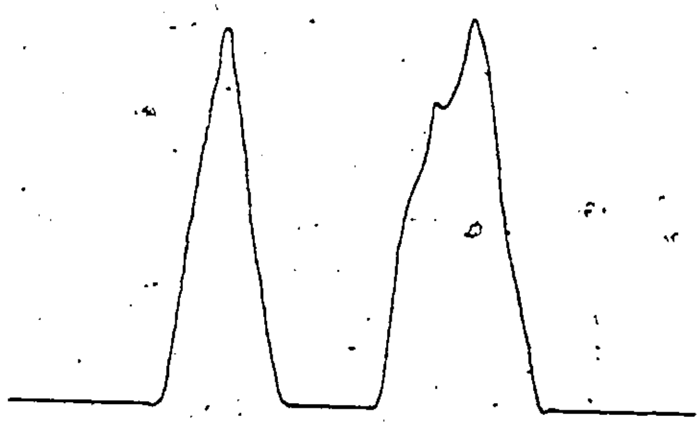

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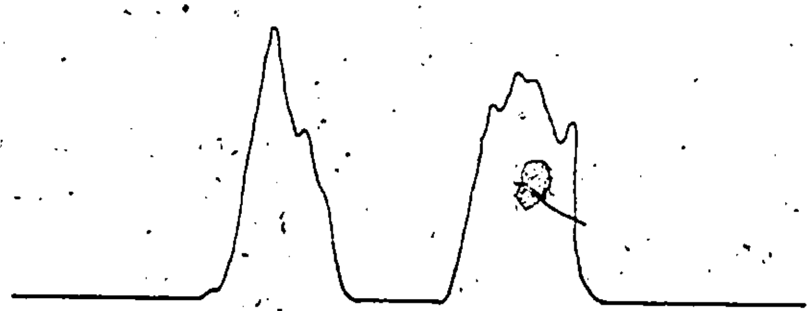


Subject D, who appeared to plant the crutches with tremendous force on the film recorded peak forces of approximately 27.3kgf per crutch or 61% of body weight when using 4 cm short crutches(102 cm). Therefore the sum of both crutch forces were approximately 120% body weight. This force was absorbed in the shoulder region of the body which is not anatomically suited for such force dissipation. Reduced peak forces were recorded when the subject was equipped with fitted crutches of 106cm. The magnitude of the reduced force was approximately 13.6 kgf which was certainly less than half of body weight. Therefore, the subject did not support his full body weight on the crutches during the double crutch support phase. Film analysis suggested the subject, being heavy in stature was unable to support his weight on crutches comfortably. To compensate, he used a swing-to gait style and relied on leg power to give the body mass vertical momentum to carry him through to the next heel stike. However, to compare this occurance with the short crutch trial (102 cm), a double peak can be observed indicating increased axial crutch force just prior to heel strike. Such an indication would suggest that this subject (D) was better able to support his body weight with the shorter crutches (102 cm) as opposed to the fitted crutches of (106 cm).

There are axial force recordings of trials with short, fitted, and long crutches with subject E. With this subject, the smallest peak amplitudes occured during the short, and fitted crutch trials. Thus, the longer crutches caused slightly larger forces to be experienced in the shoulder region due to the opposing action of the ground force. The pattern of force loading did not indicate any dissimilar characteristics between trials.

Subject F used fitted crutches and displayed similar force loading characteristics during each cycle of this trial. Between cycle peak amplitudes were similar with a distinct double peak identifying crutch strike and crutch push-off explicitly from the intermediate period of double crutch support. This trial consisted of swing-to gait so that crutch push-off could not be conceptualized on similar terms as previous swing-through trials. Rather, the pushoff peak force indicated a braking or deceleration of the body mass just prior to H-S.

The force traces of subject G showed smaller amplitudes during the fitted crutch trial(99cm) as opposed to the short crutch trial(91cm). The relative duration of time spent in double crutch support, when combined with the long distance per stride travelled, indicated the high velocity with which the subject walked.

The most noticable differance between the traces of subject H using short (91 cm) and fitted crutches (99 cm) was the discrepancy in magnitude. The peak force was 50% smaller when short crutches were used compared to the fitted crutches. During the short crutch trial, the axillary pad of the crutch was not sufficiently anchored as fore-aft slipping occurred. This may have resulted in the lower (50%) force application of the subject as one would be reluctant to apply force to an unstable support. During the fitted crutch trial, much more of the body weight was supported by the crutches which allowed the subject to perform a greater swing-through.

Subject I demonstrated visually smooth gait in both the fitted crutch (99cm) and the long crutch (104cm) trial. However, decreased axial forces were recorded during the long crutch trial. At this point the cause is unaccounted for but perhaps by combining this information with the other data collected in this study, a feasable solution will be found.

In most cases the greater peak axial forces were recorded in trials where the crutches were of the correct length, however, this was not always the case. It would stand to reason that ideal peak force would be slightly larger than 50 % of body weight per crutch. This peak force would most likely occur just before the midpoint of the support phase which was observed in most traces. Two subjects exhibited the double peaked traces of high axial loads at crutch-strike and crutch pushoff. While one subject had problems supporting the body weight with the upper body musculature, both subjects, exhibiting double peaks, preferred a swing-to style of gait. All other trails were of swing-through style by the subjects choice. Absence of adult data on axial crutch forces foiled the possibility of relative child-adult comparison.

4.6 ANALYSIS OF MECHANICAL WORK

Work is performed when the body moves as the result of muscular moments. As reviewed in chapter II, the work value which includes energy transfers both within and between segments is the absolute sum of both positive and negative energy variations in all segments of the body. Using the theory outlined in chapter II, values for work assuming energy transfer both between and within segments(Wwb), no energy transfers(Wn), transfers within segments only(Ww), and transfers between segments only(Wb), were calculated. The magnitudes of the energy transfers within and between segments, within segments only, and between segments only were also calculated and are tabulated in Table 4. Observing the work values per stride, one must consider the speed of ambulation or the time taken per stride. Note that work values assuming transfers within and between segments are the lowest work measurements due to the assumption resulting in the recycling of energy. Body mass also effects work values substantially. Therefore the elapsed time per stride was calculated in addition to the distance travelled. Wells(1979) utilized this procedure with adults which would enable adult-child comparisons. Consequently, the work values in joules/kg of body weight and distance/meter travelled were calculated as indicators of total gait efficiency. The results of this 'normalized' work value accounting for energy transfers both within and between segments are displayed in Table 5.

Observing Table 4, subject B exhibited a relatively low Wwb value while transfering relatively large amounts of energy (Et w/b) in trial B01102. The more energy transfered, the greater the energy in the system, resulting in decreases of required muscular moments for energy generation in future gait cycles. Thus subject B appears to be quite efficient based on total work values. Table 5 displays the magnitude of mechanical energy per stride in J/kg/m which permits the subjects to be compared regardless of body weight or distance travelled in one stride. Once again, subject B showed the relatively lowest energy output during crutch gait.

Subject C displayed a large discrepancy between a trial using 102cm crutches and one using 106cm crutches. Although the 102cm crutches are of the proper length, according to Cohen(1979), the work output (Table 5) was much greater than when the crutches were too long at 106cm. During the 102cm trial, subject C was not successful in anchoring the axillary pad of the crutch. However, when 106cm crutches were

TABLE 4

WORK VALUES

TRIAL	W(w/b)	W(0)	W(w)	W(b)	ET(w/b)	ET(w)	ET(b)
B01102	29.9	94.3	49.4	74.4	64.7	44.9	19.8
B06106	47.9	105.8	67.6	86.1	57.8	38.2	19.6
C02102	63.3	97.4	74.2	86.5	34.1	23.3	10.8
C07106	63.9	104.2	76.1	91.9	40.3	28.1	12.3
D03102	66.8	136.8	83.5	120.1	69.9	53.3	16.7
D08106	72.7	125.4	87.3	110.8	42.7	38.1	14.5
D12110	76.4	147.3	103.3	120.3	70.8	43.9	26.9
E04102	36.5	58.3	42.1	52.7	21.7	16.2	5.5
E09106	30.9	55.4	39.8	46.5	24.5	15.6	8.9
E14110	52.2	93.1	59.2	86.1	40.8	33.8	6.9
F05102	76.2	118.6	88.9	105.9	44.4	29.6	12.7
F11110	44.9	103.3	60.9	87.3	58.3	42.4	15.9
G16091	33.1	80.4	51.0	62.4	47.3	29.3	17.9
G20099	30.6	64.1	49.1	55.5	33.4	24.9	8.5
G24104	55.2	125.7	74.0	106.9	70.5	51.7	18.8
H21099	47.7	100.5	55.0	93.2	52.8	45.5	7.3
H25104	33.0	89.1	45.3	76.7	56.1	43.7	12.3
I19091	27.3	59.8	32.6	54.6	32.5	27.3	5.3
I22099	50.8	99.4	62.9	87.2	48.6	36.4	12.2
I26104	32.0	66.3	40.5	57.8	34.2	25.9	8.5

used, the swing-through range increased by 58.66% as seen by the distance covered, while 'normalized' energy decreased by 59%. Therefore, even though the crutches were technically 4cm too long, the energy output was smaller due to the large swing-through.

TABLE 5

NORMALIZED WORK VALUES

TRIAL	MECHANICAL ENERGY/STRIDE	WORK (w/b)	Kgf's	m's
B01102	1.14 J/kg/m	29.90	35.38	.733
B06106	1.40	47.97		.962
C02102	3.05	63.36	35.45	.586
C07106	1.80	63.91		.999
D03102	2.57	66.89	44.32	.586
D08106	4.08	72.74		.402
D12110	3.10	76.43		.556
E04102	2.95	36.55	37.73	.328
E09106	2.58	30.92		.317
E14110	2.29	52.28		.604
F05102	3.17	76.24	35.68	.674
F11110	2.39	44.88		.527
G16091	1.21	33.10	29.32	.932
G20099	1.12	30.63		.936
G24104	1.73	55.21		1.090
H21099	2.02	47.71	27.10	.870
H25104	1.27	33.03		.962
I19091	1.12	27.35	33.34	.540
I22099	2.07	50.80		.735
I26104	1.56	32.05		.617
Means	2.13			
S.D.	.845			

Subject D, being a heavy male, demonstrated difficulty in supporting his body weight on crutches which resulted in a swing-to gait. Swing-through gait requires sufficient upper torso strength. The gait pattern was very jerky with no fluidity to the motion. This combined with a large body mass, and small distance covered per stride, resulted in the large normalized work value.

Subject E demonstrated low Wwb values (Table 5) however, was unable to cover much distance per stride despite a fluid gait. This resulted in relatively high normalized work values.

Subject F expended quite large amounts of energy with crutches of 102cm. These crutches were grossly undersized which resulted in failure to properly anchor the axillary pads, hence an unsteady gait requiring high energy outputs. Work output was decreased somewhat with 110cm crutches.

The gait of subject G was interesting to explain in terms of energy output. The apparently low energy outputs were due to the extremely large distances covered per stride. Despite the fact that this subject was the shortest in stature, she exhibited a fast gait, with the greatest swing-through. However, in qualitative film analysis, the subject was so keen to ambulate quickly in a straight line, control was sacrificed by overstriding which caused gait breakdown. The subject then had to recover and once again resume forward movement.

Subjects H and I demonstrated fluid gaits. It is unusual that subject I had a higher work output when using fitted crutches of 99cm as no explanation can be found at this stage of analysis.

To generally review the normalized work outputs of all subjects dependent upon crutch lengths, there appears no steadfast consistancies or trends. Crutch length was not critical in producing overall lower energy outputs. It appeared from the normalized energy output data, that individual gait style which includes body weight and distance covered per stride, was the determining factor in energy output. However, as was explained, low energy output may be at the expense of balance and control. Wells (1979) found that energy expenditures of adult crutch ambulators compared quite closely to those of adults walking normally. The values of normalized energy expended in the present study involving children are much greater than the values obtained of adults by Wells (1979). As the energy expenditure was normalized to subject weight and distance travelled, children may be at a relative mechanical disadvantage. For a given distance, a child would have to perform more strides than an adult due to smaller limb lengths. From an internal viewpoint, it may be more costly for a child to undergo two strides to every one adult stride even though adults require larger muscle moments to compensate for the relatively larger masses and inertias involved. These are two possible explanations for the large energy expenditure values observed in these children. Perhaps if the gait was analysed under different conditions, trends may appear in normalized work outputs.

4.7 ANALYSIS OF ENERGY TRANSFERS

The major component of this study was the involvement of instantaneous energy and instantaneous power in childrens' crutch gait. Energies and powers were calculated and plotted with, each trace beginning and concluding with heel-strike(H-S). Individual interpretation of the curves improved with experience. Before a trial by trial analysis is conducted, some general findings should be noted in order that they be applied during analysis. To recapitulate, energy transfers are fundamental to mechanical efficiency in gait. If energy can be transfered, either from one form to another within a segment, or between segments, rather than lost from the system, then the work cost will be decreased. In other words, the objective in obtaining mechanical efficiency in gait would enable the body in motion to be as energy conservative as possible. To become a conservative system, energy must be transfered and maintained in other energy forms and/or in different locations in the body. Instantaneous energy curves illustrate within segment transfers of energy from one form to another. Instantaneous power curves exemplify between-segment transfers of energy when traces of adjacent body segments are compared. As was previously mentioned, analysis of the individual trials will be of the 9-segment model. Peak powers of the H-N-T segment render information that will be explained in the following sections. Table 6 shows the magnitude of the peak positive and negative powers for the H-N-T segment.

TABLE 6

PEAK POWERS OF THE H-N-T SEGMENT

Trial	Peak H-N-T Power (Watts)	
B01102	+13	-13
B16106	+18	-38
C02102	+32	-17
C07106	+31	-31
D03102	+60	-33
D08106	+61	-50
D12110	+49	-79
E04102	+39	-14
E09106	+18	-19
E14110	+34	-27
F05102	+56	-49
F11110	+27	-33
G16091	+33	-19
G20099	+39	-31
G24104	+42	-42
H21099	+20	-19
H25104	+18	-14
I19091	+14	-9
I22099	+39	-38
I26104	+25	-29

4.7.1 Subject B

Subject B was a male who used 102cm crutches in the first trial. The energy traces for the H-N-T segment typify the transfer of energy within a segment. The potential energy (PE) and kinetic energy (KE) traces have peaks that are inverse reflections of each other. As one curve experiences a positive peak, the other falls into a valley. Since the rotational energy trace remained essentially zero, there are transfers of energy between PE and KE. This demonstrates the inverse pendulum nature of the trunk during crutch gait. When the crutch tips are planted, the linear kinetic energy of the H-N-T is maximal since forward velocity is at a simultaneous positive peak. As the body weight becomes increasingly supported by the crutch, it gains height at the expense of forward velocity as the crutches act similar to a pole for a pole vaulter. Thus, KE is transferred into PE with total energy (TE) remaining essentially constant. Once the H-N-T mass is over the crutches(ie, the crutches are in a vertical orientation), then the H-N-T 'falls' down from the crutches heading toward H-S. This results in a PE transfer back to KE by decreasing the height of the centre of mass and increaseing forward velocity, hence increasing KE. This senerio of events is evident in the H-N-T energy traces (Figure 3).

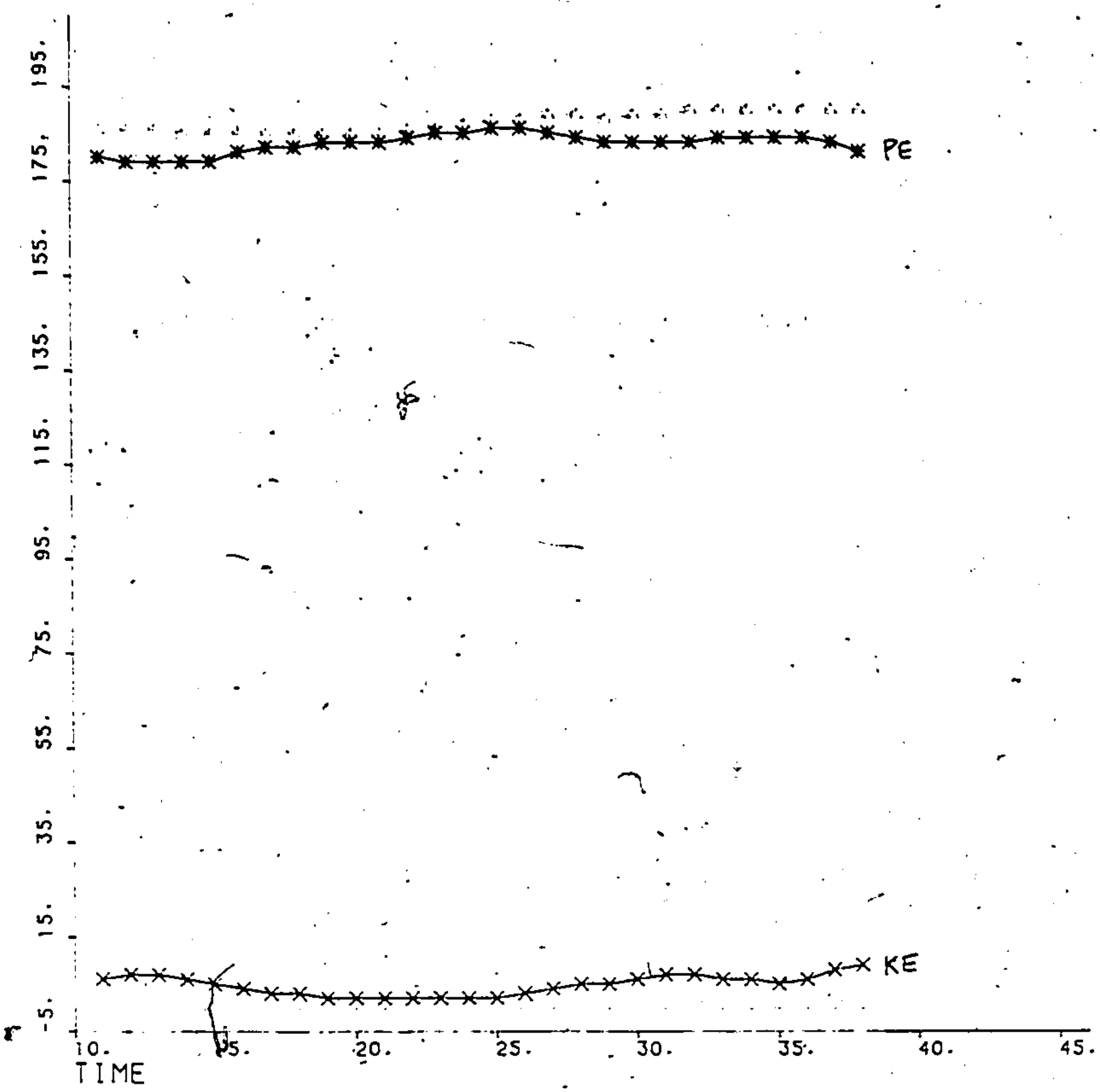


Figure 3: H-N-T BC1102

Energy traces of the arm-crutch and thigh segments infer there are no transfers of energy within these segments. As PE increases, so does KE, resulting in a phasic relationship as the arms and thighs gain height and velocity simultaneously. This suggests that the energy created by muscular moments in these segments is not conserved within the segment, however, the energy may be conserved elsewhere in adjacent segments. Analysis of the instantaneous power curves should reveal if energy is transfered and conserved in the neighbouring trunk. Figure 4 illustrates the increasing power of the arms while the H-N-T experiences decreasing power and vice versa within the region of crutch-off to crutch-strike.

Winter(1979B) demonstrated the transfer of energy from regions of high power to segments of low power at the same instant in time. Thus, while the energy created in the arms is not conserved in other forms within the segment, partial conservation is likely achieved and facilitated by the transfer of energy to the H-N-T segment. In a subsequent trial of subject B using crutches of 106cm, which were 4cm too long, it is interesting to note the differences in peak powers. Comparing the powers of the H-N-T from the two trials, peak powers of 14w(102cm) and 38w(106cm), depict the difference that unfitted, long crutches make to the power output of the gait. While most trials demonstrate close symmetry between right and left arms, the 106cm trial does

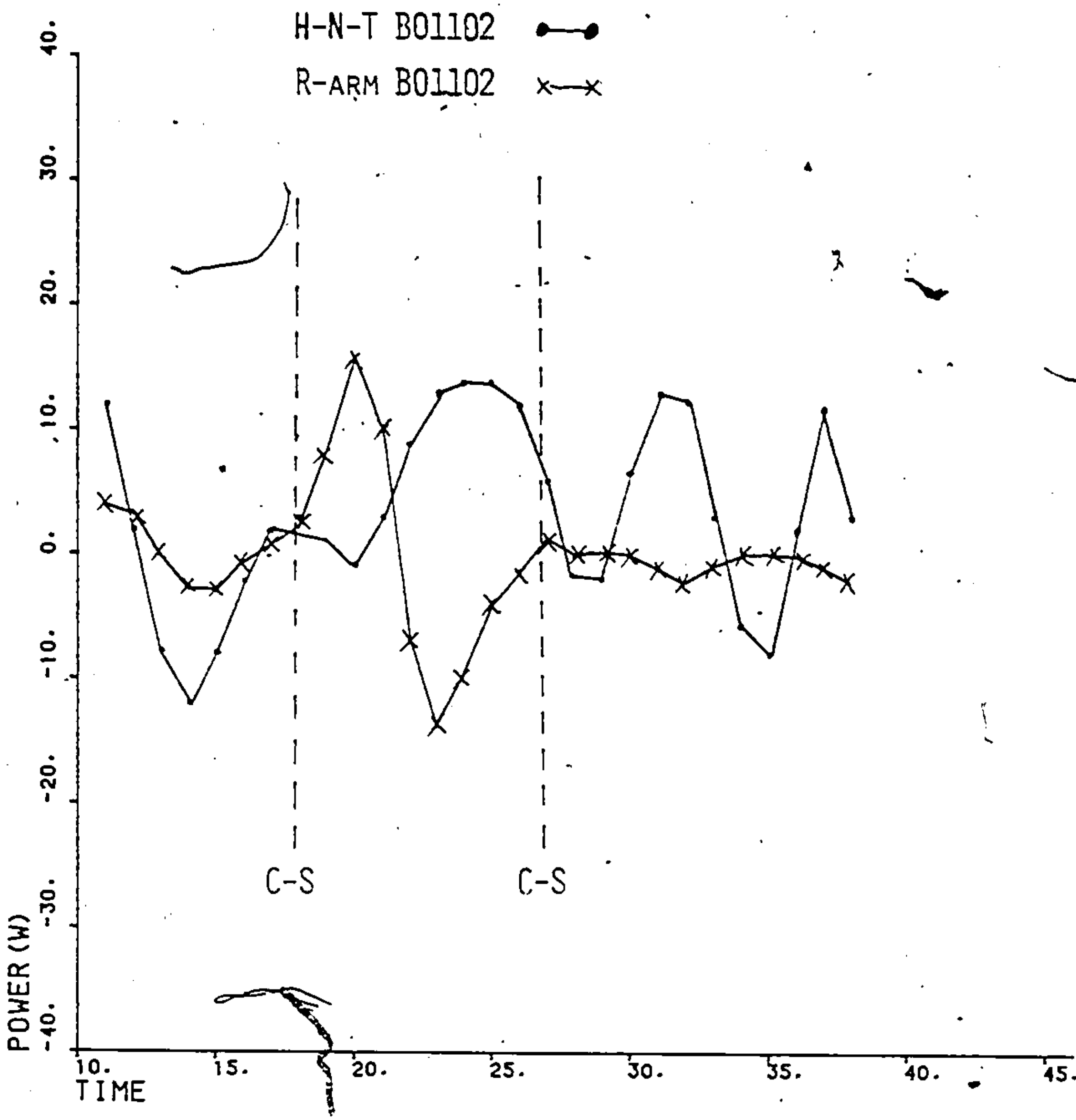


Figure 4: Arm and H-N-T Powers B01102

not. This established subject B's control problems with his arms when the crutches were too long as illustrated in Figure 5.

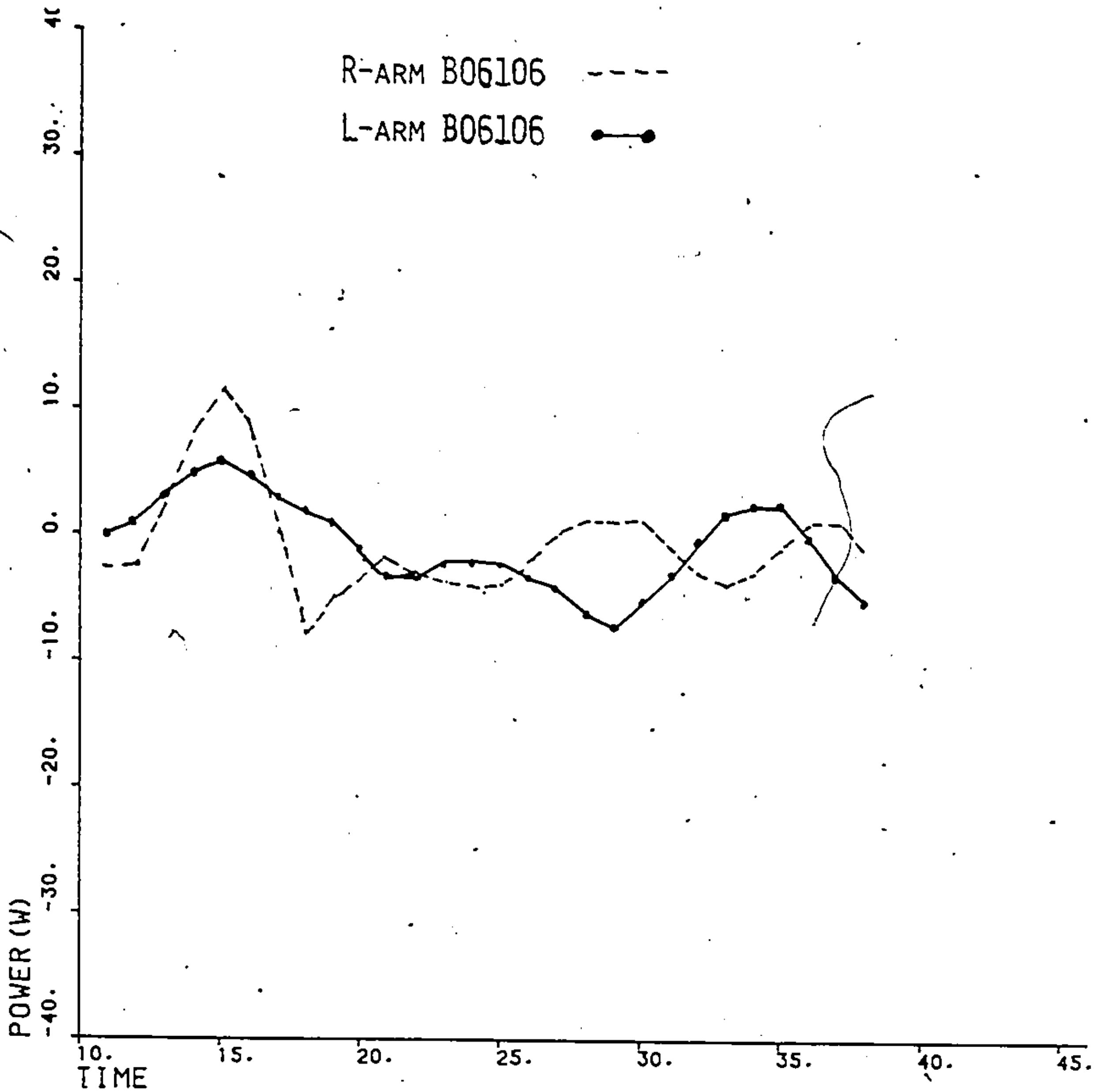


Figure 5: Right vs Left Arm B06106.

4.7.2 Subject C

Subject C was equipped with fitted crutches of 102cm during the first trial. Features worth noting were the energy traces of the thighs (Figure 6).
There was a large change in KE just after H-S which was unusual. This was caused by a very intense foot strike where the leg completely ceased motion shortly after the H-S event. The resulting jerky gait is reflected by the huge negative power output shortly after H-S.

In the C07106 trial, long crutches (by 4cm), were used. If the peak powers are compared between the two trials of the H-N-T segment, they will be noted to possess comparable magnitudes, however, the 106cm trial power curve is smooth which indicates large energy changes that are fewer in number than the 102cm trial. On film, the 106cm trial consists of a large swing-through giving rise to the observed instantaneous powers. Both legs, although different in their orientation in space, translate and rotate in unison which results in nearly identical power curves (Figure 7).
When two legs move in such a similar manner, in essence, their masses are combined into one which would increase the effect of momentum (mass x velocity). This may explain the large swing-through observed in this trial.

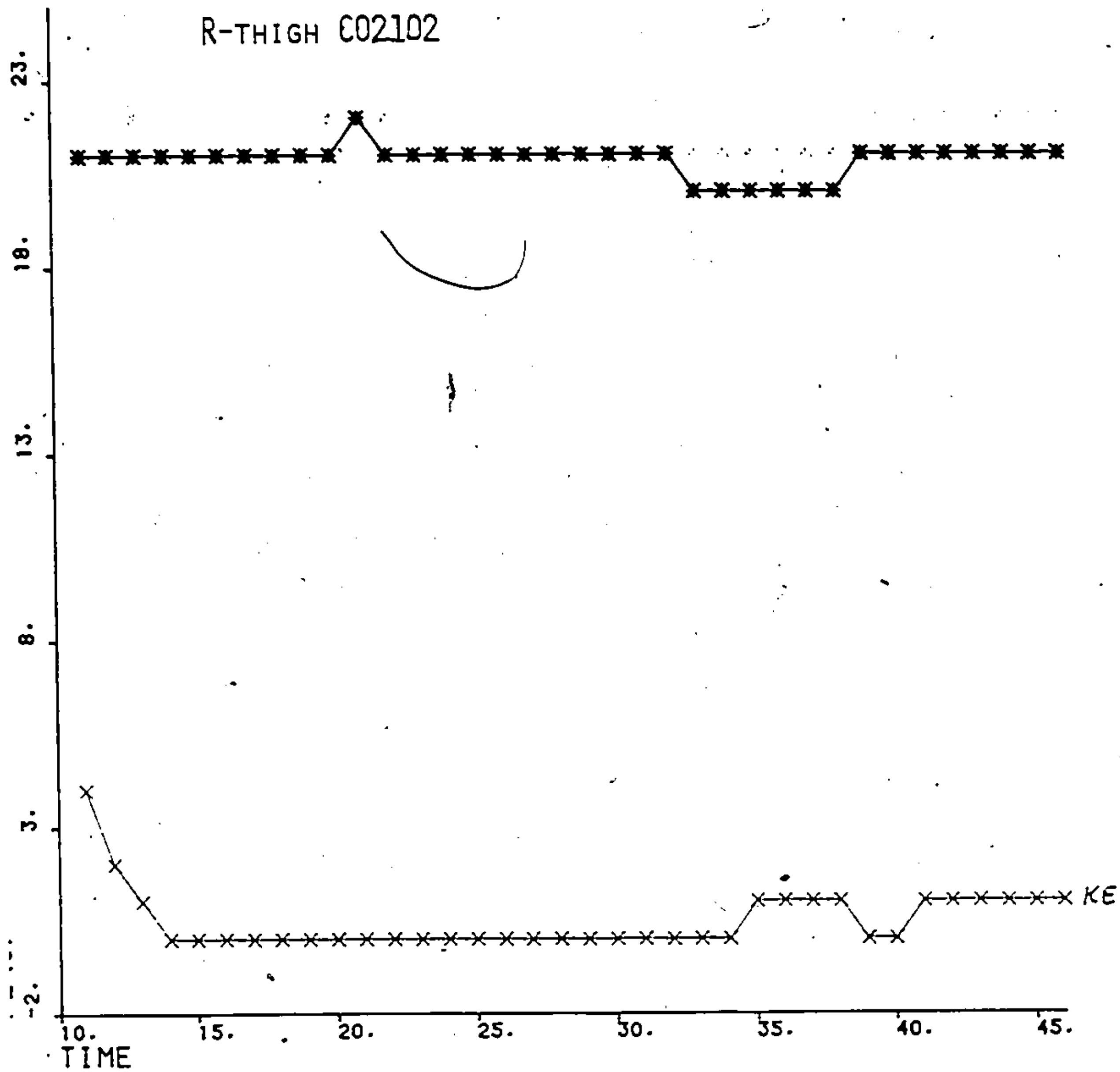


Figure 6: Energies of R-Thigh C02102

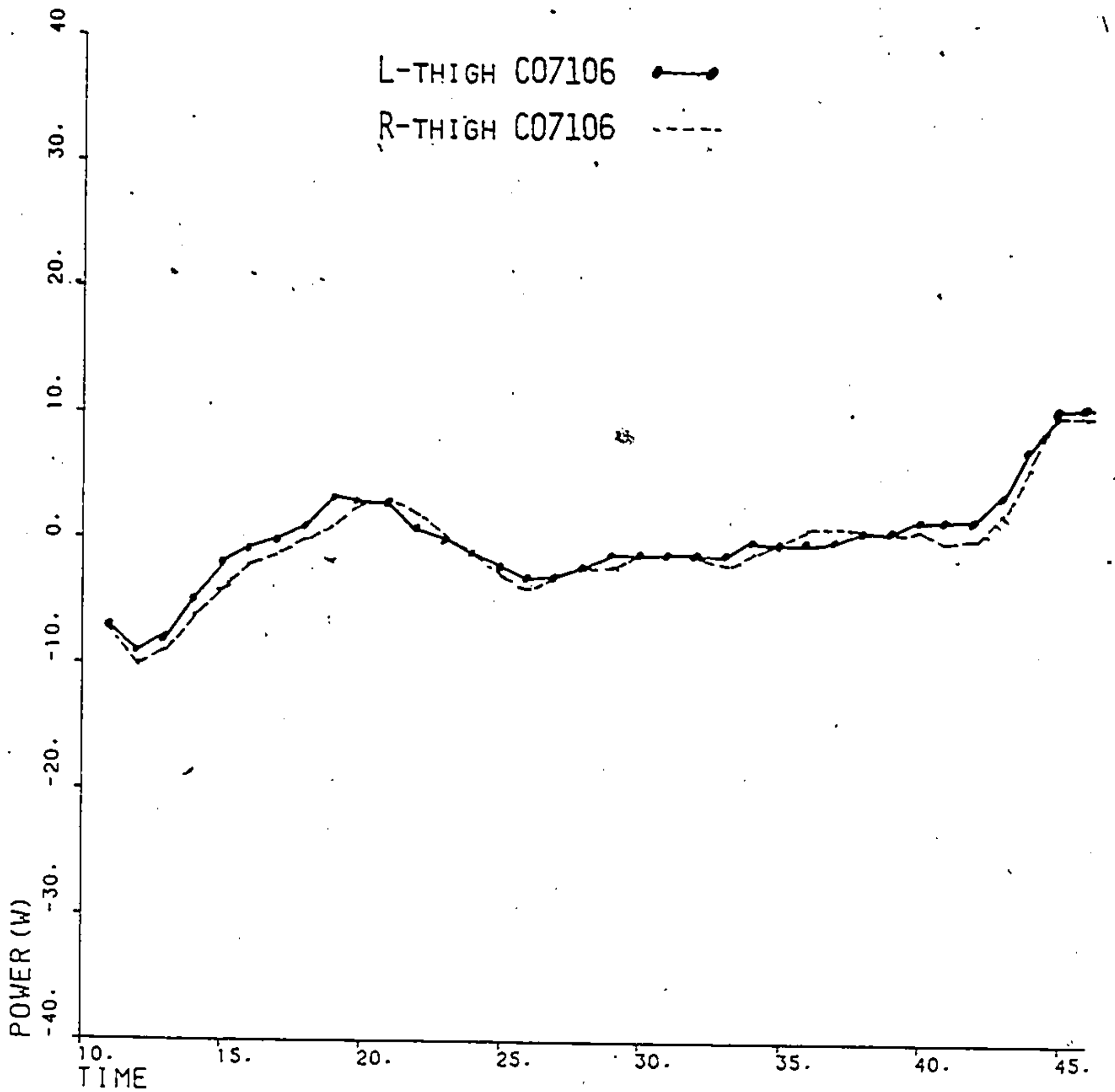


Figure 7: R vs L Thigh Powers C07106

4.7.3 Subject D

Trials of subject D were recorded, using 102cm(6cm too short), 106cm (correct length), and 110cm(4cm too long) crutches. Energy transfers between KE and PE in the H-N-T segment are readily observed in the energy traces for the 102cm crutch trial. However, pathological energy traces are observed in the arm- crutch segments (Figure 8).
It was noted that it is normal for peaks of instantaneous PE and KE to occur simultaneously in the arms during crutch gait. In this trial, the peak of KE occurs after the PE peak inferring some abnormal movement. Film analysis confirmed the cause as from the subject raising the crutches anteriorly and then proceeding to 'fall' forward onto the crutch resulting in large impacts upon crutch strike. Subject D was visibly overweight and demonstrated difficulty in supporting his body weight on crutches with his upper musculature. His gait was swing-to as opposed to the conventional swing-through due to his inability to support the body weight for the required duration of time for swing-through. The resulting 'choppy' gait of large impacts upon crutch and heel strike is illustrated in the power curves. As seen in Figure 9, the peak in the power curve of the H-N-T segment occurs very late in the H-S to H-S cycle.

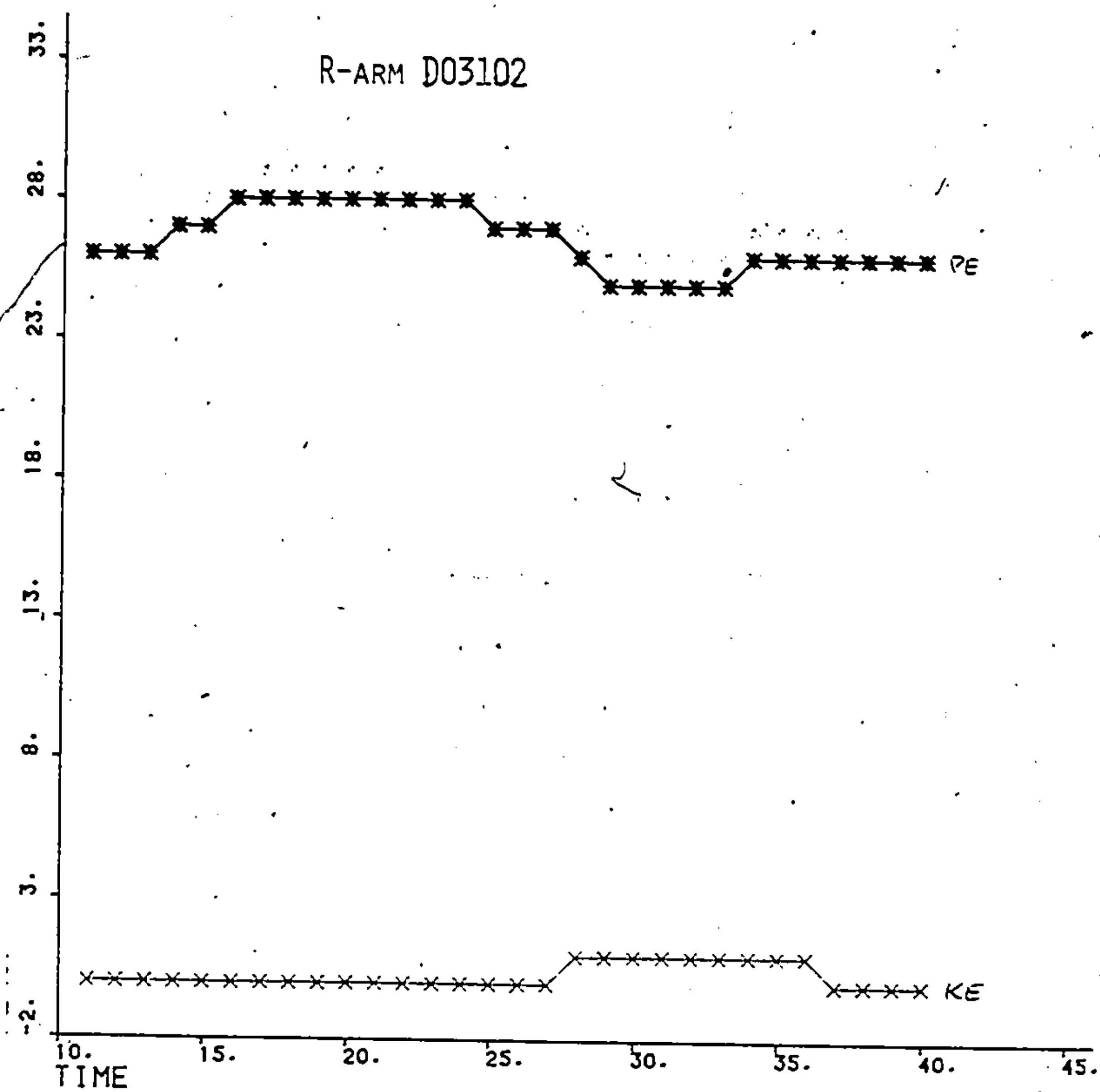


Figure 8: Energies of R-Arm D03102

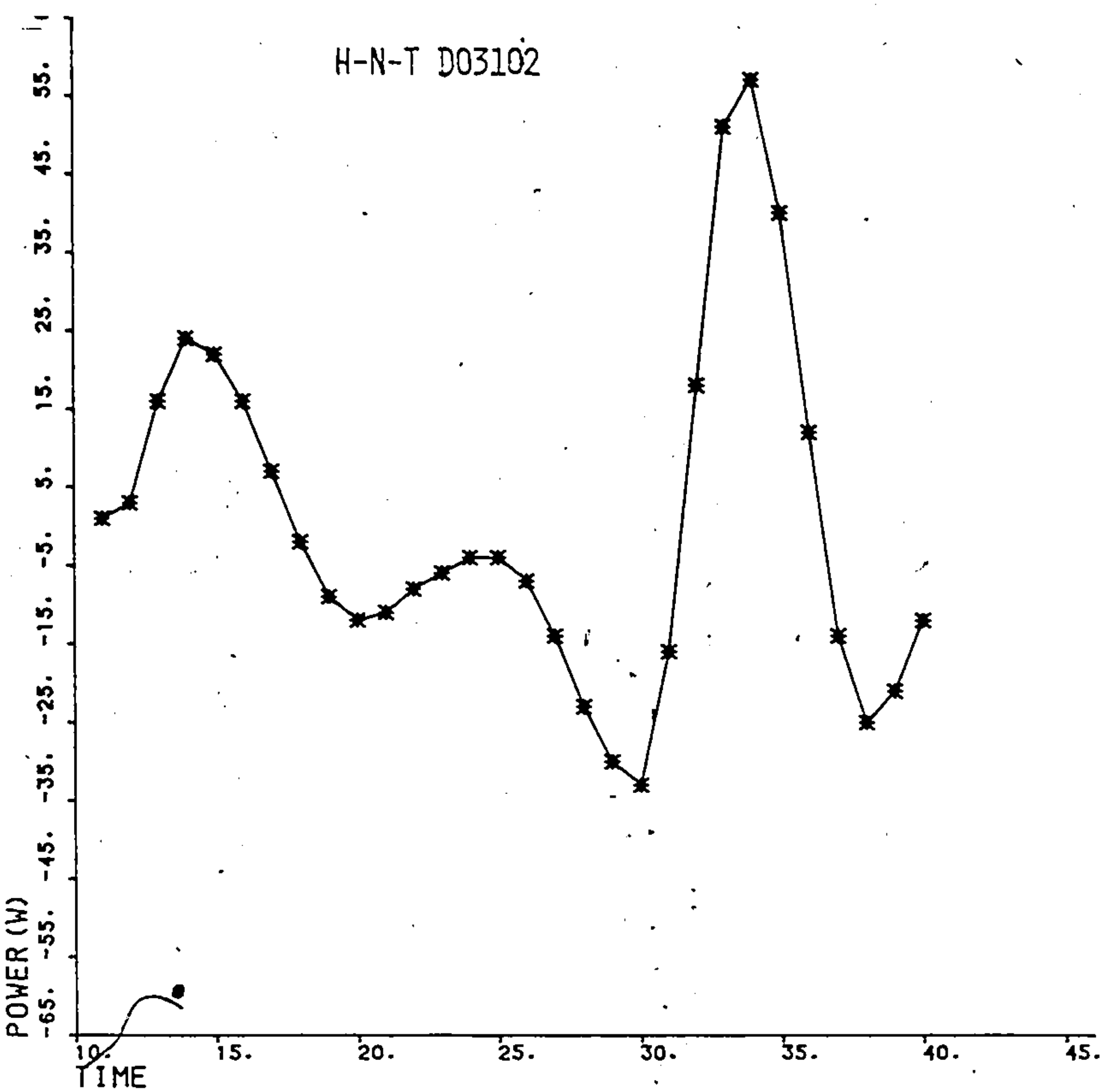


Figure 9: H-N-T Powers D03102

This peak is due to the large amount of energy absorbed by the upper trunk upon crutch-strike, which actually caused a physical 'bounce' in the segment. The subject, unable to support his body weight adequately with his arms, placed his axillas on the axillary pads of the crutches. This resulted in the loss of the shock absorbing capacity of the elbows, transfering the unattenuated energy into the trunk via the shoulder region. When the subject used fitted crutches of 108cm, similar traits were noted.

In the trial with long crutches(110cm), an interesting phenomenon was encountered. The power curve for the H-N-T was not characteristic of the previous two trials involving the short and fitted crutches. The peak power occurred earlier in the cycle which conforms to the convential standards in crutch gait(Figure 10).

A possible explation is the subject placed the axillas directly on the axillary pads with the longer crutches replaced the free space that is usually between the two. This reduced the distance that the subject had to plant the crutches and hence reduced the shock wave of energy that was previuosly transmitted up the crutch shaft into the trunk. It was also interesting to note the early cutoff of negative power in the thighs, across all trials, due to the nature of swing-to gait as opposed to the continuation of swing characteristic of swing-through gait.

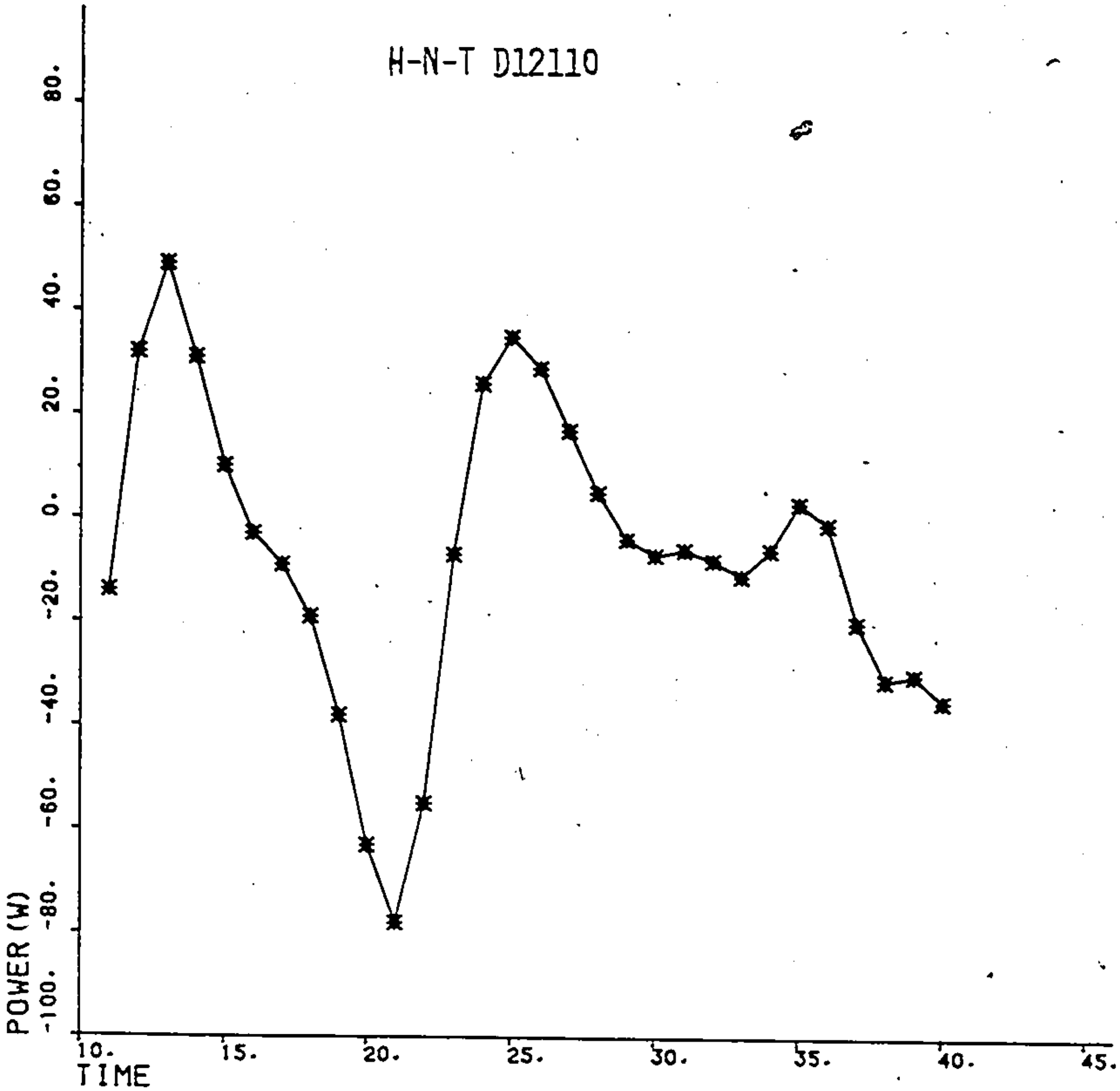


Figure 10: H-N-T Power D12110

4.7.4 Subject E

Trials of subject E were recorded with short(102cm), fitted (106cm), and long(110cm) crutches. The energy and power curves from the 102cm crutch trial were very smooth and did not infer any outright inefficiencies. The same was noted with fitted crutches. However, by comparing the power traces for the H-N-T segment over the three trials, differences are evident(Figure 11).

During the 102cm trial the peak power output is 40w. This output is reduced by 100% during the fitted crutch trial to approximately 20w. As crutch length increased beyond the optimal fitted length to 110cm, the peak power output escalated by 75% to 35w. Since the H-N-T segment contributes greatly to the total work output due to its relatively large mass, crutch length is critical, within centimeters, to the development of an efficient gait.

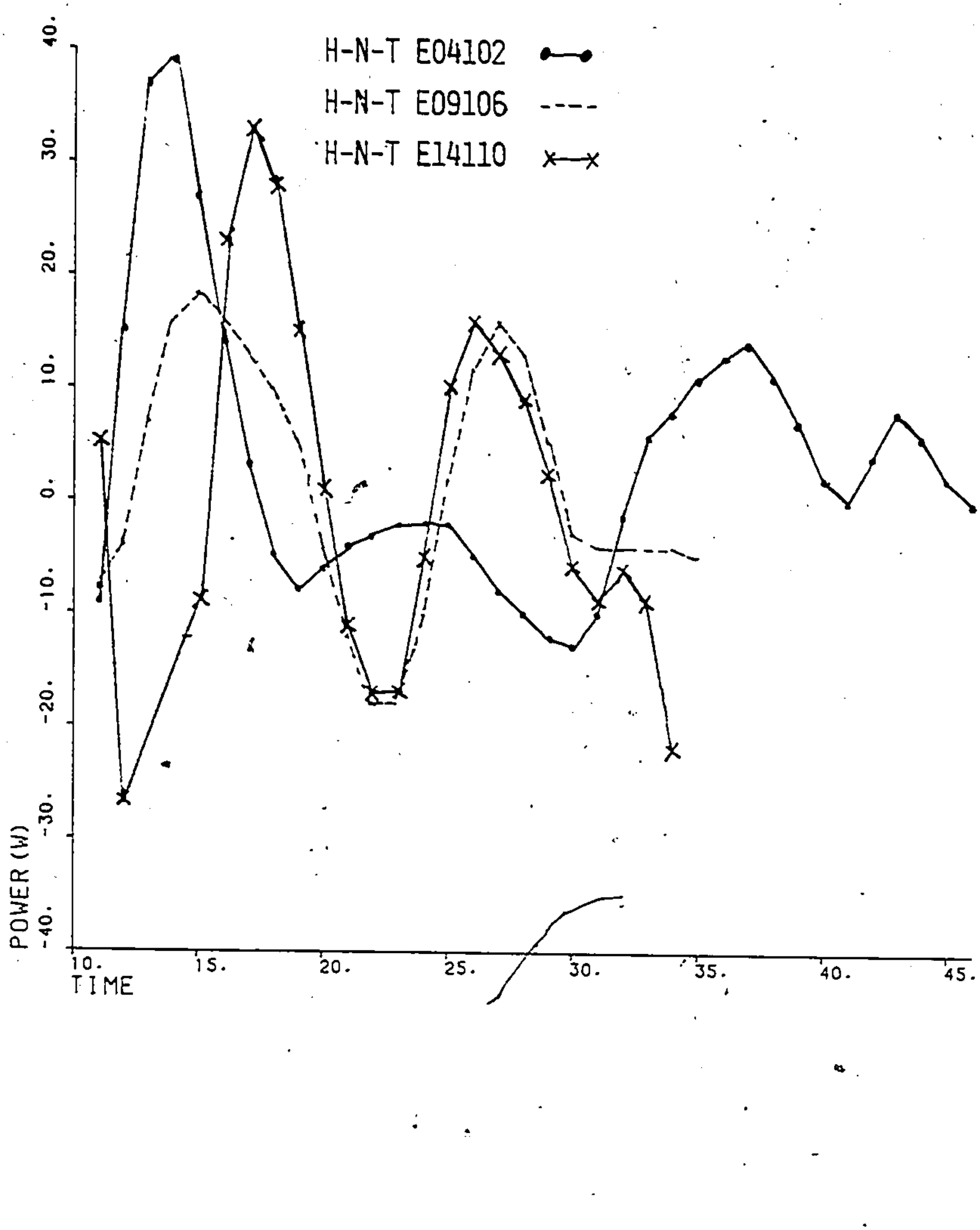


Figure 11: H-N-T Powers of E04102, E09106, E14110

4.7.5 Subject F

The trial of subject F equipped with 102cm crutches demonstrated there were no transfers of energy within the H-N-T segment." Subject F, being the tallest of all subjects, found the 102cm crutches to be much too short. The peak power output of the H-N-T segment was quite high at 56w in this trial. The power curves for the thighs were of abnormal shape and exhibited odd power peaks. Upon qualitative film analysis, it was observed that due to the grossly short crutches, the subject was unable to support his body weight and found it easier to combine a small 'hop' into the gait to carry the body through the swing-through phase. This resulted in bursts of power in the thighs giving rise to large peaks in the curves. (Figure 12)
The comparison of the left thigh power traces of subject F and the normal trace of subject B, aid in clarifying the difference in power patterns. The subject was then fitted with 110cm crutches and the peak power of the H-N-T was reduced by approximately 100% to 26w. Although the subject exhibited minor balance problems during this trial they were not manifested in the power and energy curves. This may be due to the inherent incapability of this 2-dimensional analysis method to record in the third dimension or side-to-side relative to the subject.

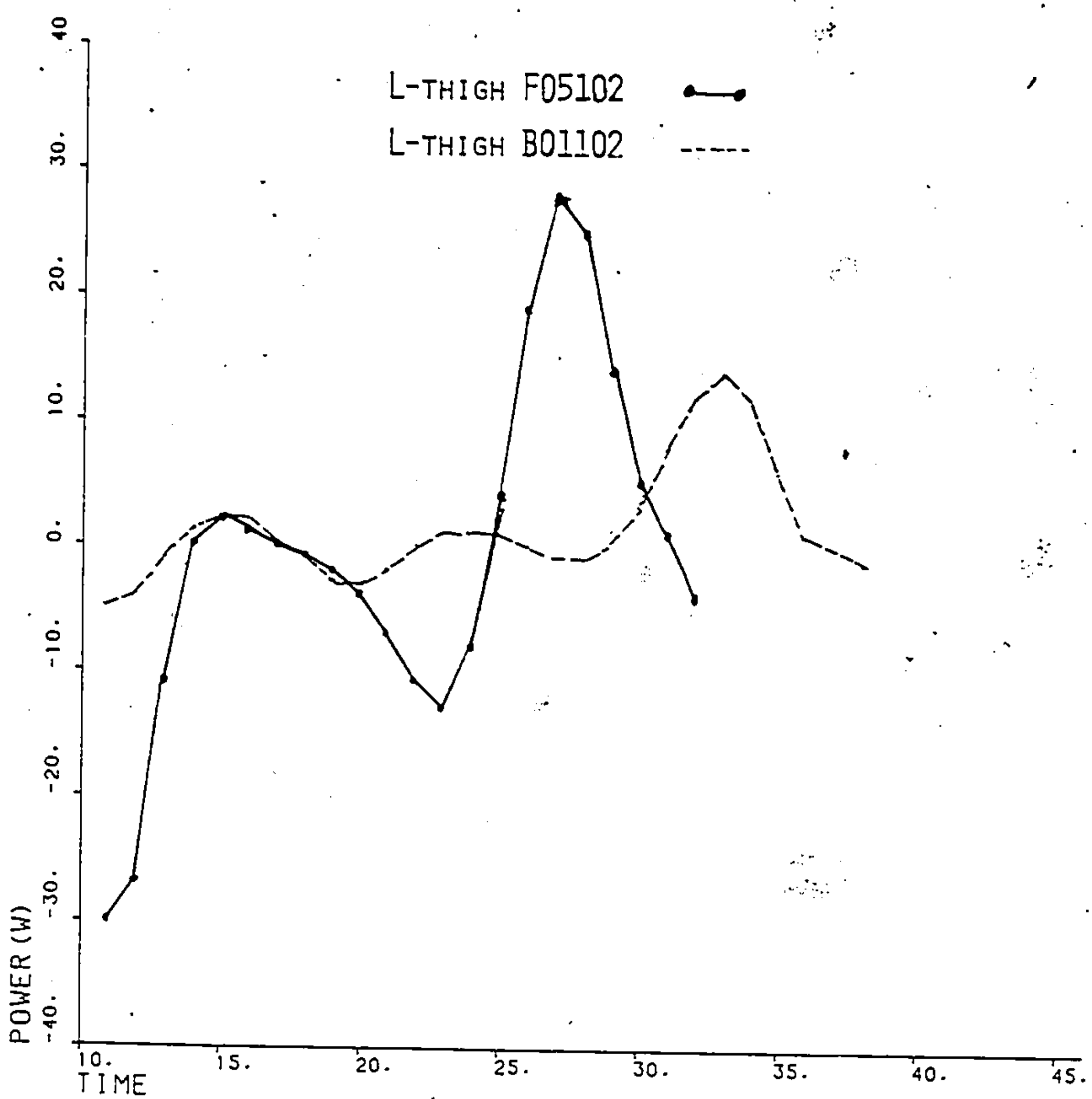


Figure 12: Powers of F05102 vs Normal Thigh

4.7.6 <u>Subject G</u>

The first female subject to be analysed was subject G. trial, using short crutches of 91cm, produced energy traces that were of short duration, thus indicating the gait cycle from H-S to H-S was carried out very quickly. The resultant high velocities accounted for the large peaks observed in the power traces. Since time is the denominator in the equation for instantaneous power calculation, a fast gait would register larger energy changes within a specified time resulting in a higher peak power. However, in this case, although high power peaks are noticed, the power curves themselves are smooth, indicating a 'smooth' gait. When the subject was equipped with fitted crutches of 99cm in length, the energy transfers within the H-N-T between PE and KE are very evident in the energy traces. The subject visually displayed this in her gait by hopping up on top of the crutches and riding swing-through and finally falling down off them just before heel-strike. She was able to manage this by keeping her arms straight, making herself an upside-down pendulum.

In the third trial, crutches were changed for a 104cm (5cm too long) pair that resulted in a higher peak power output of the H-N-T (42w compared to 38w in the fitted crutch trial). The gait was slower and more controlled which would have the effect of decreasing powers, hence, if the power increased, and the gait was slower, then the comparative power difference would be magnified by the virtue of a changing scale factor. As a result, any increase in peak powers would be comparatively larger than numerically indicated. It was intriguing to observe the foot contact in this trial. The subject was told to slow her gait, which she did for this third trial and her foot strike became a toe strike. It appeared she was very concerned with covering a large distance with every stride, since she was told to slow her gait. She maintained her attempt to achieve maximum distance from every stride by plantar flexing her foot and making ground contact with the ball of her foot. This observation is a postulate and evidence is not readily observable within the energy and power traces.

4.7.7 Subject H

The gait of subject H was very smooth and controlled, absent of any real abnormalities, using fitted 99cm crutches. When comparing the peak powers of the thighs (Figure 13) the right thigh peak occurs slightly before that of the left. This contrast is due to the injured right leg initiating an early swing-through to gain momentum which acts to pull the remaining mass through the swing.

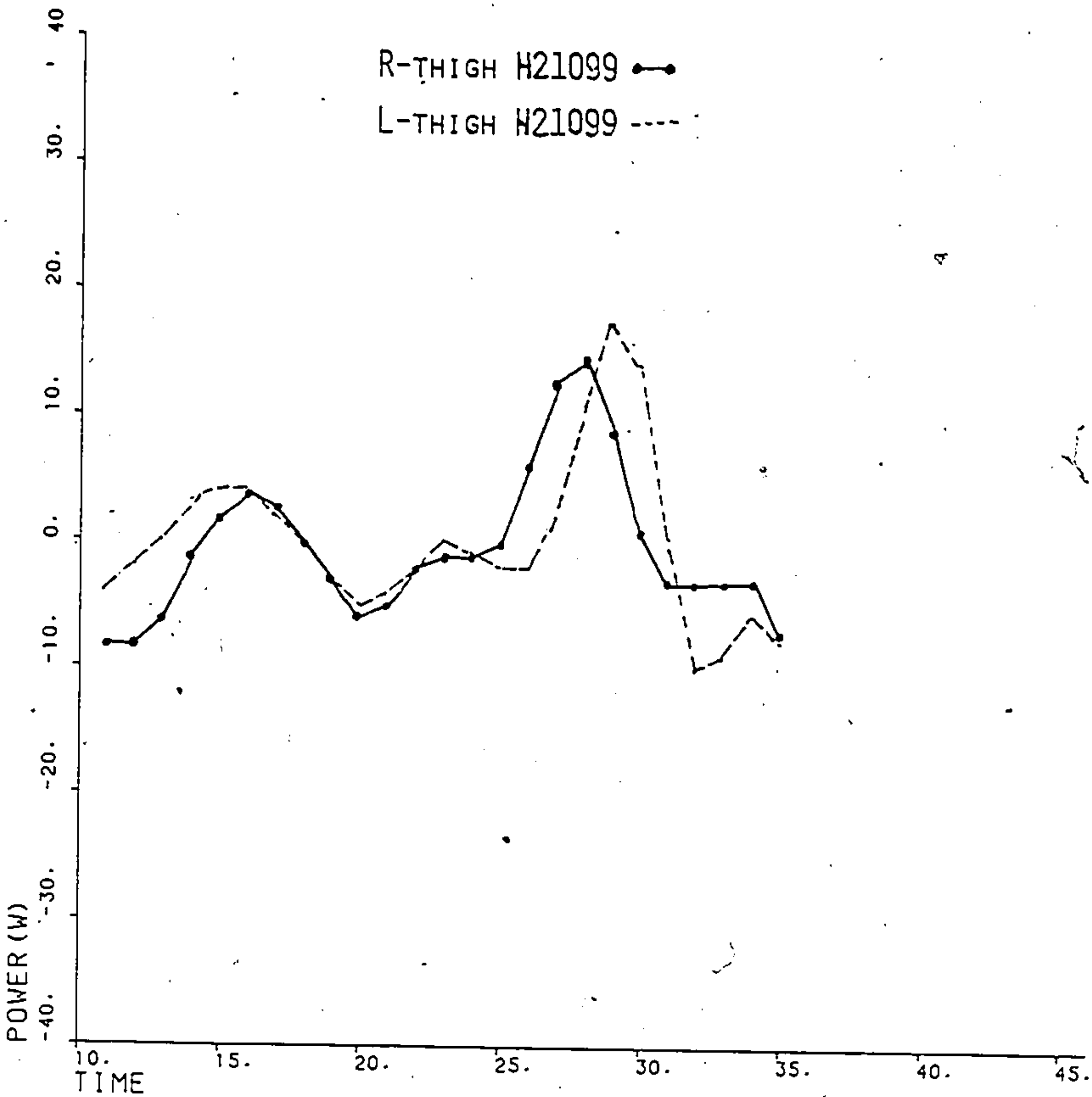


Figure 13: R vs L Thighs H21099

In a subsequent trial using 104cm crutches a comarison of the peak powers of the H-N-T established no magnitude differencs as both were in the neighbourhood of 20w. However, in one cycle, the 'injured' right leg swung-through out of line and collided with the supporting left leg and it was this cycle that was chosen for analysis to see the effect on instantaneous energy and power. By observing Figure 15, the power curve of the left shank is compared to an average power curve which highlights the positive slope just prior to H-S.

It appears that subject H had the unique ability to reach a point of maximal deceleration of the lower leg just before H-S whereby all other subjects reached a maximum deceleration at the instant of H-S. This would require additional antagonistic muscle moments which would result in a decreased striking impact. Further study would be required to analyse this mechanism for energy cost verses energy attenuation.

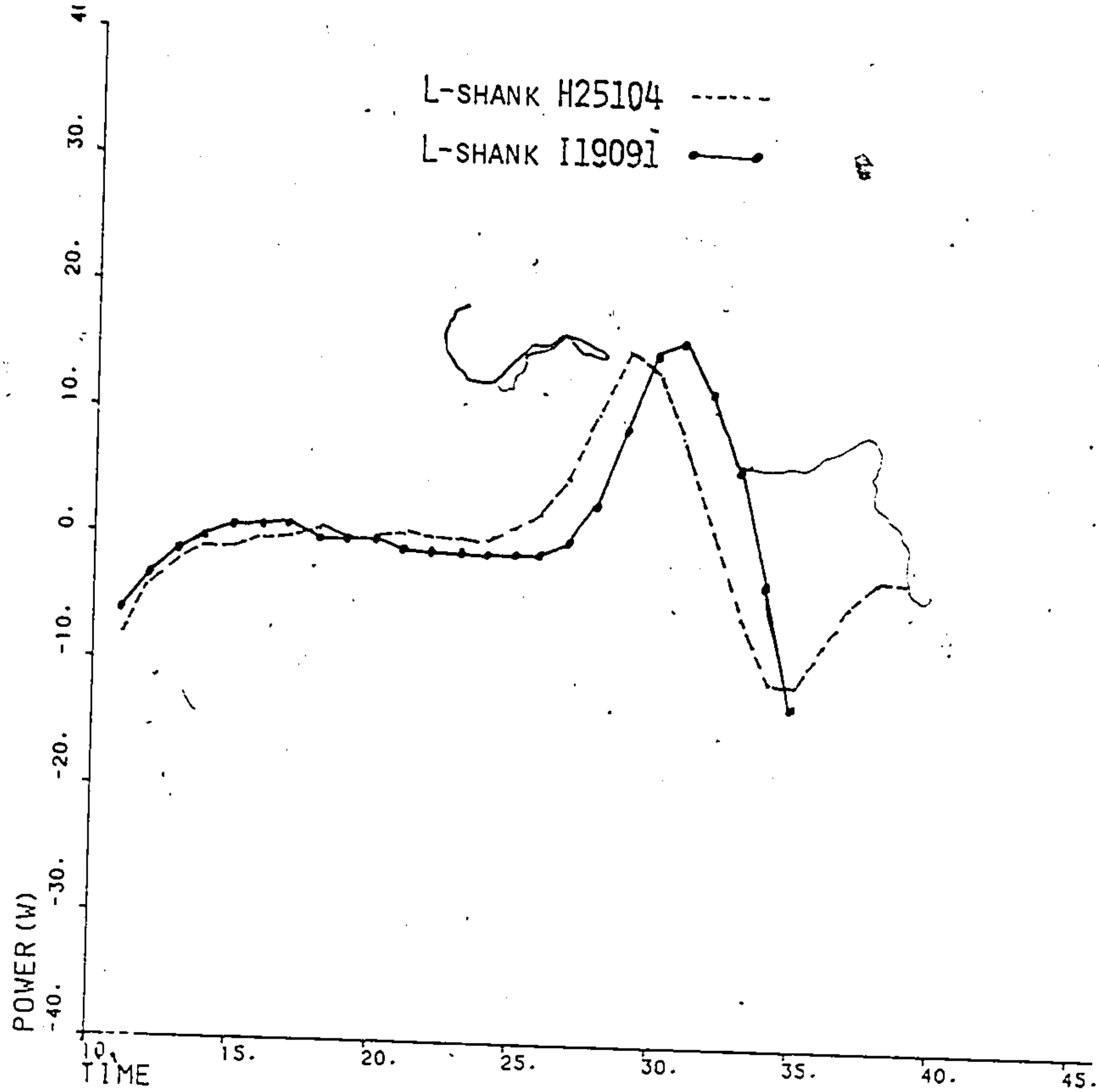


Figure 14: Powers of L-Shank H25104 vs Normal

4.7.8 Subject I

Subject I was a strong, athletic girl who was able to demonstrate good crutching skills over all trials. The first trial using short crutches of 91cm exemplified smooth, typical and symmetrical energy and power traces. The power peaks of the injured leg are small, averaging close to zero.

The second trial, using fitted crutches of 99cm, once again resulted in smooth energy and power traces with a peak power output of 38w in the H-N-T segment. The crutches were exchanged for long crutches of 104cm and the peak power in the H-N-T decreased to 28w. Technically, the crutches were too long but subject I managed an efficient style, with the head erect and looking forward, and the body weight borne on the arms with the elbow slightly bent for shock absorption. From subject I it appears that crutch length is not as critical in 'natural' crutch walkers as with children who have not developed high standards in body coordination, strength and balance.

Various mechanically inefficient mechanisms have been highlighted. Disruptions in gait are recorded in the energy and power curves for the researcher to analyse and interpret. It is hoped that this introduction into the interpretation of energy and power will aid in future gait analysis.

4.8 COMPARATIVE ANALYSIS SUMMARY

At this stage, it would be beneficial to summarize all components by a concurrent analysis and search for indicators of relationships between the comparative parameters.

Subject B displayed lower axial crutch forces, lower normalized energy/stride values, and lower peak power outputs using fitted crutches as oppossed to 4cm long crutches. From qualitative film analysis the subject had less difficulty in controlling the shorter crutches as the long ones tended to brush the ground during crutch swing-through.

Subject C appeared to perform more efficiently with slightly longer than normally fitted crutches. This is deducted from a decreased energy/stride value, and indications of smooth flow from crutch axial force, instantaneous energy and instataneous power traces when equipped with crutches of 106cm which were technically 4cm too long. The subject was able to gain a larger swing-through with the longer crutches while balance and problems of anchoring the axillary pad on the crutch were no longer evident.

It is questionable whether subject D should be prescribed crutches if he were incapacitated. Despite his age of 9 years, his large body mass hinders support by the upper body musculature. Large impact forces of both the foot and crutch cause visible shock waves (on film) through the body.

Injury to the shoulder joint or general upper limb and girdle musculature appears quite probable. Energy/stride values were the highest of all subjects. It would be recommended for this subject to gain upper body strength (which may develop with age) before assigning crutches in any treatment situation.

Subject E would benefit from fitted crutches by analysis of segment power curves.. However, energy/stride values and foot forces indicated the subject may prefer crutches slightly longer than the conventionally fitted length. Smooth progression of movement and a longer swing-through was observed with 110cm(4cm too long) crutches.

Data on subject F unanimously supported the use of fitted crutches (using Cohen's(1979)formula) for this subject as lower energy/stride values, and peak powers were recorded. Balance was improved with fitted crutches due to improved axillary pad anchoring.

Subject G demonstrated a lower energy/stride value, with fitted crutches and lower peak powers despite the fact she crutched faster than with long crutches. In addition, lower axial crutch forces were recorded during the fitted crutch trial.

Subject H exhibited comparable segment energies and powers between the fitted and long crutch trials while lower energy/stride values were recorded with the long crutches due to the achievement of a larger swing-through. However, this female subject proved highly adaptable and was able to demonstrate efficiency on the tested crutch lengths.

Subject I also possessed the ability to adapt as she exhibited smooth gait over all crutch lengths. Push-off forces of the foot were greater than those experienced at heel-strike. Smoother power curves with decreased peaks were observed with the use of longer crutches.

The relationships between the measured parameters are linked by efficiency. Those trials exhibiting low peak powers with smooth traces generally demonstrate optimal (usually lower but not always) foot-floor and axial crutch forces. This was due to the combination of achieving optimal balance, speed, force application, magnitudes and patterns, and efficient style through the correct choice of crutch length. The question of gait efficiency becomes increasingly apparent as adaptable subjects are able to perform well throughout a larger range of crutch length selection. The key to adaptability is dependant upon developmental factors as balance, agility, strength and coordination. For children who lack these prerequisites for performing efficiently, crutch length becomes a critical consideration.

4.9 CONCEPT OF IDEAL ENERGY AND POWER CURVES

From the previous analysis, the concept of 'ideal' or mechanically efficient instantaneous energy and power traces remains simply that, a concept. However, while absolute 'efficient' traces cannot be postulated for swing-through crutch ambulators, general observations suggest trends and guidelines in determining 'efficient' gait. Patterns of coordination may be preferred by one subject and rejected by another. Nonetheless, energy and power traces do locate abnormal energy levels and power which can then be evaluated with regard to a particular individual. Energy traces check within segment energy component levels and transfers. Such transfers in the H-N-T indicated a flowing gait when the total energy remained at a stable level. Energy peaks of KE and PE should occur simultaneously in the arms, in addition to a desired symmetry between right and left sides. Instantaneous power curves describe segment accelerations, as positive power accelerates the segment while negative power decelerates it (Zarrugh, 1981). In addition, the magnitude of the power combined with the given segment action, describes energy tranfers between segments. Peak powers also render the magnitude of the acceleration involved. Power traces of the H-N-T segment should be smooth and have small magnitude peaks. This is due to the large mass and conservative nature of the segment. Power peaks in the arms indicate the acceleration-deceleration pattern during the swing of the crutches. Once again symmetry between right and left sides would increase total gait efficiency. Powers in the lower limbs should correlate to some degree within the leg indicating some form of unity of the limb as a whole. Powers between sides in the lower limbs need not reflect symmetry although some people are efficient with bilaterally mirrored actions. Powers of the shank and foot should be comparable within sides with maximum power occuring at heel strike. Peak negative power occurring before H-S indicate unneccessary eccentric muscle moments. In general, peak magnitudes or bursts of power should be kept to a minimum. The speed of gait effects individual segment powers. Interpretation of energy and power traces applied to human movement improves with practice and experience. It is hoped that this paper has aided in such improvement.

4.10 GENERAL DISCUSSION

From the data derived in this study, mechanically efficient crutch gait results from flowing, full swing-through crutching. The conditions of testing were optimal for the crutch walkers during data collection. Efficiencies will change with the surface and level of ground walked on, non-linear direction changes, and in crowded situations.

If a child required gait evaluation, testing would follow the principle of trial and error in search of gaining optimal efficiency. Therefore, the subject must aim for power curves that are void of unnatural, sharp peaks resulting in smooth traces. The investigator must also be aware of such factors as crutching speed which causes shape changes in power curves. General condition of the subject is a critical measure, as crutch gait requires sufficient upper body musculature, and neuro-control development.

Modifications to the crutch itself have been suggested by Shoup(1974) which comprised of a spring-dampened energy absorber in the crutch shaft to decrease the magnitude of PE changes in the H-N-T segment. However, energy transfers between PE and KE within the H-N-T have been shown to compliment each other in efficient gait as PE is required for transfer to KE thus maintaining forward progression. If PE is decreased, forward velocity may suffer. However, as a result of this investigation crutches are in need of two improvements. The smallest increment of length adjustment is approximately 2.5cm, on all crutches regardless of whether the crutch is adult or child size. Due to the critical nature of crutch length in facilitating efficient ambulation, 2.5cm increments is too gross for children. Drilling holes in the shaft between the existing ones, resulting in adjustments of 1.25cm, would greatly enhance the versatility of the childs crutch. As child crutch shafts are the same thickness as adult crutches, the weakness resulting from the new holes would not be critical due to the smaller forces exerted by children. Thus, failure of the crutch shaft does not become a consideration. The second modification is required as children appear to lack the necessary arm adductor strength to squeeze and anchor the axillary pad of the crutch. The fore-aft slipping of the crutch was observed in many trials. Without proper anchoring, crutch walkers were reluctant to apply force to the unstable support. Styling modifications to the axillary pad are required to aid in the anchoring process with most children. It appears that this consideration increases in importance with the younger the child.

Searching for sources of error in the data began with a review of the equipment. The force plate and crutch force transducer were calibrated and checked for linearity and hysteresis. The strain guages incorperated into the axial transducer were compensated for aluminum, the material of the adhering surface of the transducer thus reducing guage error. Body segment parameters were derived from the equations for inertial and mass properties of children by Jensen (1981) rather than scaling adult data to fit children.

The use of the gleno-humeral joint as a landmark to determine trunk displacement was pointed out as a source of error by Robertson (1981). The shoulder can move independantly of the trunk thus falsely registering trunk movement. In addition, it must be remembered that a major supposition of the 2-dimensional model is an absence of motion in the third dimension. Movement does occur in the third plane which causes error in energy calculations. The magnitude of such error in this study is minimal as third dimensional movements were recorded by a second cine camera. Coordinates from the film were not always in full view to be digitized, however, the estimation of these locations were found to be accurate by performing periodic retests. The coordinate data was digitally filtered (second-order Butterworth cutting off at 3.12Hz) to increase their reliability. All functions of the computer were periodically hand checked to confirm accurate data processing. A source of concern is the fact that the energy and power values starting and finishing with heel-strike, were rarely equivalent. However, in some cases, if one was to extrapolate one more frame from the existing data, the curves would begin and end with equal values of power and energy. It is possible one could suspect variance in gait to account for unequal starting and finishing energy-power levels. Such a variance could result from a small change in style or speed. However, Dainty(1979) accounted for uneven starting and finishing cyclic data curves as a function of the digital filtering process.

At this point, it would be beneficial to note that the results obtained in this study were derived from a group of eight subjects, 9 to 11 years of age. As the subjects were selected and matched for height from a grade school class of students, there is no reason to doubt that they are a reasonalble representation of the population. Therefore the results could be applied to all prepubescent children although children younger than 8 years old are rarely prescribed crutches. For this reason, the small subject sample size of children should cover the age range for eligible crutch walkers.

The data from this study indicated the usefullness of the energy transfer technique, as a diagnostic tool in crutch gait analysis. Disruptions in energy transfers can be identified as to their segmental and temporal location in the gait cycle. Once the inefficiencies have been located, measures can be taken to correct the pattern of energy transfer, which may take the form of crutch modification, body bracing, or improvments in body movement mechanics.

Chapter V

SUMMARY, CONCLUSIONS AND RECOMMENDATIONS

5.1 SUMMARY

People using crutches have a reduced capacity for ambulation due to their disability. There is little to be found in the literature on this unique form of locomotion. Scientific investigation into crutch gait began in the nineteen sixties with an electromyographic analysis (Peacock, 1966). Basic crutch mechanics were later analysed by Shoup et al (1974). Wells (1979) conducted a study involving the concept of energy transfers with adults walking with crutches. Winter (1979B) described the use of this technique to evaluate human movement for mechanical efficiency.

The purpose of this study was to investigate swing-through crutch gait with children assisted by the development of a computer package to aid in assessment. The computer package entailed the computation of instantaneous, segment energies and powers, and work values per gait cycle. As such values are dependant upon displacement data, absolute coordinates were obtained from film and digitally filtered, of the 9-segment body model. Curves of the segmental, instantaneous energies and powers were plotted by computer and analysed for common traits and abnormalities. Body segment inertial data of children were derived from the regression equations of Jensen (1980). Center of gravity data was derived from the equations of Dempster (1955).

Inefficiencies of gait were recognized in instantaneous, segmental energy and power curves, augmented by foot-ground force and axial crutch force information. Locating energy inefficiencies was accomplished from interpretation of the instantaneous power and energy curves for each body segment. The data derived in this study has started a bank of information on children walking with crutches to be available to future researchers.

5.2 CONCLUSIONS

Within the scope of the study, the following may be concluded:

1. The 9-segment model of the body-crutch system is adequate for evaluation of subjects with no pathologies that involve abnormal movements of the head.

2. Instantaneous energy and power curve can be a useful diagnostic, clinical tool for analysing crutch gait.

The remaining conclusions were not anticipated by the statement of the problem at the beginning of the paper. However, they came to light during the course of the study and are listed below.

3. Crutch length is critical to the centimeter as 4cm differences double instantaneous power peaks and greatly increase energy expenditure. However, determination of optimal crutch length follows no formula and some individuals benefit from slightly longer crutches. Therefore the formula discribed by Cohen(1979) is only a gross approximation of crutch length and by no means an indicator of the most mechanically efficient length.

4. Peak axial crutch forces vary from 40% to 100% body weight per crutch during the double crutch support phase.

5. Foot contact forces vary greatly but shape and peak magnitudes reveal important information such as force application patterns, time in contact and balance problems.

6. Normalized work values for children (corrected for body weight, and distance covered per stride) of mean=2.13 J/kg/m, S.D.=.845, are much greater than those found in adults(.88 to 1.72 J/kg/m) by Wells(1979).

5.3 RECOMMEDATIONS

1. Crutch modifications of redesigning the axillary pad to fascilitate improved anchoring capabilities for use with children.

2. Crutch modifications of drilling additional holes in the crutch shaft between existing ones to decrease the mini-

mum length increment from 2.5cm to 1.25cm for use with children.

3. More emphasis should be placed on general subject evaluation to determine if the prerequisite strength is developed in the upper body musculature to fascilitate proper crutch gait.

4. Further study into the area with developments in three dimensional anaysis as opposed to two dimensional-planar analysis, and development of immediate feedback systems for clinics to gain coordinate data for computer processing.

5. Further study with force transducing instrumentation on both crutches and shoes to gain additional force information is recommended. In addition, physiological parameters should be combined with the biomechanical data for deeper understanding into crutch gait.

BIBLIOGRAPHY

Abbott, W.M., and Darling, R.C., (1973), Axillary Artery Aneurysms Secondary To Crutch Trauma, Am. J. Surg., 125:515

Atlas Of Orthotics: Biomechanical Principles And Applications, American Academy Of Orthopedic Surgeons, C.V. Mosby Co., St Louis, 1975.

Baxter, M.A., Allington, R.O., Koepke, G.H., (1969), Weight Distribution Variables In The Use Of Crutches And Canes, Phys. Ther. 49:360-365

Beckett, R., & Chang, K., (1968), An Evaluation Of The Kinematics of Gait By Minimum Energy, J. Biomech. 1:147-159

Bergmann, G., (1979), Walking With Walking Aids III, Control and Training Of Partial Weight Bearing By Means of Instrumented Crutches, Z. Orthop. 117 (3):293-300

Bergmann, G., Kobel, R., Rohlmann, A., (1978), Walking Aids: The Effect On Forces Transmitted At The Hip Joint And Proximal End Of The Femur, Biomechanics VI-B, Proceeding Of The Sixth International Congress Of Biomechanics, Copenhagen, Denmark, University Park Press, Baltimore.

Bernstein, N., (1930), Untersuchung Der Korperbewegungen Und Korperstellungen Im Raum Mittels Spiegelaufnahmen, Int. Z. Angew. Physiol. 3:179-206

Braune, W., and Fischer, O., (1889), In Krogman and Johnston (eds.) Human Mechanics- Four Monographs Abridged. Wright-Patterson Air Force Base, Ohio, 1963, (AMRL-TDR-63-123)

Bresler, B., Radcliffe, C.W., Berry, F.R., (1957), Energy and Power in the Legs of Above Knee Amputees During Normal Level Walking, Prostheic Devices Research Project, Institute of Engineering Research, University of California, Berkley, Series III, issue 15.

Bresler, B., and Barry, F.R., (1951), Energy And Power IN The Leg During Level Walking, Prostheic Devices Research Project, Institute of Engineering Research, University of California, Berkley, Series II, issue 31.

Brooks,A.L.,and Fowler,S.B.,(1964), Axillary Artery Thrombosis After Prolonged Use Of Crutches, J. Bone Jt. Surg. 46(A):863-4

Burnett,C.N., Johnson,E.W.,(1971), Development of Gait In Childhood:Part II, Dev. Med. Child. Neurol. 13:207-215

Cappozzo,A., Figura,F., Marchetti,M.,(1976), The Interplay Of Muscular And Internal Forces In Human Ambulation, J. Biomech. 9:35-43

Cappozzo,A., Leo,T., Pedotti,A., (1975), A General Computing Method For The Analysis Of Human Locomotion, J. Biomech. 8:307-320

Casper,R.M., Jacobs,A.M., Kenney,E.S., McMaster,I.B., (1971), On The Use Of Gamma Ray Images For Determination Of Human Body Segment Parameters, Paper Presented At Quantitative Imagry In Biomedical Sciences, Houston, Texas.

Cavagna,G.A., Saibene,F.P., Margaria,R., (1963), External Work In Walking, J. Appl. Physiol. 18:1-9

Cavagna,G.A., Margaria,R., (1966), Mechanics Of Walking, J. Appl. Physiol. 21:271-278

Cavagna,G.A., (1975), Force Platform as Ergometers, J. Appl. Physiol. 39:174-179

Cavagna,G.A., Thys,H., Zamboni,A.,(1976) Mechanics of Walking And Running, J. Physiol. 262:639-657

Cerza,P.F., (1978), Device To Assist Ambulation, Phys. Ther. 58 (10):1218-9

Childs,T.A., (1964), An Analysis of the Swing-Through Crutch Gait, Phys. Ther. 44:804-7

Clauser,C.E., McConville,J.T., Young,J.W., (1969), Weight, Volume and Center of Mass of Segments of the Human Body, Wright-Patterson Air Force Base, Ohio, (AMRL-TR-69-70)

Cochran,G.V., et al. (1973), Force Measurement Device For Crutches And Canes, Arch. Phys. Med. Rehabil. 54:43-44

Cohen,S., (1979), Teaching A Patient How To Use Crutches, Am. J. Nurs. 79(6):1111-26

Cotes,J.E., and Meade,F., (1960), The Energy Expenditure And Mechanical Energy Demand In Walking, Ergonomics 3:97-119

Dainty, D.A., (1979), PhD Thesis- Validation of a Technique for Determining Ground Reaction Forces in Three Dimensions. Penn State University, Aug.

Dempster, W.T., (1955), Space Requirements Of The Seated Operator, WADC, Tech. Report 55-159, Wright-Patterson Air Development Centre, Ohio, USA.

Drillis, R., Contini, R., (1966), Body Segment Parameters, School Of Engineering and Science, New York University, (PB 174 945:Tech. Rpt. No. 1166.03)

Elftman, H., (1939), Forces and Energy Changes in the Leg During Walking, Am. J. Physiol. 125:339-356

Epstein, S., (1937), Art, History and the Crutch, Ann. Med. History. 9:304-313

Ettien, J.T., (1980), Crutch Induced Aneurysms of the Axillary Artery, Am. Surg. 46(4):267-9

Fenn, W.O., (1930), Frictional and Kinetic Factors in the Work of Sprint Running, Am. J. Physiol. 92:538-611

Fischer, O., (1906), Theoretische Grundlagen Fur Eine Mechanik Der Lebenden Korper Mit Speziellen Anwendungen Auf Den Menschen Sowie Auf Einge Bewegungsvorgange An Maschinen, Leipzig, Teubner.

Fisher, S.V., (1981) Energy Cost Of Ambulation With Crutches, Arch. Phys. Med. Rehab. Jun:62(6):250-256

Foley, C.D., Quanbury, A.O., Steinke, T., (1979), Kinematics of Normal Child Locomotion- A Statistical Study Based On T.V. Data, J. Biomech. 12:1-8

Gersten, Orr, W., Sexton, A.W., Okin, D., (1969), External Work In Level Walking, J. Appl. Physiol. 26:286-289

Hatze, H., (1981), The Use Of Optimally Regularized Fourier Series For Estimating Higher Order Derivatives Of Noisy Biomechanical Data, J. Biomech. 14:13-18

Jensen, R.K., (1978), Estimation Of The Biomechanical Properties Of Three Body Types Using A Photogrammetric Method, J. Biomech. 11:349-358

Jensen, R.K., (1981), paper presented to the american society of biomechanics. Case Western Reserve University, Oct 19-20

Klenerman,L., Hutton,W.C., (1973), A Quantitative Investigation of the Forces Applied to Walking Stichks And Crutches, Rheum. Phys. Med. 12(3):152-8

Lamonda,J., (1979), From The Clinic: Dorsal Hand Cuff for Lofstrand Crutches, Am. Correct. Ther. J. 33(6):192-3

Liu,Y.K., Laborde, J.M.; Van Buskirk, W.C., (1971), Inertial Properties of a Segmented Cadaver Trunk: Their Implicatibns in Accelerayion Injuries, Aerospace Med. 42:650-657

Lukin,L., Polissar,M., Ralston,H., (1967), Methds Of Studying Energy Cost And Energy Flow During Human Locomotion, Human Factors. 9:603-608

Lukin,L., Ralston,H.T., (1969), Energy Levels Of Human Body Segments During Level Walking, Ergonomics. 12:39-46

Miller,D., Nelson,R., (1973), Biomechanics of Sport, Lea and Febiger, Baltimore

Murray,M.P., Seireg,A., Scholz,R.C., (1967), Centre of Gravity, Centre of Pressure, and Supportive Forces During Human Activities, J. Appl. Physiol. 23:831-838

Muybridge,E.,(1955), The Human Figure In Motion, Dover, New York

Norman,R., Sharratt,M.T., Pezzack,J.C., Noble,E.G., (1976), Reexamination of the Mechanical Efficency of Horizontal Treadmill Running, Biomechanics V-B (ed. P.V. Komi)(Baltimore, University Park.Press)

Peacock,B., (1966), A Myographic and Photographic Study Of Walking With Crutches, Physiotherapy London. 52:264-268

Pezzack,J., Norman, R.W., Winter, D.A., (1977), An Assesement of Derivative Techniques Used For Motion Analysis, J. Biomech. 10:377-382

Pierrynowski,M.R., Winter,D.A., Norman,R.W., (1980), Transfers of Mechanical Energy Within The Total Body and Mechanical Efficiency During Treadmil Walking, Ergonomics. 23:147-156

Plagenhoef,S., (1968), Computer Programs for Obtaining Kinetic Data on Human Movement, J. Biomech. 1:221-234

Plagenhoef,S., (1971), Patterns Of Human Motion- A Cinematographic Analysis, Englewood cliffs, N.J., Prentice-Hall.

Quanbury,A.O., Winter,D.A., Reimer,G.D., (1975), Instantaneous Power and Power Flow in Body Segments During Walking, J. of Hum. Movmt. Studies. 1:59-67

Rader,C.M., Gold,B., (1967), Digital Filter Design Techniques in the Frequency Domain, Proc. IEEE. 55:149-171

Robertson,D.G.E., Winter,D.A., (1980), Mechanical Energy Generation, Absorption and Transfer Amongst Segments During Walking, J. Biomech. 13:845-854

Robertson,D.G.E., (1981), Paper presented to CASS, Halifax, N.S., November.

Sankarankutty,M., et al. (1979), The Relative Efficiency of 'Swing-Through' Gait on Axillary, Elbow and Canadian Crutches Compared to Normal Walking, J. Biomed. Eng. 1:55-7

Saunders,J.B., et al. (1953), The Major Determinants in Normal and Pathological Gait, J. Bone. Jt. Surg> 35:534

Shoup,T.E., Fletcher,L.S., Merrill,B.R., (1974), Biomechanics of Crutch Locomotion, J. Biomech. 7:11-20

Smith,A.J., (1975), The Kinetic Energy Of The Human Body, J. of Hum. Movmt. Studies. 1:13-18

Stallard,J., Sankarankutty, M., Rose, G.K., (1980), One Leg Swing Through Gait Using Two Crutches, Acta. Orthop. Scand. 51:71-7

Sydenham,P., (1980), Transducers in Measurement and Control, Adam Hilger Ltd., Bristol.

Townsend,M.A., Seireg,S., (1972), The Synthesis of Bipedal Locomotion, J. Biomech. 5:71-83

Tucker,P.G., (1979), Adaption of Axillary Crutches, Phys. Ther. 59:884-5

Vaughan,J.,(1975), Application of B & K Equipment to Strain Measurements, Bruel & Kjaer, Ottawa

Virga,A., (1979), Adapted Crutches For Patients With Little Or No Handgrasp, Phys. Ther. 59-37

Wells,R.P., (1979), The Kinematics and Energy Variations of Swing Through Crutch Gait, J. Biomech. 12:579-585

Winter,D.A., Sidwall,H.G., Hobson,D.A., (1974), Measurement and Reduction of Noise in Kinematics of Locomotion, J. Biomech. 7:157-159

Winter,D.A., Quanbury,A.C., Reimer,G.D.,(1976), Analysis of Instantaneous Energy Of Normal Gait, J. Biomech. 9:253-257

Winter,D.A., (1978), Energy Assesments in Pathological Gait, Physiother. Canada. 30:183-191

Winter,D.A., Robertson,D.G., (1978), Joint Torque And Energy Patterns in Normal Gait, Biol. Cybernetics. 29:137-142

Winter,D.A., (1979A), A New Definition Of Mechanical Work Done in Human Movement, J. Appl. Physiol. 46:79-83

Winter,D.A., (1979B), Biomechanics of Human Movement, John Wiley and Sons, Toronto.

Zarrugh,M.Y., Radcliffe,C.W., (1979), Computer Generation of Human Gait Kinematics, J. Biomech. 12:99-111

Zarrugh,M.Y., (1981), Power requirements and Mechanical Efficiency of Treadmill Walking, J. Biomech. 4:157-165

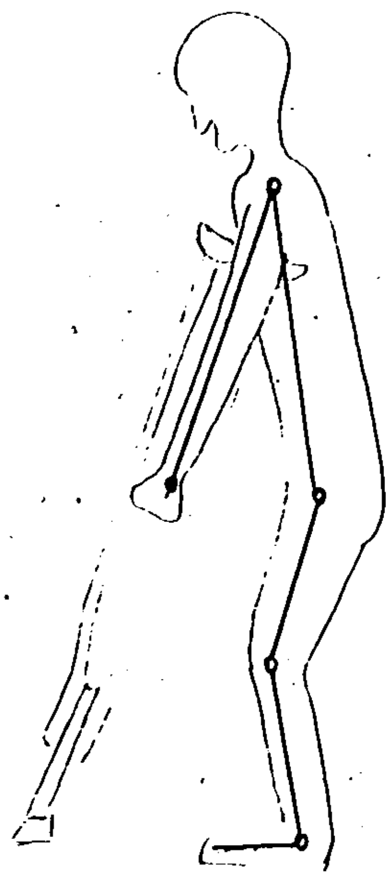
Zatsiorsky,V., Seluyanov,V., (1981), The Mass and Inertia Characteristics of Main Segments of Human Body, Presented to VIII Internatinal Congress of Biomechanics. July 1981, Japan.

Appendix A

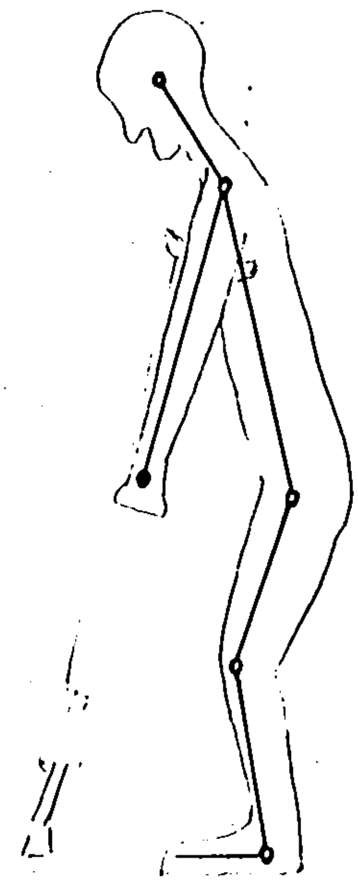
LINK SEGMENT MODELS

Link Segment Models

9 Segments



10 Segments

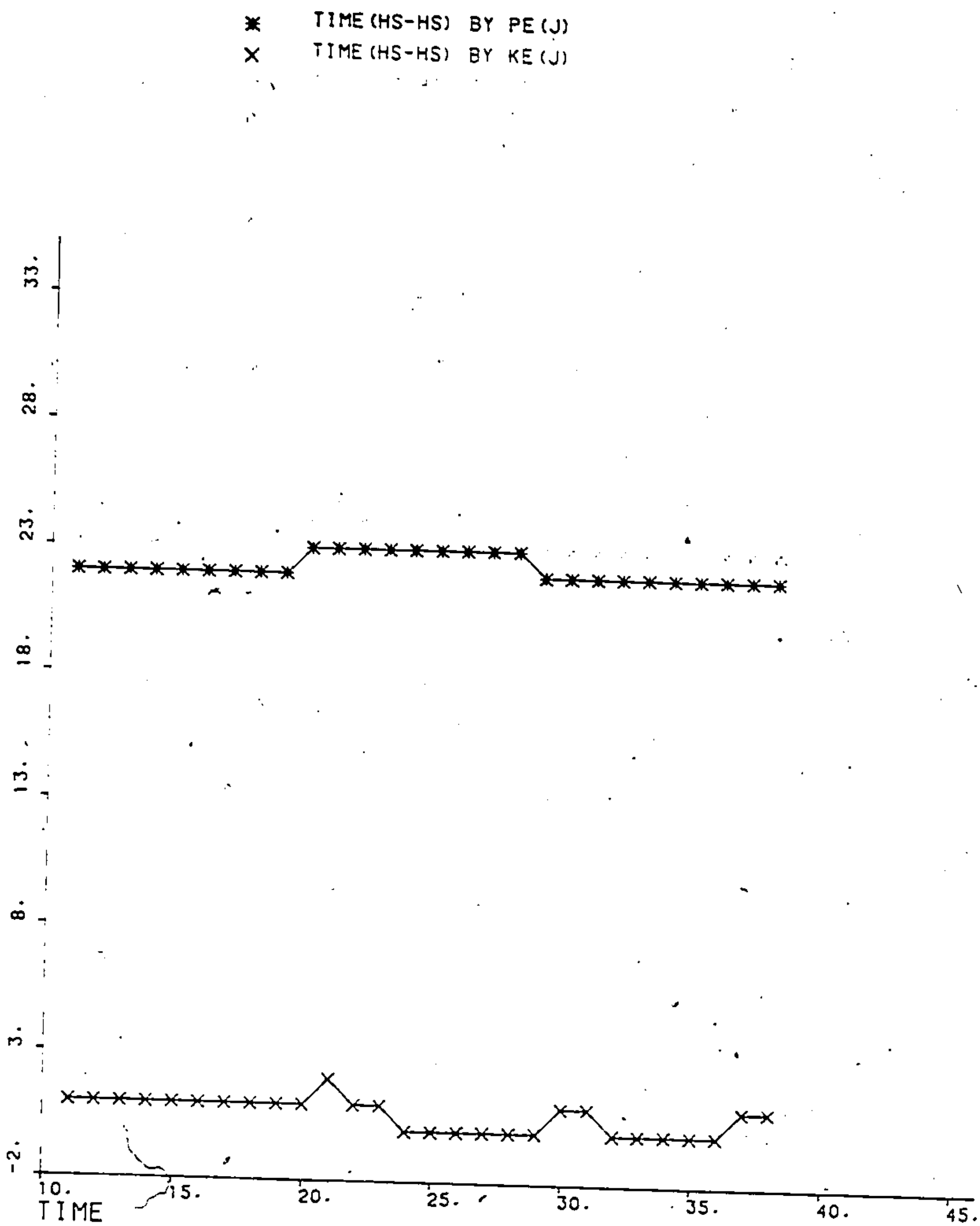


Appendix B

9-SEGMENT ENERGY AND POWER CURVES

The following traces are of one representative trial. The traces of the rest of the subjects and trials are available upon request of the author.

INSTANTANEOUS ENERGY LEVELS
SUBJECT-TRIAL/SEGMENT: B01102 R-ARM



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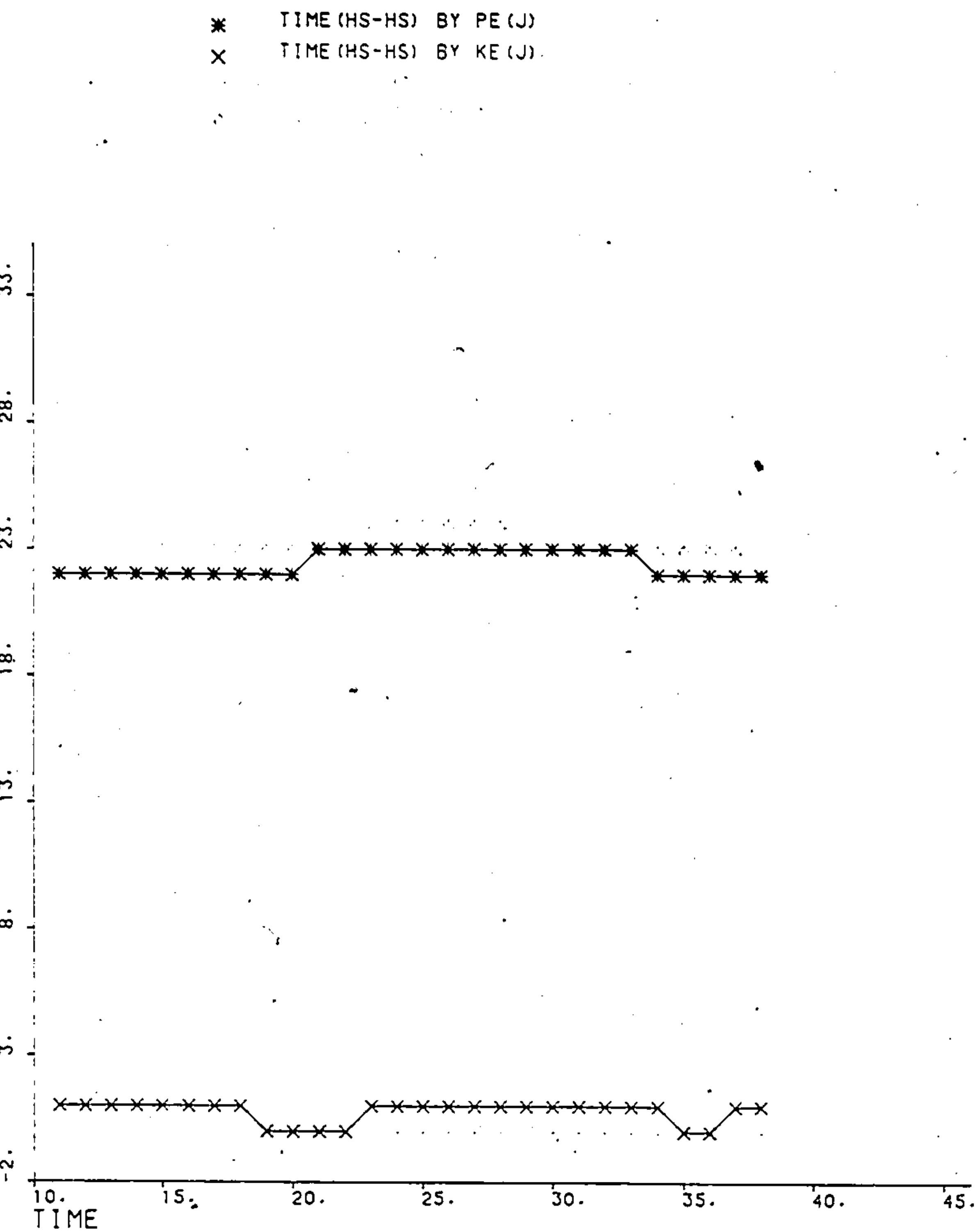

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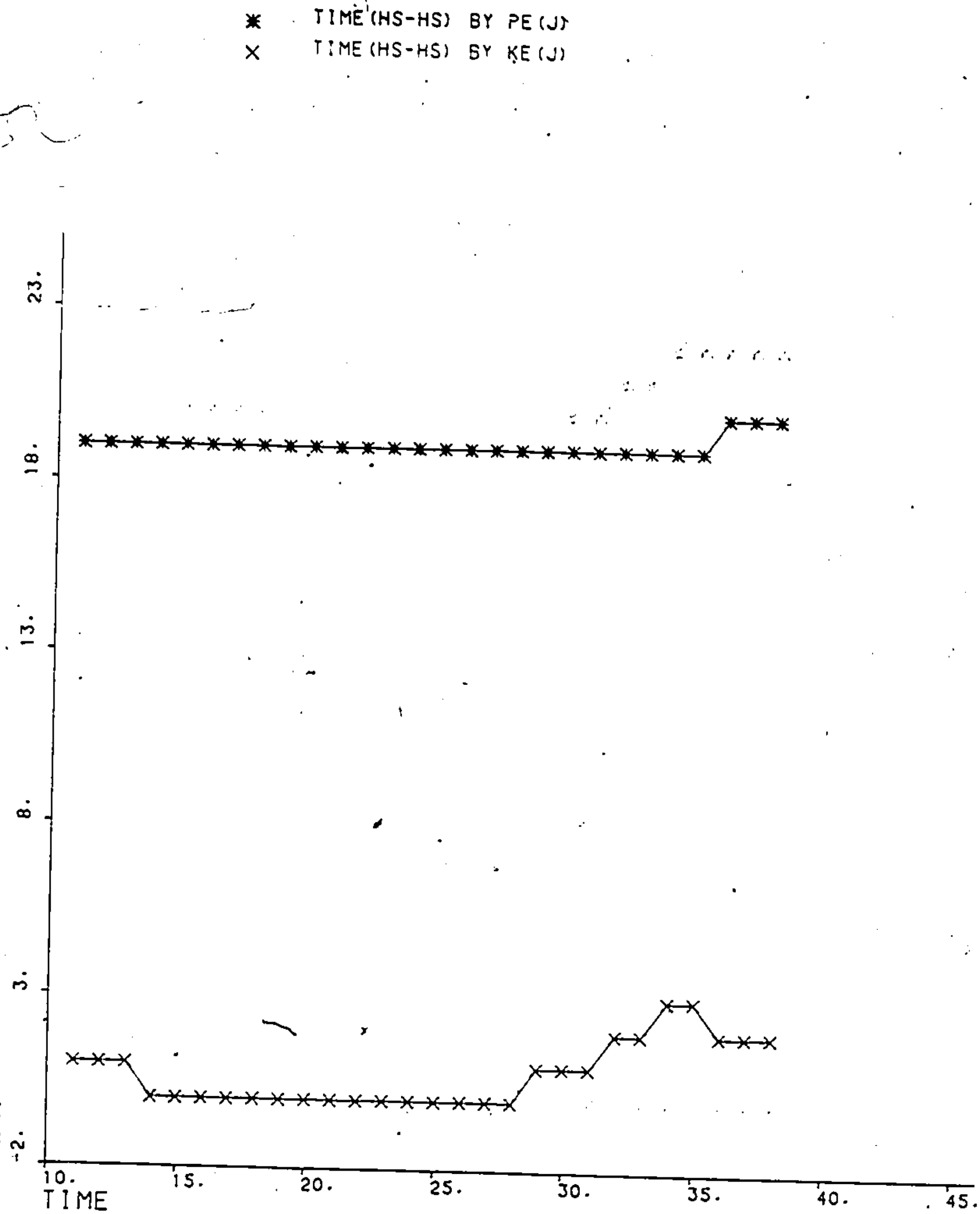

INSTANTANEOUS ENERGY LEVELS
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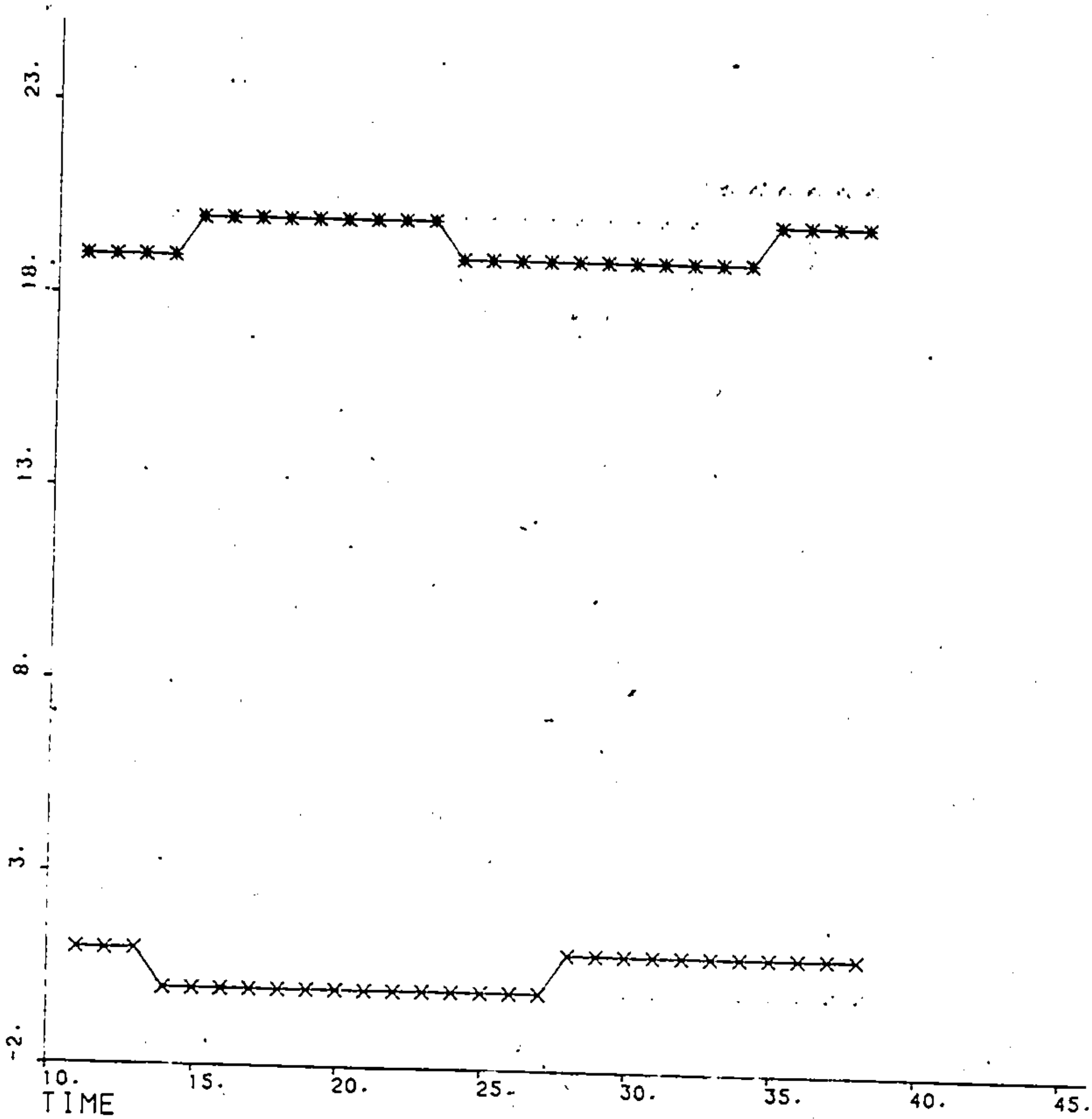
X TIME (HS-HS) BY KE (J)



INSTANTANEOUS ENERGY LEVELS
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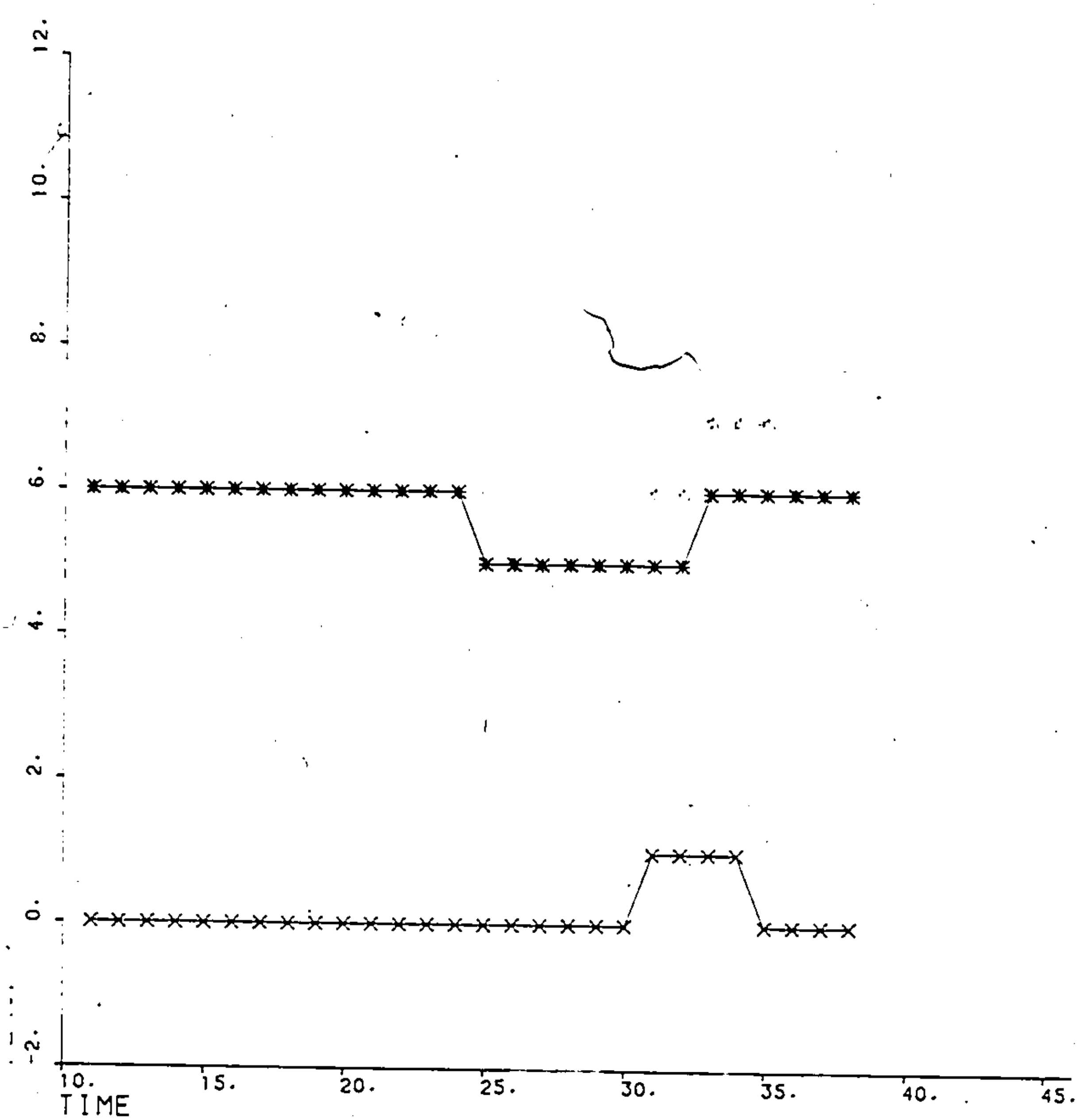
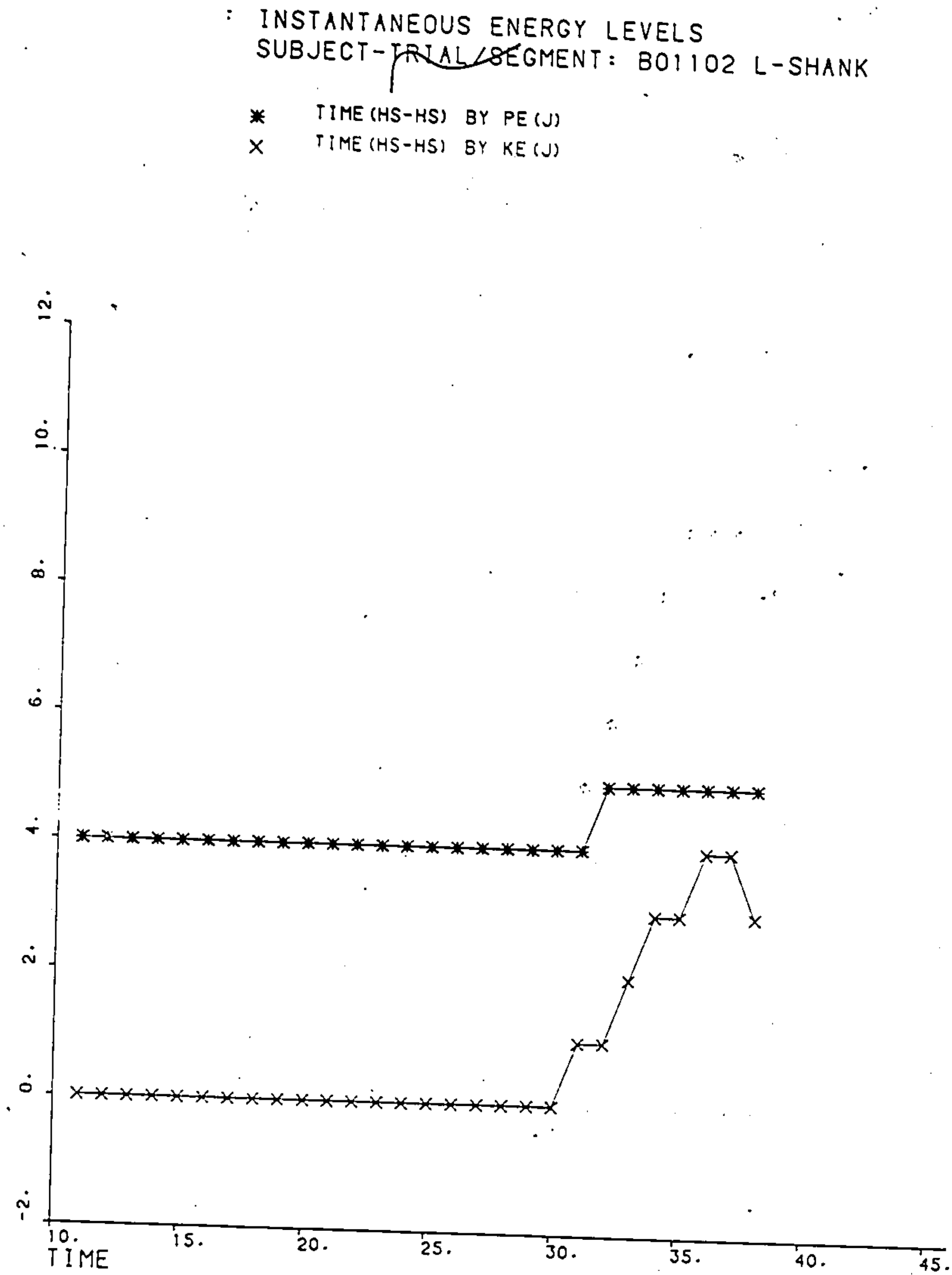
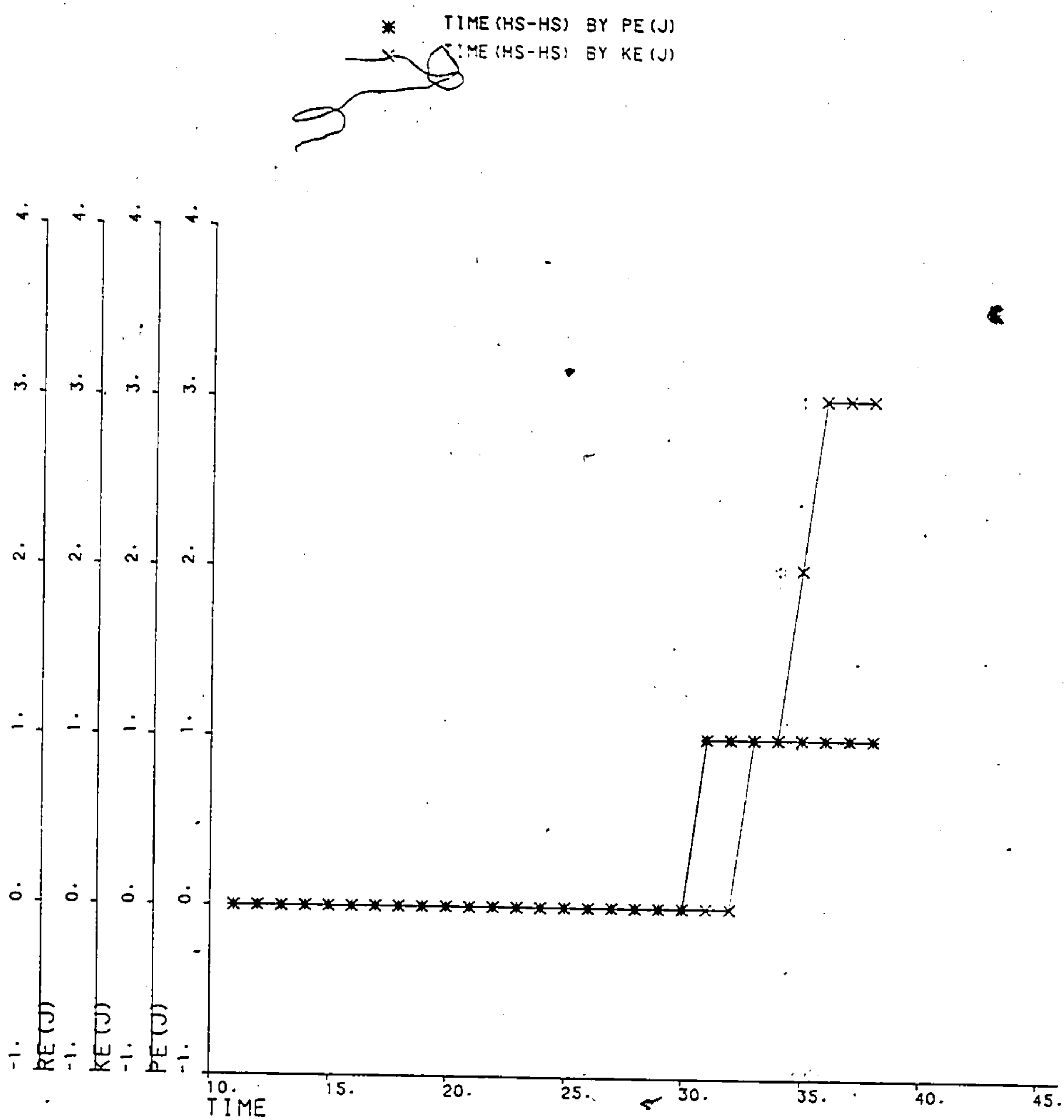
✱ TIME(HS-HS) BY PE(J)
× TIME(HS-HS) BY KE(J)




INSTANTANEOUS ENERGY LEVELS
SUBJECT-TRIAL/SEGMENT: B01102 L-SHANK
TIME (HS-HS) BY PE (J)
TIME (HS-HS) BY KE (J)
12.
10.
8.
6.
4.
2.
0.
-2.
10.
15.
20.
25.
30.
35.
40.
45.
TIME

INSTANTANEOUS ENERGY LEVELS
SUBJECT-TRIAL/SEGMENT: B01102 L-FOOT



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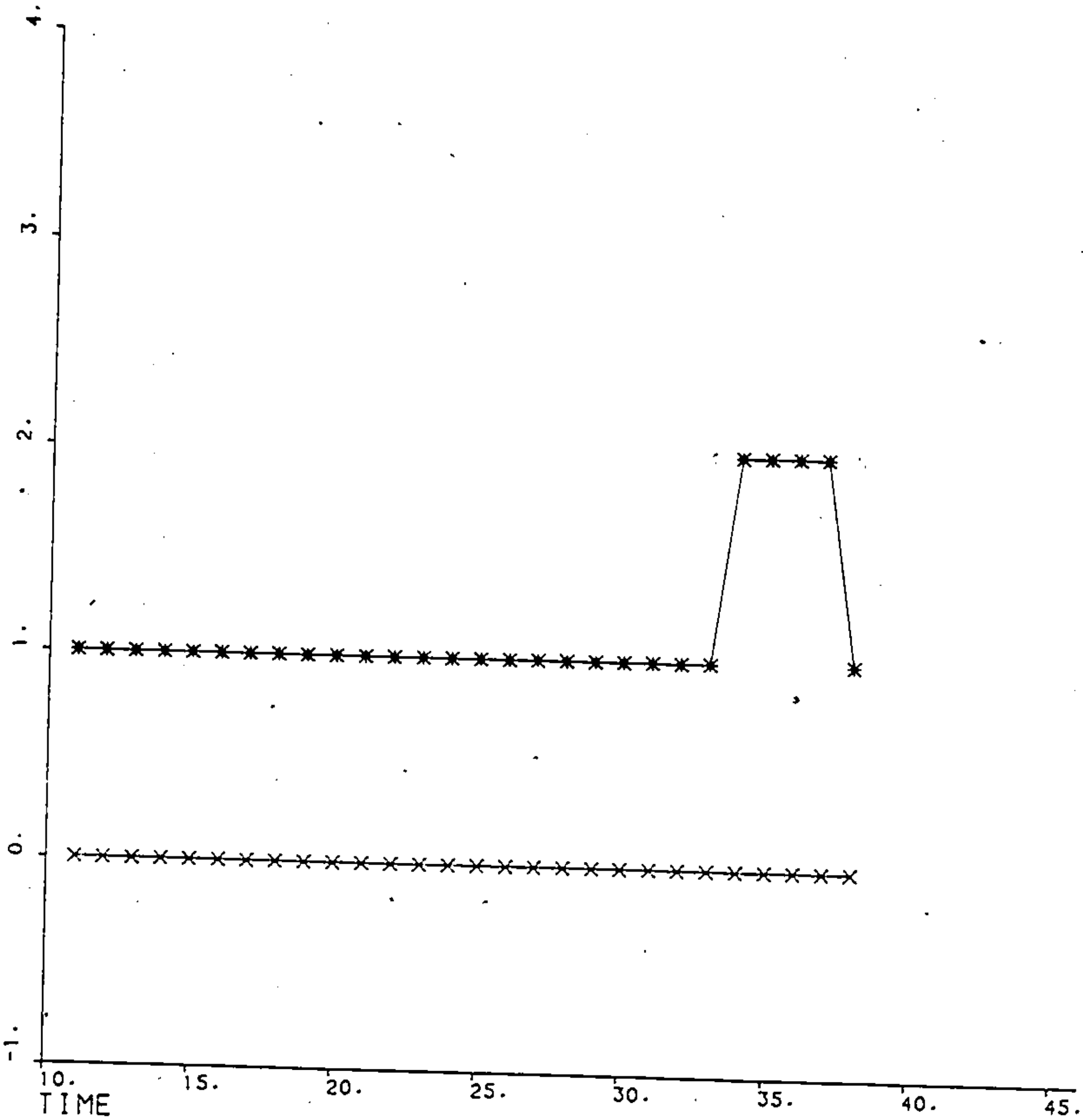
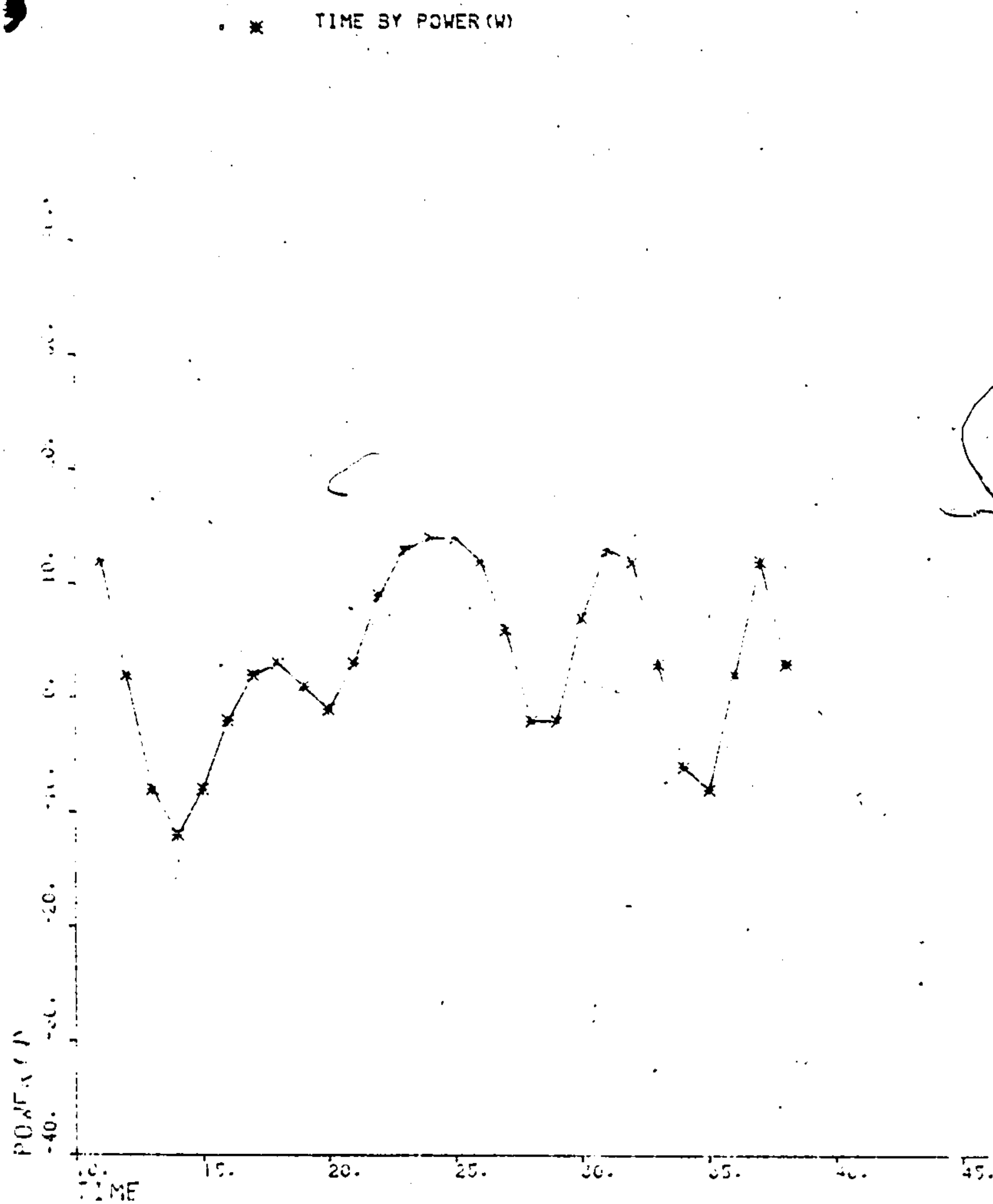
* TIME(HS-HS) BY PE(J)
× TIME(HS-HS) BY KE(J)



INSTANTANEOUS POWER LEVELS
SUBJECT-TRIAL/SEGMENT: B01102 H-N-T



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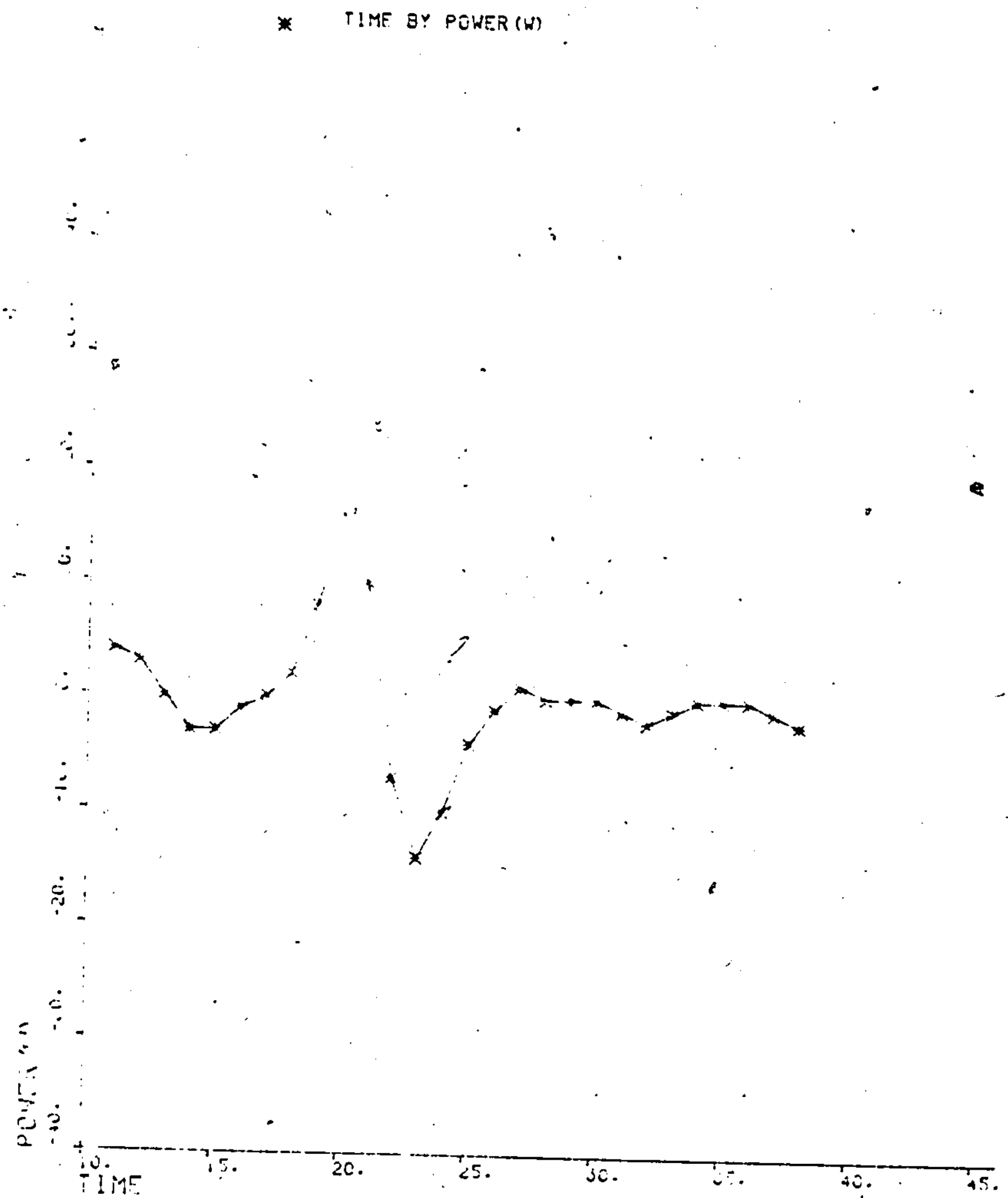

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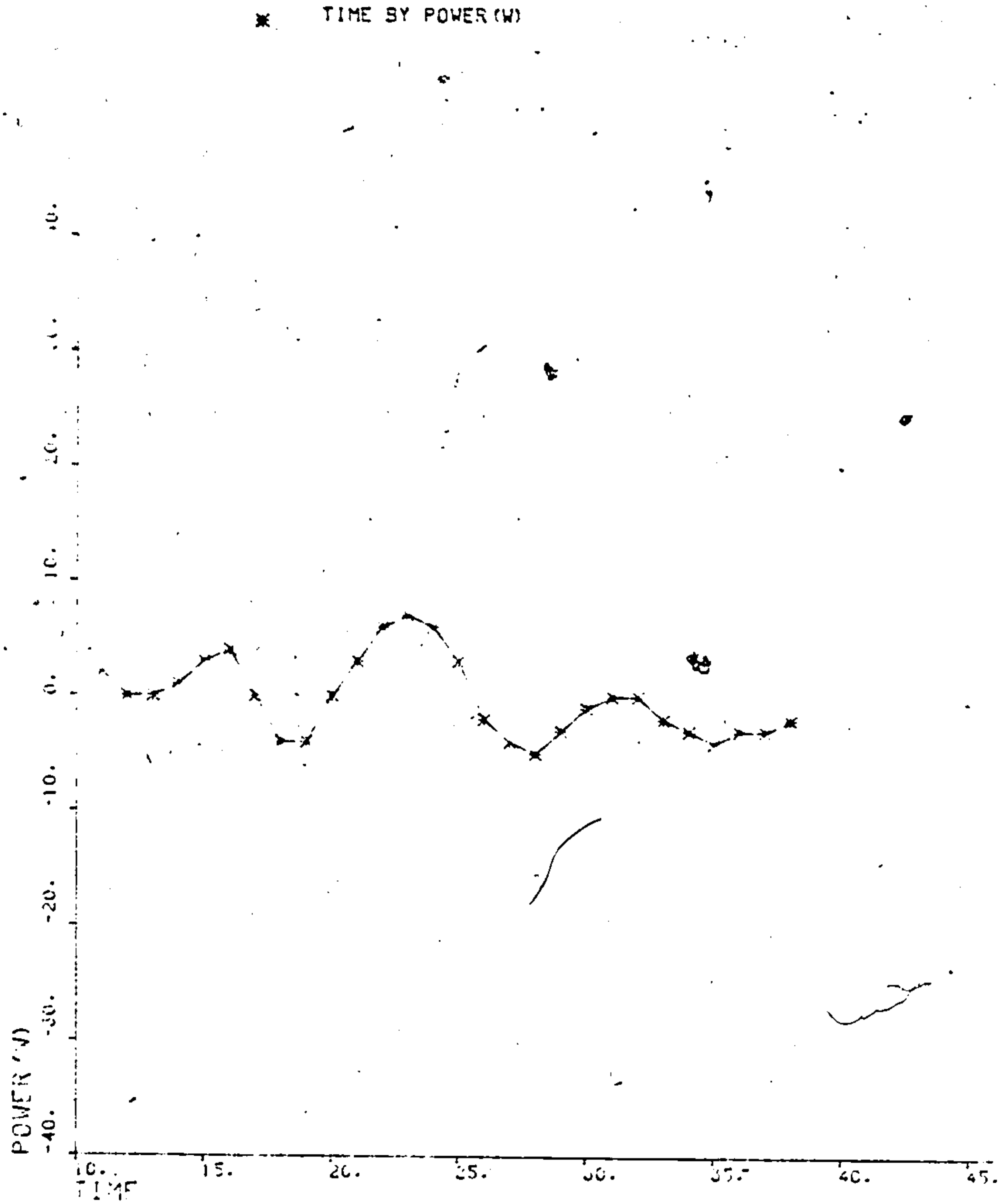

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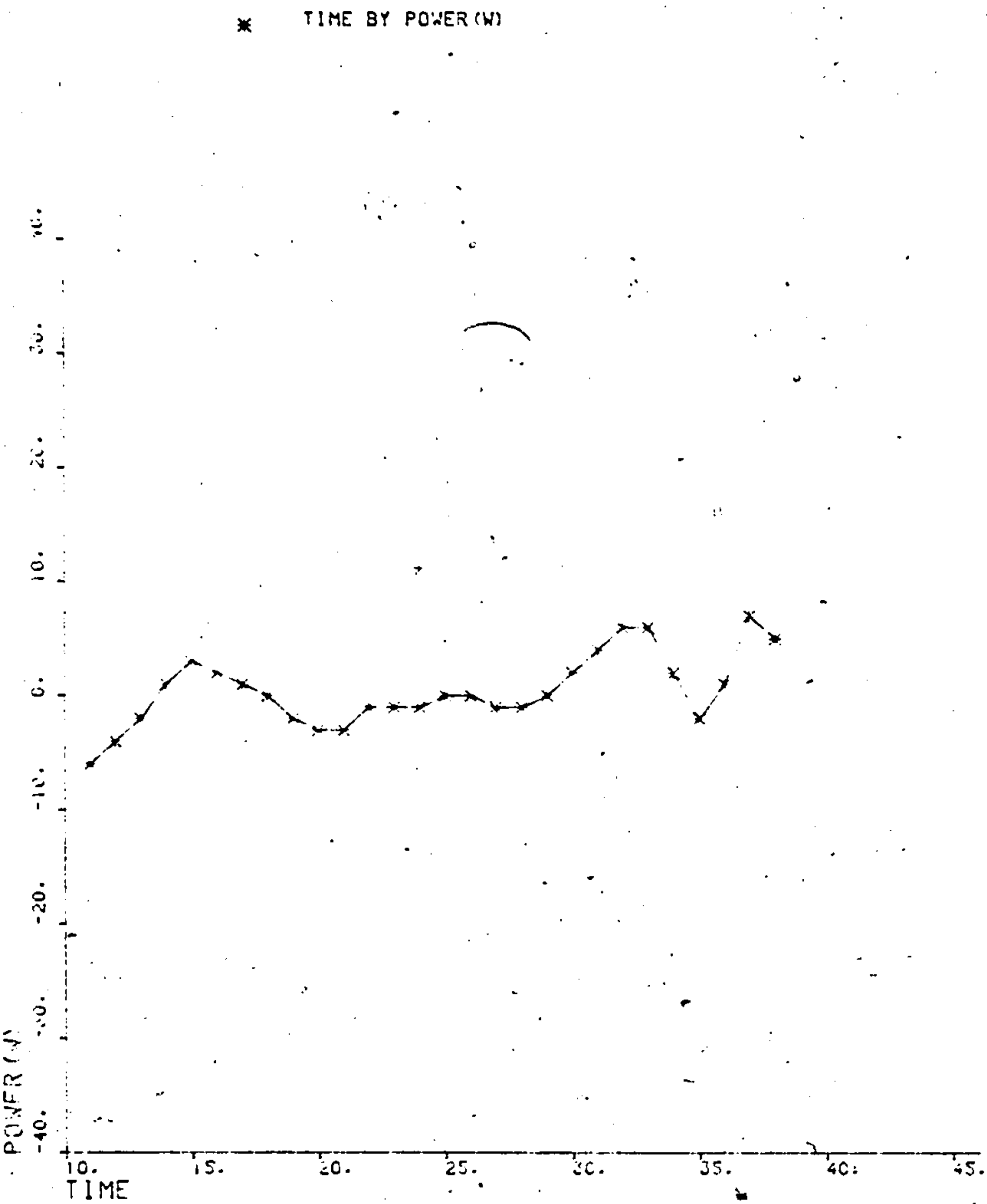

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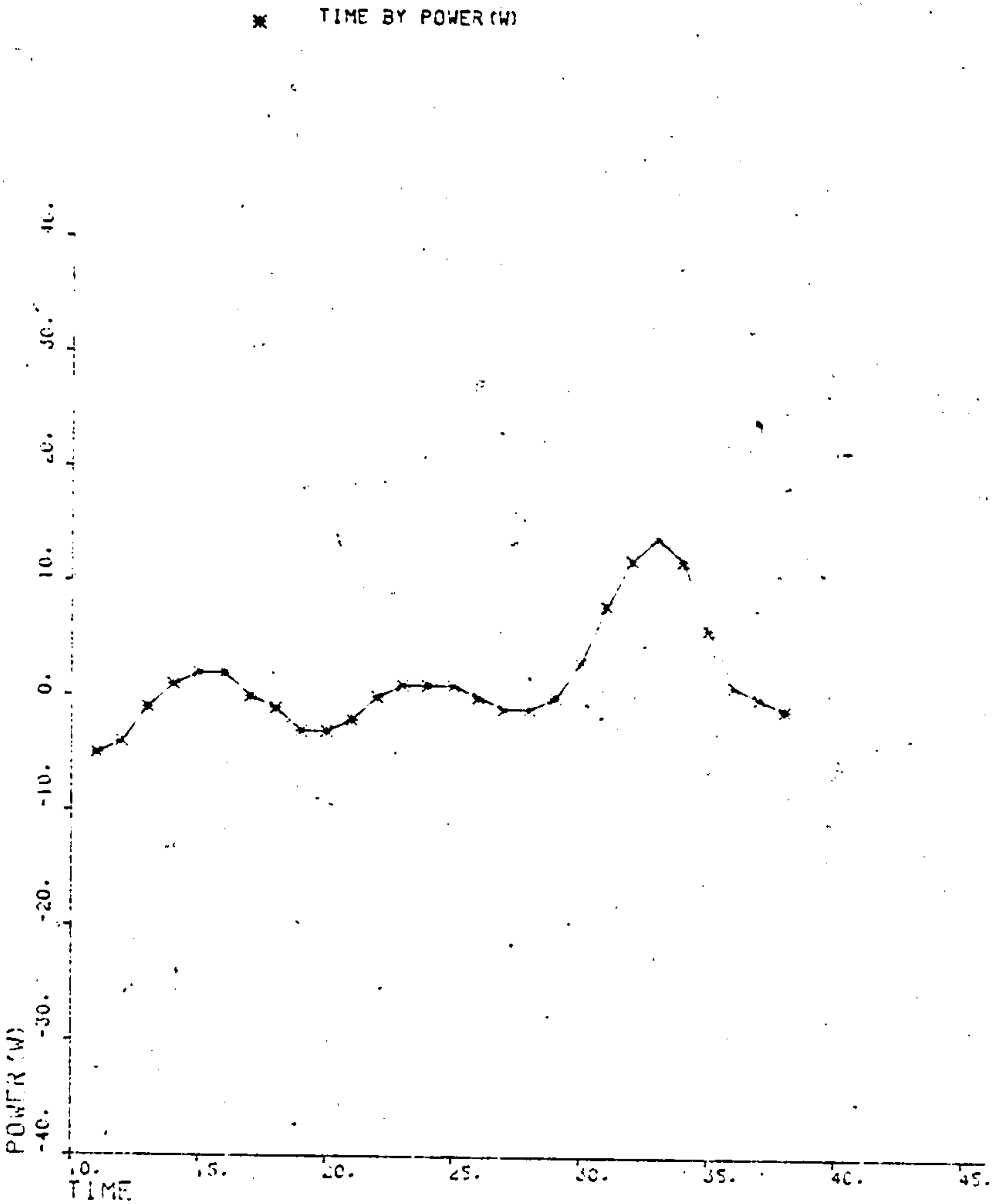

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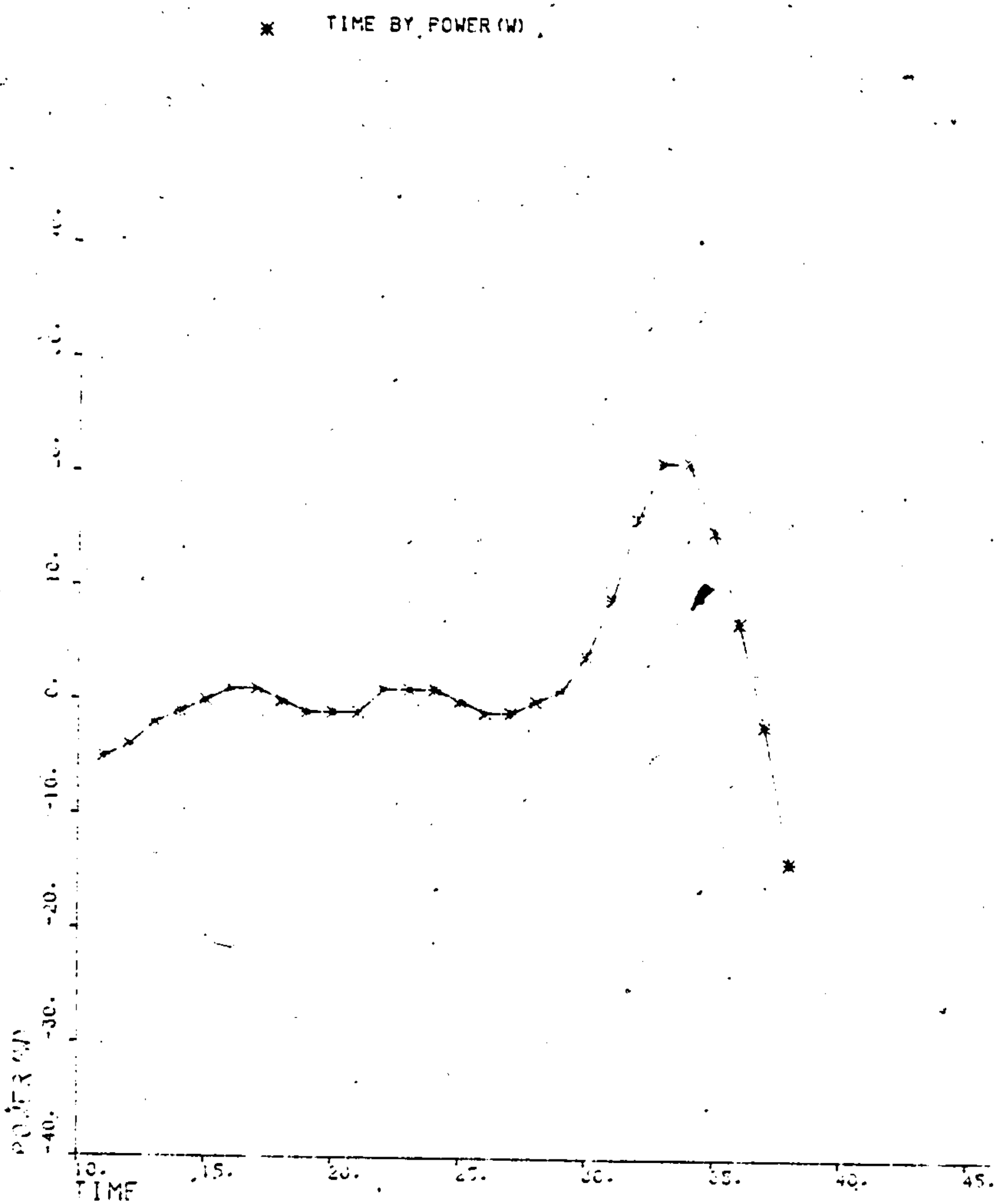
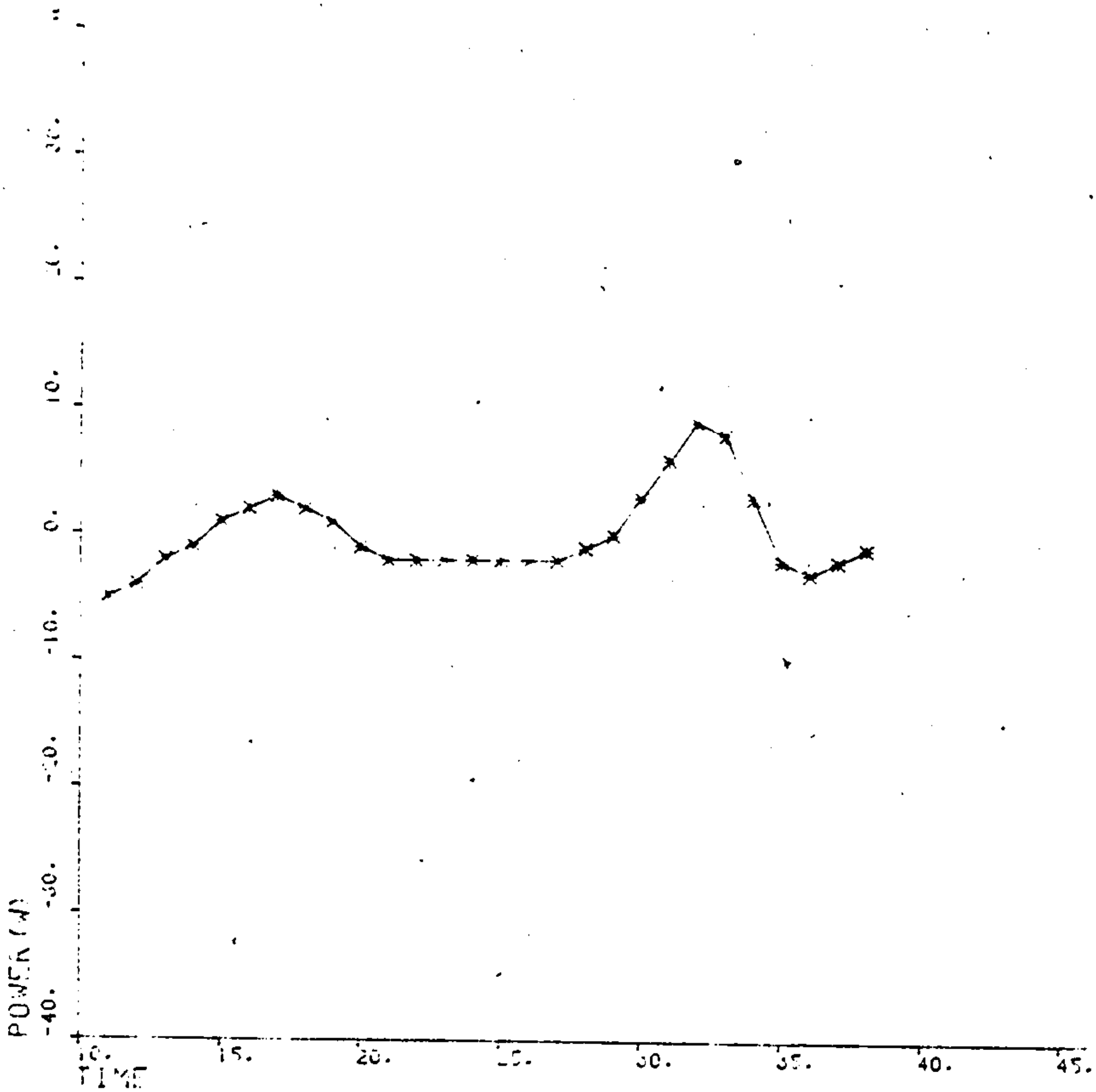


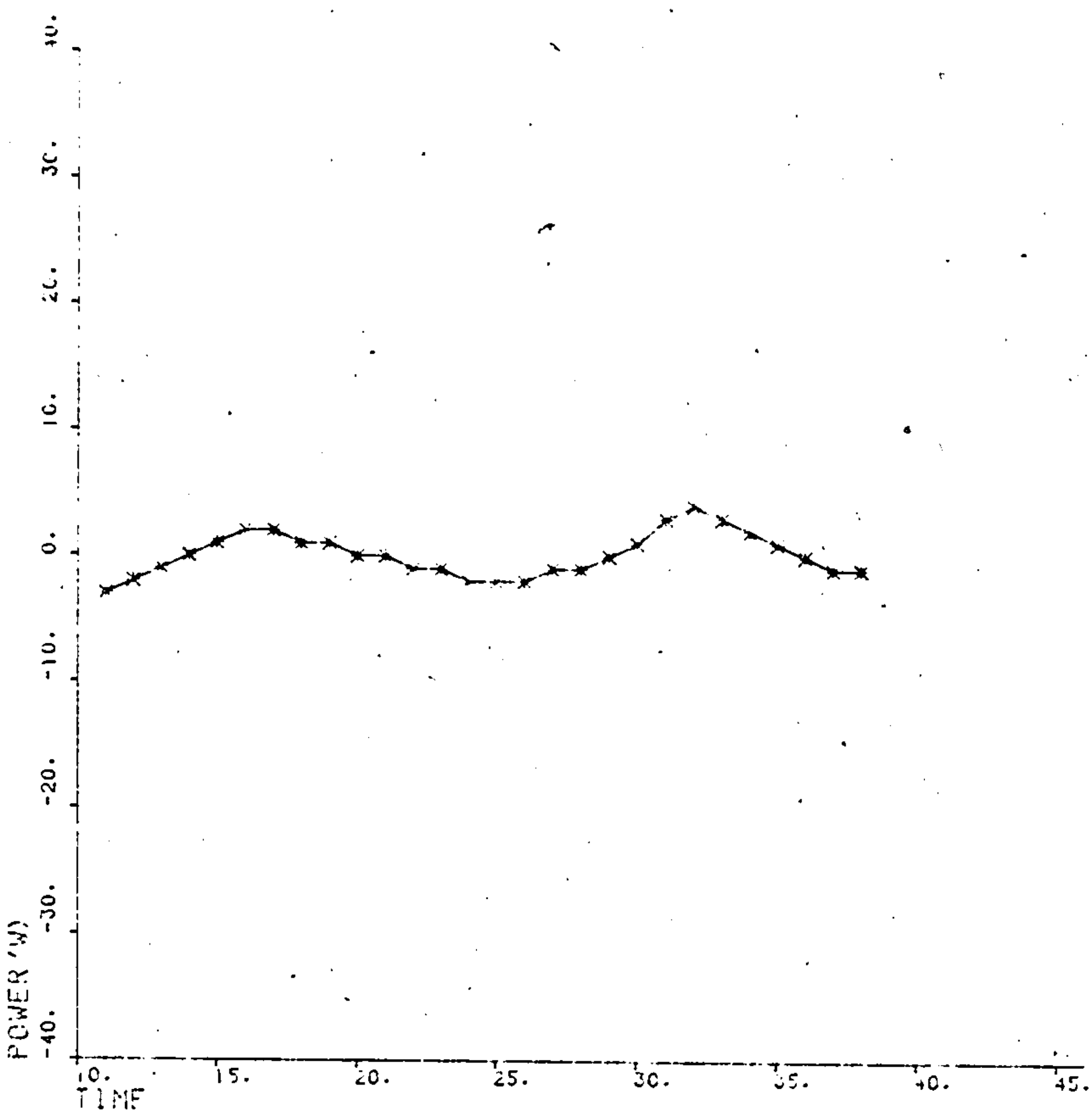
INSTANTANEOUS POWER LEVELS
SUBJECT-TRIAL/SEGMENT: B01102 R-SHANK

* TIME BY POWER(W)



INSTANTANEOUS POWER LEVELS
SUBJECT-TRIAL/SEGMENT: B01.102 R-FOOT

* TIME BY POWER(W)



INSTANTANEOUS POWER LEVELS
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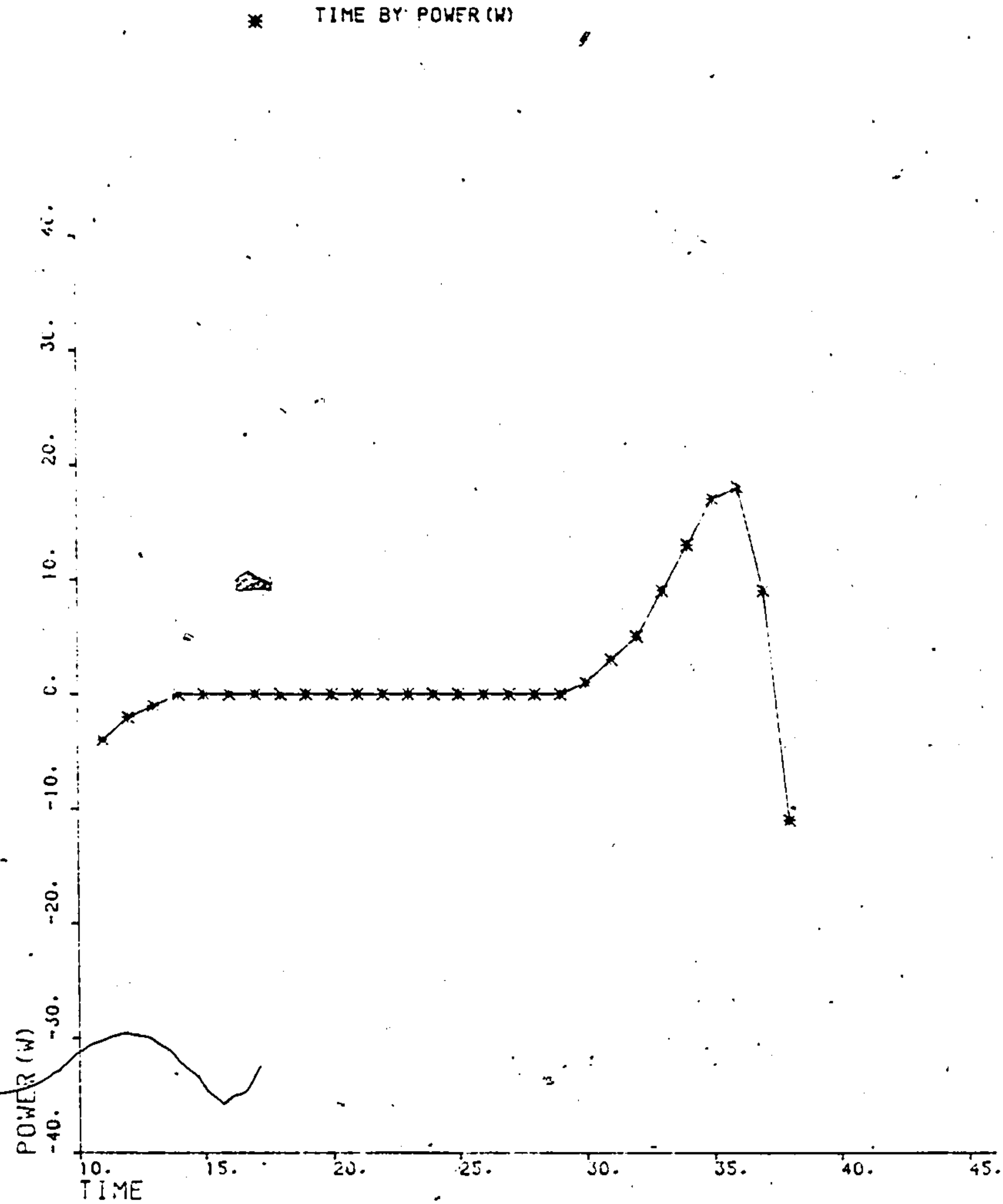


Appendix C

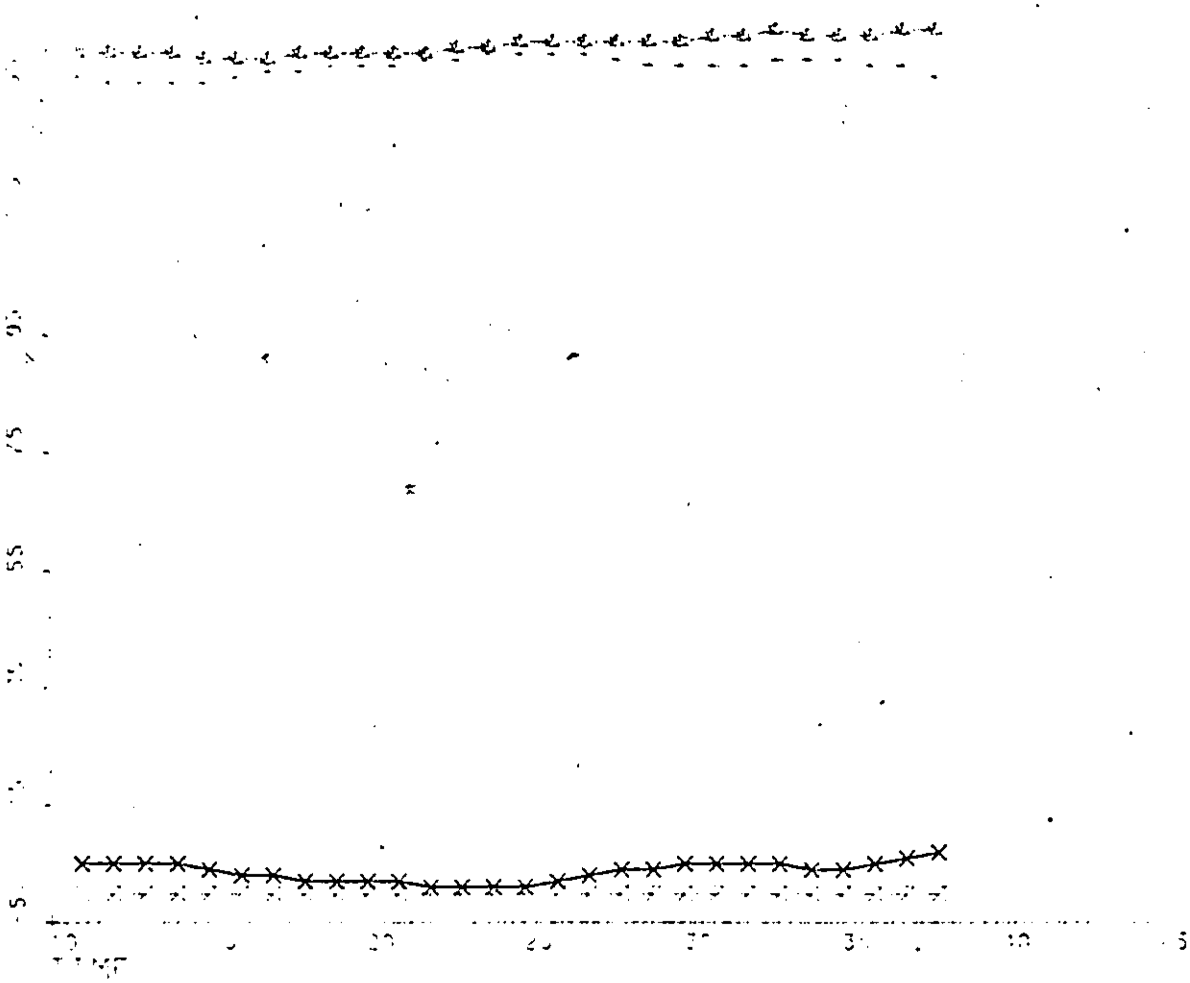
10-SEGMENT ENERGY AND POWER CURVES

The following traces are of one represenative trial. The traces of the rest of the subjects and trails are available upon request of the author.

INSTANTANEOUS ENERGY LEVELS
SUBJECT-TRIAL/SEGMENT: BS 108 TRUNK

X TIME (HS-HS) BY KE (J)

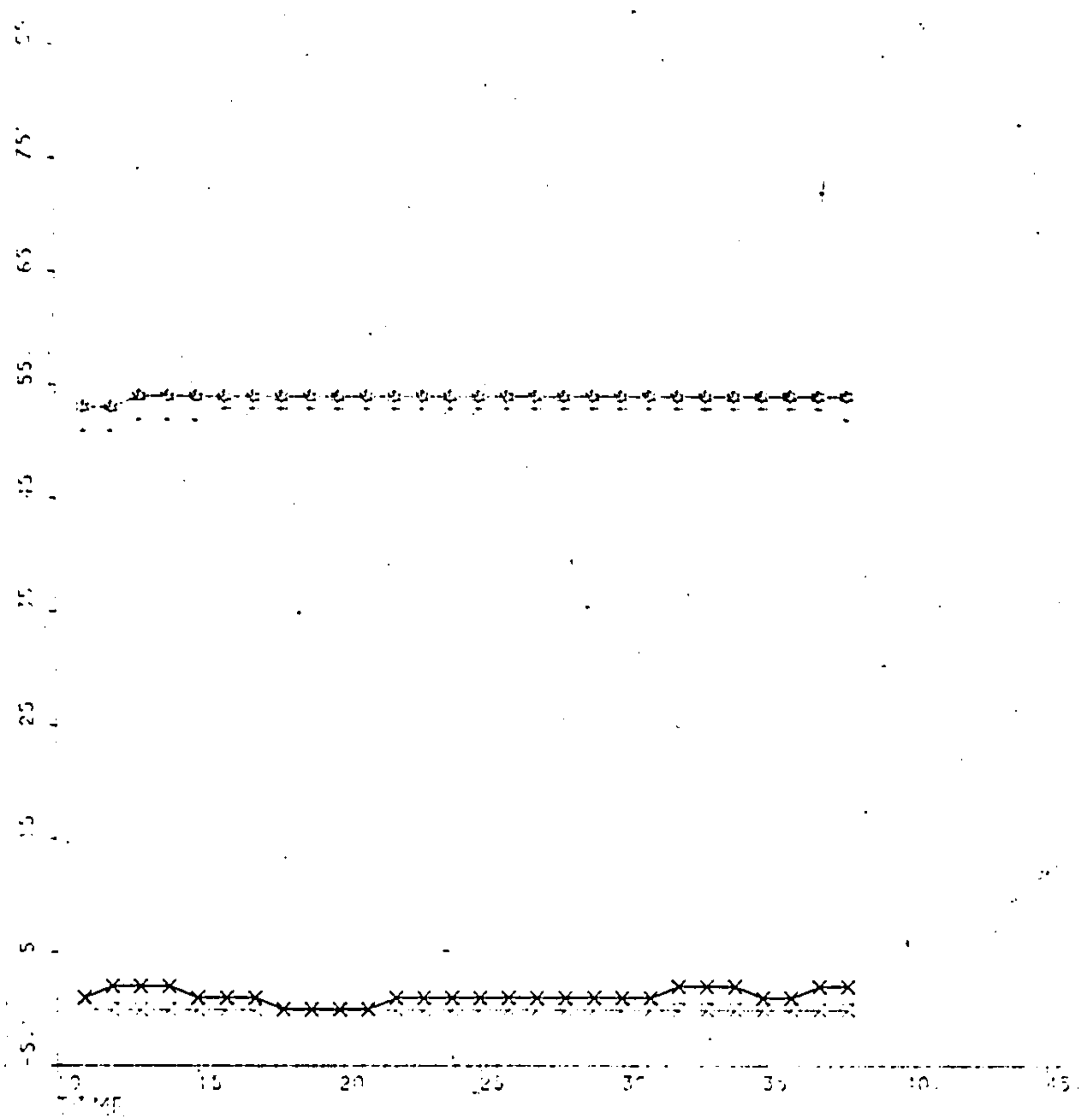
TIME (HS-HS) BY TE (J)



INSTANTANEOUS ENERGY LEVELS
SUBJECT-TRIAL/SEGMENT: BO, 102 HEAD-NECK

TIME(HS-HS) BY KE(J)

TIME(HS-HS) BY TE(J)



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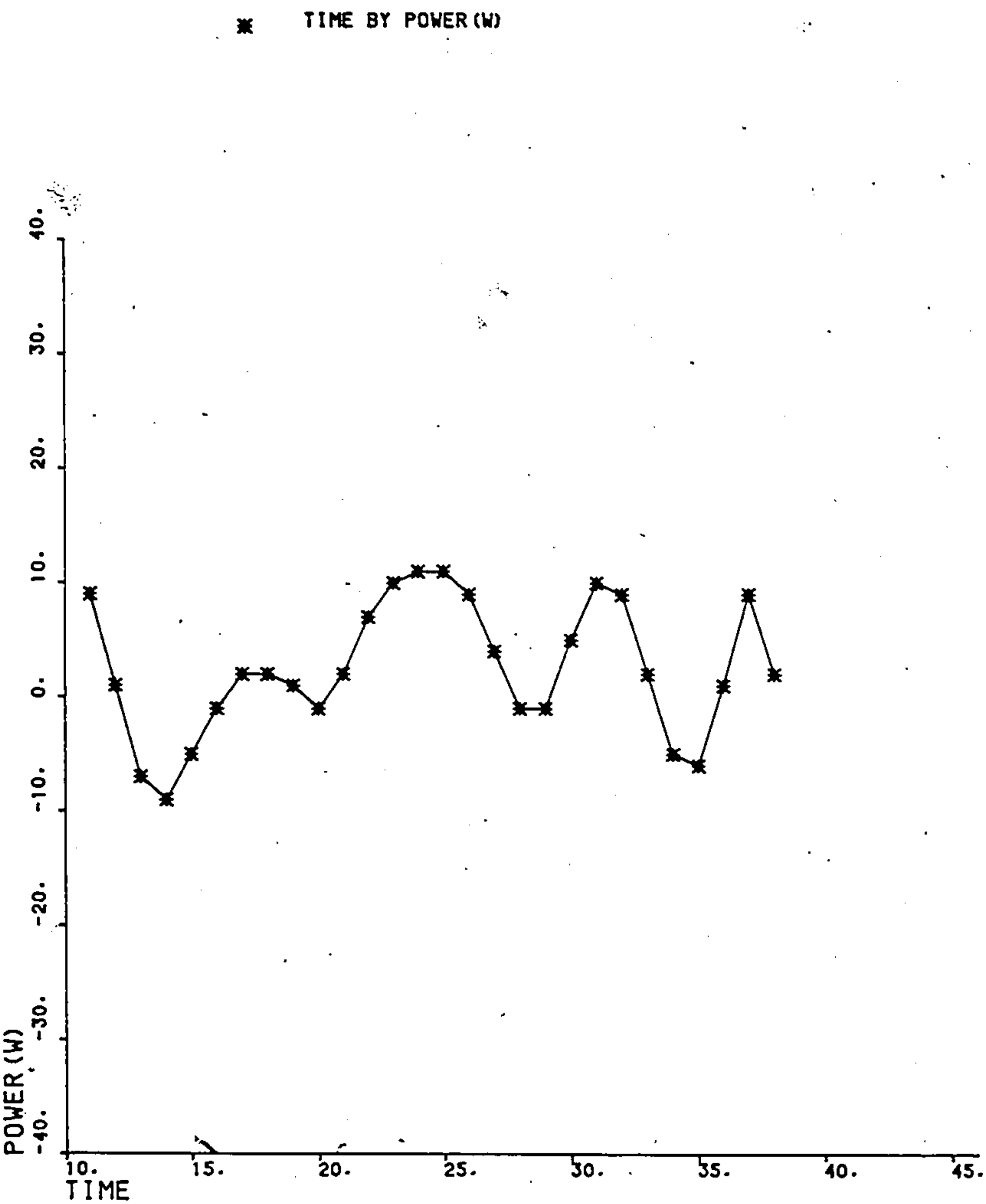
SUBJECT-TRIAL/SEGMENT: B01102 TRUNK



INSTANTANEOUS POWER LEVELS
SUBJECT-TRIAL/SEGMENT: B01102 HEAD-NECK