MODELING OF THE BIOMECHANICS OF POSTURE AND BALANCE

by

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Abstract: This research has developed a technique to be used in the study of the relationship of posture to balance. In order to investigate this relationship quantitatively, a model of the body was developed which treats the body as being made up of eleven rigid body segments, each with six degrees of freedom. The input to this model was wholebody kinematic data and force plate data provided by a bilateral Selspot/TRACK data acquisition system. The model was used to quantitatively evaluate "static" posture. The model was extended to permit the estimation of the position, velocity and acceleration of the body center of gravity and of the individual body segment centers of gravity. In this form the model is a useful tool for evaluating postural control, i.e. dynamic posture. With suitable technical modifications this approach should be useful in the study of the coordination of other complex movements such as gait. The accuracy of the system was evaluated using appropriate mechanical model "body segments", and by analysis of the reproducibility of human subject data.

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I wish to acknowledge that this work owes a great deal to the efforts of R. S. Fijan who, in the early days of the MGH Biomotion Laboratory, laid a very sound foundation upon which much of this work was built. Also, M. C. Murphy's adaptation of the Woltring b-spline smoothing technique to TRACK data processing was crucial to successful kinematic study of posture as it permitted the more subtile adjustments to be observed.

I wish to thank my wife, Merrill, for inspiring me to pursue this goal. I wish to thank my son David whose struggle to master standing and walking taught me the importance of the topic of this thesis. I also wish to thank my son Kevin for providing by his imminent arrival in this world, the impetus needed to complete this work. In addition I wish to thank my entire family for their patients during the long years at MIT.

CHAPTER I

INTRODUCTION

The goals of this research effort were to develop a model which quantifies the biomechanical relationship between posture and balance and to evaluate the model's potential usefulness in the clinical study of disorders of posture and balance. It is first necessary to define the terms posture and balance. In this study the term posture is defined as the position and orientation of the body segments. The term balance is defined as the control of the position of the body center of gravity [CG]. In the literature the term posture is sometimes used to refer to what is in this work called balance. For example, the use of force plates to study the movement of the center of pressure and make inferences about the movement of the center of gravity is called, among other things, posturography.

Given the definition above, posture must be studied primarily by the aquisition of kinematic data defining the position and orientation of the body segments. The kinematic data must be sufficiently complete. Since all body segments participate in defining the body's posture, wholebody kinematic studies are required. In this work the body is modeled as consisting of eleven segments. Each segment is considered a rigid body with six degrees of freedom. While it can be argued that this model is not completely adequate, that an even more detailed model is required, this model represents the most complete model of posture currently achievable. The usefulness of this model in quantifying the essential features of posture and postural control will become apparent.

In order to study posture, the kinematic data must also be sufficiently precise. Postural deformities are often comparatively subtle. The reactions to postural disturbances are also frequently quite subtle. In order to be useful in the study of posture the kinematic data must have a precision and resolution of on the order of one degree. The evaluation of the system accuracy will, therefore, be a recurrent theme in this work.

Posture in this study is defined as the alignment of the body segments. The term "static posture" herein refers to the alignment of the body segments in stance. Such posture is not truly static as force plate center of pressure data indicates that the CG and, therefore, the body are in constant motion. The term "dynamic posture" herein refers to the alignment of the body segments during a disturbance to static posture. The disturbance may be either active or passive. An example of an active disturbance to posture studied in this and other research is rapidly raising one arm. A passive disturbance to posture is a disturbance generated by the environment, e.g. the movement of the supporting surface. Passive disturbances may be anticipated and regular, e.g. the regular rolling of a ship, or anticipated but irregular, e.g. a ride on the MBTA Red Line, or completely unanticipated, e.g. an airplane hitting a air pocket. Although not the subject of this study, from the point of view of posture and balance, gait may be considered a series of active CG disturbances, i.e. deliberate coordinated body segment movements complicated by passive disturbances due, for example, to terrain irregularities.

In order to define an approach to quantitatively characterizing posture it is useful to presuppose what constitutes "good" posture. "Good" static posture can be characterized in part as posture which maintains the CG in an optimal location. For stable static stance the CG must be positioned so that its projection on the support surface lies within the base of support, i.e. the area bounded by the footprints. There are indications mainly from force plate studies the CG location is controlled within a much smaller volume over the center of support [15, 18, 25, 37, 49, 57]. Maintaining the projection of the CG near the center of support provides a margin of stability to accommodate disturbances.

"Good" static posture should be efficient, that is it should also somehow minimize the effort (muscle activation and therefore metabolic cost) and/or loading (joint torques and forces) required to maintain stance. An inverted pendulum model of stance is probably too simple, but it may help identify the most basic issues. One prediction of this model is that very small torques are adequate to maintain balance near the centered position, but correction of larger deviations from equilibrium requires disproportionately larger torques. We may hypothesize that the musculoskeletal system has evolved so that good posture not only minimizes the joint torques but also minimizes the muscle force required for small corrections or the maintenance of continuous small amplitude oscillations about the equilibrium position. Taken together the hypotheses that good posture maintains the center of gravity in a favorable position and that it minimizes joint loading and muscle fatigue provide a functional rather than an anatomical definition of good posture.

"Good" dynamic posture can be characterized in part as posture which controls the position of the body CG and of those CGs of the more massive body segments with minimum acceleration. This implies a minimization of net force [F=ma]. The response to a disturbance should minimize muscle action, joint forces and metabolic cost. In the case of normal gait, for example, the forward velocity of the body CG does not vary much from the mean velocity despite very significant changes in the velocities of the various limb segments. This has long been recognized as a significant determinant of efficient gait, but gait analysis has not in general attempted to quantify this characteristic of gait.

"Good" dynamic posture should also be functional. The position and orientation of the body segments are not only adjusted to control the CG location; they are also adjusted to facilitate function. When an arm is manipulating a wrench, postural control is essential to facilitate this action. When vision tracks a moving object, posture is continuously adjusted to position the eye relative to the inertial frame of reference. Thus while study of the movement of the hand and the arm is necessary to understand how the brain functions to control manipulation as is study of the vestibulo-occular reflex relative to visual tracking, in both cases a very interesting and vital part of the brain's control problem is neglected if overall postural control is not considered.

In summary, one criterion for defining good static and dynamic posture has to do with the control of kinematics of the CG, that is, the relationship of posture and balance. Another attribute of good posture is energy or metabolic efficiency. This is an extremely complex issue, in this work approached by evaluating the net joint forces and moments thus deferring the difficulty of relating the associated muscle activity to energy consumption. Finally there is the issue of posture control to achieve function, perhaps the most complex issue. This study is limited to observing quantitative and objective aspects of this last process.

This study postulates a simple relationship between posture and balance as its working hypothesis. Mechanically posture and balance are coupled; most alterations in the alignment of the body segments change the CG location. The brain is presumed to control the body segment alignment in order to control the CG location, i.e. the brain employs posture as a means of maintaining balance. While the working hypothesis of this research, it is not a critical assumption; other hypotheses or assumptions are possible. Much vestibular-based balance research assumes that the orientation of the head is being stabilized. Gurfinkel has hypothesized that the orientation of the trunk is being stabilized [12]. A significant advantage of the approach pursued in this work is that it is basically independent of these kinds of assumptions. Since the posture of the whole body is being evaluated, the results are equally meaningful whether the brain is stabilizing the CG, the trunk orientation, the head orientation, or all three concurrently.

On the surface, the relationship between posture and balance seems trivial. Each limb segment has a certain mass and its own center of gravity. Moving the limb segment a certain amount will cause a change in the body center of gravity which can be calculated simply from the total mass and the mass and displacement of the limb segment. If we look at the control problem from the perspective of the brain, however, we see that the problem is far more complex, even if we suppose that the brain adjusts posture only to maintain balance with no consideration of the alignment of the body segments. If the brain controlled the position and orientation of only one limb segment, balance control would be simple. When the CG is disturbed from its setpoint position by a given amount, the positionable segment would be moved an appropriate amount to bring the CG back to the setpoint. The brain, however, controls a large number of body segments. Holonomic or linkage constraints reduce the total number of degrees of freedom to much less than six times the number of segments. It is, however, safe to say that for a reasonable disturbance in CG position, the brain has a large repertoire of postural adjustments from which to choose, any one of which would restore the CG to its setpoint position. How the brain selects an optimal postural adjustment, what criteria it employs, are questions biomechanics and neuroscience are only starting to address. This selection process, called coordination, is clearly tremendously important. The first step in understanding any process is to be able to assess it quantitatively.

Most previous studies of balance have used force plates to measure the stability of the position of the center of gravity (CG) during postural maintenance with and without disturbances [9, 10, 14, 15, 18, 22, 23, 24, 25, 28, 30, 31, 32, 38, 39, 40, 41, 42, 49, 50, 51, 52, 57]. Electromyographic studies (EMG) have been used to determine when and where muscles have been activated to produce the torques necessary to control posture and balance [14, 23, 35, 36, 39, 40]. Limited use has been made of photographic techniques, goniometers and accelerometers to obtain kinematic data while studying posture and/or balance [12, 23, 32, 39]. However, since no systematic attempt has yet been made to obtain complete three dimensional wholebody kinematics, little has been done to correlate posture (some function of the position and orientation of the body segments) and balance (the control of CG position).

Clinical evaluation of posture has remained purely subjective based on anatomical criteria for "good alignment".

The manner in which we model posture and balance has clinical relevance in that it biases our interpretation of clinical data. In a recent study [40] of normal children and adolescents and an age-matched group of cerebral palsied patients it was shown that a certain type of postural disturbance elicited a pattern of muscle activation (shown by surface EMG) which was fairly consistent among the normals. Several different "abnormal" patterns were observed in the cerebral palsied population. The normal pattern was interpreted, applying the inverted pendulum model, as being effective in providing the necessary level of rigidity while the "abnormal" patterns were taken to be indicative of neuromuscular pathology. However, while the origin of cerebral palsy is a neuromuscular disorder, some patients develop orthopaedic complications such as contractures which affect joint ranges of motion and, therefore, posture and balance; thus the neuromuscular control system must regulate an altered system. Perhaps some of the "abnormal" patterns of muscular activity are in fact adaptive, i.e. appropriate to maintain the balance and posture of the modified system. If so, the cerebral palsied group whose current balance and posture problems are primarily orthopaedic rather than neuromuscular might benefit from different interventions and treatment modalities.

The inverted pendulum model of postural maintenance appears to be too simple to describe this complex process adequately. It assumes that balance is maintained by the ankle functioning as a torque generator and is therefore only valid where that is the case. EMG studies have shown that the "ankle strategy" (which can be equated to inverted pendulum type control) is used only over a limited range and that other more complex (multi-joint) strategies come into play [14, 23, 39]. The inverted pendulum model does not incorporate the body's ability to control actively the relative position of a large number of independent but linked segments. A standing person can alter the CG position without flexing the ankle simply by moving the head or arms or trunk. While a given change in the CG position could be produced by a very large number of different combinations of movements of the body segments, a smaller repertoire of movement patterns which are in some way optimal is in all likelihood employed to control the CG position and therefore to maintain balance.

This work attempts to provide a more inclusive representation of the system which models more of the salient features of balance and posture control. The model developed herein treats the body as eleven independently positionable segments, a number sufficient to represent most available movement patterns. The model includes estimates of the mass and the inertial properties of these segments so that the effect of their movement on CG position and joint loads and torques can be quantified.

In order to assess the potential clinical usefulness of the posture model the technique was used to evaluate the posture and balance of a group of children diagnosed as having cerebral palsy who were considered to have postural deformity and/or balance deficits. An age matched group of "normal" children was also studied. Specific clinical hypotheses were not evaluated in the course of these studies. Rather, the studies were used to explore the nature of the posture and balance problems present and to evaluate the kinds of information which might be obtained using this approach. Data from these studies will be presented here for the purpose of demonstrating the technique's application to real world situations. This work will assume that the precise quantitative evaluation of posture and balance is a worthwhile undertaking apriori and will not attempt to prove this point by, for instance, showing how an individual's clinical treatment should be modified based on the results of these studies. Clinical studies using this approach are important and are underway; they are not, however, considered to be a integral part of this work.

This research was conducted at the Massachusetts Institute of Technology [MIT] Newman Laboratory of Biomechanics and Human Rehabilitation and at the Massachusetts General Hospital [MGH] Biomotion Laboratory. All testing involving human subjects was conducted at the MGH Biomotion Laboratory.

CHAPTER II

THE CURRENT STATE OF POSTURE AND BALANCE RESEARCH

The scientific/technical and clinical literature on posture and balance is quite extensive. There is, however, little which addresses posture as defined here. Almost all quantitative studies, particularly in the clinical literature, have involved measuring the stability of the center of pressure [CP]. Clinically, postural deformity and balance impairment are associated with a wide range of neuromuscular pathologies such as parkinsonism, cerebellar ataxia, cerebrovascular disease, cerebral palsy, craniospinal injuries, vestibular disorders, organophosphate pesticide intoxication, alcohol intoxication and multiple sclerosis. In addition, there is evidence that in the elderly population there is a link between unexplained falls and occult balance deficits [7, 44].

Posture and balance have been studied using three basic technologies; 1)force plates and balance platforms, 2)electromyography [EMG], and 3)kinematic data acquisition systems, used alone and in combination. Both static balance and dynamic posture control and balance have been studied. Dynamic posture control and/or balance have generally been quantified by observing the CP excursions associated with human subject responses to "controlled" disturbances. As will be discussed below, in these studies posture control is assessed using very limited kinematic data and very simplified models. Alternatively posture control may be evaluated via the timing of EMG activity with little or no kinematic data. This study is based on the contention that approaches which ignore or greatly simplify the biomechanics are inadequate.

II.1 FORCE PLATE STUDIES

The primary application of force plates has been to quantify the variability of the CP location, the center of action of the ground reaction or support forces. In the static case the ground reaction force vector is the force vector obtained by integrating the support surface pressure over the whole support area (Figure 2.1). The CP is the point where the line of action of this vector passes through the horizontal support surface and which must be specified in two dimensions.

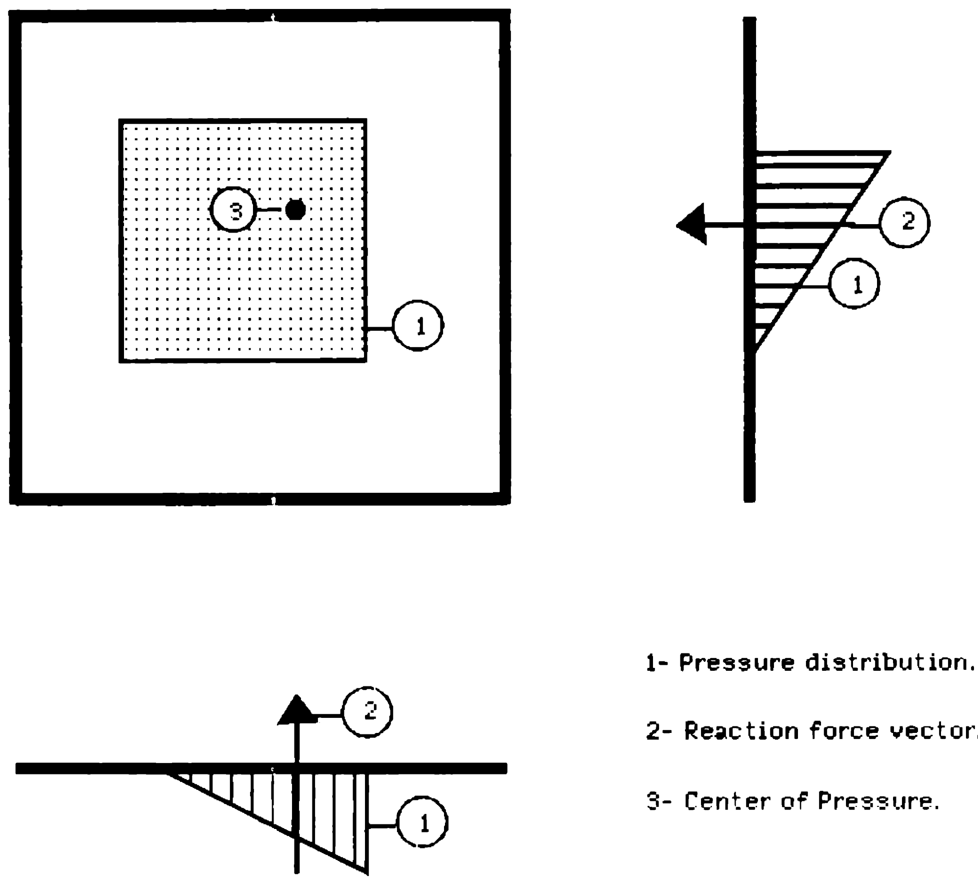


Figure 2.1 Center of Pressure

In the real non-static case the ground reaction has shear components in addition to the surface normal or pressure components. In this case the resultant vector is not vertical. The CP is still the intersection of the resultant force vector line of action with the support surface and is still specified in two dimensions.

The center of gravity [CG] is that point in space at which the entire mass of the body can be considered located and must be specified in three dimensions. The location of the CG is defined by the following integral equations:

$$X_{cg} = \int x \cdot \rho \cdot dv \,/ \int \rho \cdot dv$$

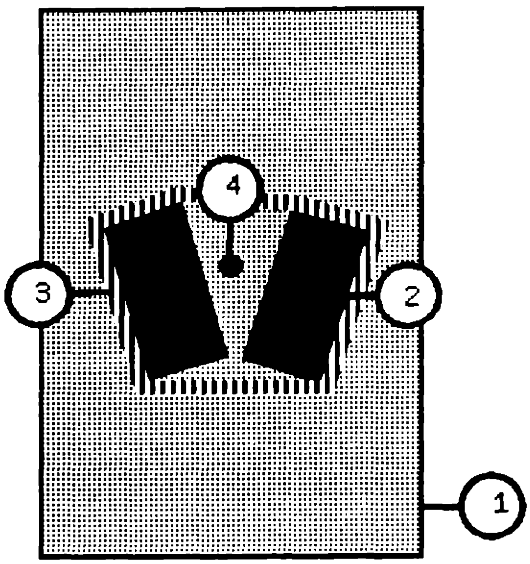
$$Y_{cg} = \int y \cdot \rho \cdot dv \,/ \int \rho \cdot dv$$

$$Z_{cg} = \int z \cdot \rho \cdot dv \,/ \int \rho \cdot dv$$

If stance were truly static the CP location would coincide with the vertical projection of the center of gravity location to the support surface. The inevitable CG motion is important because the net forces associated with linear motions of the body are a product of the body mass and the corresponding linear accelerations of the CG. Thus, acceleration of the CG implies net force generation and metabolic load. Stance or movement which minimizes CG acceleration minimizes this component of metabolic load.

In general, force plate balance studies postulate that balance is the control of the center of gravity position (as does this work) and that CP location and movement is a direct indicator of CG location and movement. Assuming that the CP is the projection of the CG neglects dynamic effects. Most authors consider this simplifying assumption acceptable in studies of "static" balance, but a few have used modeling techniques to attempt to refine the estimates of CG motion derived from CP motion information [30, 31, 50]. It can, however, be argued that the CP location rather than the CG location is the more appropriate parameter to use in evaluating balance; that balance is maintained if the CP is maintained within the base of support, i.e. within the foot print, regardless of dynamics (Figure 2.2).

Force plate CP studies sometimes claim to be characterizing "postural sway". The inverted pendulum model of stance is generally assumed implicitly if not explicitly, in these analyses. The "sway," i.e. the variation of the CP position, is assumed to be the result of a rigid body rotation about the ankle joint. There appears to be no real justification for this presumption; no definitive kinematic evidence is offered that relate the CP motion observed to ankle joint angle changes only. Given the large number of degrees of freedom provided by the numerous joints in the human body it seems highly unlikely that only one or a few are involved in the maintenance of balance.



1- Force plate.

2-Foot print.

3-Base of support.

4-Center of Pressure

Figure 2.2 Center of Pressure within Base of Support.

In some cases the force plates are equipped with actuators that permit controlled platform translation and/or rotation [14, 23, 28, 39, 40, 51, 52]. Such devices are called disturbance platforms. The platform is rotated or translated, disturbing the balance of the subject standing on the platform; the effect on CP position is then observed. Nashner and coworkers have used these devices to study posture control in normals and in subjects with neuromuscular pathology [14, 23, 39, 40, 52]. Young and coworkers have investigated the balance problems associated with prolonged space flight and exposure to reduced gravity [28, 61]. Since these tests actually attempt to evaluate postural control by observing how effectively the CP position is controlled, for reasons discussed below, they cannot assess balance. The result is a quantitative assessment of overall postural control at least as it relates to CP position control, with little information as to the posture control strategies used to maintain balance.

The results of the force plate CP studies are presented through a variety of techniques, all of which attempt to illustrate the variability of CP location. Graphic techniques include display of the CP path during trials of a given duration [see Chapter V] and 3-D histograms which show the amount of time the CP is located in a given position relative to the mean position [18]. Various statistical measures are also used to characterize the variability of CP position [9, 10, 15, 18, 25, 38, 41, 57], with the standard deviation of the CP position one of the most common and meaningful [9].

Frequency domain analysis of the data is also widely used [10, 32, 41]. The results of this type of analysis are frequently referred to as the frequency or power spectra of postural sway. Again it is difficult to justify this interpretation literally, in that there appears to be no kinematic evidence that the body is, in fact, rotating around the ankle joint, and that therefore this putative rotation has a frequency distribution that corresponds to the frequency distribution observed in CP position data.

There is, however, one very important result from the frequency domain analyses. All studies indicate that the frequency content of CP motion is concentrated at quite low frequencies, with most power below 5 Hertz and with significant peaks in the spectra in the fractional Hertz range [10, 32, 41]. This implies that balance can be assessed only by observing long time histories. If there is significant power near 0.1 Hertz then the time history must be at least ten seconds long to observe the complete spectra [5, 43, 55]. Since ensemble averaging in the frequency domain is frequently necessary to obtain an adequate signal to noise ratio [SNR], several times ten seconds are required. In this work 90 second trials were used.

These temporal considerations raise two significant implications for the disturbance platform type of studies. First, because the time course of disturbance studies are general quite short, a few seconds in length, they cannot assess balance. This may be a matter of semantics since even though the studies are called "balance" studies, the authors seem well aware that they are dealing with transient postural responses. There is, in fact, little overlap or confusion between the disturbance platform literature and the static force plate balance literature. The second implication has more practical significance. The disturbance platform studies are in fact observing a transient response of a system which is probably nonlinear and probably not at steady-state at the onset of the disturbance. There are several approaches to managing this problem. The researcher may attempt to observe the subject and initiate disturbances during periods when the subject appears to be at steady state [14, 23, 28, 39, 40, 51, 52], improbable since much balance activity is at an almost imperceptable level. Repetitive tests may be used to average out the variability of results due to the non-stationarity of balance. This raises difficulties with human subjects because fatigue and learning effects are being introduced which are equally difficult to assess quantitatively. Alternately, the researcher may employ disturbances which elicit responses that are of significantly greater magnitude than the movements maintaining normal balance. This is implicit in most disturbance platform research and is adopted in this thesis. A disadvantage of this approach is that it makes the study of subjects with very severe balance problems difficult. For such subjects normal balance maintainence activity is likely to involve quite large postural adjustments. Very large disturbances which perturb posture greatly would be necessary to insure that the disturbance response is significantly greater than the baseline activity. For obvious reasons with these subjects, just the opposite is done; the magnitude of the disturbance is reduced to compensate for subject impairment and minimize subject anxiety, physical stress etc. As a result with subjects whose baseline balance is poor, it is more difficult to determine what portion of observed activity during a disturbance trial is truly response to the disturbance and what portion is part of the subject's normal background activity. For this reason, the pretest "static" balance and posture of these subjects must be carefully assessed before undertaking disturbance response observation.

The quality of instrumentation used in force platform studies is highly variable. Precision six component force plates are generally used in research laboratory settings for static force plate CP studies. In clinical applications the instrumentation varies from precision six component force plates to very basic balance platforms with only two or three single axis force sensing elements [4, 6, 32]. The clinical community, at least in Europe, appears to recognize the need for standards [4]. Unfortunately, however, there appears to be a tendency to standardize on rather imprecise instruments. This tendency may well reflect the clinical community's inability to make full use of the results of these studies due to the current incomplete understanding of the biomechanics of balance. In the absence of an adequate biomechanical model to permit interpretation of the CP data, there is little incentive to obtain high quality data.

II.2 EMG STUDIES

The study of posture and balance using EMG is typified by the work of Marsden et al [35, 36]. EMG data were obtained from subjects performing various active balance disturbance maneuvers. For example, the subject would perform a rapid arm raise maneuver. EMGs were recorded fro the muscles involved in raising the arm and for various trunk and lower limb muscles. EMG activity which appeared to correlate with the arm raise maneuver was detect throughout the trunk and lower limbs. The onset activities in the trunk and lower limb muscles, which presumed to reflect postural control, often preceded onset of EMG activities in the shoulder muscles which rais the arm. These observations hint at the complexity postural control. A seemingly simple single limb moto:

is shown to actually involve the whole neuromuscular system. This dramatizes the point that a biomechanical study of posture and balance must be a study of the whole body.

These EMG studies raise more questions about the biomechanics of posture and balance than they answer. In particular what is the function of this wide-spread muscular activity; what is happening biomechanically? A great deal of muscle co-contraction is evident. This could imply that the rest of the body is stiffening so that the dynamics of the arm movement maneuver has a minimum effect on position. Alternately, the rest of the body could be performing a subtile realignment to compensate for the effect of the arm raise maneuver on, say, the CG position. EMG data alone cannot answer these questions. The basic contention of the research described herein is that only precise, whole body, kinematic data interpreted via a whole body model of posture can address these important questions.

II.3 KINEMATIC STUDIES

The kinematics of posture and balance have been studied using various approaches. Static posture has been assessed qualitatively using photographs of subjects standing in front of a grid, providing crude evaluation of the alignment of body segments [58]. A somewhat more quantitative approach photographs the subject standing on a balance beam. A vertical line called the line of gravity is projected throught the CP defined by the balance beam instrumented to define the CP in one dimension. The horizontal distance from this line to the estimated position of certain joint centers is then measured and factored into a postural index [58] based on the assumptions that the ground reaction force vector is vertical and that the joint centers should lie on the line of action to minimize joint moments (Figure 2.3). Tail [56] has developed a more technically sophisticated implementation of this technique which superimposes the force vector on video recordings of stance and gait.

One research group [24] has used kinematic data to study postural sway. A marker placed on the side of the pelvis is monitored; the change in its position relative to a known fixed ankle joint position is interpreted as postural sway. A planar inverted pendulum model is assumed with the body rotating as a single rigid link about the ankle in the sagittal plane, an overly simplified model considering the number of degrees of freedom which actually exist.

Kinematic data has also been used to evaluate disturbance responses, usually employing photography or cinematography. The positions of certain anatomical landmarks are then estimated manually from the photos, usually only in two dimension [12, 14, 23, 39, 40, 51, 52]. The studies of Soviet cosmonauts performed by Clement et al [12] are typical of this kind of study. Shortly before an

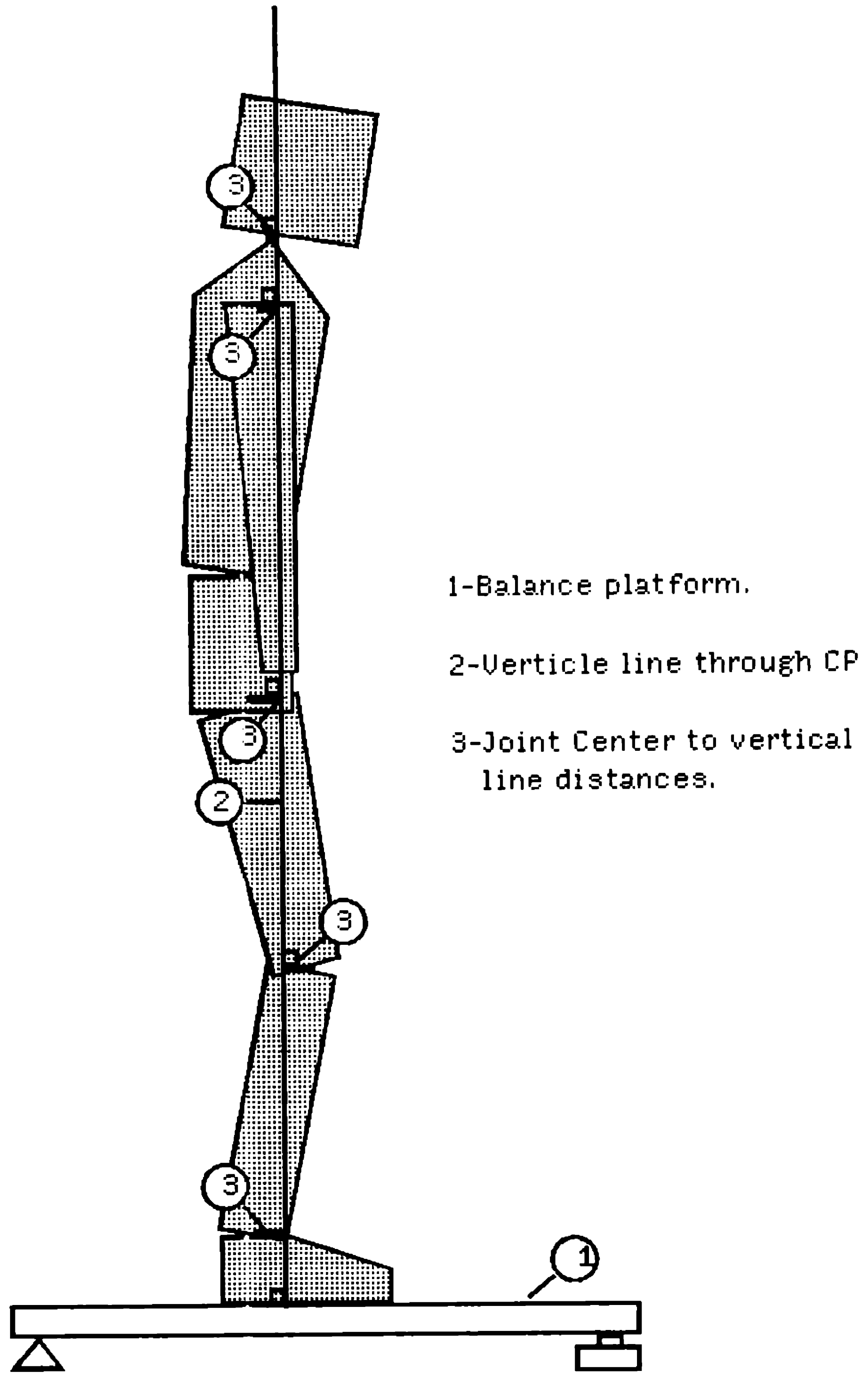


Figure 2.3 Posture Assessment Using Balance Beam Photograph.

extended space flight the subjects were filmed performing rapid arm raise maneuvers. From lateral views the movements of a number of markers on the lower leg, pelvis, arm and head were defined approximately in the sagittal plane. Similar recording were then made in space, under near zero gravity conditions, and again shortly after the subjects returned to earth. Since this maneuver corresponds to the right arm raise maneuver performed by subjects in this thesis, Clement's pre-flight data will be compared to our results for normals in Chapter V. Discounting questions of accuracy, the movements of only a limited number of markers (associated with anatomical landmarks) in two dimensions does not provide a very complete description of the subject's posture. Only very simplified quantitative biomechanical models, with a very few degrees of freedom, can be developed using this limited data. Since only a limited number of the body's many degrees of freedom are defined, it is not possible to say with certainty whether observed muscle activations (concurrent EMG studies are typical) were used to produce joint stiffening or segment movement.

II.4 MODELLING

The inverted pendulum model is, as has been indicated, implicit in much of the research to date. Nashner formalized this model and has expanded on it somewhat [39], treating compound as well a simple inverted pendulums. This description of posture is extremely simplified and is only appropriate to describe a very limited set of circumstances. Nashner and his coworkers have tested posture control using standardized disturbances which elicit inverted pendulum responses [14, 23, 39, 40, 51, 52]. However, even under these conditions, departures from the simple inverted pendulum response are observed and attempts are made to determine the extent to which the subjects respond with a "hip strategy" as opposed to an "ankle strategy." Under more natural, and variable disturbance conditions one would expect to elicit a multiplicity of strategies.

Hemami and coworkers have explored the control of posture using multilinked models of the human body [11, 19, 20, 21]. In general the body segments are modeled as links rather than solid bodies and holonomic constraints further reduce the number of degrees of freedom. Each joint angle is controlled by a simple control loop with a set point reference trajectory based on kinematic data obtained from various sources. The goal of the modelling is to evaluate how effective various joint angle control schemes are in causing the model to follow the set points. These models, although still greatly simplified, more accurately reflect the complexity of the control problem and can be said to be true posture models. This approach to controlling model biological systems is analogous to the technique for controlling multidegree of freedom non-biological systems, e.g a robot arms, by using individual position-intergral-derivative [PID] controllers to control each degree of freedom. Asada and Slotine [3] discuss the advantages and limitations of this approach in robot control, generally concluding that such an approach is workable, i.e. that it provides stable control in a broad range of cases, but is restrictive and does not adequately control the dynamics of motion.

The human body performs an extrodinary repertoire of movements, most of which are exquisitely coordinated. Further, the human body is controlled to respond adequately to a seemingly infinite variety of disturbances. Anyone who has observed the development of motor ability in very young children must acknowledge that a great deal of motor pattern learning is evident. However, it seems unlikely that tables of joint angle set points are stored away in the brain sufficient for the whole range of intended human movements and responses to disturbances. In order to explore the nature of human posture control the Hemami model must be extended considerably.

II.5 MOVEMENT CONTROL

The field of movement control, i.e. how the brain controls movement, is to some extent interested in posture. Interest centers on the question of what the posture control system stabilizes. There are several hypotheses. Posture may be controlled to stabilize the position of the center of pressure or the center of gravity, the working hypothesis of this work. Posture may be controlled to stabilize the head position. This hypothesis is implicit in models of the ankle joint angle control loop which include input from the vestibular system, as in Nashner's models. Gurfinkel has postulated that posture is controlled to stabilize the trunk position [12]. In order to test and choose among these hypotheses observation of the posture control system functioning under a variety of conditions will be necessary. The techniques developed in this work will make possible such studies.

II.6 CLINICAL APPLICATION

In the clinical realm forceplate or balance platform studies have been conducted to quantitatively assess balance impairment [15, 22, 25, 42]. These tests are commonly referred to as quantitative Rhomberg test because they are analogous to the qualitative Rhomberg test which has been a part of the neurological physical examination for over 100 years. Njiokiktjien [41] and Cernacek [10] review the clinical application of these techniques.

Disturbance platform studies are coming into use clinically. Recently, for example, a study was was conducted of normal children and adolescents and an age-matched group of cerebral palsy patients using the disturbance platform techniques [40]. The children were subjected to standard postural disturbances under varying sensory feedback conditions. The CP position deviations were observed and lower limb kinematics were estimated in two dimensions from film records. The control of the CP position was less precise in the cerebral palsy population. Coordination aspects of posture control were evaluated primarily by analysis of the EMG data. Postural disturbances elicited patterns of muscle activation (shown by surface EMG) which were fairly consistent among the normals. Several different "abnormal" patterns were observed in the cerebral palsy population. The "normal" pattern was interpreted, employing the inverted pendulum model, as effective in providing the necessary level of rigidity, while the "abnormal" patterns were taken to be indicative of neuromuscular pathology. While cerebral palsy is a neuromuscular disorder some patients develop orthopaedic complications such as contractures which affect joint ranges of motion and, therefore, posture and balance; thus the neuromuscular control system is called upon to regulate an altered system. Perhaps some of the "abnormal" patterns of muscular activity are in fact adaptive, i.e. appropriate to maintain the balance and posture of the modified system. If so, the cerebral palsy group whose current balance and posture problems are primarily orthopaedic rather than neuromuscular might benefit from different interventions and treatment modalities.

Similar studies have been conducted comparing normal children to children with Downs syndrome [52]. The results of these studies were qualitatively similar to the results obtained in the study of cerebral palsied children, the CP was observed to be less well controlled during the disurbances and the patterns of muscle activation were abnormal. The clinical conclusion of this study was that the Downs syndrome population would probably benefit from physical therapy training to improve their postural control stratagies. However, more complete kinematic information must be obtained before it is possible to assess the extant postural control stratagies and determine what improvements are needed.

One of the goals of the disturbance platform testing is to determine the role of various sensory information sources in posture and balance control and for those with neuromuscular pathology to identify possible deficits in sensory integration. The platform kinematics are designed to generate ankle joint angles (hence proprioception) which do not match the body's true orientation (here again the inverted pendulum model is assumed). Additionally, the visual surround is manipulated to provide false visual clues as to orientation [14, 23, 39, 40, 51, 52]. The presumption, if inverted pendulum behavior is assumed, is that control is based on vestibular function in the absence of correct visual and ankle joint input [39]. However, these tests seem to ignore a source of information which may be very important. The scientific and medical communities now evaluate balance by observing the motion of the CP. It is possible that sensory input on the CP position contributes to the control of balance. The brain may be "aware" of the center of pressure location based on the numerous mechanoreceptors on the soles of the feet and within the foot structures. It seems likely that this information is used for posture control and balance in at least certain regions of the control system bandwidth requirements. While generating false CP feedback would be quite difficult, such tests are needed if issues of sensory integration are to be completely evaluated.

CHAPTER III

KINEMATICS OF POSTURE

Since posture is defined as the alignment of the body segments, it follows that posture is quantified by measuring the alignment of body segments. In order for this quantification to be complete two conditions must be met. First, the body must be modelled by a number of body segments sufficient to adequately describe the significant features of posture. Second, the position and orientation of each body segment must be described completely and precisely so that changes in alignment effecting posture and balance can be detected and quantified. This chapter will describe the techniques developed to meet these conditions.

Determining the number of body segments necessary to adequately describe posture is not a simple task. The number of segments used by various researchers depends on the effect being studied or modeled. For instance, the inverted pendulum model of posture treats the body as a two segment, one joint, one degree of freedom system. The body above the ankle rotates about the ankle joints as a rigid body while the feet are a stationary base of support. On the other extreme, groups studying spinal mechanics treat each vertebra, at least each lumbar vertebra, as an independent segment.

For the purposes of this research a body segment is significant to posture if its position and orientation are independently controllable and if its mass and inertial properties are such that the position and dynamics of the segment will significantly affect the center of gravity and/or the center of pressure. If the position and orientation of a possible segment are fixed relative to another possible body segment then these two segments are not independently controllable. Since the body segments form a linkage, their positions and orientations are not, of course, completely independent. Defining the position and orientation of the pelvis and shank, for instance, defines the position and orientation of the thigh. However, the orientation of the thigh can be controlled relative to the pelvis or the shank in at least some degrees of freedom. The thigh does not have to move as a rigid body with either the shank or the pelvis. The fingers or toes are independently controllable according to this definition, but the movements of these segments are not likely to significantly effect the position of the center of gravity or the center of pressure; at least not within measurement ability. The orientation of the head, on the other hand, is independently controllable and would significantly effect these parameters. The head, therefore, must be considered as a body segment in this study.

This research has taken an empirical approach to defining the necessary number of body segments. This does not mean that the number of body segments was determined experimentally. Rather, the number of body segments employed is determined by the the requirements of the posture model under development and by the limitations of the kinematic data aquisition system. That is, while the number of segments chosen should adequately describe the possible body configuration modes, it makes no sense to define a segment whose kinematics cannot be measured because, say, array fixation is impossible. It also make little sense to define a body segment if no data is available to permit estimating the mass or other properties of that segment since that segment could not be adequately modeled. It follows that if a possible mode or adjustment of body configuration is not described by the set of segments chosen, one must preclude the use of that mode or at least define the experiment so that the overall results are unaffected by adjustments in that mode.

These criteria lead to an eleven segment model to describe the body's posture (Figure 3.1). The segments are the right and left feet, shanks, thighs and arms; and the pelvis, trunk and head. The trunk above the level of the second lumbar vertebra is treated as a single rigid body. Each arm is also treated as a single rigid body. Each such segment is treated as a rigid body with six degrees of freedom. Ten joints exist between the segments. However, no holonomic constraints are imposed, i.e. no linkage constraints are defined. Each segment is permitted six degrees of freedom. The entire system, therefore, has a total of sixty-six degrees of freedom.

III.1 DATA AQUISITION SYSTEM

In order to measure the kinematics of eleven body segments, one must have at least one marker array mounted on each segment. Because posture is not static, data to define the position and orientation of all segments must be obtained essentially simultaneously by sampling at an adequate rate. For example, it would not be acceptable to acquire data for one side of the body and determine the segment alignments, then look at the other side while assuming the the alignments of the first side remained unchanged. As a practical matter this means that each array must be observed by the detection system in every, or nearly every, sample cycle and that sample cycles must be

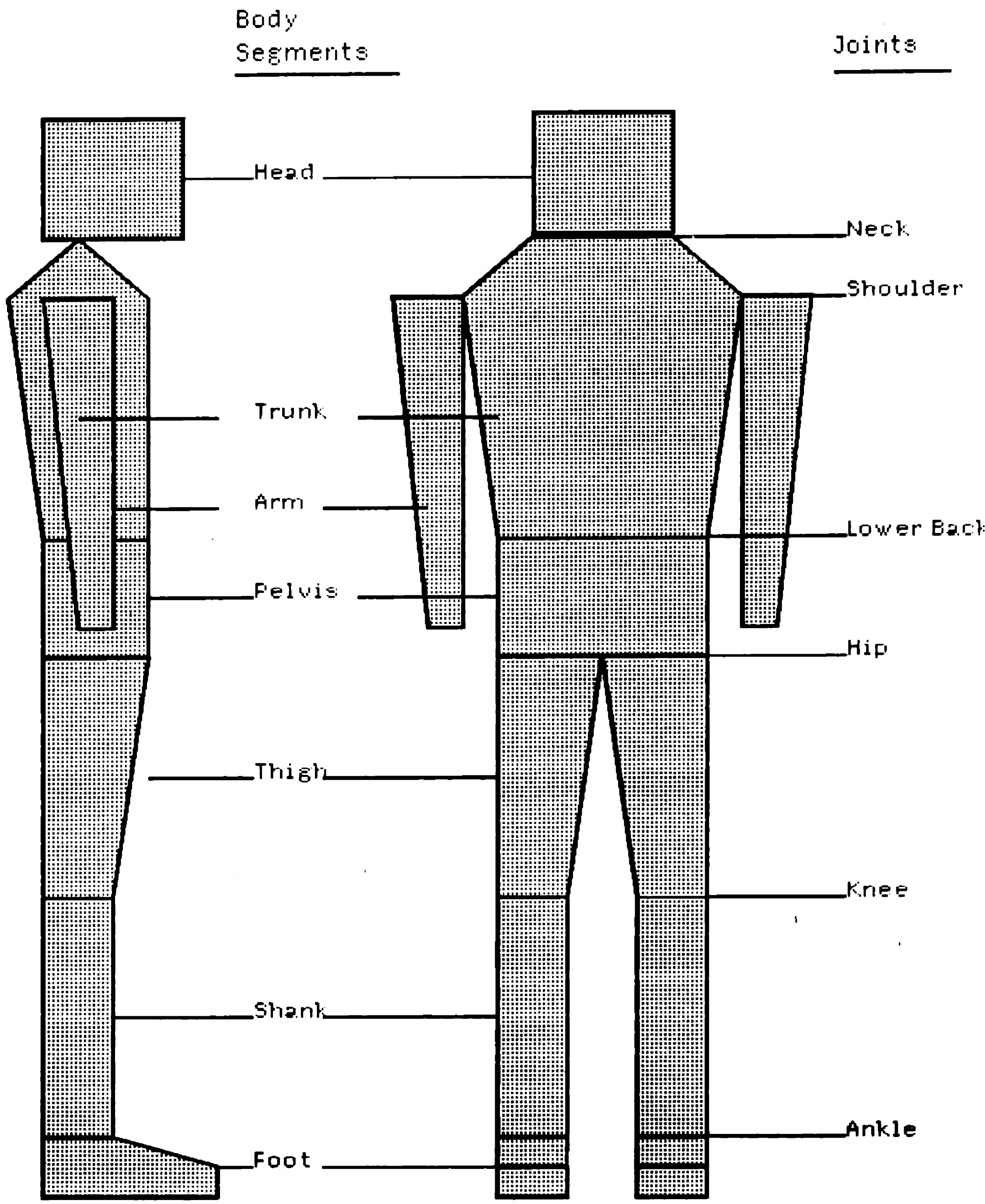


Figure 3.1 Posture Model

sufficiently brief so that the sample frequency is at least twice the frequency of the highest significant component of the kinematic activity of the respective bodies. More realistically the sample frequency should be about ten times the frequency of the fastest component. While most postural adjustment activity appears to occur at frequencies below five hertz or so, it is more conservative to assume that activity can have components on the order of the frequencies seen in gait, that is up to fifteen hertz or so [2]. These requirements imply that the kinematic data aquisition system must have the capability of tracking eleven body segments at a sampling frequency on the order of 150 hertz.

The bilateral Selspot II kinematic data aquisition system at the Massachusetts General Hospital Biomotion Laboratory permitted meeting these requirements. Features of the laboratory relevant to this research are described in detail in Appendix A. Figures 3.2 and 3.3 show the principle features of the laboratory in diagram and block diagram form. Here a brief summary of the laboratory facilities will be given before describing the instrumentation developed specifically for this research.

The basic features of the Selspot kinematic data aquisition system and the **TRACK/NEWTON** processing system used by the MGH Biomotion Laboratory are described by Antonsson [2]. Antonsson specifically described the Selspot I system while a Selspot II system was used for these

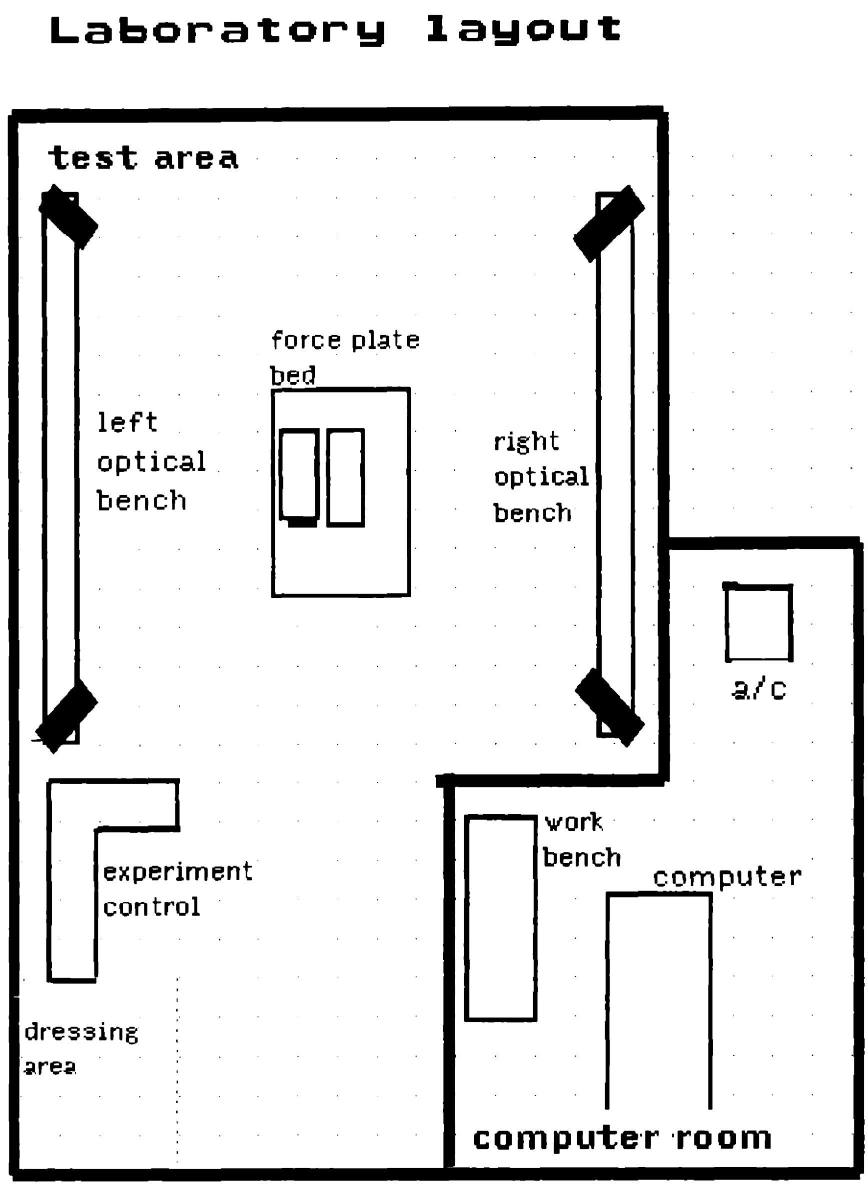


Figure 3.2 Biomotion Laboratory Diagram

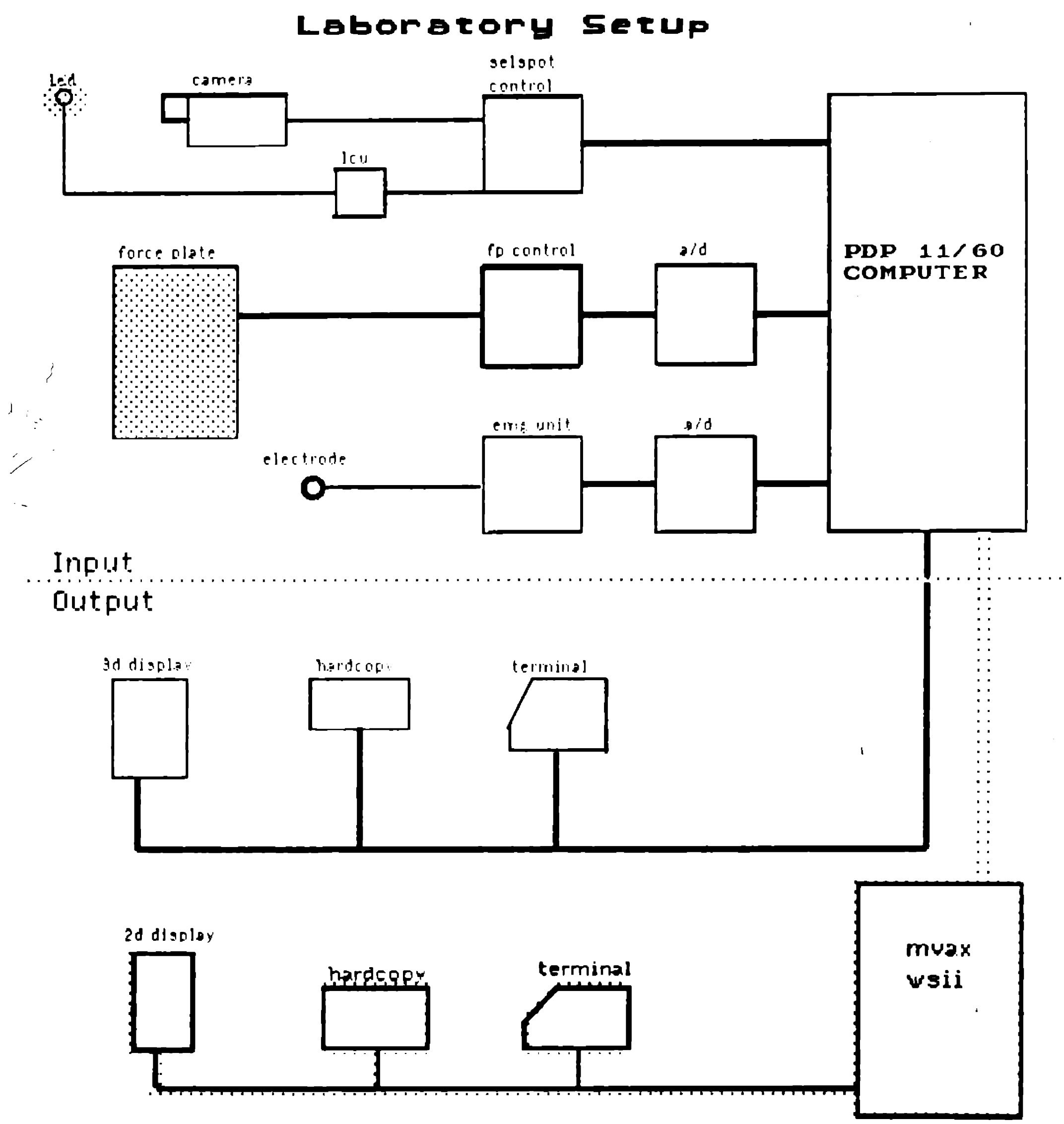


Figure 3.3 Biomotion Laboratory Block Diagram

studies (see Appendix A). The laboratory is bilateral, that is, it is equipped to view both sides of the body simultaneously. Four Selspot II cameras are used for this purpose. The cameras are configured as two sets of two cameras. Each set of cameras is mounted on an optical bench, one on either side of the study area. Two Kistler piezoelectric force plates are located approximately in the center of the study area. Their function will be discussed later.

The Selspot II cameras are really infrared detectors. The output of the cameras is the digitized (12-bit precision) 2-D position of the center of intensity of the infrared source as seen on each camera image plane. TRACK employs arrays of infrared light emmiting diodes (LEDs) used as infrared sources. The LEDs are sequenced by the Selspot system so that only one LED or cluster of LEDs is seen by the system during a given sample interval. The fundamental sample frequency of the Selspot system is 10,000 hertz. The sample frequency for a given set of LED is then:

$$Freq_{sample} = 10000 / (N_{LED} + 1)$$

In this study a total of sixty-four LEDs were used resulting in a sample frequency of slightly over 150 Hertz.

In order to derive enough information to define movement in six degrees of freedom from an array of LEDs, the array must contain enough LEDs or clusters of LEDs to define a plane, i.e. at least three. Better results are obtained if excess LEDs are present. Most arrays used in this study contained five LEDs.

The two Kistler piezioelectric force plates are located in the approximate center of the testing area. The are mounted so that their locations can be varied independently. This permits the forceplates to be located in positions optimal for the specific type of test being conducted. The force plates are located side-by-side near the center of the camera viewing volume for posture and balance tests. The vertical and horizontal components of the ground reaction force are measured to an accuracy of one percent of full scale. The data is diqitized to twelve bit precision. This permits the ground reaction force vector to be completely defined in the coordinate system of each force plate. Once the location and orientation of the force plates are determined in the camera ("global") coordinate system, a simple transformation permits the ground reaction force vectors to be defined in that coordinate system. Calculations of combined parameters, e.g. the combined center of pressure, are also then possible.

III.2 ARRAYS AND ARRAY MOUNTING

The instrumentation designed specifically for the posture study consisted primarily of the sets of LED arrays and associated mountings used to define the position and orientation of the eleven body segments. To track eleven body segments at least eleven arrays must be used, one on each segment. In this study thirteen arrays were used. Two arrays (left and right facing) were used on the pelvis and trunk segments. This was done primarily to assess how effectively separate left side and right side data are transformed into the common global coordinate system.

To some degree having each side see a complete body at least from foot to shoulder also simplified the processing and display of results. Joint angles for one side of the body, for example, could be determined from only a single data file for that side without reading data from the opposite side data files. This was important when all data processing was being done on the Biomotion Laboratory's PDP 11/60 computer because of that machine's rather severe memory constraints.

Figure 3.4 illustrates approximate array placement. The arrays used are described in detail in Appendix B. In general, the arrays were made from plexiglass sheet 1/8 inch or 1/16 inch thick, depending on the application. The array surface was planar and painted with a low reflective black paint to minimize reflection of infrared light from the array's own or other array's LEDs. The array mass was reduced by machining away excess material from the back surface of the plexiglass piece. The front surface was maintained flat, however, to minimize edge reflections which seem to occur when dealing with infrared. It must be stated that it is very difficult to determine what type of reflections most significantly degrade the data. The LEDs are mounted in the array so that the highest point of the lens is just flush with the array surface. This effectively recesses the LEDs to minimize self reflection from the array surface, but minimally reduces their field of visibility.

Initially, the foot, shank, thigh, and pelvis arrays were fixed to their associated body segment using high density polyethylene pieces molded to fit and move with the body segments. Where possible the polyethylene molds were contoured to fit over and be held in place by boney prominances. They were secured with elastic bands. Cut-out reliefs were provided so as not to interfer with muscle buldge or tendon lines of action.

This arrangement had several disadvantages. For static posture studies the principle disadvantage was the bulk of the molds, particularly the thigh mold, which slightly distorted normal posture. In dynamic posture control studies the molds tended to interfer with segment motion. As a consequence during test involving large ranges of

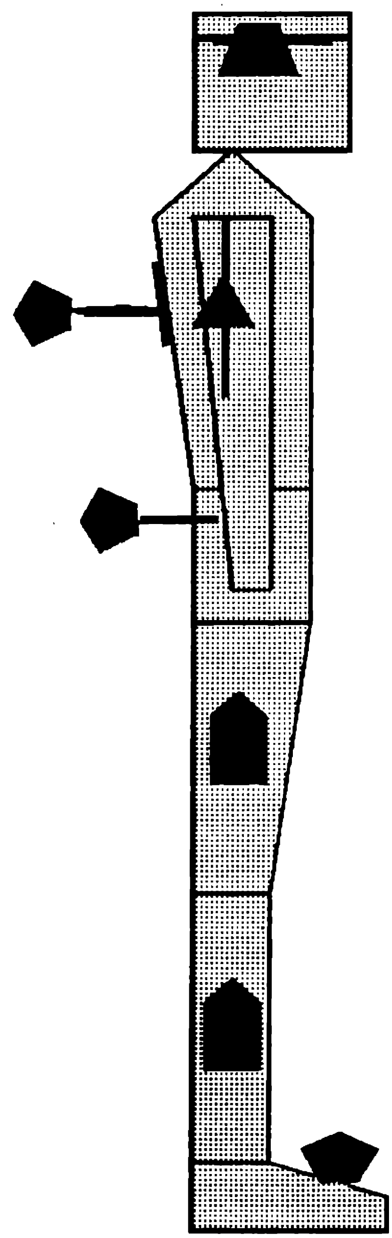


Figure 3.4   Lateral View with Arrays

motion such as rising from a chair, the molds either restricted the subject, producing unnatural movements or moved on their segment changing their relation to the joint centers and invalidating the initial condition determination (see below). For this reason the molds were replaced by an elastic lower-body suit. Small molded plastic plates under the suit provided the actual array fixation. The suit served the fix these plates to the skin. The plates were mounted on sites selected to minimize soft tissue movement. This arrangement provided better, although by no means perfect, array fixation while interfering much less with the subjects normal range of motion.

The pelvic arrays were mounted to the pelvic mold via a outrigger. The outrigger was used to prevent line-of-sight obstruction of the arrays when the arms hang naturally at the side. The trunk arrays were also mounted on their base mounting plate by an outrigger. Again this was done to make these center-line mounted arrays visible to cameras on both sides of the subject. Initially plexiglass outriggers were used reinforced to increase their stiffness. Later aluminum outriggers were employed stiffened by bending the upper and lower edges to develop an "S" cross-section. Because the trunk and pelvis arrays were mounted on outriggers it was especially important that their mass and inertia be minimized. For this reason these arrays were made of the thinner 1/16 inch thick plexiglass and as much material as possible was removed from the back surface of the arrays.

The mounting base of the trunk arrays presented a particularly difficult problem. The thoracic spine is really rather rigid and can be considered to move as a rigid segment. However, there are few boney prominances suitable for fixation in the rigid area of the spine. Further, although the spine and posterior portions of the rib cage are comparatively rigid, these subcutaneous structures are in great part covered by the scapulae. The scapulae are part of the shoulder girdle and translate a great deal relative to the spine. Initially, a mounting plate sized to fit over the thoracic spine between the scapulae was used to mount the trunk arrays. The inner surface of the plate was covered with a plastic foam to improve congruence to the body and to provide a more adherent surface. An elastic band around the thorax holds this plate in place. The band was fixed to the plate by a standoff block so that the stretched band clears the scapulae. Movement of the scapulae under the band does not move the band itself and, hence, does not disturb the array mounting. Excess material is removed from the mounting block to minimize weight. Additionally, suspenders are used to keep the mounting from sliding down the trunk to a lower, less stable position. The suspenders are crossed in the back above the mounting plate and cross the shoulders at the base of the neck where they are minimally disturbed by movement of the shoulder girdle or neck.

Subsequently, a molded polyethlyene mount was employed to fix the trunk arrays. The mount has three wings. One vertical wing cover approximately the same area over the thoracic spine as the original mounting plate. The other two wings wrapped around the lower rib cage. These were about three inches wide with an elastic front closure to allow for expansion and contraction with respiration. Elastic shoulder straps were employed as with the previous mount. This arrangement was more convenient and appeared to provide a more stable base for the array outrigger.

The arm arrays are mounted on polyethylene bars which for children extend from the top of the shoulder to below the elbow joint. Ace-bandage-wraps are used to secure the bars to the arm. By extending the wrap across the elbow joint, the joint is immobilized making the arm a single segment corresponding to the model. The upper portion of the bar is contoured to just engage the curve of the shoulder without interfering with shoulder abduction, preventing the assembly from sliding down the arm during activity.

The head array is mounted using a plastic head band which is adjustable to a broad range of sizes. The band includes a crown band which is also adjustable. There is only one head array which is visible to the right side cameras.

III.3 INITIAL CONDITIONS AND JOINT CENTERS

In order to study posture one must precisely specify the initial conditions. In terms of the mechanics of data analysis, initial conditions are the transformation from a given array coordinate system to the corresponding body segment coordinate system. This transformation involves a translation from the origin of the array coordinates to the origin of the body segment coordinates. The origin of the array coordinates is defined for each array by the segment file, the description of the array geometry used by the **TRACK/NEWTON** system. In this study the origin of the body segment coordinate system was either the proximal or distal joint center. The proximal joint center was used for the feet, arms and head. The distal joint center was used for the shanks, thighs, pelvis and trunk.

The transformation also involves a rotation from the array coordinate system to the segment coordinate system. This rotation might have been defined by specifying a standard method of array mounting. For example, the y-axis of the array might be visually aligned with the long axis of its segment and the plane of the array might be aligned to indicate the internal/external rotation of the array. This approach was not used for two reasons. First, it did not appear to provide sufficient accuracy; it is difficult to say how accurately a small array could be aligned with a segment. Second, when studying subjects with severe postural deformity, it proved to be physically impossible to align the arrays with the body segments and simultaneously keep then visible to two cameras. Indeed, in such cases it is difficult to achieve an array configuration which provides an adequate range of visibility. For this reason alone, one must maintain maximum freedom in aligning the arrays.

If the rotation from the array coordinates to the segment coordinates is to be arbitrary, the transformation information must be specified to the system. All degrees of freedom need not be specified explicitly. For example, for most segments the location of the proximal and distal joint centers defines the major axis of the segment and only one additional angle must be specified. The method used to specify the undefined rotation utilized a large alignment/pointing array. "Large" means a major axis approximately 0.2 meters in length and a minor axis approximately 0.09 meters in length; approximately twice the size of the typical segment-mounted arrays. The array was fitted with alignment aids to facilitate identifying its axes and aligning these with the body segments. While this approach still required the "eye-ball" alignment of an array to a segment, the large size and design of the array helped the accuracy to the process. This approach is considered at least as good as any other contemporary clinical angle measurement technique. Based on repeated application to the same subject by the same clinician, this technique is repeatable to between one and three degrees.

The location of the joint centers is a separate, but related issue. Since the body segments are modeled as six-degree-of-freedom rigid bodies, there are a number of possible methods for determining each segment's center of axis of rotation from the segment kinematics and estimates of its velocities and accelerations [17, 46, 47 48]. However, as Fijan pointed out [17], the available methods are sensitive to data noise, and require minimum changes in the kinematics and/or minimum angular velocities to achieve reliable and repeatable results. While there is no such thing as purely static posture, the degree of movement seen in normals is quite small, frequently nearly imperceptible. Even subjects with severe balance impairments move comparatively little. Our data shows that the maximum changes in given joint angles for undisturbed five second trials are usually only a few degrees. Under these conditions it is not feasible to use instantaneous axis of rotation methods to define the joint centers.

Since it is not practical to determine instantaneous axes of rotation or joint centers, the joint center locations must be approximated. Two methods of locating approximate joint centers are used in this study. In one method the alignment/pointer array is used to indicate the location of anatomical land marks from which the approximate joint center location is inferred and the segment alignment is derived. This method yielded acceptable (repeatable) results only when performed by highly trained persons.

This basic approach places the point of the large pointer array on the lateral surface of the limb segment at a location which, based on the anatomical landmarks and knowledge of "normal" anatomy, approximated the joint center. The circumference of the limb at this location was determined and the radius estimated by dividing by 2Pi. The joint center was then estimated to be one joint radius medial to the locator array point. This basic procedure was developed by Fijan. The method is illustrated in Figure 3.5. This method was used for locating the ankle, knee, neck and shoulder joint centers. Similar techniques were employed to find the hip and back joint centers. However, in these cases the error in estimating the joint radius from the circumference was found to be excessive; therefore, modified approaches were developed. For the hip, the pointer array was used twice, once to define a plane through the greater trochanter in the direction of the femoral head, and a second time to point at the estimated location of the femoral head from the frontal plane (Figure 3.6). In the case of of the back, the joint center was defined as the mid point of a line connecting two pointer arrays, one placed on the right illiac crest the other on the left. The repeatability of this procedure for the hip joint was no better than two centimeters when performed by skilled persons. The results were highly variable when these procedures were attempted by persons less familiar with skeletal anatomy or the concepts underlying the technique.

In the present method the alignment/pointer array is used only to define sagittal planes approximately through the centers of the joints. Separate trials in which the joints undergo comparatively large ranges of motion are used to define average joint axes of rotation. These axes of rotation are defined by the kinematic data; no human intervention is involved except requiring the subject to perform a suitable maneuver and selecting suitable data for kinematic analysis. The maneuvers used yield axes of rotation which are approximately normal to the sagittal plane. A joint center is then defined by the intersection of the axis of rotation with the joint sagittal plane defined by the pointer array(Figure 3.7). This approach minimized the error in joint center location due to personal

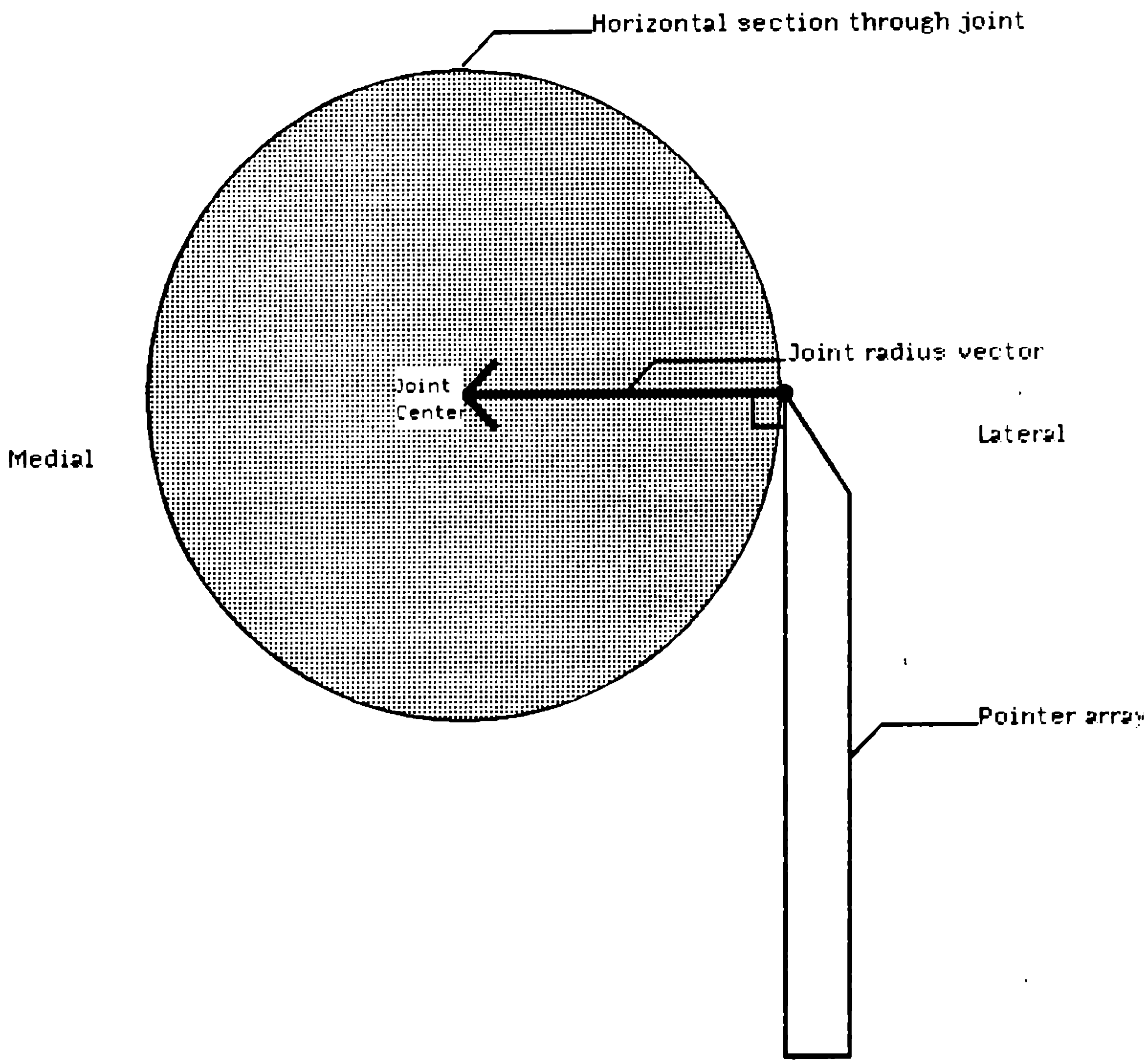


Figure 3.5 Joint Center from Anatomical Landmarks

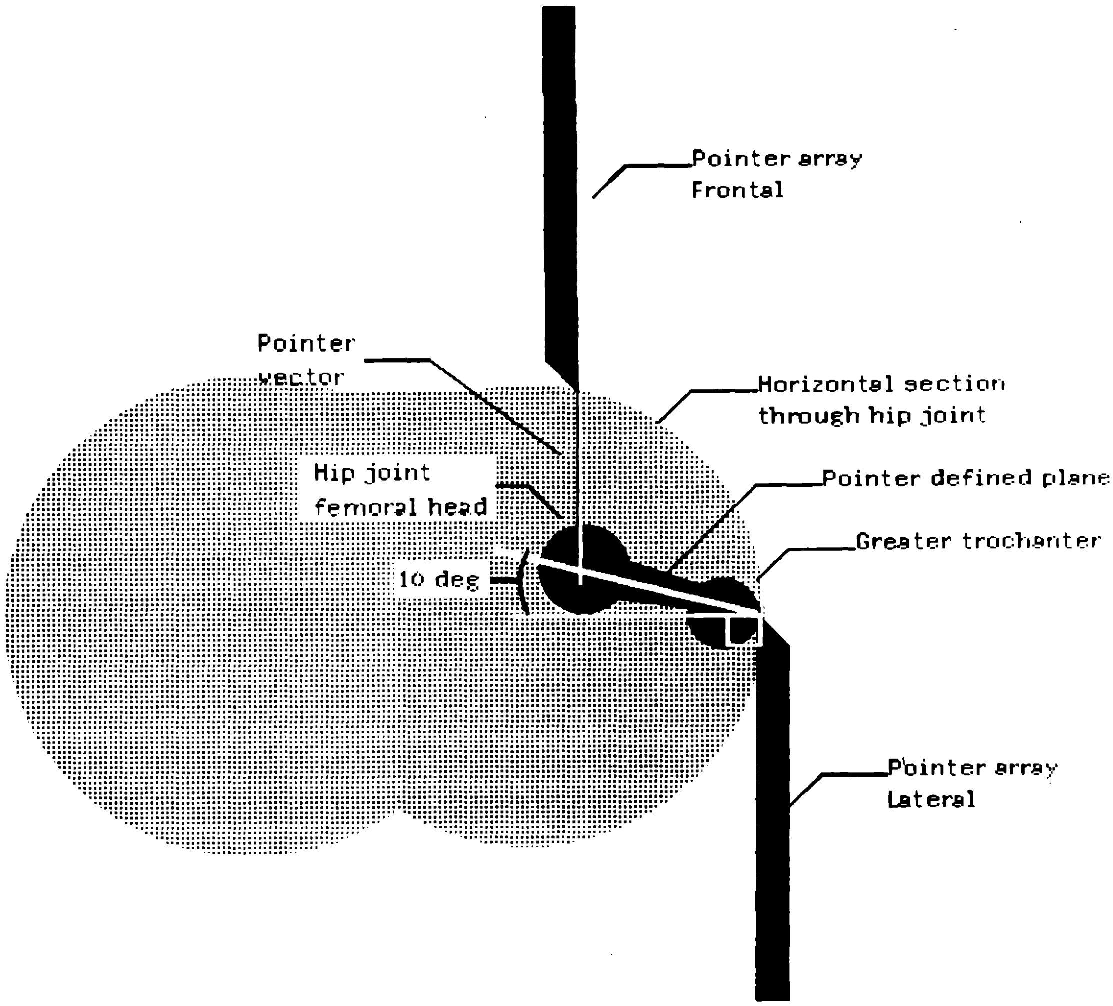


Figure 3.6 Hip Joint Center from Anatomical Landmarks

ability since the person manipulating the pointer array was only required to be able to define the approximate center plane of the joint.

This second approach was suggested by Fijan [17] but was first implemented in this study (Appendix C). Fijan's finite displacement method for locating the axes of rotation was used. Axes of rotation were determined for the ankles, knees and hips. Generally the necessary range of motion was obtained by observing the subject rising from a chair. Usually one such maneuver was adequate to define all six axes. However, provision was made for using several sets of data in which different maneuvers were performed in order to obtain the necessary axes. While the finite displacement method finds an average rather than an instantaneous axes, care was taken to use the smallest possible range of motion and to use that portion of the range of motion where the normal stance position was being approached. In this way it was possible to obtain reasonable estimates of the axes of rotation appropriate to the normal postural adjustment movements. Fijan showed this method to be repeatable to an accuracy of better than 1 cm for the knee; Lesczynski showed a similar level of reproducibility for the ankle [33]. The method did not, however, yield reproducible results when applied to the back, shoulder or neck joints. The movement of these multi-segmental joints is quite complex and obtaining reproducible results would have required

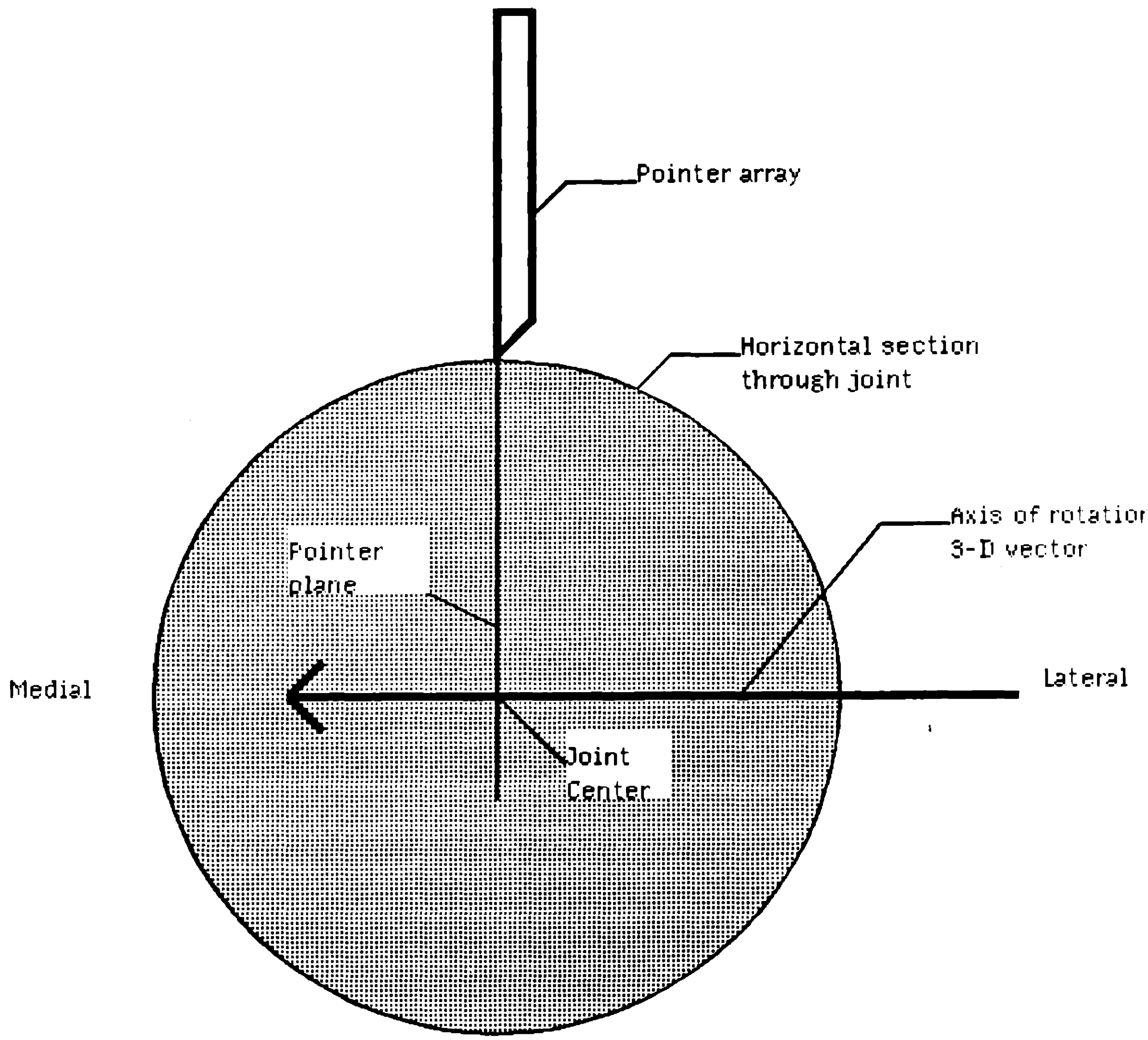


Figure 3.7 Joint Center from Axis of Rotation

developing means of constraining the movements. Axes of rotation defined using such constrained movements may not be valid for actual postural adjustment movements. For this reason the first method was used to define the joint centers for these complex joints.

In either case, once the joint center is located using the special trials, its position relative to the segment array is assumed to remain constant through out the series of trials. This assumption makes adequate array fixation vital. Since the segments have six degrees of freedom, it is a simple matter to determine qualitatively if the arrays have moved significantly and the joint center determination is no longer valid. If this occurs the segments are no longer conjugate; i.e. if the thigh array slips down during a trial then when viewing the data on a 3D display the upper thigh will no longer meet the pelvis and the lower thigh will extend into the shank. Unfortunately, given current processing capabilities, it is not possible to detect this effect in real-time; i.e. at the time of data aquisition.

III.4 ACCURACY OF KINEMATIC DATA

Antonsson [2] discussed the sources of error in TRACK data and estimated their magnitude and the magnitude of the total system errors. He showed that the system is capable of yielding accuracies of approximately 1 mm in position and

1 degree of rotation under the condition of this study. This level of accuracy is adequate for the study of posture.

For the data acquired in this study, however, there is an additional source of error. In a four camera bilateral system the position and orientation of all cameras must be specified in a single reference coordinate system. To assemble a complete wholebody representation of posture, right and left side kinematic data must be combined. In this process any errors in the position and orientation of cameras relative to the reference coordinate system will result in relative position errors in the assembled body segment configuration. For example, if there was a two centimeter error in the location of the left side cameras in the reference coordinate system X-axis with no other errors the position of the segments defined by the left side cameras would be in error by two centimeters in the X direction. The left arm would, for instance, be displaced two centimeters along the X-axis relative to the trunk position defined by the right side cameras. The geometric configuration of the bilateral system must, therefore, be precisely determined in order to express all position and orientation data in a single global coordinate system.

The four Selspot II cameras are set up as two sets of two cameras. Each set of two cameras is mounted on an optical bench using precision translators and rotators. However the geometric relationship of the two optical benches to each other is extremely difficult to determine to the necessary level of precision (on the order of 1 mm translation and on the order of 1/100 degree of rotation about all three axes). Moreover, due to misalignment of camera optical and mechanical axes and apparent fractional degree torsion of the optical bench main beams, even the geometry of the two cameras on each optical bench is not adequately defined using simple mechanical measurements. This resulted in unacceptable skew-ray errors (distances between lines of sight at the closest point of approach, Figure 3.8). Mechanical shimming to mitigate these errors was impractical because a number of different camera configurations would be used depending on study type. Because the viewing volume was quite large and because the Selspot system is very sensitive to reflections, building a 3D "calibration" array and using a direct linear transform (DLT) geometry calibration was not practical. Instead techniques were implemented to use the kinematic data aquisition system to estimate its own geometry.

The problem was divided into two parts. First, the relative geometry of the two camera configuration on each optical bench was estimated. The distance between cameras and the rotation of each camera about the vertical were taken as known. The remaining four degrees of freedom for each camera were then estimated by using a gradient search algorithm which minimized the total skew-ray error for a

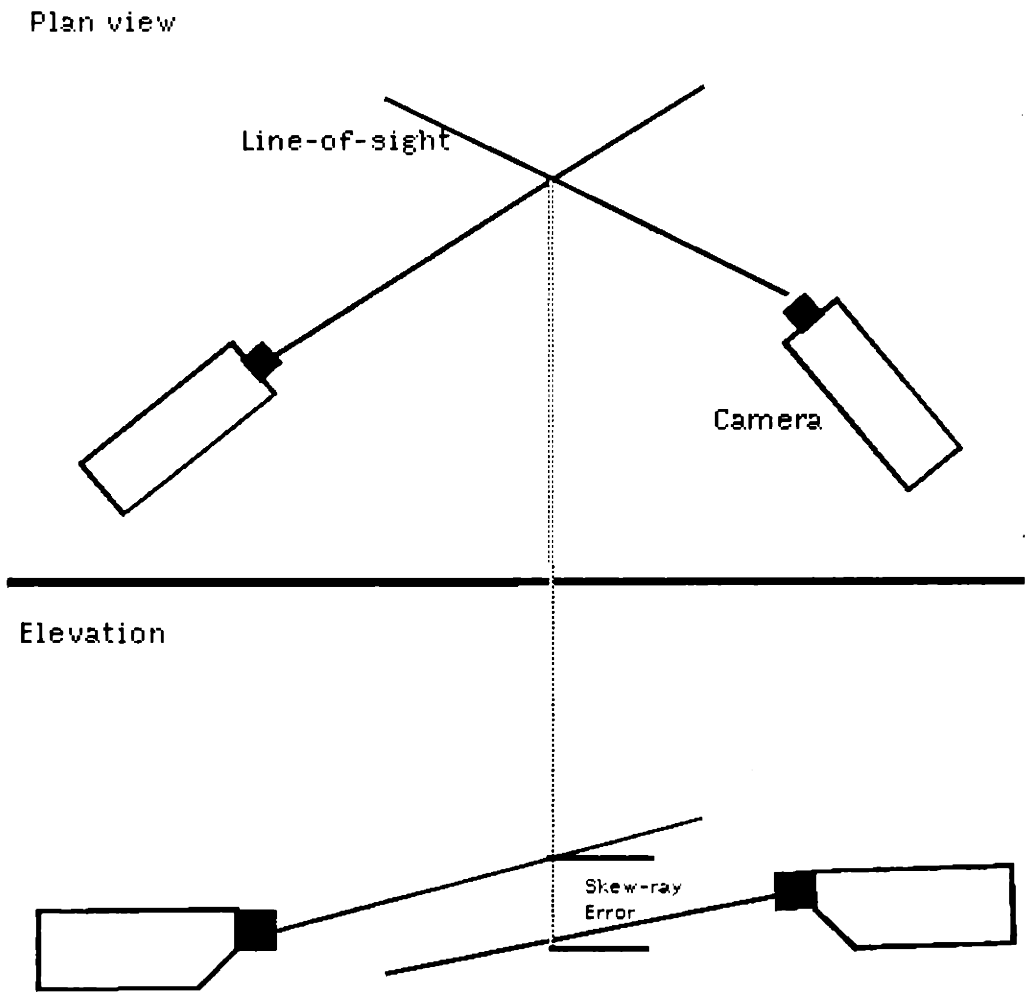


Figure 3.8 Skew-ray Error

large set of data (several hundred points). This method yielded results which are reproducible to on the order of one mm in camera position and 1/100 degree of camera axis rotation. Processing data using the geometry determined with this technique results in mean skew-ray errors of 1 to 2 mm and maximum skew rays errors of 2 to 4 mm for data sets in which the infrared markers are always within the viewing volume. Fijan was primarily responsible for the development of this technique.

The second problem was to determine the position and orientation of the two sets of cameras relative to each other. A large bilateral planar array of infrared markers, half of which are seen by each side, was used. Bilateral data was taken with the array positioned in a large number [~30] of locations throughout the common viewing volume. Using the geometry determinations arrived at in the first problem, the locations of the array center were determined. This process yielded a set of array center positions in the coordinate system of the right side [reference] set of cameras and a corresponding set of array center positions in the left side coordinate system. Using the TRACK software the 4X4 rotation matrix which describes the translations and rotations necessary to make the left side set of positions "fit" the right side set of positions was determined. This rotation matrix is the rotation matrix which defines the position and orientation of the left side set of cameras in the right side [master] coordinate system (Appendix D). Applying this rotation matrix to the left side camera geometry, as determined in part one, yielded the geometry of all 4 cameras in one coordinate system (Figure 3.9) Processing bilateral data using this geometry yields left side-right side agreement on the position of commonly seen objects to within 1 to 2 mm in the central viewing volume, with discrepancies nearer the edges on the order of 3 to 4 mm. Thus the overall technique resulted in a geometry determination which permitted the system position accuracy to be maintained at on the order of 1 mm.

The device shown in Figure 3.10 was used to test the the axis of rotation determination technique. Four arrays were mounted on the tripod, a left and right foot array were mounted on the moving portion, and left and right shank arrays were mounted on the base. The foot arrays were approximately 0.45 meters from the z-axis joint. The shank arrays were approximately 0.25 meters from the joint. Data was taken while the moving part was rotated about the z-axis joint. Between tests the upper assembly was rotated about the y-axis joint to vary the direction of the axis of rotation. Axes of rotations were determined for ten degree rotations about the z-axis joint. Two axes of rotation were determined for each side, the axis of rotation of the foot rotating relative to the shank and the axis of rotation of the shank rotating relative to the foot. A total of four

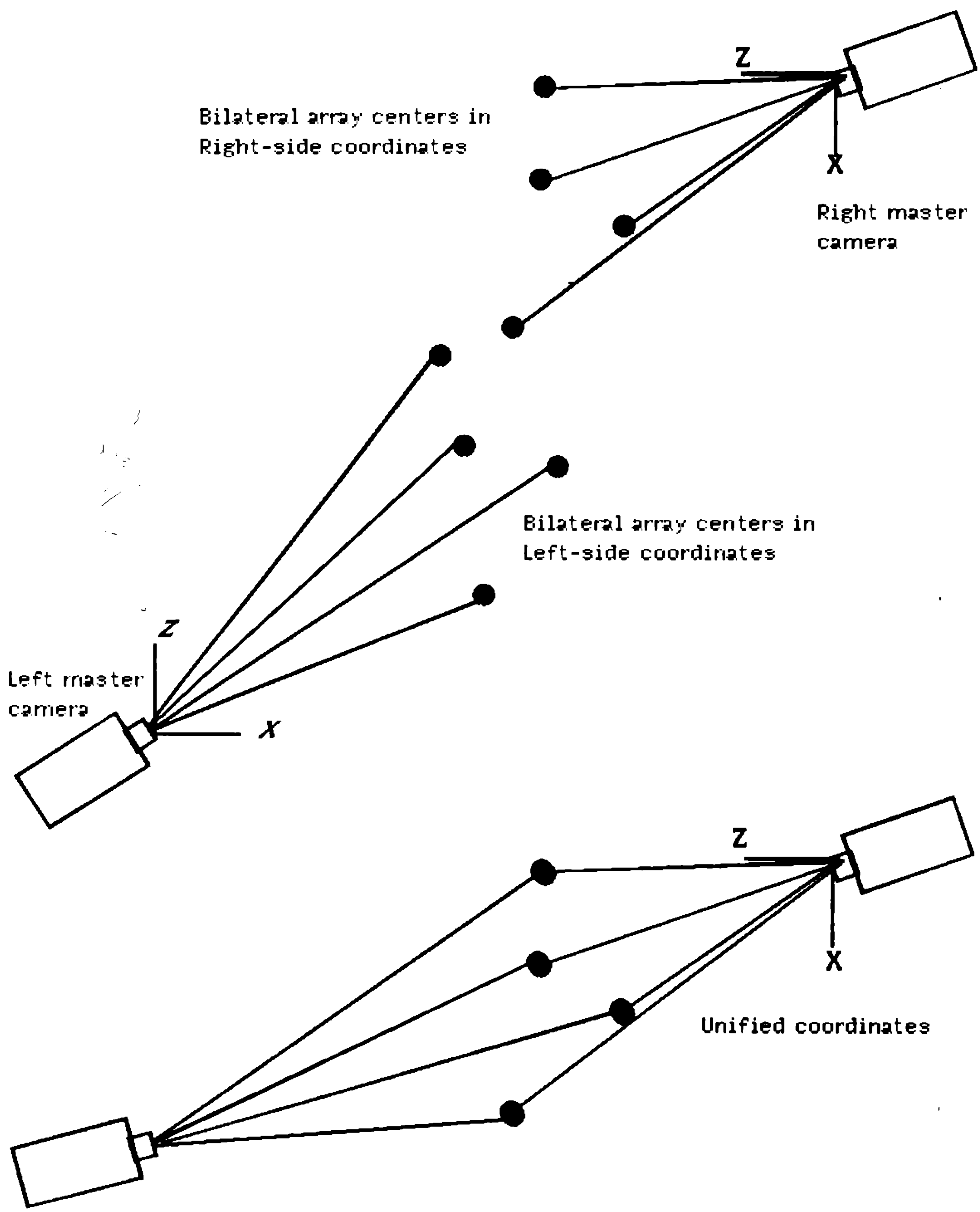


Figure 3.9 Bilateral System Geometry Determination

axes of rotation were determined for each data set. Four data sets were obtained. The two axes on a given side agreed exactly as indeed they must if the algorithm is mathematically correct. The maximum error between sides was two degrees as defined by the direction cosine for the global Z-axis. This test verfies that the axis of rotation determination is computationally correct and consistent. It also set the upper bound on the possible misalignment of the left and right side coordinate systems at two degrees.

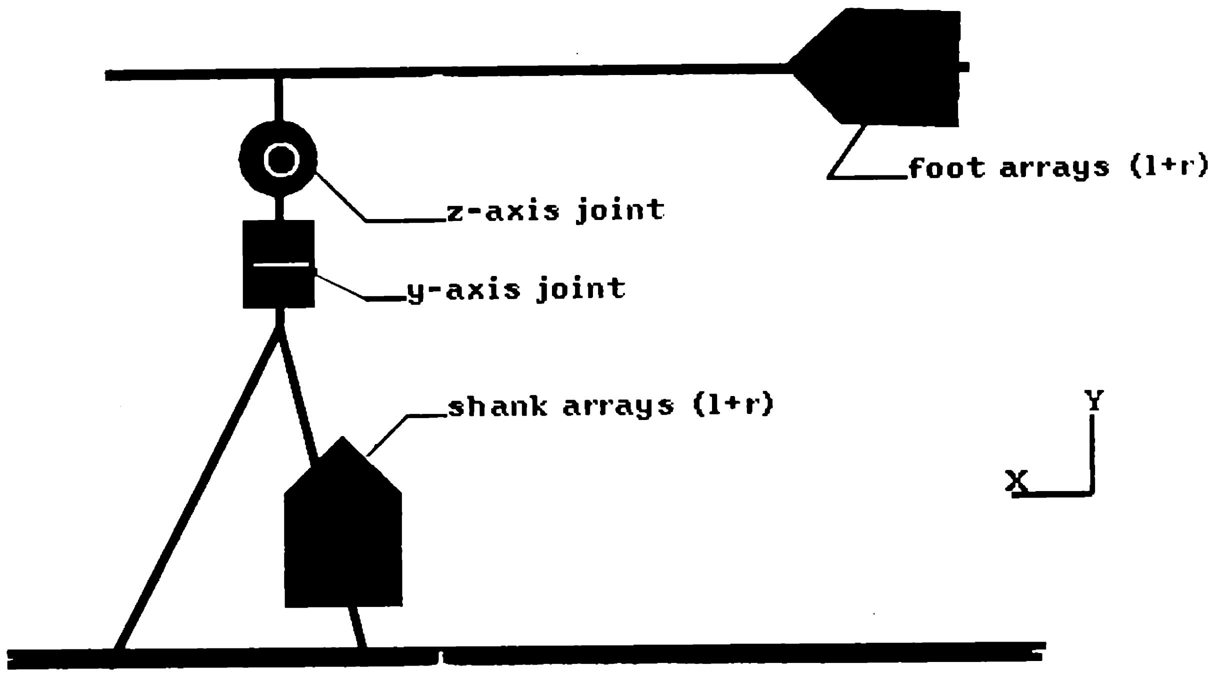


Figure 3.10 Axis of Rotation Test Device

CHAPTER IV

CENTER OF GRAVITY ESTIMATION TECHNIQUE

Assessment of postural control requires the quantitative observation of responses to disturbances. As previously stated, postural control has been studied by others by applying a standardized disturbance to elicit a response in terms of muscle activation as detected by EMG and/or center of pressure movement measured using force plates. This approach provides no explicit data on body-segment kinematics and kinetic changes in response to the disturbance. Further, these traditional methods were considered inappropriate for the study of patients/subjects with severe postural deformity for a number of reasons. Because of the biomechanical consequences of postural deformity, the <u>appropriate</u> response to a disturbance for a patient/subject with significant pathology might be significantly different from the <u>normal</u> response. Predicting the appropriate postural response from any possible set of initial conditions is not possible given the current level of understanding of the postural control system.

Since the ability of subjects to handle postural disturbances varies widely, the use of one standard disturbance for all subjects is not appropriate; a disturbance sufficiently mild to be tolerated by more severely disabled subjects would not elicit an adequate response from controls or less severely disabled persons. The use of one standard disturbance is also undesirable because it precludes assessing the most significant feature of postural control, the ability of the neuromuscular system to select and coordinate an appropriate response to an extremely wide variety of disturbances. In order to study the biomechanics of posture and postural control without these restrictions, a technique using kinematic data for the eleven segment 66 DOF model of the body was developed.

Postural control should be assessed in terms of a parameter or set of parameters which are functionally related to the coordination of movement. "Good" static posture (henceforth defined as stance) can be defined as posture which controls the location of the center of gravity (CG) in a small volume over the base support. "Good" dynamic posture (of which gait is an excellent example) is characterized by contol of the overall center of gravity and the centers of gravity of the more massive body segments (e.g. the trunk) so that overall body translation is achieved with minimal acceleration, i.e. the net force and metabolic cost required are minimized. Such evaluation of the CG kinematics could be a useful tool in evaluating the coordination of postural control.

In this study the position of the CG and the mass of each of eleven body segments are estimated. The location of the wholebody CG is calculated from the segment masses and CG locations. The projection of this estimated CG location onto the horizontal plane is compared to the measured center of pressure location on the force plates during "static" tests to evaluate the quality of the CG position estimate. The kinematics of the segment CGs, the combined CG and the center of pressure, all during disturbance responses, can then be used to evaluate the coordination of postural control responses.

Figure 4.1 illustrates the method used to measure the segment and body CG kinematics. The Selspot II/TRACK system acquires kinematic data from sets of infrared LED arrays sufficient to define the kinematics of the eleven body segments corresponding to the wholebody posture model. The body segment kinematics are evaluated from the array kinematics using transformations defined by data acquired specifically to establish the initial conditions. The positions of the body segment centers of gravity are estimated from the segment kinematics and estimated "body segment parameters," i.e. estimates of the segment CG

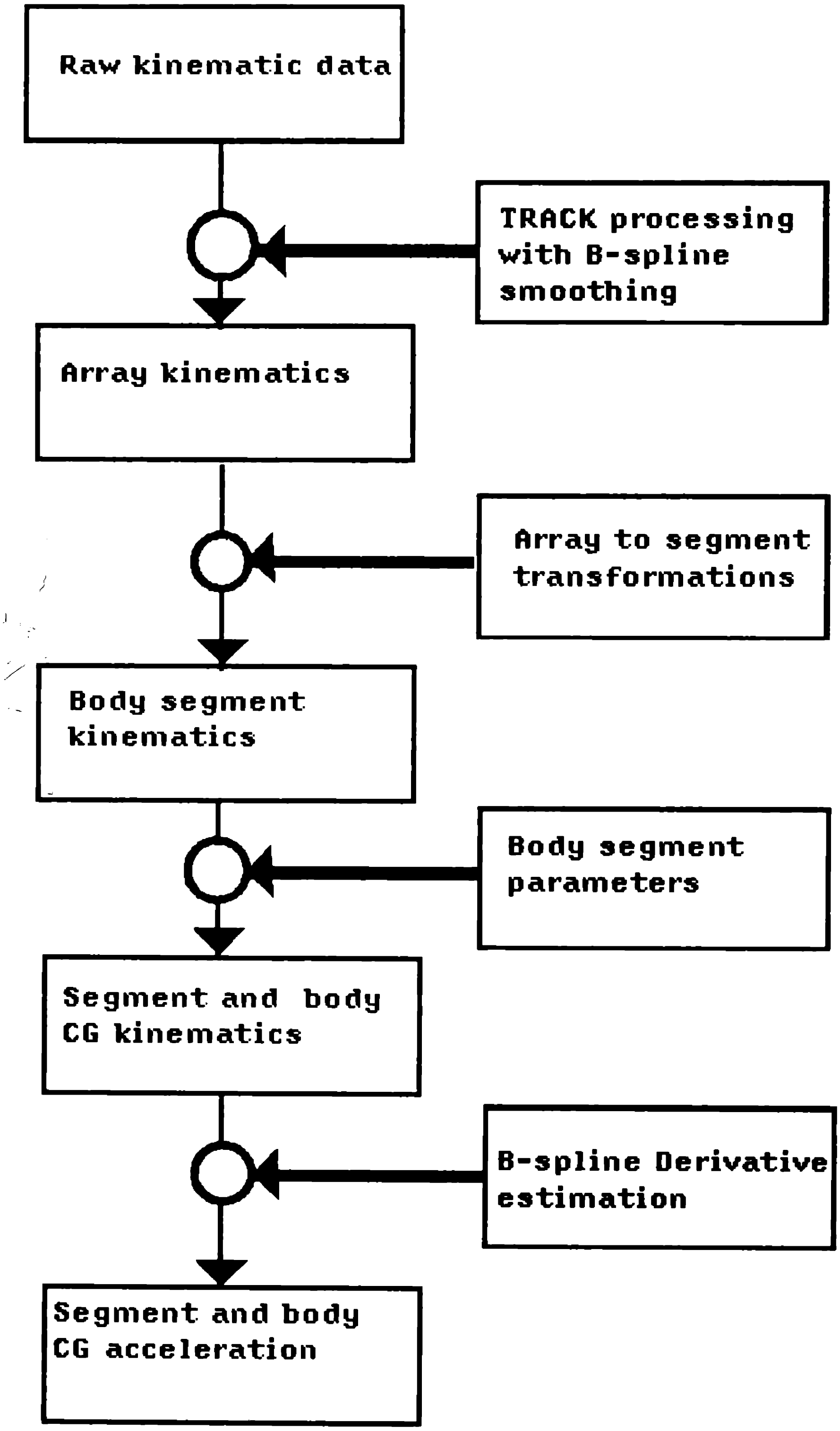


Figure 4.1 Procedure for Estimating CG Acceleration

locations within the segment. For the subject population composed of normal and cerebral palsied children age 7 to 17 the data for normal male children in the 5 to 15 age rangeof Jensen [26] was used. This is the only available data which accounts for the change in segmental mass distribution with age. For adult subjects the data compiled from a number of sources by Contini [13] was used. The position of the combined CG, i.e. the wholebody CG, is also estimated. The estimated CG positions are then processed to determine the CG velocities and accelerations.

IV.1 SEGMENT PARAMETERS

The determination of body segment kinematics was discussed in Chapter II. As Figure 4.1 indicates, in order to estimate the segment CG kinematics an additional transformation is necessary. The segment CG location relative to the origin of the segment coordinate system must be located. In this model, a 3-D translation must be specified from either the segment's proximal or distal joint center to the segment CG in segment coordinates. This statement makes a major assumption, namely, that the segment CG is at a fixed location in the segment. This is not generally true; during movement the contraction of muscles continuously changes the distribution of mass within a body segment. Thus the complete transformation would be a function of three spatial dimensions, the segment dynamics and other factors which determine muscle activity. However, defining this multi-dimensional transformation is beyond the state of the art.

An additional assumption is made to further simplify model development. At some level of precision the transformation we are discussing is unique to each individual; that is, if segment CG locations could be defined exactly they would be different for each individual. However, this model will assume that the human form is sufficiently standardized so as to permit the segment CG locations to be approximated using data from normals and a limited number of subject specific parameters such as height, weight, sex or age. At this time this assumption must be justified on grounds of necessity rather than on any evidence of its validity. There are, in fact, at least two reasons for questioning its validity; the clear variability of the human form particularly where neuromuscular pathology is present and the inconsistency of available normal data [13, 16, 26]. For this reason there are efforts underway to develop techniques for defining patient-specific body segment parameters. For example, computed tomography (CT) and magnetic resonance imaging (MRI) data are being used to estimate these parameters [8]. Although currently performed on cadaver specimens, the techniques may be under certain circumstances, be applicable to living subjects.

In addition to estimating the segment CG location it is also necessary to estimate the segment masses. While not required for the current model, the software developed for this model also estimates the segment inertial properties. The same assumptions and caveats apply to all of these estimated parameters which are collectively referred to as body segment parameters.

Jensen estimated the segmental and whole body mass and inertial parameters using what he calls an eliptical zone technique. Using lateral view and frontal view photographic records of the subjects the body segments were divided into 2 cm thick slices and major and minor axes lengths were estimated for each slice. Assuming that each 2 cm thick zone had a uniform elliptical crossection, its volume was then estimated. Using segment density data from the literature, zone masses were estimated. The segment mass and inertial properties were then estimated using numerical integration and the parallel axis theorem. Twelve boys ages 4 to 15 were studied once a year for three years, to determine the change in mass distribution with growth. The properties of fifteen body segments were estimated with segment mass results reported as fractions of total body weight. The estimation technique was verified by comparing total body mass obtained by summing the segment mass to the measured total body mass. The segment CG radius and radius of gyration are reported as fractions of segment length, i.e. in only one dimension.

The results indicate that the change in the mass distribution with growth and maturation are significant. Specifically the head mass fraction decreases with age while the lower extremity segment mass fractions, particularly the thigh, increased with age. These trends correspond to the trends in anthropometric data reported for this age group [54]. When Jensen compared his results for the fifteen year-olds with the results of adult cadaver studies [Dempster and Clauser] he found major differences. We will consider how these differences affect the posture model later in this chapter.

The segments reported by Jensen were the foot, shank, thigh, trunk, upper arm, forearm, hand, and head. For our purposes the data for the upper arm, forearm and hand was combined to estimate the parameters for the single segment arm of the posture model. Jensen's trunk segment combined both the trunk and the pelvis segments of our posture model. Accordingly 85% of the Jensen trunk mass was assigned to the trunk of the posture model, and 15% to the posture model pelvis. This distribution approximately corresponds to the volume distribution of the two segments.

Jensen defines the segment CG radius and radius of gyration in terms of fractions of the segment length. The other two dimensions required for the CG position are specified for the posture model by assuming that the CG lies on the segment centerline, which for most segments is the line connecting the joint centers. Note that the inertial properties do not enter into the CG estimation; however, the software (Appendix E) includes inertial parameter estimation in anticipation of future needs. Axial symmetry was assumed in estimating the inertial parameters.

Contini presents data only for the upper and lower extremity body segments. For those body segments not covered by the Contini data, parameters are estimated using the Jensen values for age 15, except that the mass of the combined pelvis and trunk is calculated as the difference between the total mass and the sum of the estimated arm, leg and head masses. Again, a 15/85 split of the torso mass between the pelvic segment and the trunk is assumed.

IV.2 CENTER OF GRAVITY ESTIMATION PROCEDURE

As Figure 4.1 indicates, the center of gravity is estimated using a multi-step procedure. The first step is the determination of the array kinematics using TRACK processing. The version of TRACK used employs b- spline smoothing using software developed by Woltring [60] and adapted for TRACK by Murphy [37]. While the data can be smoothed at a number of different points in the processing, smoothing of the individual LED position data (referred to as 3-D data) was the most beneficial. The position of each LED is specified in three global coordinates with each LED treated as a point. The coordinates are smoothed individually; i.e. a spline is fit to the X position data, then to the Y position data, then to the Z position. No attempt has been made to fit 3-dimensional splines to the data although such an approach might be applied to smoothing when the data is in this form. After smoothing the individual LED position data, TRACK then solves for the positions and orientations of the arrays of LEDS. The array positions and orientation parameters may also be smoothed. If this is not done interpolation is necessary fill in gaps in the data before velocities and accelerations are estimated. Processing a five second bilateral data set using the b-spline smoothing requires approximately one hour. This includes the time required to transfer the raw data from storage on the Biomotion Laboratory's PDP 11/60 to the MicroVAX and to store the processed data back on the 11/60.

Having determined the array kinematics, the next step is to determine the body segment kinematics. This is done by using the array coordinate to limb segment coordinate transformations described in Chapter II. At this point the kinematic data, in the form of 3-dimensional wire frame wholebody figures using a vector graphic display, is reviewed to insure that array fixation has been adequate during the trial, i.e. that adjacent segments remain contiguous. This whole body display was developed by by author using the approach originally developed by Fijan for displaying the lower limb segments. The visual display also facilitates determining which portions of the data are sufficiently complete to permit modeling the whole body. For example, analysis of a stair climb would have to be terminated before the head went out of camera view, since the interpolating algorithm leaves the segment in the last discerned position. Hence, if the model continued to estimate the body CG after the head had disappeared the head would be treated as an object fixed at its last visible location. This would, of course, cause a significant error in the combined CG estimate.

Once the segment kinematics are determined the next step is estimating the segment CG locations. The preliminary step in this process involves two programs. The first creates a file which defines the location of the segment CG in segment coordinates relative to the proximal and distal joint centers. The second program provides an estimated mass for each segment. The input data for these programs consists of information such as subject height, weight, sex and age. Additionally, certain measurements are provided, which are either supplied at the time of execution (e.g. trunk circumference) or read from an appropriate subject segment file (e.g. distance between joint centers). The segment file provides the physical description of the subject used for generating 3-D graphics of the kinematic data. These measurements are used primarily to estimate inertial parameters which are not used in the posture model in its current form. Both of these programs are based on programs in the MIT **NEWTON** software package which estimated the lower limb segment parameters only.

A check that the algorithm for estimating the segment CGs was computationally correct was needed. Multiple coordinate transformations are involved, introducing the possibility of calculation errors; e.g. premultiplying by post multiplying transformation matrix. Therefore, the program which displays the kinematic data in the form of 3-D anthropomorphic figures was modified by the author to also display marks indicating the position of segment CG's. This permits visual verification that the estimated segment CGs are at least reasonable. In practice, errors in the algorithm were identified and corrected by this means. Although developed to check the software, the display provides an interesting graphical visualization of the individual models of each subject. The paths of the segment CGs during disturbance responses also prove interesting.

Given the segment kinematics and the estimated body segment parameters, the determination of the segment CG kinematics and the body CG kinematics are straight forward. The segment origin to segment CG vector is transformed from segment to global coordinates and linearly added to the position of the segment origin, also expressed in global coordinates. This transformation is performed using the segment rotation matrix determined for each data frame by **TRACK**. The combined CG location is then given by the following equations:

$$X_{CG} = \sum_i (x_{cg}(i)*m(i)) / \sum_i m(i)$$

$$Y_{CG} = \sum_i (x_{cg}(i)*m(i)) / \sum_i m(i)$$

$$Z_{CG} = \sum_i (x_{cg}(i)*m(i)) / \sum_i m(i)$$

The segment CG kinematics are stored in files which are the input for the next stage of processing and for display programs. The next step is to estimate the CG velocities and accelerations. In the sequence shown in Figure 4.1 this is done using the fifth order b-spline smoothing algorithm. If the initial kinematic data has been adequately smoothed the velocities and accelerations may be estimated using a fifth order Lagrangian estimator with essentially identical results.

IV.3 ACCURACY OF THE MODEL

A simple static test was devised to test the CG kinematics estimation routine. Figure 4.2 illustrates the two segment body physical model used to find the mass and array-to-CG vector for each segment. The force plate CP can define only two coordinates of the center of gravity. Each segment was first placed on the force plate in the horizontal position to define the global X and Z coordinates of the CG, then the body was rotated 90° about the Z axis to determine the global Y and Z coordinates. Each determination was repeated five times. The global position of the array was determined for each trial so that the global array-to-CG vector could be completely defined. However, once the array-to-CG vector is defined in global coordinates it must be transformed into array coordinates for input into the model. This presents a problem since two sets of data, each with its own array rotation matrix, are used to define the CG in three dimensions. An acceptable solution is to evaluate the magnitudes of the global components and obtain for the smallest component an average value over several sets of data. Global vectors are then formed with that average value as one component and measured

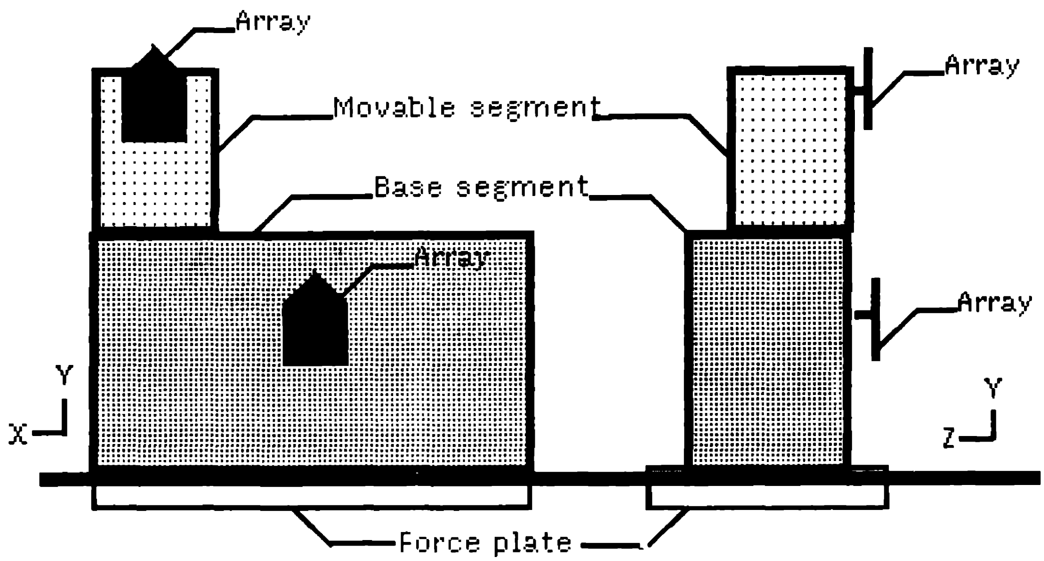


Figure 4.2 CG Estimation Test Assembly

values as the other two components. These vectors are transformed using the global to array coordinate transformation associated with the measured values. For both segments the X-component of the array-to-CG vector was at least an order of magnitude smaller than the Y and Z components, so this approximation was satisfactory. Five determinations were performed for each segment; the weights are the average of ten values, the vector components are the average of five values. Table 4.1 lists the mass and array-to-CG vector components for the test segments. The small standard deviations indicates that the values have been estimated to the limits of system accuracy; $\pm1\%$ of full scale (200Newtons) for the weight and ±1 mm for positions.

Table 4.1 CG Test Assembly
Segment Parameters

Segment #	Mass [kg]	Array to CG vector X[m]	Y[m]	Z[m]
2	13.18 $\pm$0.28	0.0056 $\pm$0.0014	0.0700 $\pm$0.0008	0.1926 $\pm$0.0016
3	5.43 $\pm$0.05	-0.0015 $\pm$0.0005	-0.1324 $\pm$0.0021	0.1325 $\pm$0.0016

A series of six static tests was then performed with the smaller moveable block placed in different locations on the larger base block. The top block positions were right rear, right forward, left rear, left forward, center rotated counter-clockwise, and center rotated clockwise. The CG estimation algorithm was then used to estimate the combined

CG of this two segment system. With the two segments used it was possible to change the combined CG location by 0.13 meters in the X-direction and 0.05 meters in the Z-direction. Since the test are static the X and Z coordinates of the combined CG should agree with the X and Z coordinates of the combined center of pressure. Table 4.2 shows the X-coordinate results for the six static test.

Table 4.2 Estimated CG vs CP

X-coordinate

Test	X-CG [m]	X-CP [m]	dif [m]
CGTS30	1.497568	1.494298	0.003270
CGTS31	1.372115	1.376497	-0.004382
CGTS32	1.492237	1.494399	-0.002162
CGTS33	1.371474	1.374174	-0.002700
CGTS44	1.444174	1.442564	0.001610
CGTS45	1.423925	1.435081	-0.011156
		mean dif	-0.002587
		mean|dif|	0.004213

Table 4.3 show the results for the Z coordinate.

Table 4.3 Estimated CG vs CP

Y-coordinate

Test	Y-CG [m]	Y-CP [m]	dif [m]
CGTS30	2.852119	2.848616	0.003503
CGTS31	2.843919	2.841022	0.002897
CGTS32	2.892822	2.888682	0.004140
CGTS33	2.901051	2.897485	0.003566
CGTS44	2.854364	2.853671	0.000693
CGTS45	2.823965	2.825328	-0.001363
		mean dif	0.002239
		mean|dif|	0.002694

The errors are of approximately the same magnitude as the uncertainty of in the array-to-CG vectors. The fact that the mean errors are non-zero would tend to indicate that a non-random error exist. The combined CG computations were checked and are correct; therefore, the mean error is most likely due to residual errors in the segment masses and array-to-CG vectors and the comparatively small number of data points averaged. When the segment masses are known to ±2% and the array-to-CG vectors are known to ±0.002 meters the location of the estimated combined CG agrees with the CP location to within ± 0.004 meters. The X-coordinate data for the rotated configurations, especially set CGTS45 is considerably poorer than the unrotated data sets. The array-to-CG vector is, however, being transformed properly. The unusually large error may be due to reflection of the upper array LEDs by the upper surface of the base block. If this data set is discounted the mean magnitude of the CG-CP difference is 0.0028 meter X and 0.0030 meters Y.

The magnitude of these errors may place limits on the application of the model. The model was developed to facilitiate evaluation of the coordination of movements by quantifying the contribution of segment kinematics to the control of the wholebody CG kinematics. The model is, given adequate body segment parameters, sufficiently precise for this purpose. However, if sufficient accuracy were achievable it would be possible to extend the model and develop a technique for improving body segment parameters estimates using kinematic data and the model. The practicality of this approach would depend on whether the errors in CG location due to errors in the segment parameters are large compared to the errors in the kinematic data, force plate data and the model.

In order to assess the effect of errors in estimating the body segment parameters, the difference in the results produced by using the two different body segment parameter estimation algorithms was evaluated. A subject was tested and her body segment parameters were estimated using the Contini data for adult females and the Jensen data for a boy of fifteen years of age. The difference in these results should be fairly representative of the variation between subject body types which is being neglected when using standard or reference data rather than individual specific body segment parameters.

Table 4.3 Estimated Segment Masses

Segment	Jensen Mass[kg]	Contini Mass[kg]
foot	0.836	0.521
shank	2.059	1.987
thigh	4.261	3.673
pelvis	2.501	2.953
trunk	14.171	16.734
arm	2.272	1.744
head	4.363	4.363

Table 4.3 lists the segment masses estimated by each algorithm. The Jensen data places a larger portion of the body weight in the lower extremity and a correspondingly smaller portion in the torso. This is notable because Jensen reports a significant shift in body mass from the head and upper extremities to the lower extremities with

age; possibly he is over estimating this trend. However, it is also possible that the Contini data is biased by over-weight subjects. In both studies the number of data points is sufficiently small that the discrepancies may simply reflect the individual differences among the subjects.

The effect of the difference in the estimated segment masses on the posture model were then evaluated by estimating the combined CG for a set of static data using both sets of parameters. The X and Z coordinates of the combined CG were different by approximately three millimeters. As would be expected from the differences in the estimated segments masses the error associated with the Y coordinate is larger; the Jensen's segment parameters yield an estimate of the body CG which is about three centimeters lower than that found using the Contini parameters. The X and Z coordinate errors are on the order of the errors inherent in the kinematic data and the model calculations. The Y coordinate error is larger than inherent system errors, but still on the order of the segment position errors when the joint center estimation error is taken into account.

If the model were a dynamic model rather than a quasistatic model; i.e. if the model attempted to estimate the CP position instead of the CG position, the effect of errors in estimating the mass and inertial parameters would be more significant. For this reason the simplier model appears to be justified at this time.

A dynamic model would be required to estimate body segment parameters using the force plate and wholebody kinematics. The errors in segment kinematics, due to the errors in joint center determination, are on the order of centimeters. This appears to be the same order of magnitude as the errors in estimating the CP position, caused by errors in the segment parameters. Thus improving segment parameter estimates using this method does not appear practical. Obviously, it would be desirable to be able to obtain subject-specific body segment parameters to eliminate this source of error in the model.

CHAPTER V

HUMAN POSTURE STUDY RESULTS

In this chapter typical results obtained when studying the posture and balance of human subjects will be presented. The results of tests on inanimate objects performed specifically to check the precision and accuracy of the techniques developed in this thesis are presented in Chapters III and IV. This chapter applies the posture and balance techniques to the evaluation of human subjects to illustrate the application of the methods to clinical problems. This thesis does not address any particlular hypothesis about human balance or posture control.

The human studies are organized in two catagories; tests to evaluate "static" posture and balance, and disturbance response tests to evaluate posture control. Chronologically the first round of experiments primarily assessed static posture and balance, including only a limited number of disturbance response trials. The study population for the initial static posture and balance test was comprised primarily of normal and cerebral palsied children and adolescents in the 7 to 17 age range. Later, disturbance response tests included normal adults as well. The results of the static posture and balance tests will be presented first.

V.1 STATIC POSTURE AND BALANCE STUDY RESULTS

Static postural kinematics and balance were assessed using two separate protocols. In most cases the evaluations were performed during separate sessions. Although it might seem that the study of static posture and balance could be combined in the same test, because, as pointed out in Chapter II, the evaluation of balance requires long time histories this is not possible. Trials of 90 seconds duration were necessary to quantify balance. Kinematic data sets of such length would require an extremely large data storage capacity, beyond the practical capacity of our system. Therefore trials of 5 seconds duration were used to evaluate static posture. To use longer trials to evaluate static posture and balance simultaneously it would be necessary to reduce the Selspot sampling rate significantly. This presents some technical problems.

In order to establish a lower limit on the sampling rate the primary need is defining the frequency content and the upper limit of the power spectrum of postural kinematics. This is not possible using the Selspot data alone since the frequency content would have to be determined from unfiltered, unsmoothed data for static posture kinematics where the signal to noise ratio is quite low. The significant noise content in the power spectra would make the determination of an appropriate cutoff frequency impossible.

The frequency content of the long balance data sets permits evaluation of the power spectrum of center of pressure (CP) movement, but again the signal to noise ratio for CP position in the raw static data is rather low. Also, we are not absolutely justified in assuming that the CP movement power spectra is the power spectrum of postural kinematics. High frequency low amplitude movements could be present which would not cause detectable displacement of the CP. Conceivably there could also be high frequency large amplitude movments of multiple limb segments whose net effect is little or no CP motion. Perhaps accelerometer data can be used to determine the frequency content of postural kinematics, but there is no indication in the literature that this has been done for the body segments defined in this model and this technique was not attempted in this study.

From these arguments until the frequency content of postural adjustment kinematics can be established it is reasonable to assume an upper bound similar to that of gait, i.e. significant power up to about 15 hertz [2]. This implies that kinematic data must be aquired at a rate greater than 30 hertz, with a rate of 150 hertz as used here being highly desirable. For this reason, even if the system were sufficiently sensitive, it is not possible to do kinematic studies of postural sway at this time.

The test protocols, given in Appendix F, were designed to evaluate posture and balance under conditions of normal stance, and coached or corrected stance with and without orthotics. The effect of altering sensory inputs--visual, tactile and auditory--was also studied. In addition kinematic data were acquired during two types of postural disturbances, active and passive. The active disturbance was a rapid arm raise maneuver. The passive disturbance was a push from behind at shoulder level. This test was usually performed twice, first with no warning to the subject and a second time after the subject had been alerted. The arm raise tests will be discussed in more detail later. The passive disturbance tests proved to be highly irreproducible and are not further discussed.

Figures 5.1 through 5.4 are 3-D graphic representations of a single time-frame of static posture of four different subjects. This demonstrated capacity to generate, via kinematic acquisition and computer processing, quasi-realistic, solid segment representations of the human figure represents a significant advance in the study of posture. The eleven segment wholebody model is the most

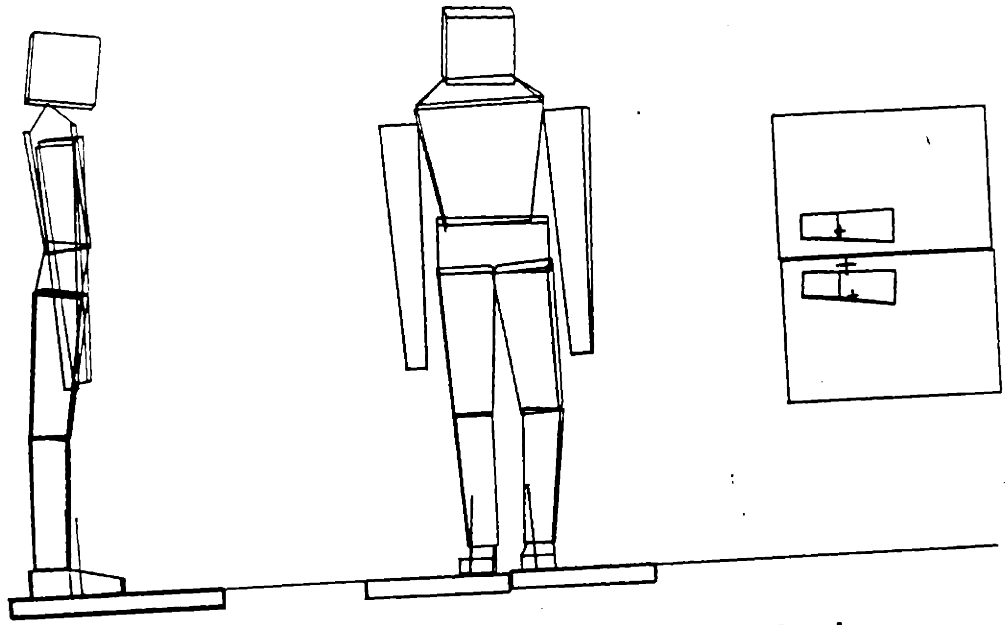


Figure 5.1 Static Posture Control Subject

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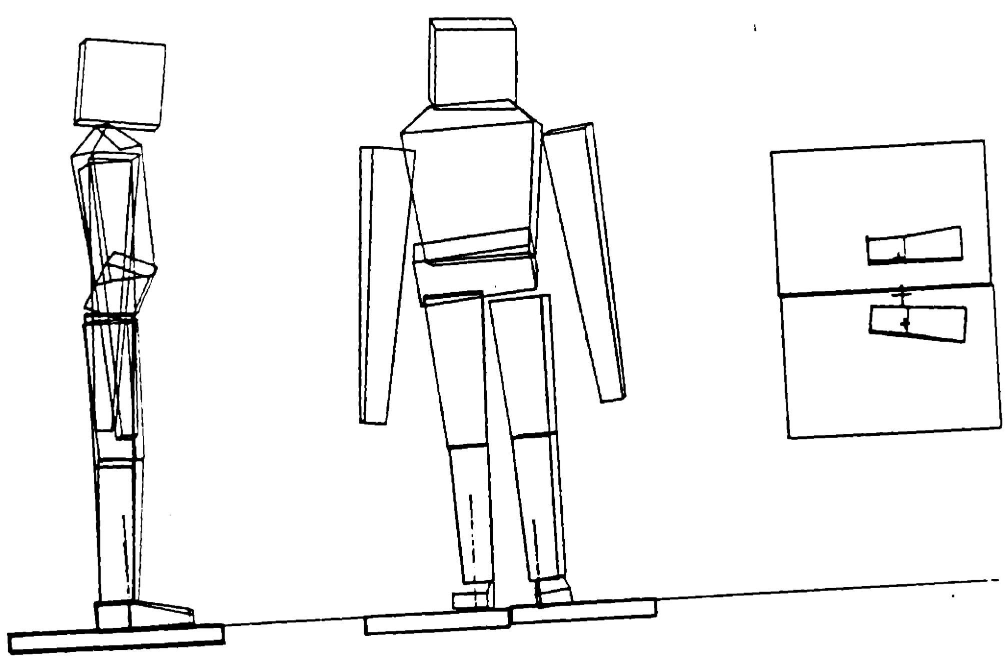

Figure 5.2 Static Posture Diplegic Subject

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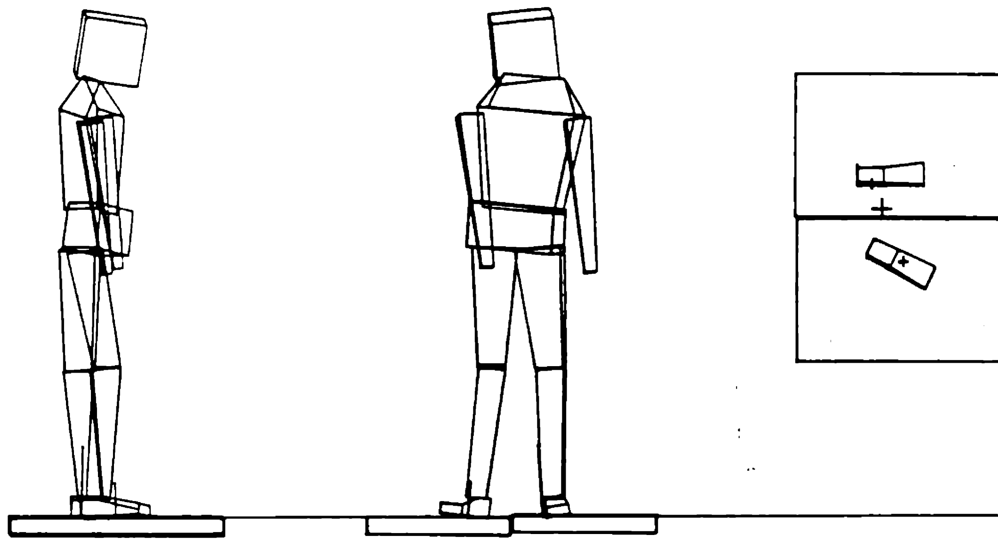

Figure 5.3 Static Posture Hemiplegic Subject

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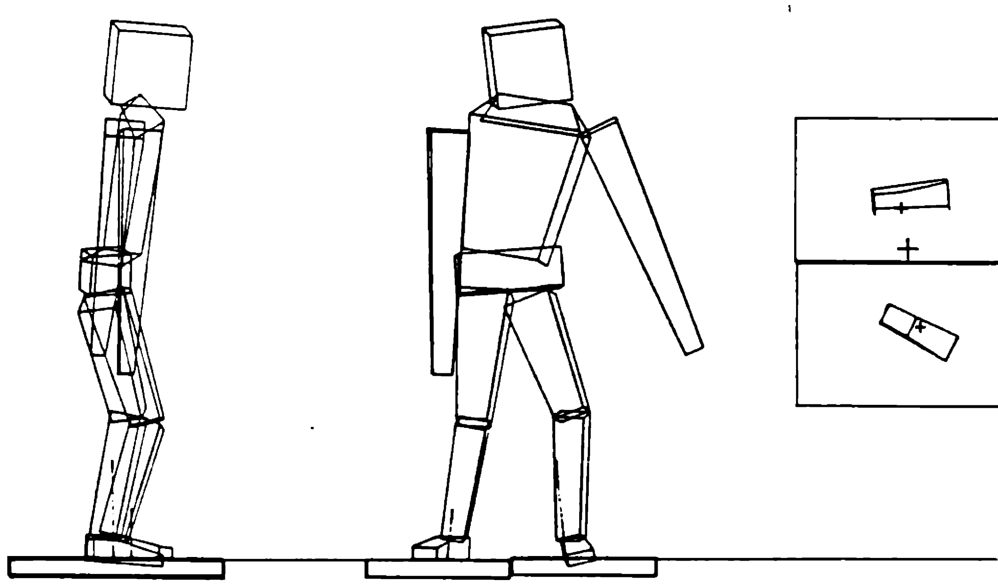

Figure 5.4 Static Posture Quadraplegic Subject

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complete model of posture based on actual kinematic data currently available. Since each body segment has a full six degrees of freedom the model is capable of representing and displaying visually much of the actual range of postural adjustment kinematics.

Figure 5.1 summarizes the kinematics of posture of a typical control subject. The figure shows the frontal and side views of the eleven segment model of the subject's posture, from one frame of kinematic data. A 3-D dynamic display is used to select a representative frame of data. If a subject's posture were highly time variable multiple frames would be required to display the range of postures. However, the overall posture of cerebral palsied children is surprisingly constant even though balance studies indicate abnormally large variations in the center of pressure location. Therefore the static posture of our cerebral palsied subjects could be characterized using a single frame of data. The posture of the control subjects was even more stationary. The force plates and their ground reaction force vectors are also shown, useful for characterizing assymetry in stance; e.g. certain subjects carry much more than 50% of their weight on one leg. The third portion of each figure shows a top view of the force plates and the location of the subject's feet for the same frame of data. The two small crosses indicate the location of each center of pressure while the large cross indicates the location of the combined center of pressure. All center of pressure locations are the average location for the entire five second kinematic data set. The display of mean CP positions together with foot positions from one data frame is justified for only those tests in which the feet did not move. The relative consistency of the kinematics of posture over the entire data set for both the normal and cerebral palsied subjects is further support for this combination.

Figures 5.2 through 5.4 show the static posture of cerebral palsied subjects, clinically described as spastic diplegic, spactic hemiplegic, and spastic quadraplegic. Since cerebral palsy is not a specific disease but a general designation for all motor dysfunction resulting from injury to the immature developing brain, there are a number of classifications and subclassifications and great variablility within these classification groups. In general, a quadraplegic is more severely involved than a diplegic or hemiplegic with impairment of function of all four limbs. Hemiplegics are more involved on one side; while diplegics are most involved in the lower extremities. The posture shown in Figure 5.4 is fairly typical of quadraplegic subjects and is refered to as a crouch deformity. The posture shown Figure 5.3 is particular to this hemiplegic subject with the most notable feature being the hyperextension of the knee on the left weight bearing leg. The postures shown in Figures 5.1 and 5.2 are grossly similar indicating that the static posture of this particular diplegic subject is close is normal. As Figure 5.1 indicates the static posture of the control subjects in the adolescent age group is not particularly "good." This subject is somewhat lordotic which is fairly typical of the age group, and has a degree of genu varum (bowleg). Weight bearing is also somewhat assymetric.

These figures, which represent but one frame of data from one data set under one test condition, contain a great deal of information. The opportunity to inspect and evaluate subject's posture, even in this static form, should be useful in the clinical study of postural deformity. The dynamic 3-D graphics which animate serial time frames are even more interesting and informative. However, the veritable flood of information in quantitative form mandates some concise but informative way of characterizing static posture in a quantitative objective manner using only a single or limited number of parameters. For this reason three kinematic indices, defined below, were developed and used to evaluate the posture data.

$$ANG = \sum_{i} |(\text{Flex Ang} - 0.0)|$$

$$ROM = \sum_{i} |(\text{Max Flex Ang} - \text{Min Flex Ang})|$$

$$MOM = \sum_{i} |\text{Flex Moment}| / \text{Weight} * \text{Height}$$

The angle index (ANG) is the summation over the joints of interest of the absolute values of the differences between the joint flexion angles and 0.0, the assumed normal anatomical flexion angle for "good" posture. The shoulder and neck joints were not included so that the index calculated for tests in which head or arm position was deliberately manipulated could be compared to other tests. The flexion angles are average values for the 5 second data record. The range of motion index (ROM) is the summation over the same joints of the difference between the minimum and maximum value of the flexion angle for the data record. This index provides a crude measure of the variability of posture. The moment index (MOM) is the summation of the flexion moments at the six lower extremety joints, normalized for subject height and weight. Again the flexion moments are average values for the data record. As such it is a kinematic, not a dynamic index, being basically a measure of the average moment arm at each joint; i.e. the distance from the joint center to the ground reaction force vector.

Table 5.1 Angle Index ANG for the Normal Stance Eyes Open Condition

Control Group

LMB	EVM	DFH	ANH	ESC	DJW
73.4	36.7	24.8	62.6	14.6	48.4
42.1			55.9		

Diplegic/Hemiplegic Group

MPC	MPC	SNC	KAD	ANM
52.4	72.1	109.4	73.7	87.5
58.8	82.0		78.6	

Quadraplegic Group

CSB	VMR	MJR	JSK
128.3	156.2	118.9	199.4
109.6	154.0	134.8	

Table 5.1 list the angle index values for all normal stance, eyes open, tests for all subjects examined during the initial round of experiments. Multiple values for a given subject represent repeat tests during the same session; repeated listing of the subject (MPC) indicates a second session. Clearly the values for the quadraplegic subjects are larger than for the controls. The diplegics and hemiplegic subjects also generally have values greater than the controls. However, the range of values for the controls is fairly large and somewhat overlaps the range of values for the diplegic and hemiplegic subjects. The obvious question that arises when reviewing this data is

what is the cause of the variability, particularly among controls.

The answer is actually multifactorial. The posture of the cerebral palsied subjects is highly variable even within a given subgroup. This is why a method of objective quantitative evaluation of posture is needed. Therefore, we should expect to see a broad range of posture scores in this population. The variation among the normals is of more concern. Certainly there is some variation in normal posture especially in the adolescent population. The protocol restricted the normal subjects to stand in approximately the position of attention; i.e., with the feet together and the back straight, see Figure 5.1. The cerebral palsied subjects stood in their "usual" posture except for one test in which an attempt was made to coach then into the "normal" posture. In hind sight the position of attention may not have been the best normal posture to select. The base of support is abnormally narrow which probably made the subjects feel somewhat awkward [29]. Also, all subjects in this testing phase were tested using the plastic molds as the array mounting technique. The bulk of these molds was large enough to perceptibly alter normal posture, particularly when attempting to stand with the feet together. This may well account for some of the bowleg stance seen in Figure 5.1. In additional, subjects in the adolescent age group are not particularly responsive to direction and in some cases we were not successful in motivating then to maintain their best posture through all of the trials.

However, I believe that the greatest single contributor to the subject to subject variability in normal joint angles in these early tests is error in the joint center identification procedures. During this first phase of experiments the anatomical landmark method (see Chapter II) was used. The magnitude of this error is difficult to assess. The reproducibility of separate determinations and the left vs. right symmetry in the same determination for appropriate subjects provide some indication of the quality of the joint center determinations. In some early tests with normals the discrepancy in shank length (knee height) was observed to be about 2.0 centimeters. In the case of one quadraplegic cerebral palsied subject one shank had a length of 0.44 meters the other a length of 0.48 meters; i.e., a 4 centimeter length discrepancy. In this case the subject was severely involved and some of the discrepency may be real; however, when the subject was tested one year later when the joint centers were determined using the axes of rotation method, the shank length discrepancy was less than 1.5 centimeters.

Identifying joint centers using anatomical landmarks is also extremely difficult with cerebral palsied subjects both because of their abnormal anatomy, and because it is difficult to maintain the pointer on the desired location when the subject cannot stand still. When an inexperienced person used the anatomical landmark method on the same normal subject on two different occassions, leg length discrepancies of on the order of 3 to 4 centimeters were observed between left and right side and between the successive determinations on the same side. Therefore, a possible error of 4.0 centimeters represents a reasonable upper limit for the anatomical landmark method, and errors of approximately 2.0 centimeters are probably fairly typical.

Figure 5.5 illustrates a simple model of the influence of joint center location errors on the angle determination. A 4.0 centimeter error on a segment of 0.44 meters in length results in an angle error of over 5 degrees. A 2.0 centimeter error corresponds to an angle error of about 2.5 degrees, while a 1.0 centimeter joint center error produces an angle error of about 1.3 degrees.

For the wholebody posture model an angle error of 2.5 degrees at each joint would produce about 20 degrees of variability in the angle index. If we add an additional degree of variation in each joint angle due to the system's inherent 1.0 degree angular resolution there would be about

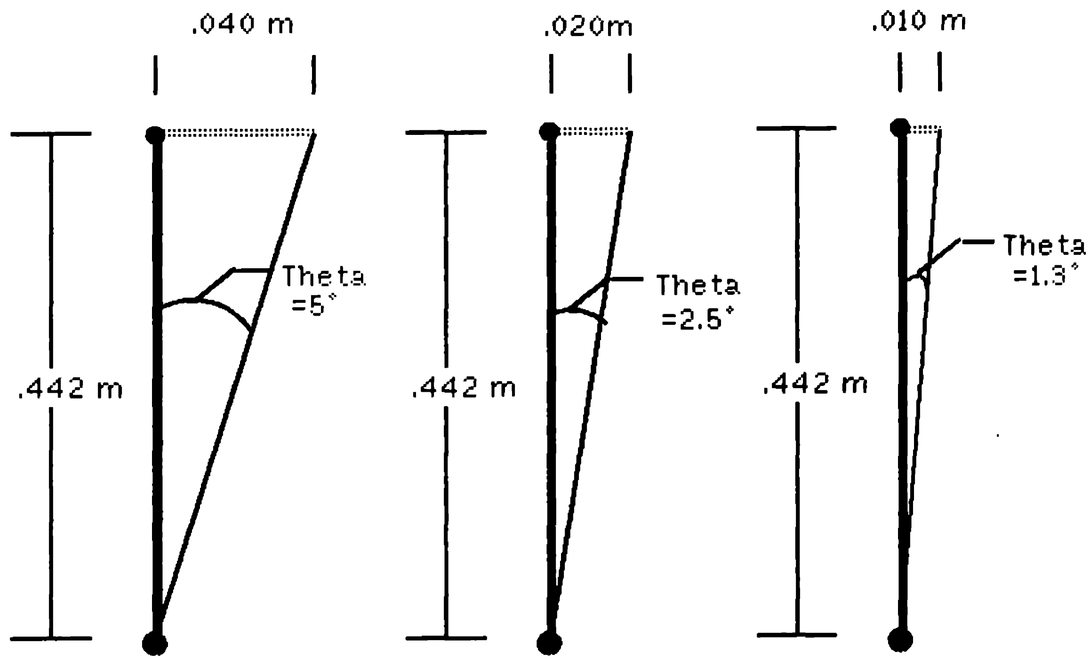


Figure 5.5 Angle Errors Due to Joint Center Error

30 degrees of variability in the angle index. Therefore, these sources could account for most of the variability observed in the angle index among and within normals. The two columns for diplegic subject MPC represent two seperate sets of test for this subject with different joint center determinations, both using the anatomical landmark method. The discrepancy between the two sets of data could be explained largely by joint center location errors of the magnitude discussed.

The angle error associated with a 1.0 centimeter error in joint center location is included because this appears to be the level of reproducibility of the axis of rotation method. Table 5.2 show the shank, thigh and total leg lengths determined by using the this method under the same conditions on six different data sets on the same subject. Consistency to within 1.0 centimeter seems to be achievable.

Table 5.2 Limb Segment Lengths from Independent Joint Center Determinations Using Axes of Rotation

Right Shank Length [m]	Left Shank Length [m]
.3648	.3634
.3771	.3707
.3711	.3708
.3709	.3642
.3946	.3672*
.3665	.3588
$\overline{.3742}\pm.0109$	$\overline{.3658}\pm.0047$

Right Thigh Length [m]	Left Thigh Length [m]
.4031	.4054
.3881	.3986
.3936	.4030
.3863	.3867
.3898	.3850
.4041	.4025
$\overline{.3941}\pm.0077$	$\overline{.3969}\pm.0088$

Right Leg Length [m]	Left Leg Length [m]
.8560	.8446
.8328	.8430
.8433	.8526
.8446	.8146*
.8344	.8312
.8511	.8376
$\overline{.8437}\pm.0091$	$\overline{.8373}\pm.0132$

*Data would probably not be used for joint center determination due to left/right mismatch.

Table 5.3 lists the posture study results as quantified using the ROM parameter.

Table 5.3 Range of Movement Index ROM for the Normal Stance Eyes Open Condition [degrees]

Control Group

LMB	EVM	DFH	ANH	ESC	DJW
19.9	12.6	26.4	20.1	12.1	20.1
		13.1			22.2

Diplegic/Hemiplegic Group

MPC	MPC	SNC	KAD	ANM
20.1	15.1	13.2	25.0	59.3
29.1	18.3		40.6	

Quadraplegic Group

115.3	42.6	69.5	36.5
53.9	32.9		119.9

The principal disadvantage of this parameter is that it is not particularly sensitive. There is a trend with the quadraplegic group, showing higher values than the controls and with some high values among the diplegic/hemiplegic group. In the quadraplegic group there are two data sets where the value of ROM is over 100 degrees. In both cases the subject nearly fell during the course of the trial. A better index of the variability of the joint angles might be developed based on a summation of the variances of the joint angles rather than the difference between the maximum and minimum angles.

The ANG index and the ROM index were calculated based on flexion angles only, i.e. sagittal plane angles. There were two reasons for doing this. First, using a combination of flexion angles, abduction angles and external rotation angles would have required developing a weighting scheme to define the relative value of the different angles. Such a scheme seemed unduly complex and arbitrary. Second, the use of only flexion angles makes it feasable to compare our results with results obtained by others with simplier 2-D kinematic data acquisition systems. The principle disadvantage of this approach is that the clinical term "flexion angle" is not well defined mathematically and, thus, must be approximated rather arbitrarly. Perhaps a better basis for these indices would be the angle between the segment axes. By this I mean that the two segment axes, unless colinear, define a plane, and an angle between the axes may be determined in that plane. This is well defined mathematically and would incorporate the clinial abduction/adduction angle as well as flexion/extension angle information. It would not include the internal/ external rotation angle information, but the absolute values of these angles are probably not well specified under any conditions.

The moment index (MOM) results are summarized in Table 5.4.

Table 5.4 Moment Index MOM for the Normal Stance Eyes Open Condition

Control Group

LMB	ANH	DFH	EVM
0.1011	0.0933	0.0758	0.0766
		0.0682	0.0661

Diplegic/Hemiplegic Group

MPC	MPC	SNC	KAD	ANM
0.0556	0.0508	0.0556	0.0845	0.1027
	0.0561	0.0587	0.0728	
			0.0627	

Quadraplegic Group

CSB	VMR	MJR
0.0906	0.1484	0.1253
0.0949	0.1330	0.1268

Since the flexion moments are normalized to subject weight and height the values listed in Table 5.4 are dimensionless. The general trend is similar to that seen with the previous two indices. There is more overlap of the control and hemiplegic/diplegic groups and even some overlap with the quadraplegic group. This is to be expected since the moment calculations are more sensitive than the angle calculations to the to errors in joint center determination. Recall that all data presented here was acquired while the anatomical landmark method was used for joint center determination. For normal stance or near normal stance the actual moment arms are typically only a few centimeters; i.e., of the same magnitude as the joint center position errors. For this reason, the moment index is of little value in the study of static posture and will not become useful until the joint center positions are defined to significantly greater precision and accuracy.

Table 5.5 lists the balance study results for the normal stance eyes open condition.

Table 5.5 Standard deviation of the Center of Pressure SDCP for the Normal Stance Eyes Open Condition [millimeters/meter]

Control Group

ANH	EVM	DFH	DJW	ESC	LMB
2.4	14.1	7.8	7.5	2.9	4.3
3.6	5.5				5.0
					4.9

Diplegic/Hemiplegic Group

MPC	SNC	KAD	ANM	SLR
6.0	6.1	9.4	11.0	7.6
5.6	4.6	7.0	.	9.7
	4.0			

Quadraplegic Group

MJR	JSK	MAH
18.6	27.4	41.2
13.5	32.8	19.1

Again the trend is for the lowest values to occur in the control group and for the highest values to occur in the quadraplegic group. The values for the diplegic/hemiplegic group are comparable to the control group, perhaps indicating that with normal sensory input the undisturbed balance of these subjects is near normal. A number of factors contribute to the variability of the results. Subject compliance to instruction is an especially important problem in these tests; standing still for 90 seconds is quite difficult for children and adolescents. Figures 5.6 and 5.7 illustrate the time history of the CP position for a normal stance eyes open trial of a control and a CP subject.

The results of the static posture and balance studies indicate that the posture model is a useful tool for assessing posture to objectively and quantitatively and for corelation with measures of balance stability. However, to relate posture to balance dynamic or disturbed posture and balance must be studied. The remainder of this chapter will deal with the study of posture control using the center of gravity estimation technique presented in Chapter IV.

V.2 RESULTS OF POSTURE CONTROL TESTS

Along with the static posture tests a limited number of posture disturbance or posture control tests were performed, 1 or 2 trials in each session. One such test was a right arm raise maneuver. The subject stood, usually with one foot on each force plate, facing in the -X direction. The line from the right to left shoulder was generally parallel

CENTER OF PRESSURE DATA: LMBF11

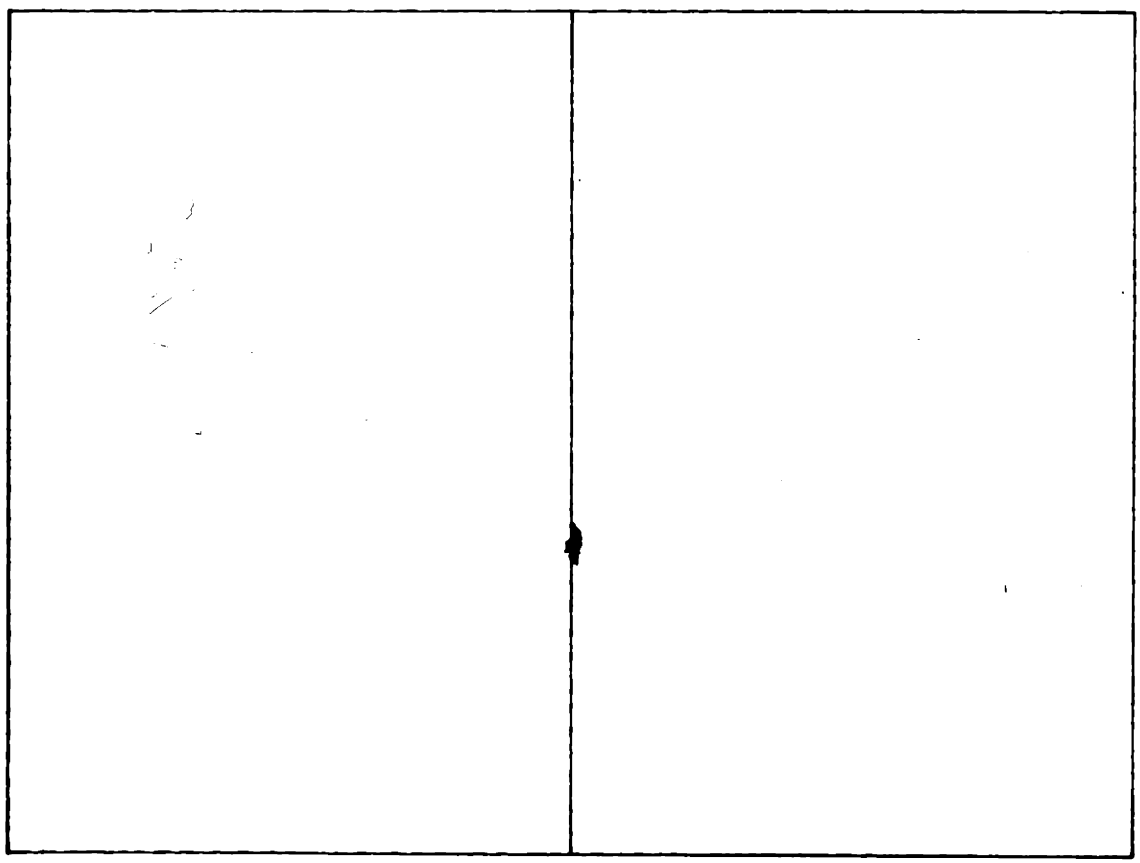


Figure 5.6

Center of Pressure Pattern

Control Subject

CENTER OF PRESSURE DATA: MJR110

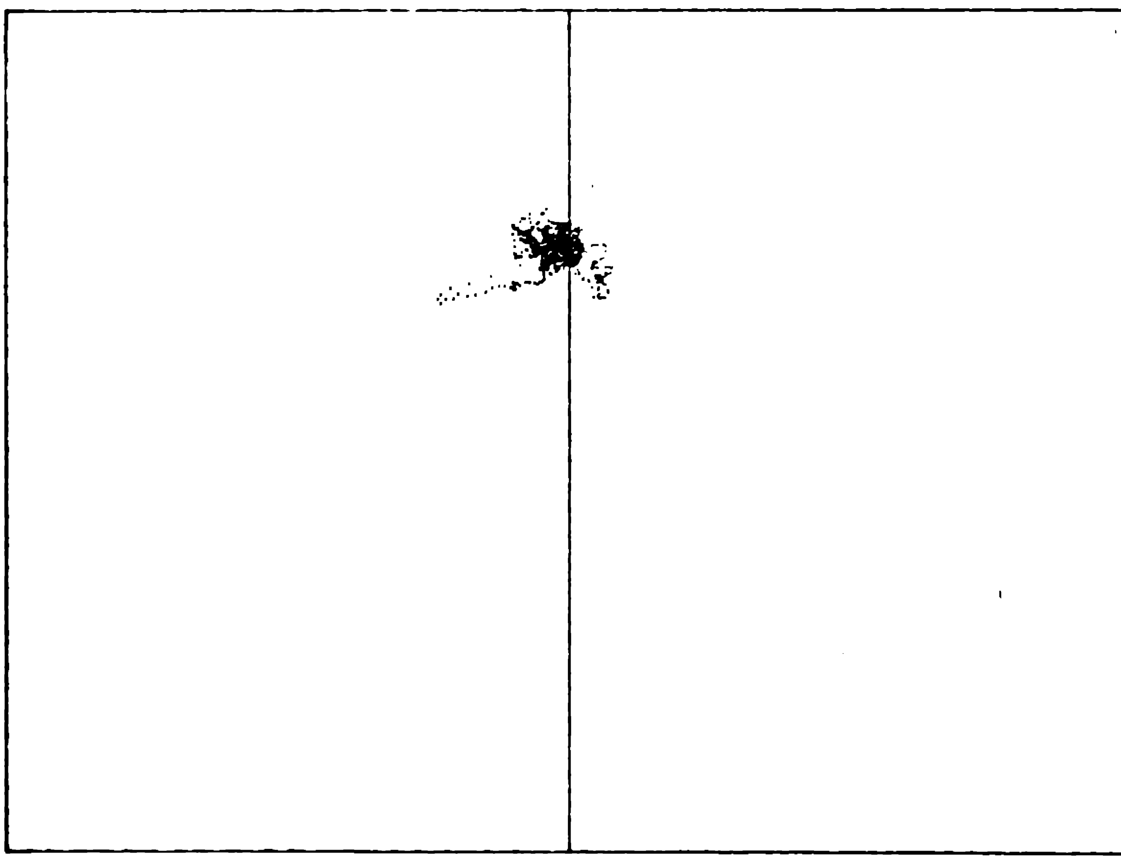


Figure 5.7

Center of Pressure Pattern

CP Subject

to the global Z-axis, and the line from the feet to the head was parallel to the Y-axis. A researcher stood two arm lengths in front of the subject and extended his/her hand out at the subject's eye level with the palm down. The subject started with both arms at his/ her side in the normal stance position. On signal, the subject extended the designated arm as fast as possible to strike the researchers hand with a "clap". The subject then stood as still as possible with the arm in the raised position for the remainder of the test. Test duration was five seconds; typically the arm raise started at about +1 seconds and was completed by +2 seconds.

Figures 5.8, 5.9 and 5.10 illustrate the data obtained during these tests. The four 3-D figures show the initial position, the final position, and two intermediate positions equally spaced in time between initiation and completion of the arm raise. Also shown are the time histories of the combined CP locations for the trials. Figure 5.8 illustrates the performance of a control subject. In the kinematic data the only apparent postural compensation for the arm raise maneuver is a rotation of the trunk with the shoulders translating back slightly and the right shoulder rotating further back than the left. Biomechanically this seems to be an approach response to compensate for the

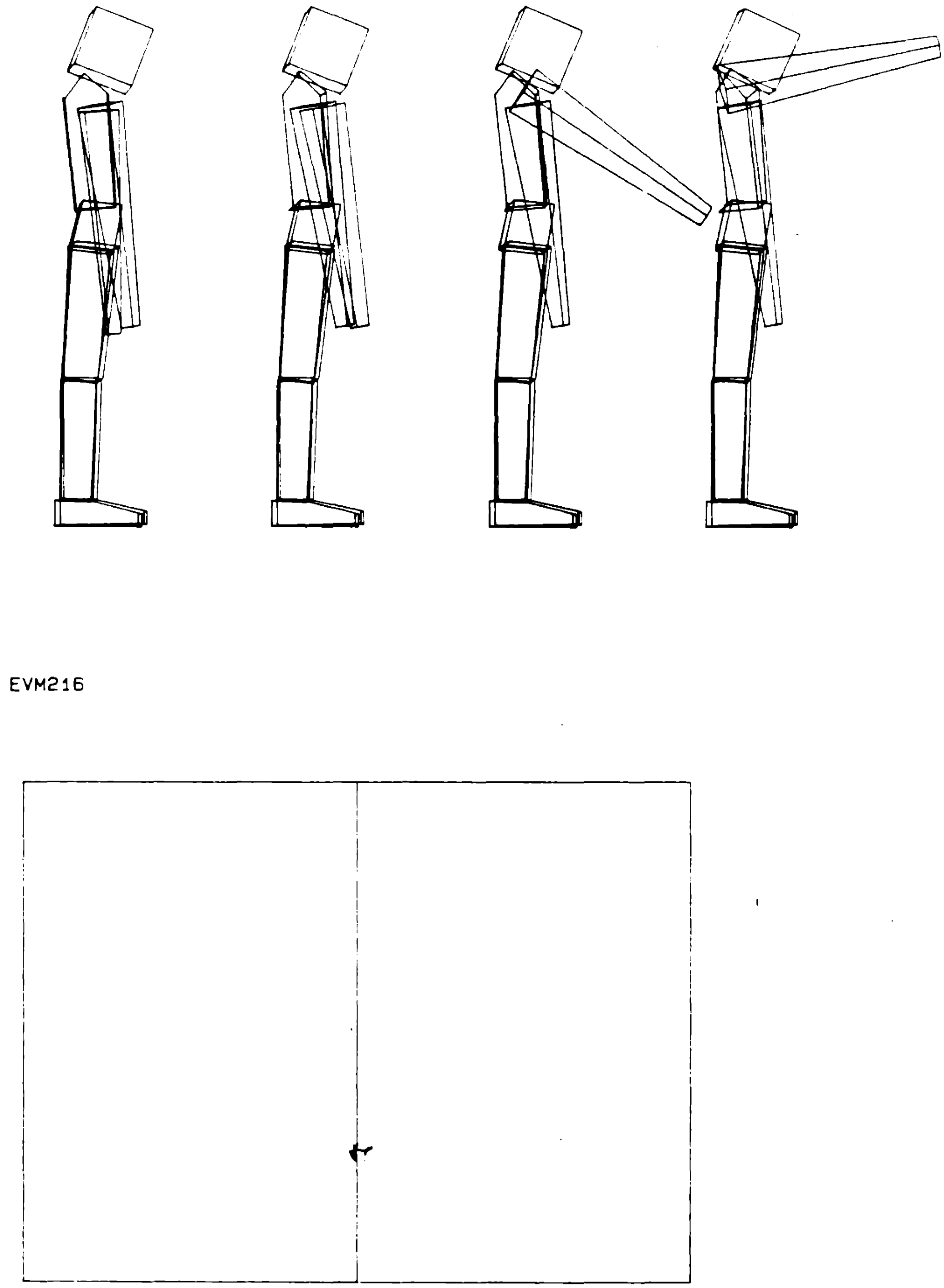
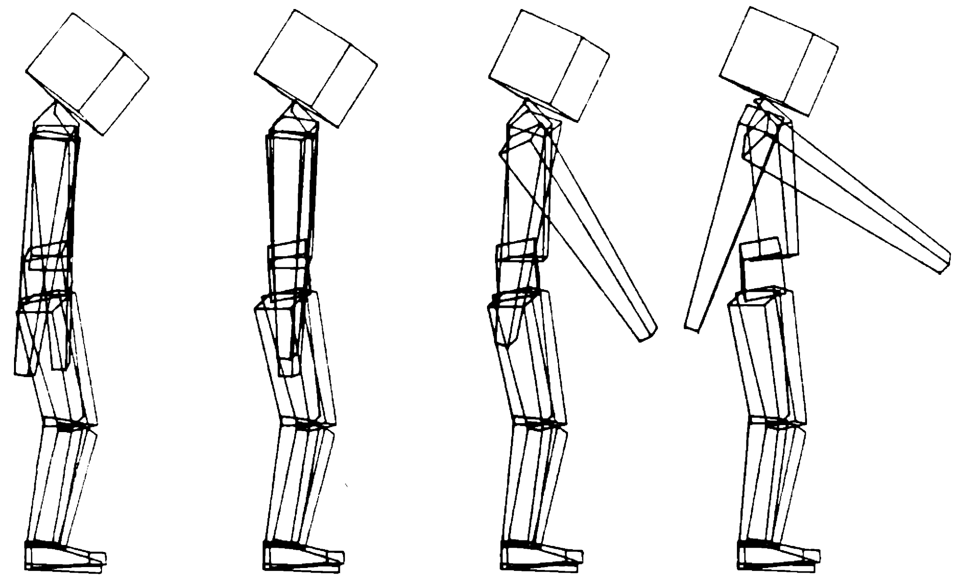


Figure 5.8 Right Arm Raise, Control Subject



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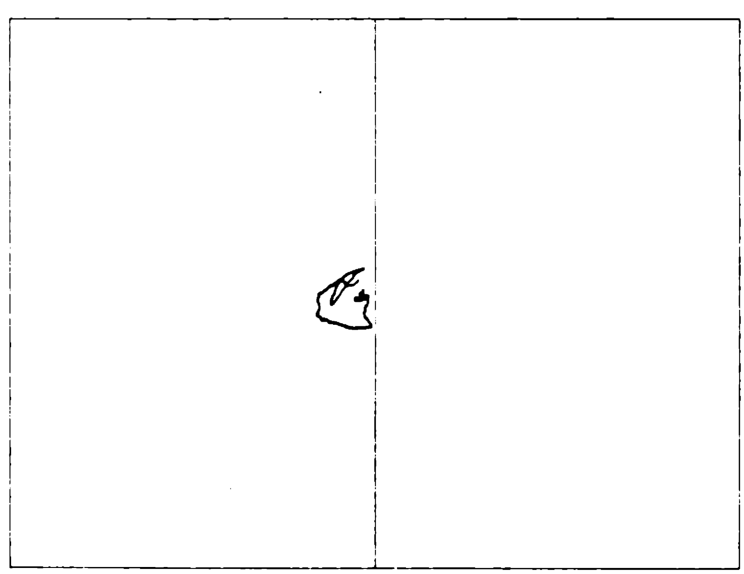

Figure 5.9 Right Arm Raise, Quadraplegic Subject

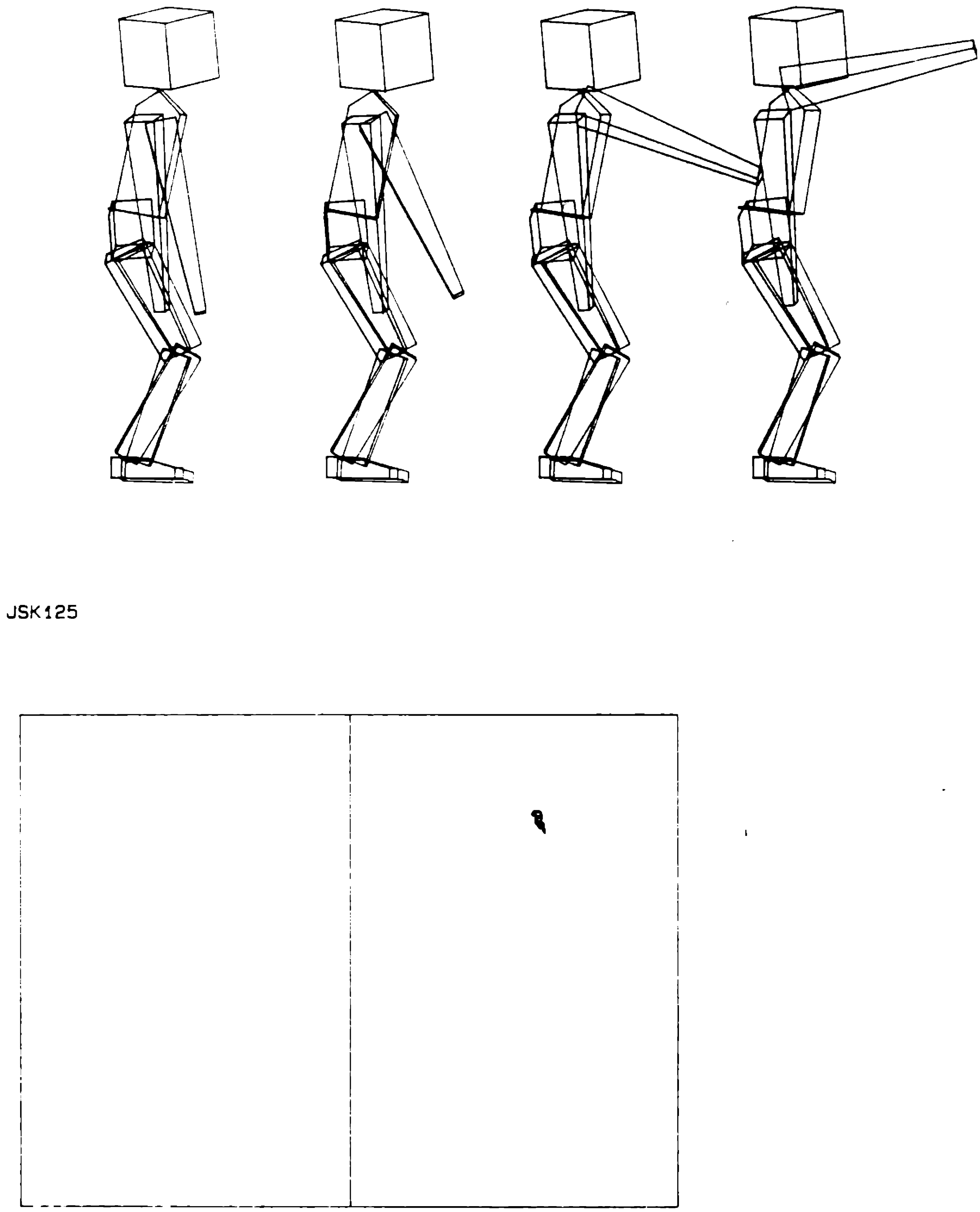

Figure 5.10 Right Arm Raise, Quadraplegic Subject

effect of the arm raise on the CG position. The CP traces indicate that the compensation is quite successful; the CP position is well controlled.

Figures 5.9 and 5.10 show the performance of two quadraplegic cerebral palsied subjects. The basic trunk rotation compensation mechanism seems to be present also. In Figure 5.9 there is also a perceptable movement of the left arm. For this subject the CP location is less well controlled during the actual arm raise maneuver and during the seconds following the maneuver. The subject in Figure 5.10 performed better. Only the trunk moves and the CP excursion is not large. However, there is an important difference in the way this subject performed the right arm raise maneuver. The control subject shown in Figure 5.8 and the quadraplegic subject subject shown in Figure 5.9 each completed the arm raise maneuver in about 100 frames of kinematic data, about 2/3 seconds. The subject shown in Figure 5.10, despite several practice tests and coaching to emphasize the need for speed, required approximately twice as long to complete the arm raise phase of the maneuver. This performance significantly reduced the dynamics of the maneuver and accounts for the ability to control the CP location so well.

These first tests including dynamic posture with the static posture to evaluate the feasibility of studying posture control yielded only modest hard information but indicated that further study was warranted. In general, the controls responded mainly with slight trunk adjustments and controlled the CP location well. The cerebral palsied subjects tended to perform the maneuver more slowly despite our best efforts to coach for more speed and controlled the CP position less well. While the cerebral palsied subjects seemed to use trunk rotation compensation, there was evidence, principally from joint angle plots, that postural adjustment activity was more wide spread. There were also indications that deficits in postural control could be detected even among cerebral palsied subjects whose static posture and balance seemed near normal. The wholebody CG model was developed primarily as a tool to investigate the coordination of postural control during such disturbance response tests.

In the remainder of this chapter results of the CG model will be presented in graphic form. Since these figures are complex and contain a large amount of information a brief explanation of their form and content is necessary. Each figure will consist of a set of three data plots. The CG/CP displacements or accelerations are shown in the three plots with the values for one global coordinate plotted against frame number, i.e. time. The upper plot will be labeled "Body CG and CP" and will show the data for the model's estimate of the combined or body CG and sometimes the combined CP from force plate data.

The second plot will be labeled "Segment CGs" and will display the data for the model's estimate of the segment CG locations. The left and right pelvis and trunk are averaged and the resulting value is displayed. For displacement data, all values are presented as changes from the initial position; i.e., the initial positions of each segment CG and the body CG are set to 0.0 and displacements from this initial position are plotted. The difference between the body CG and CP in the upper plot is the actual difference in global coordinates, but plotted with the initial CG position set to zero. The segment CG positions are plotted with vertical offsets to facilitate distinguishing each segment. Since the position data is plotted relative to the inital position and offsets are included, the zero and maximum and minimum values indicated on the vertical axis only indicate the plot scale. In the case of acceleration plots the absolute values are plotted. Since the data for multiple segments are shown in this plot it is quite complex. At the scale presented here, only the gestalt of the coordination of the segment movements can be appreciated, but at this stage that is what is important.

The third plot show the data for a single selected segment CG. This plot is labeled "Disturbance." In the case of a right arm raise maneuver test the right arm data would be plotted here and the label would be appropriate. In most other cases trunk data, as the most massive segment is plotted and the label is perhaps not appropriate. The following abbreviation are used to identify the data:

CG	Body CG
CP	Body CP
rf	Right foot CG
rs	Right shank CG
rt	Right thigh CG
ra	Right arm CG
lf	Left foot CG
ls	Left shank CG
lt	Left thigh CG
la	Left arm CG
pv	Pelvis CG
tk	Trunk CG
hd	Head CG

Figures 5.11 through 5.14 show the data for a normal stance, eyes open, test for a control subject. Figures 5.11 and 5.12 show X-axis and Z-axis displacements respectively; Figures 5.13 and 5.14 show the corresponding accelerations. Recall that the subject is facing in the -X direction and that the right to left shoulder line is approximately parallel to the Z- axis. The nearly constant differences in the body CG and CP data are due to model and body segment parameter errors as discussed in Chapter IV. The joint centers for this set of figures were determined using the axes of rotation method. The body CG and CP plots have the same general form since there are little dynamics involved

in normal stance; i.e., the quasi-static model applies well. The body CG data and the trunk data ("Disturbance") have very similar forms. This will in general be true; the trunk, by far the most massive segment, dominates in defining the body CG. There are, however, small but probably highly significant differences which will be more apparent in subsequent data.

The X-axis displacement data appears to be consistent with an inverted pendulum model of fore-and-aft sway. The quite small segment displacements display approximately the same shape and the magnitude seems to increase proportionally as distance from the foot (ankle joint) increases. Very little Z-axis displacement is observed. In the acceleration plots the initial and final transients are due to the differentiating algorithm and should be ignored. Recall that the absolute value of acceleration data is plotted. Again the similarity of form of the X-axis acceleration plots is consistent with an inverted pendulum model of sway.

Note that the body CG acceleration data is similar to the trunk acceleration data but has a smaller magnitude. The similarity of form is to be expected since, as previously stated, the trunk mass is very significant in determining the combined CG location. The fact that the combined or body CG undergoes smaller accelerations than the trunk raises the possibility that the movement of the other

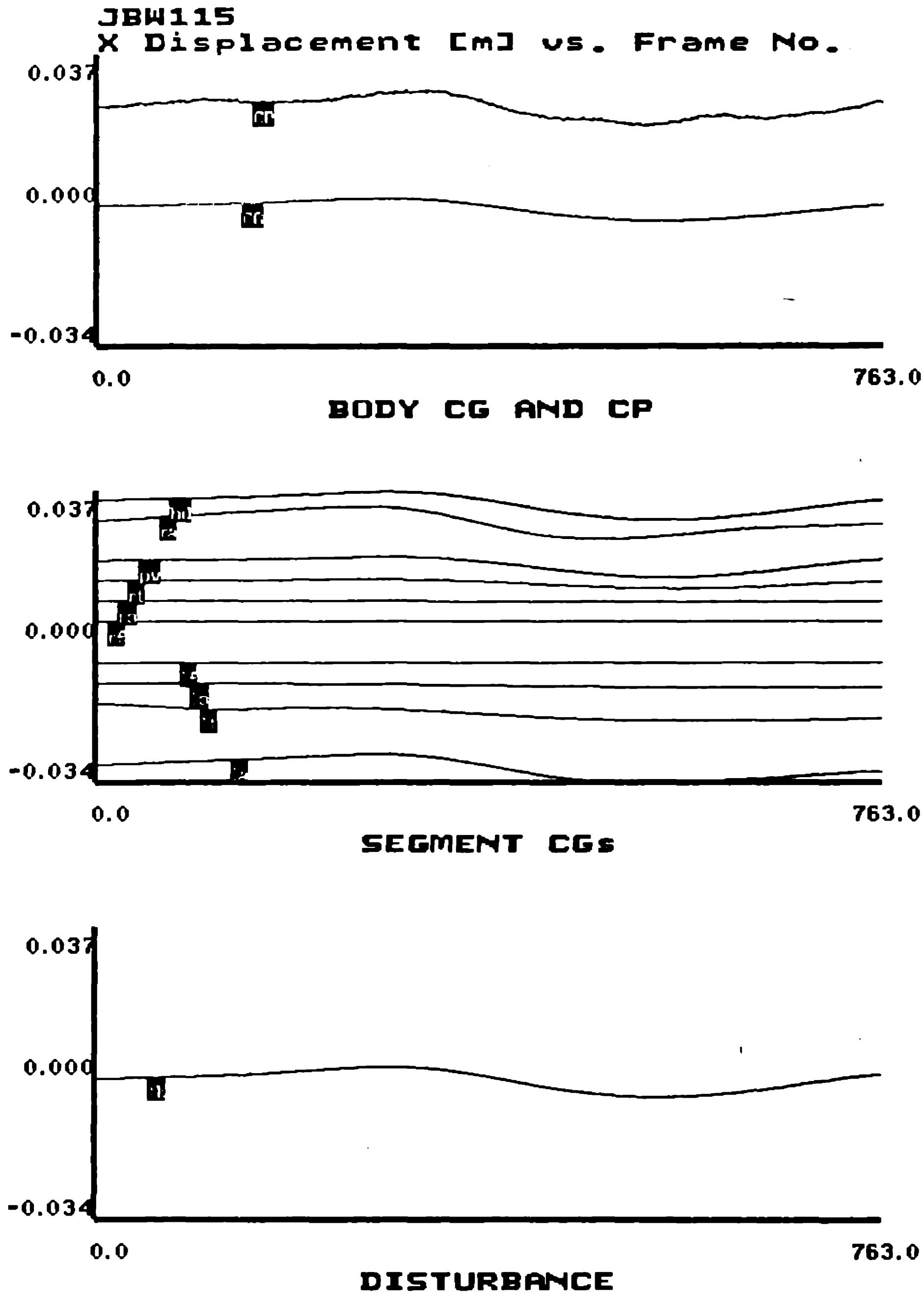

Figure 5.11 Control normal stance, eyes open
X displacement

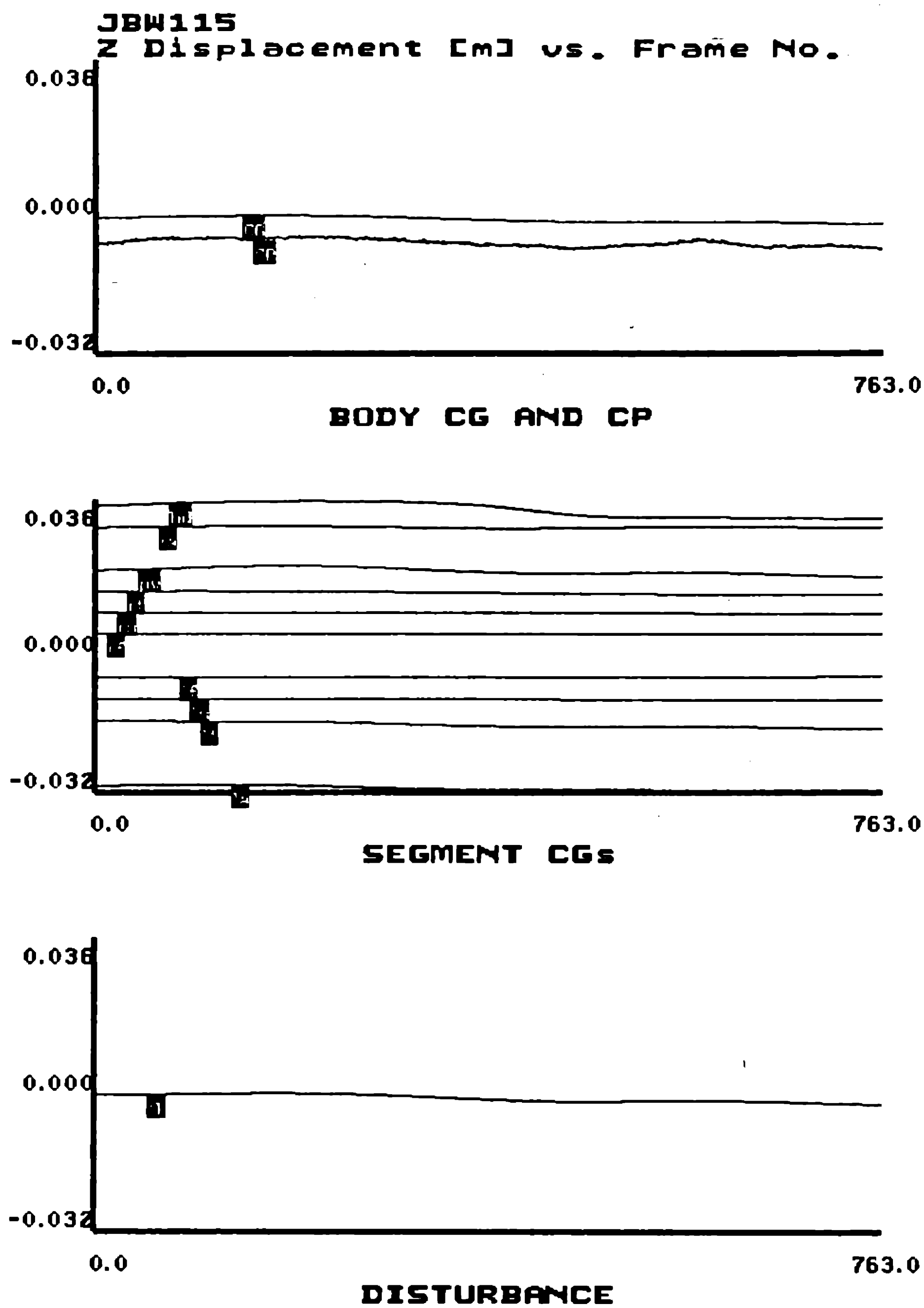

Figure 5.12 Control normal stance, eyes open
Z displacement

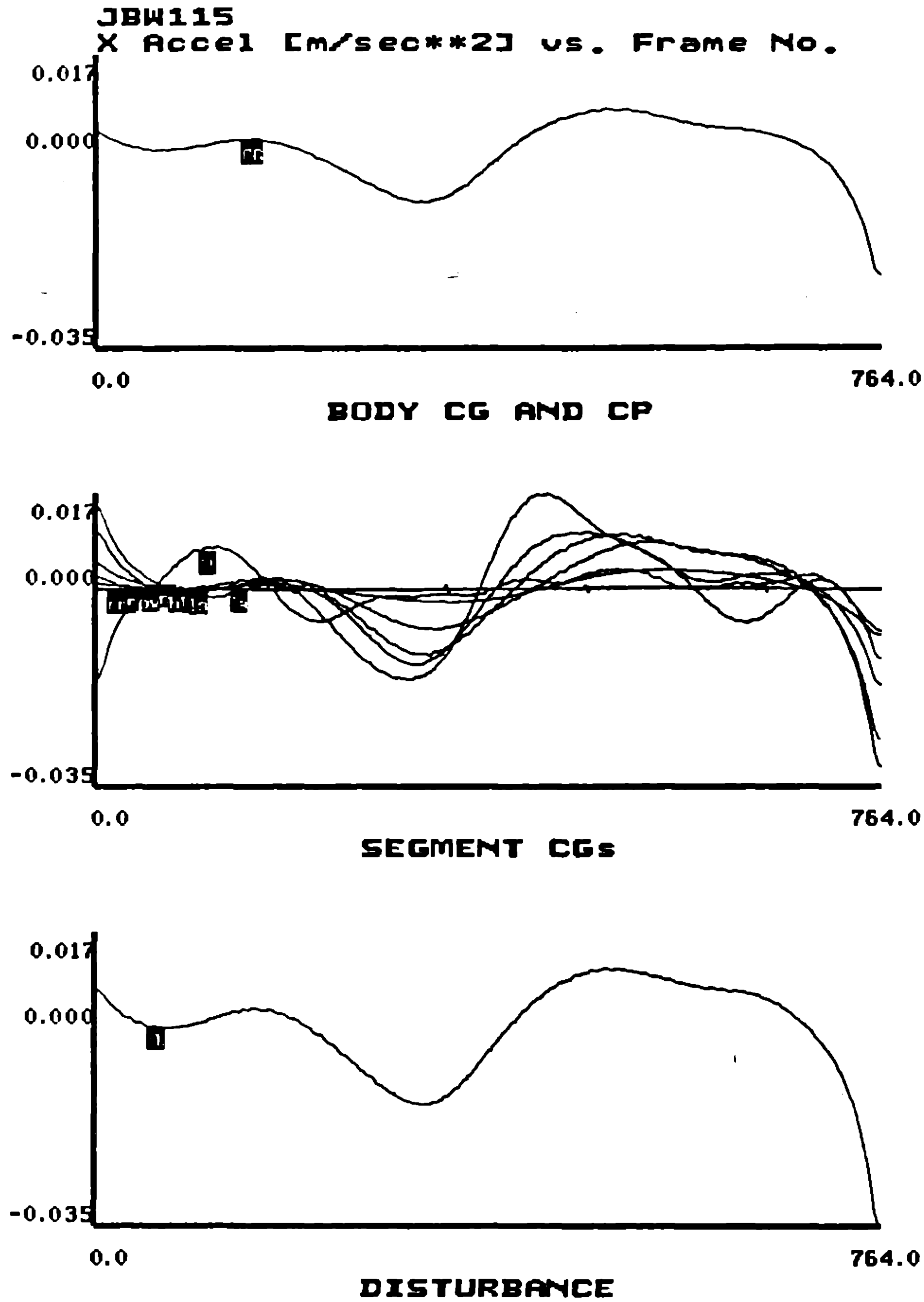

Figure 5.13 Control normal stance, eyes open X acceleration

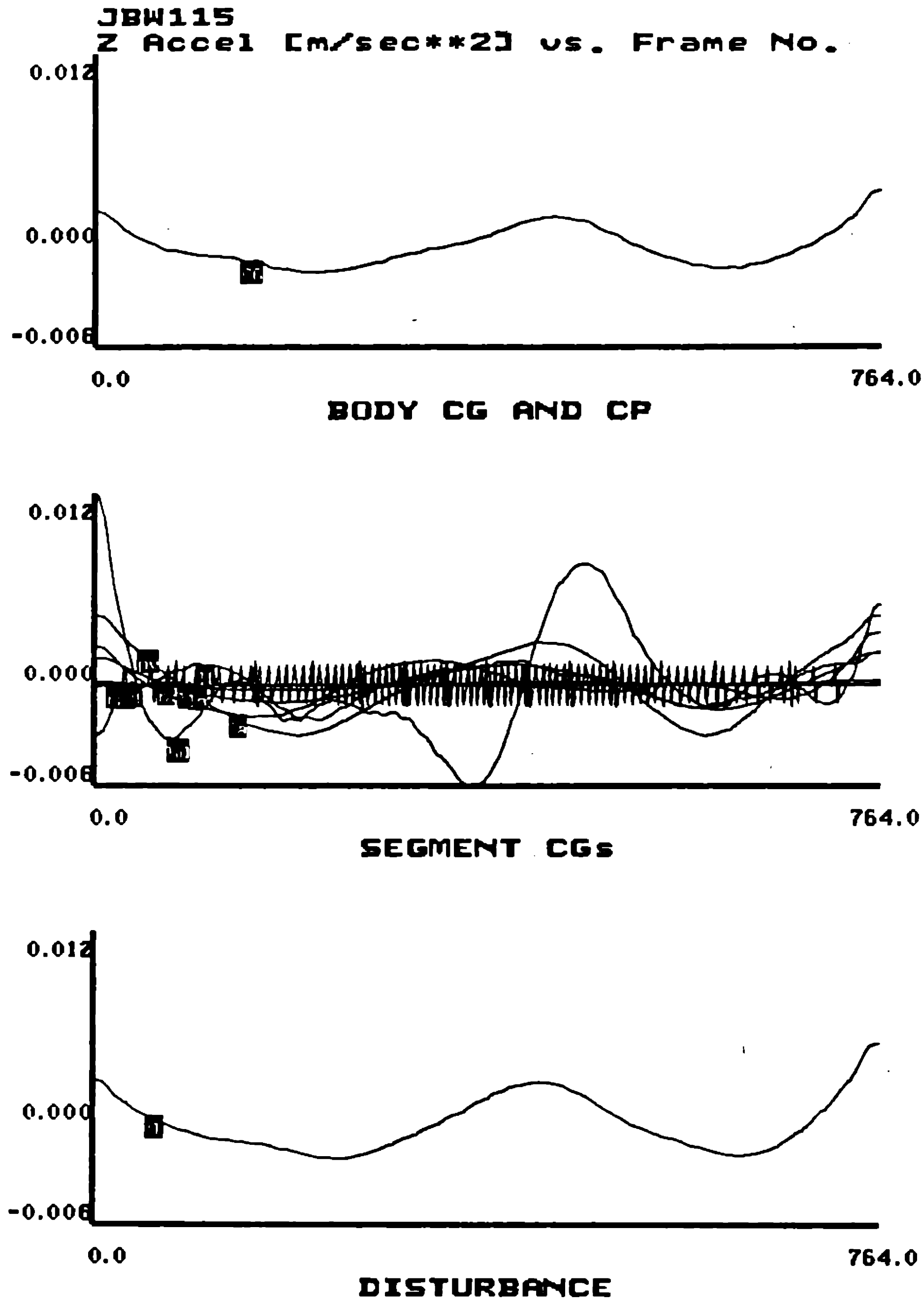

Figure 5.14 Control normal stance, eyes open Z acceleration

body segments is somehow coordinated to minimize the effect of trunk displacements on the body CG. This effect will become more apparent when data involving large displacements such as gait and rising from a chair are shown. In these cases the effect will be seen in the displacement data as well as the acceleration data.

Figures 5.15 through 5.20 show similar data for a normal stance eyes open trial of a spastic quadraplegic cerebral palsied, subject. The joint centers for this set of data were determined using the anatomical landmark method. In this case the Y-axis data is also plotted. This data was not plotted for the control because no significant Y-axis displacements are seen in normals. Y-axis displacements and accelerations are only observed in quadraplegic subjects with significant crouch deformity. The X-axis and Z-axis displacements and accelerations are greater than observed in the normal. There is less similarity of form in the segment data. It does not appear that the body is simply rotating about the ankle joint. The difference in the form of the body CG data and the CP data suggest that the movements in standing posture are significant enough to introduce dynamic effects.

Figures 5.21 and 5.22 show the Y-axis body CG acceleration and total vertical force plots for another quadraplegic subject with a severe crouch deformity. The vertical oscillation of the CG is similar to that seen in

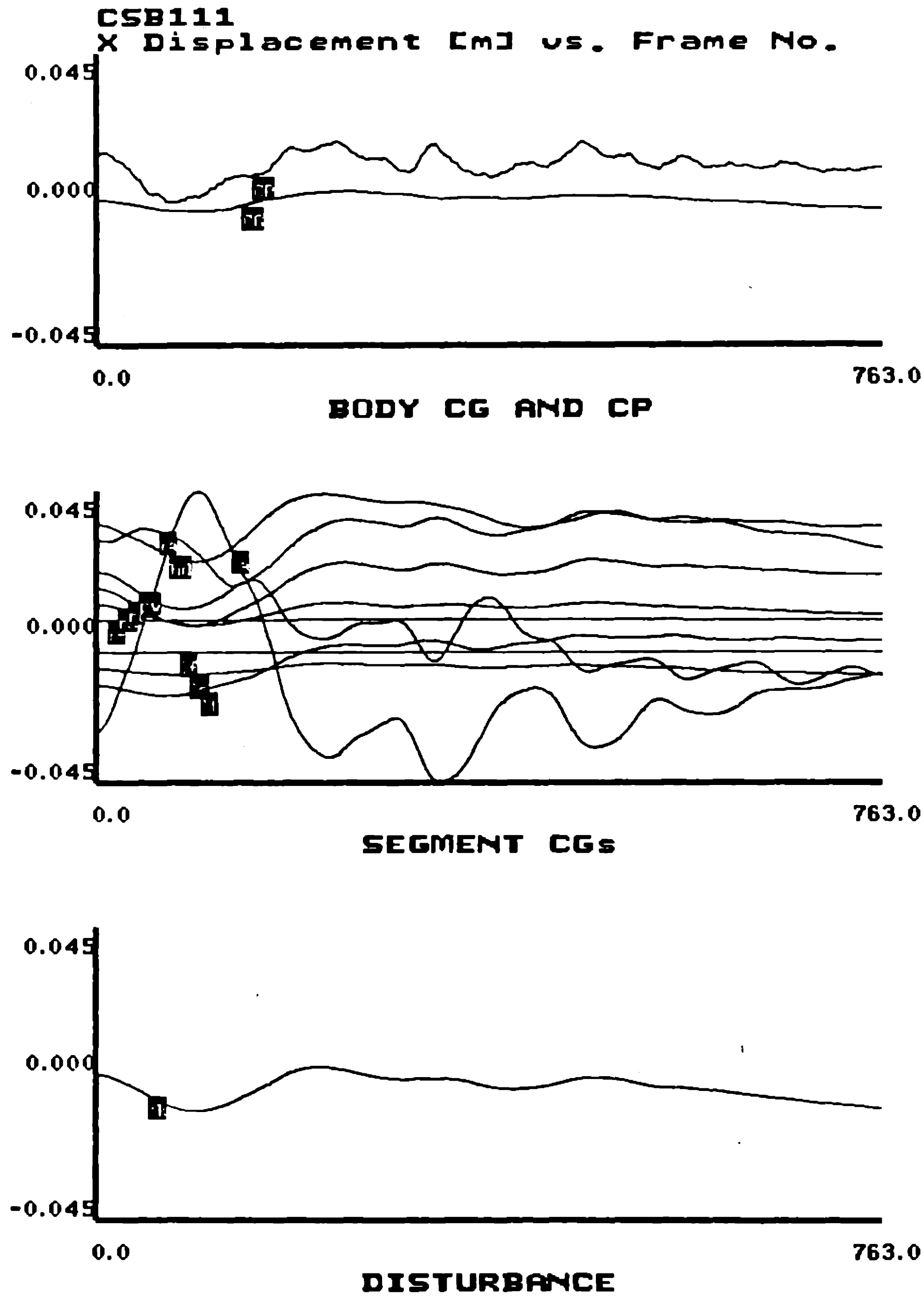

Figure 5.15 Cerebral Palsey subject, normal stance, eyes open, X displacement

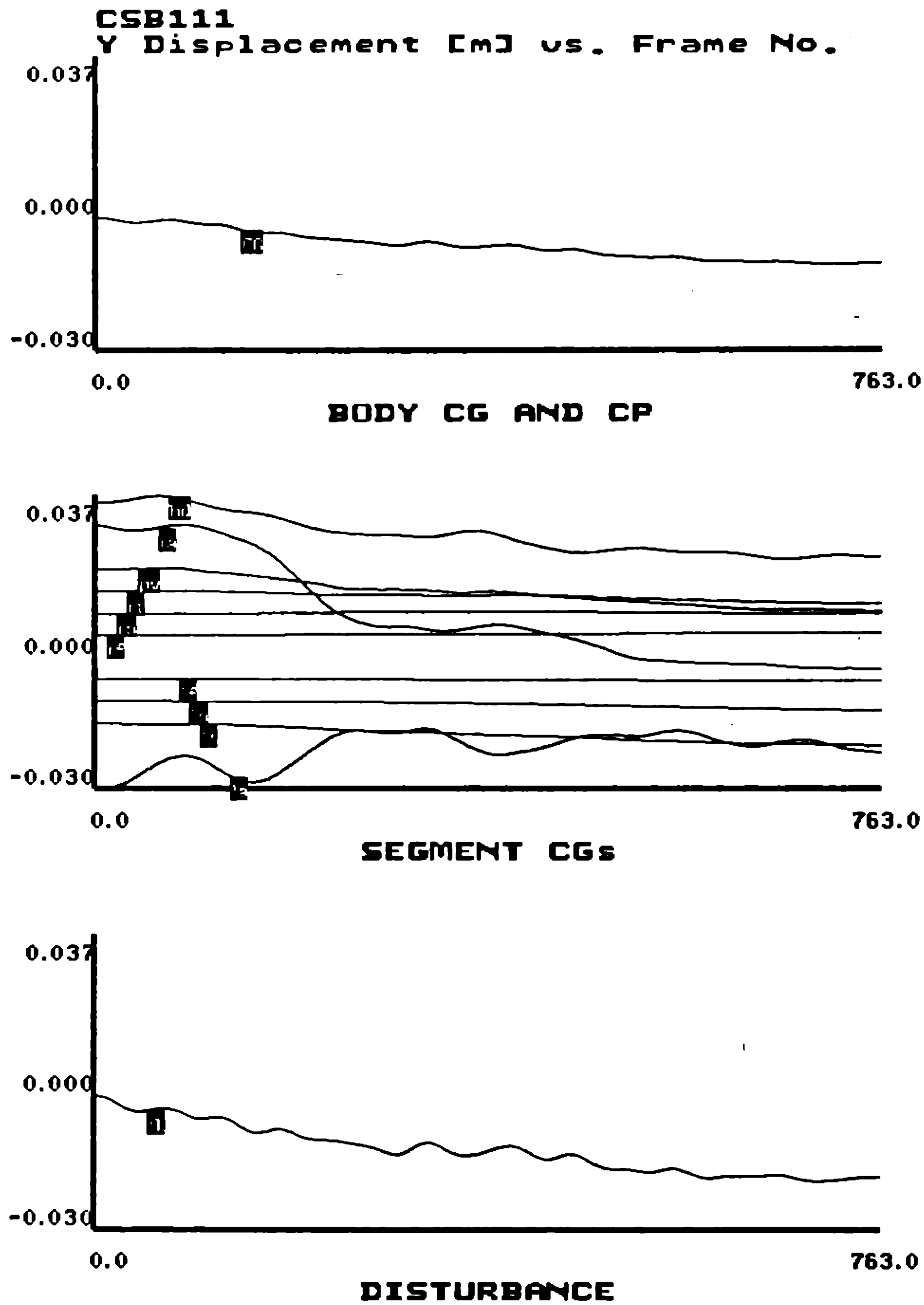

Figure 5.16 Cerebral Palsey subject, normal stance, eyes open, Y displacement

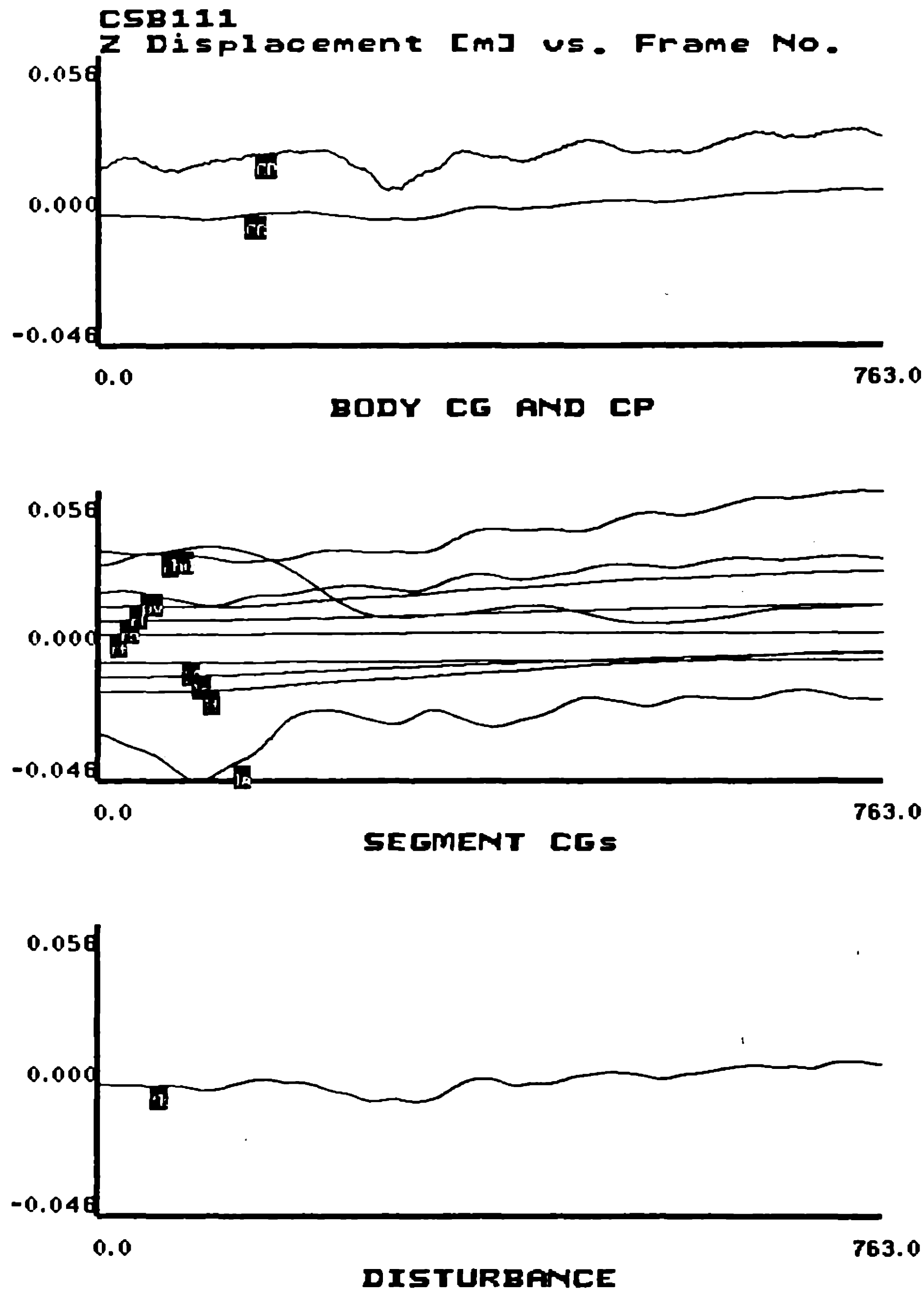

Figure 5.17 Cerebral Palsey subject, normal stance, eyes open, Z displacement

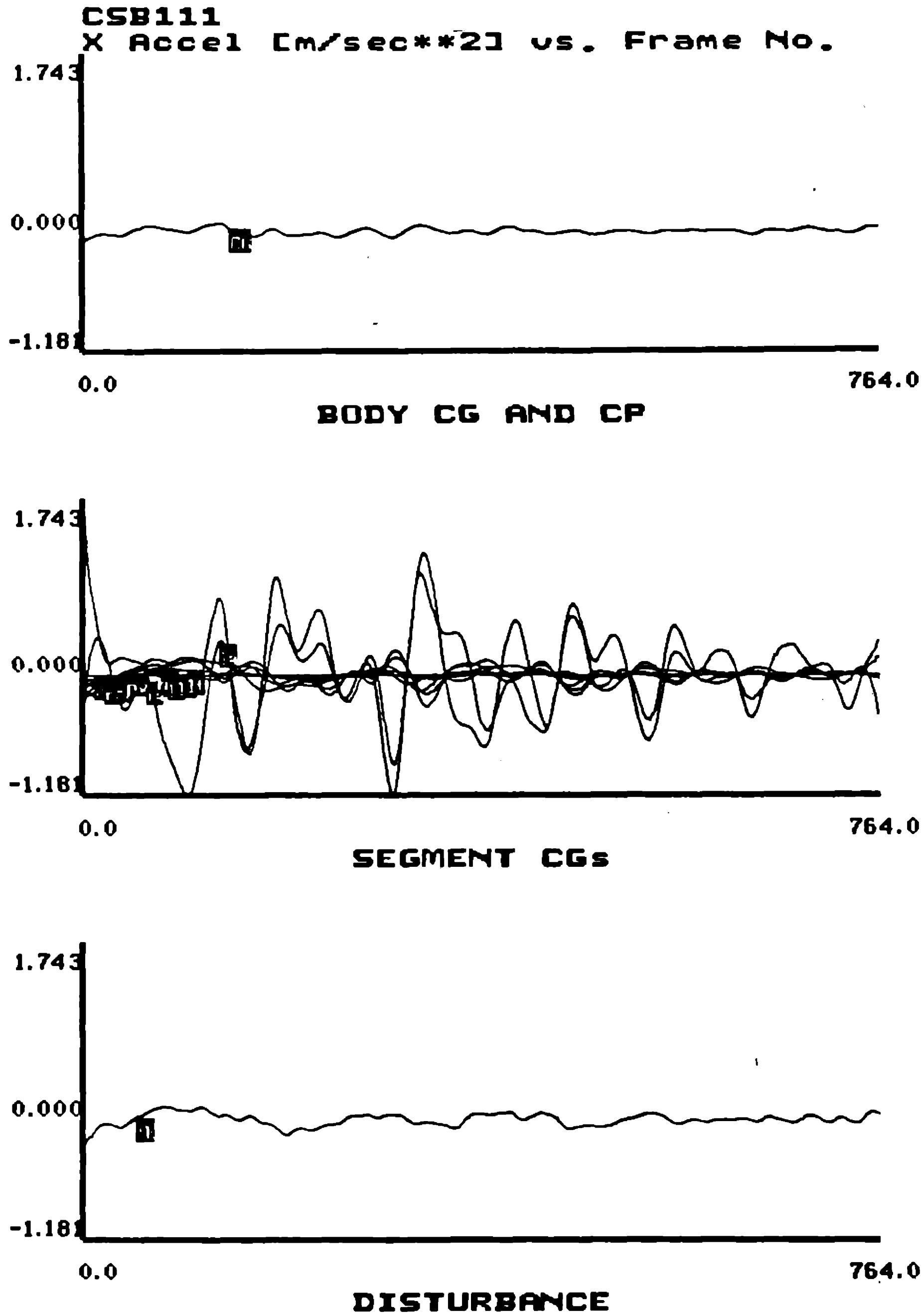

Figure 5.18 Cerebral Palsey subject, normal stance, eyes open, X acceleration

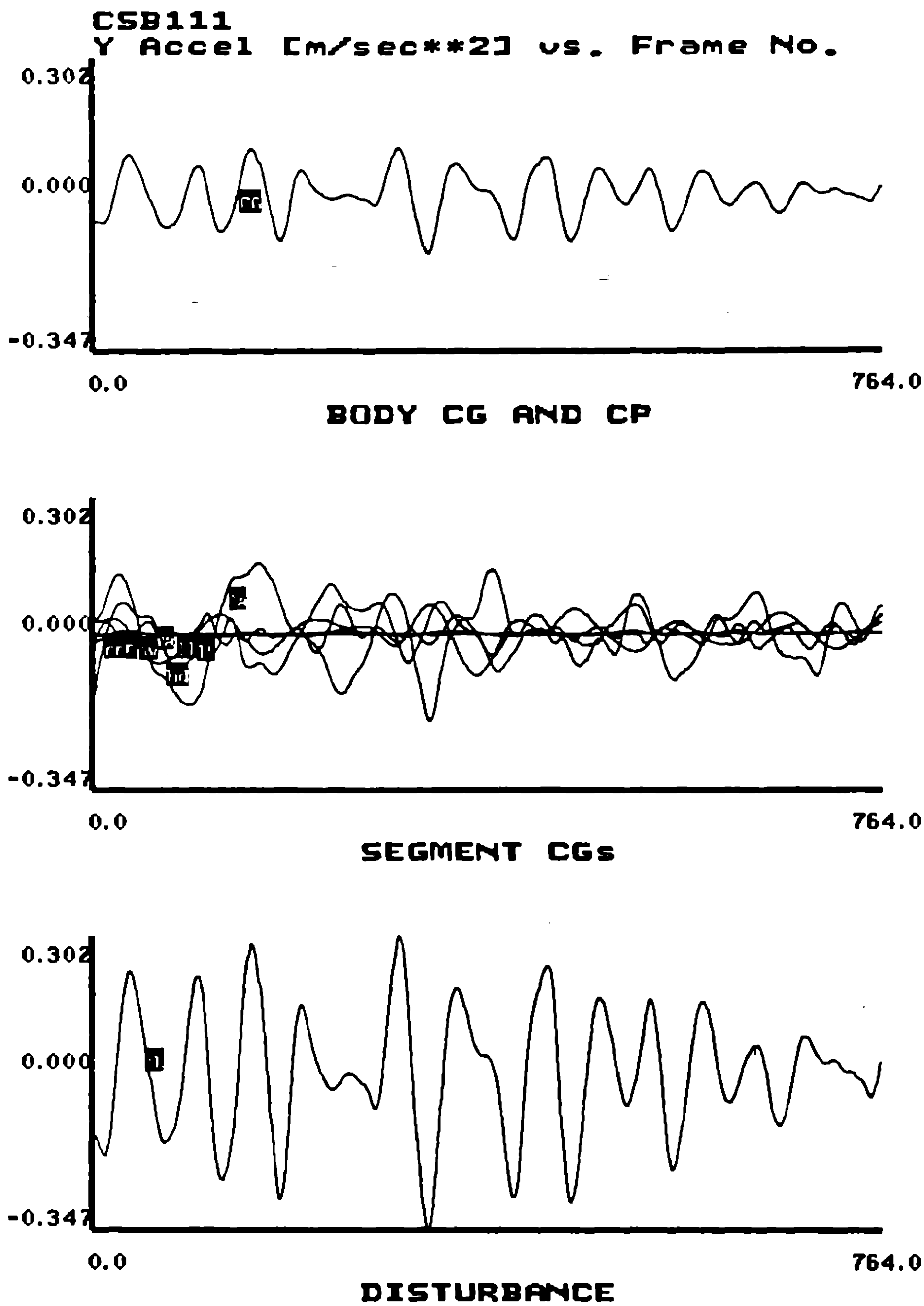

Figure 5.19 Cerebral Palsey subject, normal stance, eyes open, Y acceleration

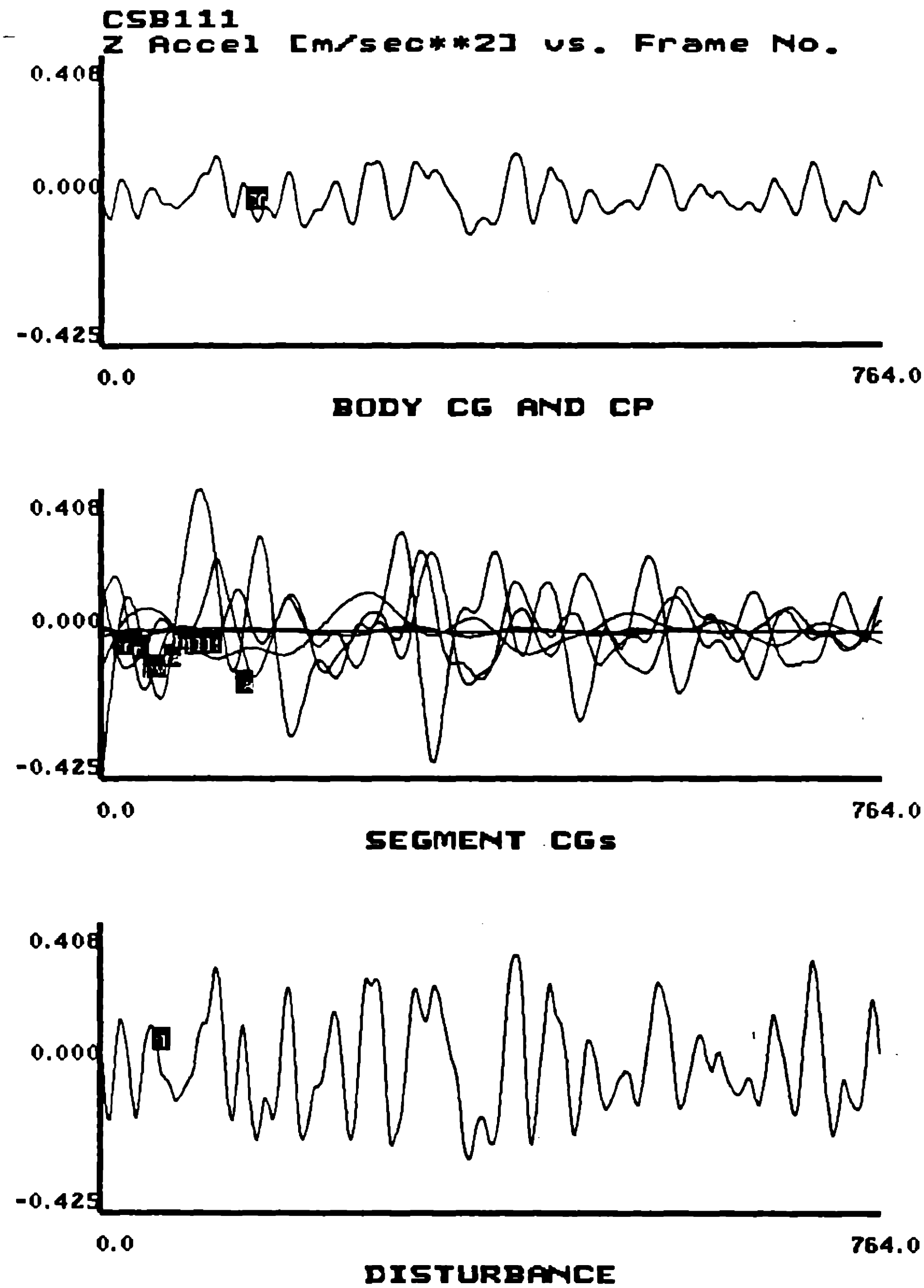

Figure 5.20 Cerebral Palsey subject, normal stance, eyes open, Z acceleration

the previous subject. The similarity of the CG oscillations and the vertical force oscillations is reassuring and suggests that the estimated CG oscillations are real. An upward CG acceleration produces an increase in the negative vertical component of the ground reaction force; therefore, the plots should be inverted images of each other. The correspondence is not, of course, perfect. We are at the limits of precision of both the kinematic and force plate data; a few millimeters of displacement and force changes of about 1% of full scale. If body weight is estimated using mass = delta acceleration divided by delta force for the oscillations which correspond in time the results are accurate to within a factor of two which is quite good given that the force oscillation are only a percent or so of body weight.

Figures 5.23, 5.24 and 5.25 show the displacement data for a single stride of gait for a normal subject whose joint centers were defined from axes of rotation. Figure 5.23, the X-axis displacement data, highlight the exquisite symmetry of normal gait, the perfectly matched lower extremity displacement produce a near straight line, i.e. uniform velocity, displacement of the trunk and the body CG. The Y-axis data shows a slight rise and fall of the trunk and body CG during the stride. Note, that the rise and fall of the body CG is slightly less than the rise and fall of the trunk, again indicating the possibility of coordination

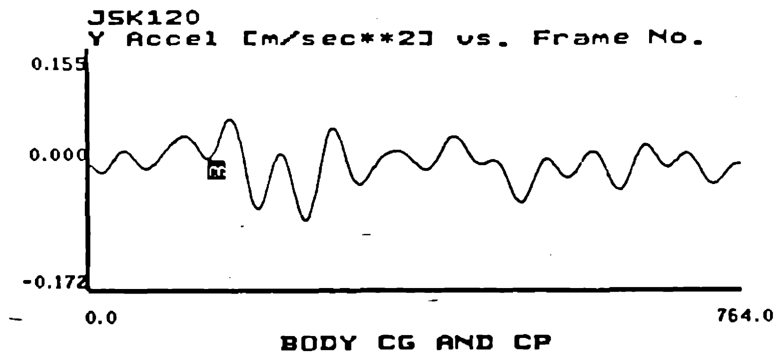

Figure 5.21 Subject JSK vertical body CG acceleration

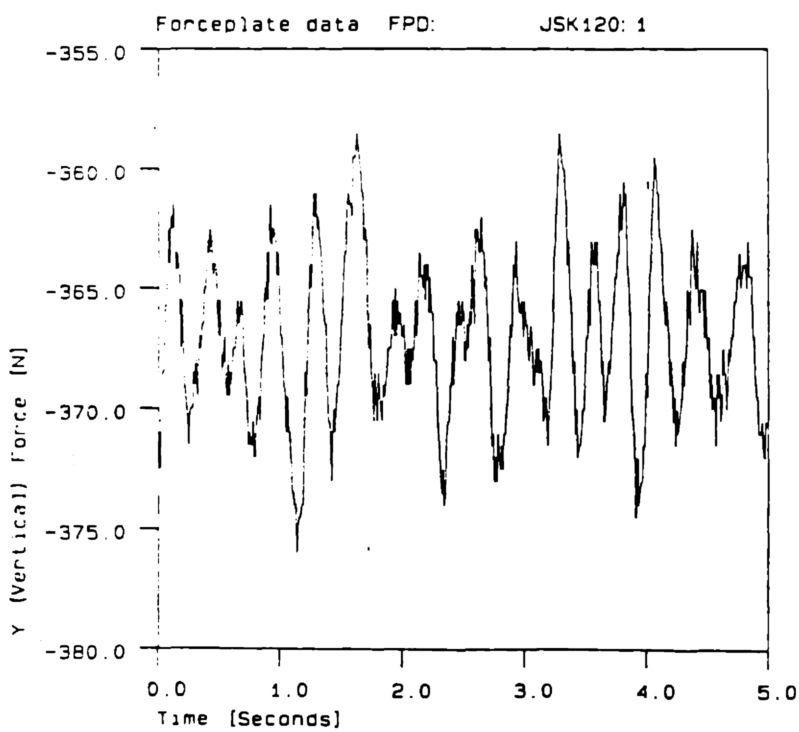

Figure 5.22 Subject JSK total vertical force

of segment movements to smooth the CG path. The Z-axis data show the shift of the CG to the side of the weight bearing lower limb. Again the CG path appears to be slightly smoother than the path of the trunk.

Figure 5.26 shows the X-axis displacement data for a step taken by a quadraplegic cerebral palsied subject. While the form is basically similar the symmetry of the lower limb motion is less perfect. The result is a small but preceptible deviation in the trunk path from a straight line. The effect of the trunk deviations on the body CG is not apparent. Unfortunately the quality of these data sets did not permit comparing the acceleration data which would be a more sensitive indicator of the effects with which we are concerned.

Figure 5.27 which shows the X-axis velocity data for the control subject gait data indicates that there is some variation in the X direction velocity. Conventional wisdom is that in normal level walking the X velocity varies out of phase with the body CG Y-axis position; so that as the CG location varies from the highest to the lowest a portion of the lost potential energy is transferred to forward kinetic energy. Recalling that the subject is traveling in the -X direction so that the maximum forward velocity is the lowest point on the body CG X-axis velocity plot, we see that indeed the peak in the forward velocity does occur at the same time as the lowest Y position of the body CG.

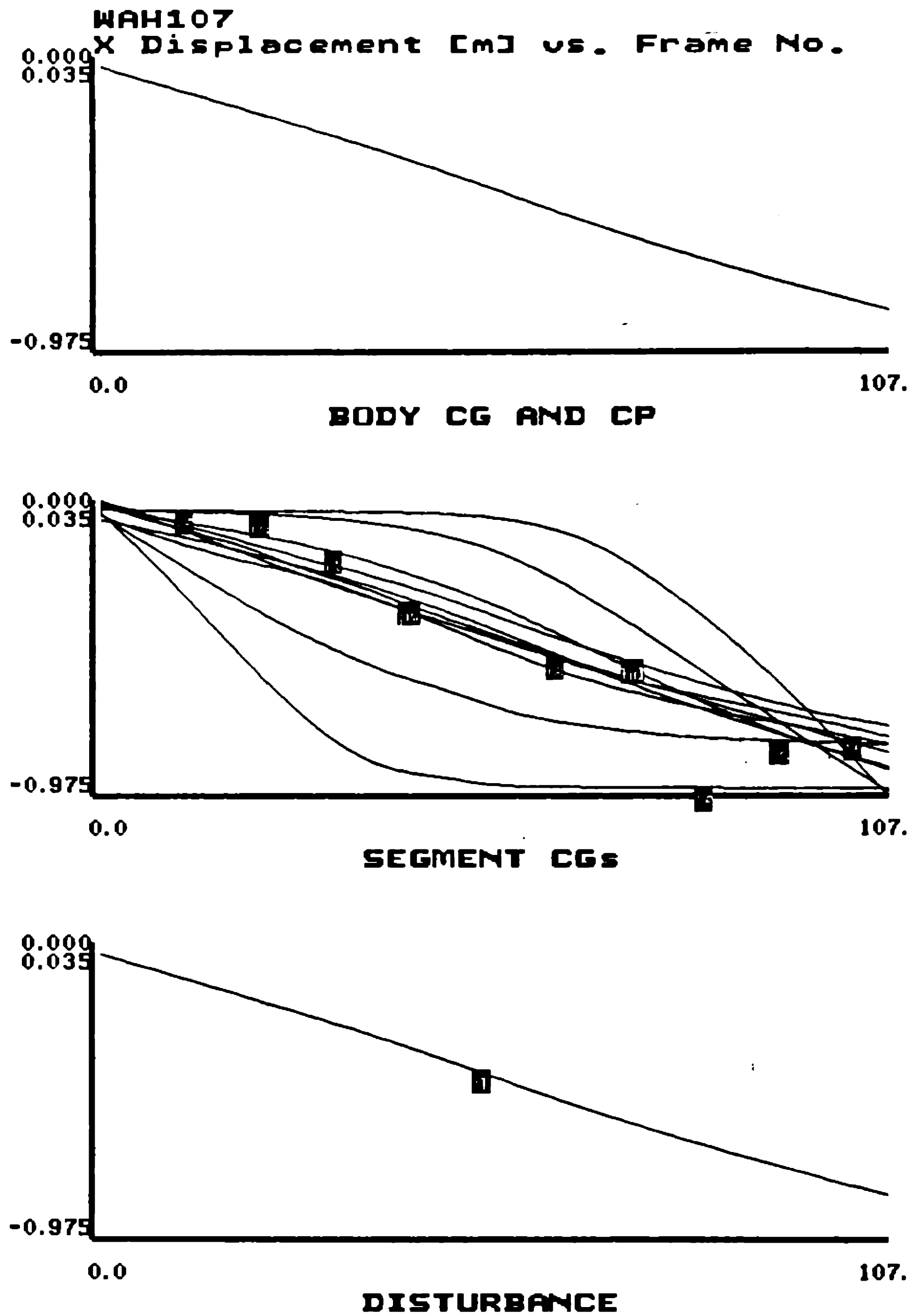

Figure 5.23 Control subject, gait X displacement

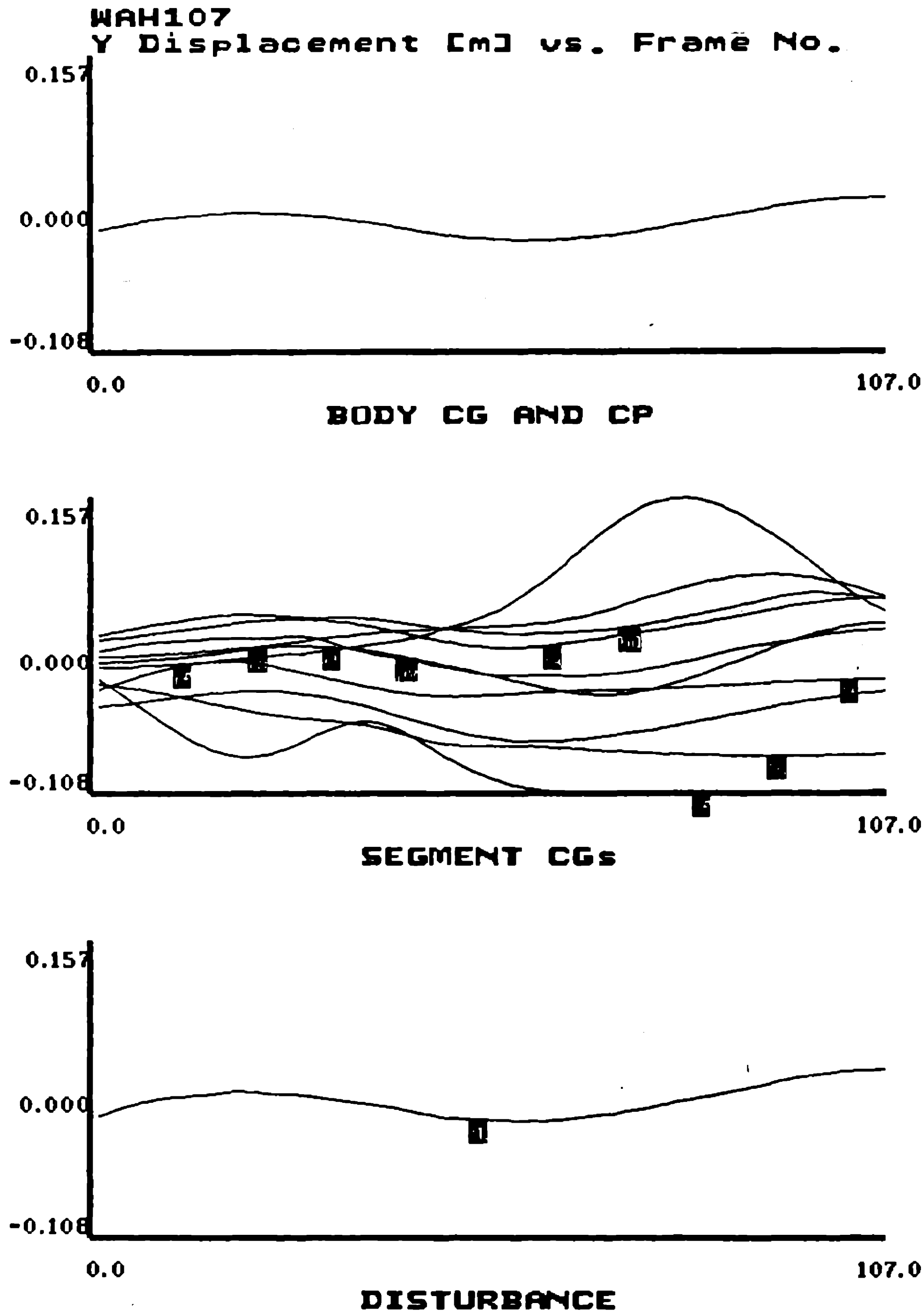

Figure 5.24 Control subject, gait Y displacement

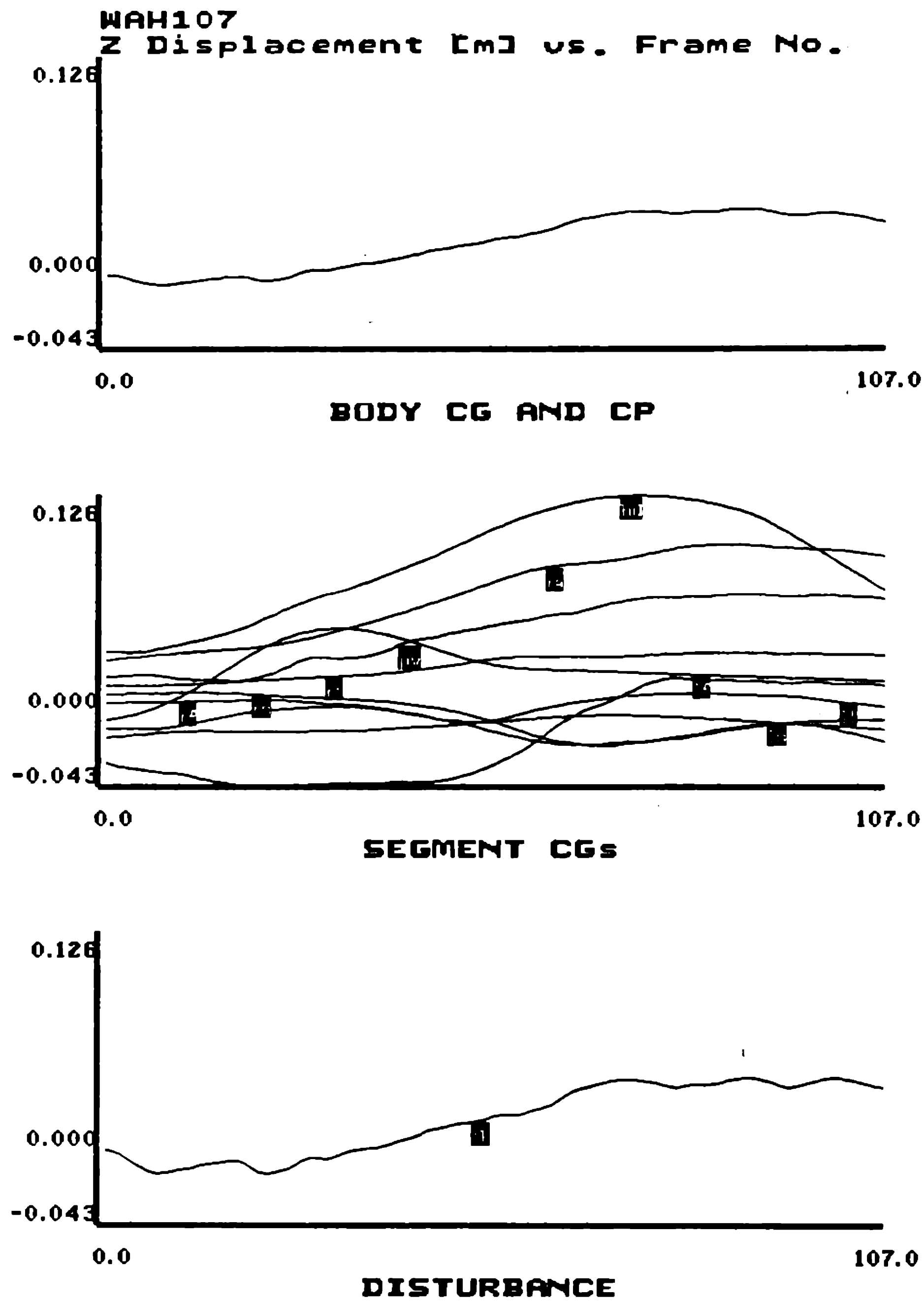

Figure 5.25 Control subject, gait
Z displacement

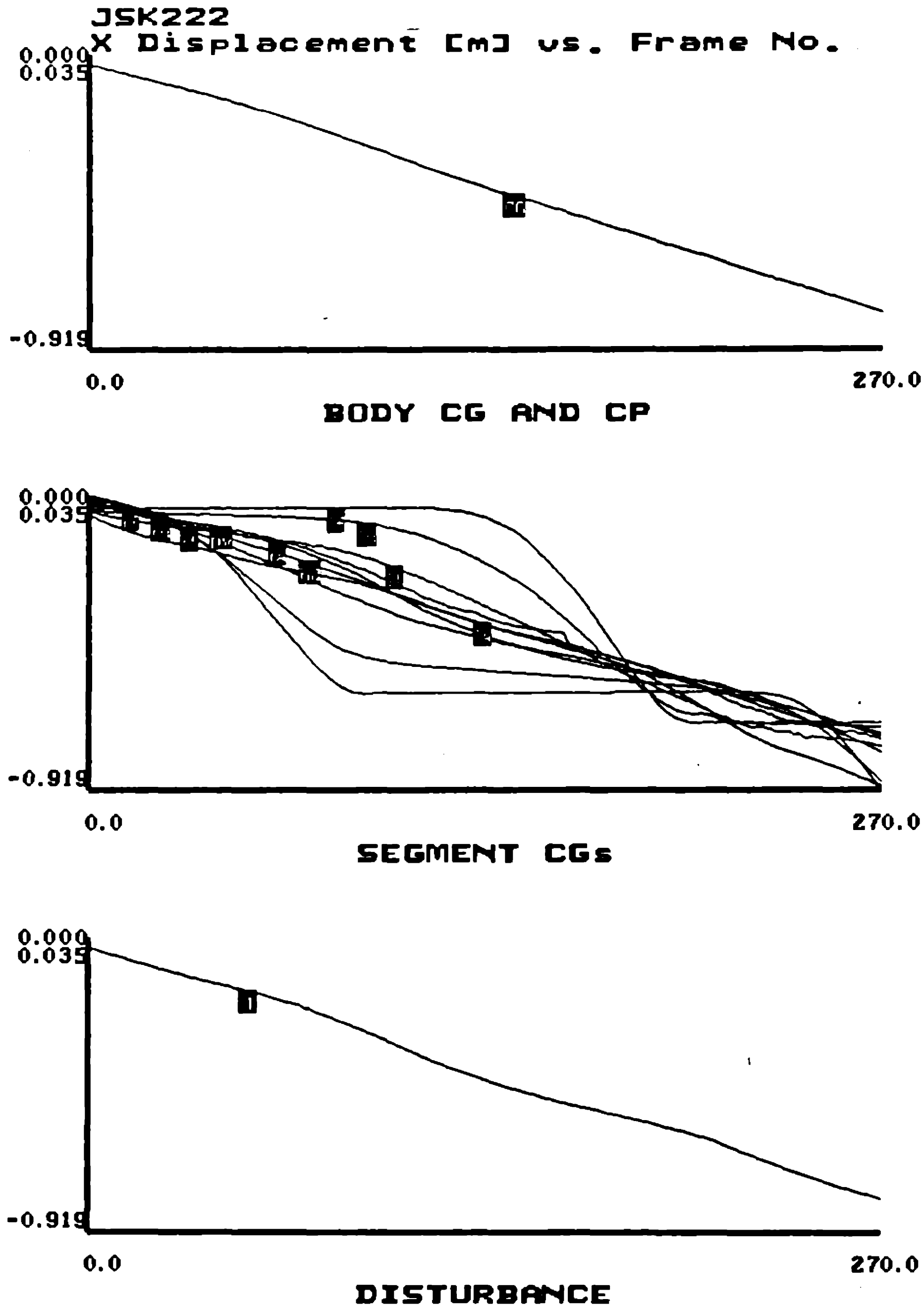

Figure 5.26 Cerebral Palsy subject, gait X displacement

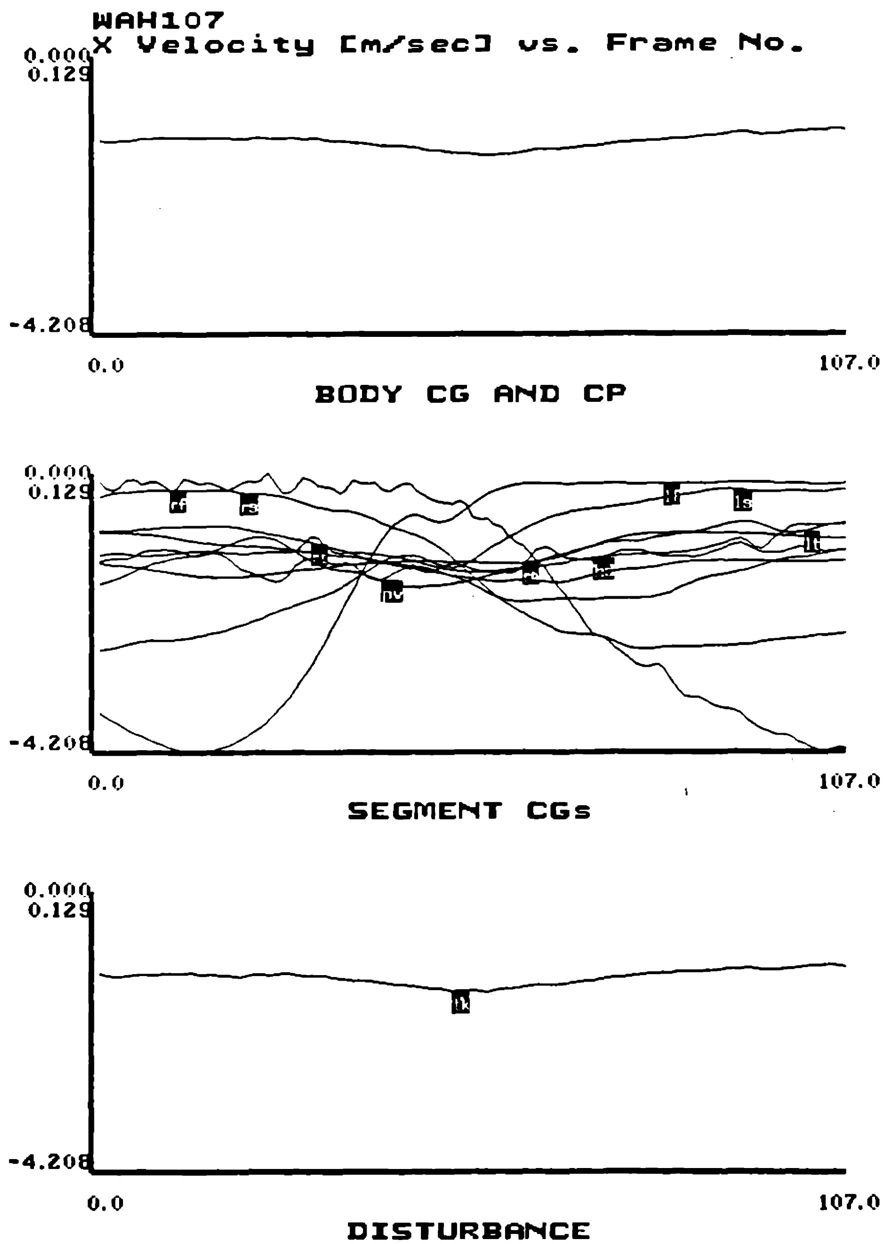

Figure 5.27 Control Subject, gait X velocity

Figures 5.28 through 5.31 show the X-axis and Y-axis data for a control subject doing a chair rise maneuver. Since initially only a portion of the subject's weight is on the force plate there is a large difference between the body CG and the CP. As the subject rises to an erect standing position all weight is transfered to the force plates and the CG and CP positions converge. Note that the trunk and body CG data are similar and comparatively smooth and that the body CG displacements and accelerations are slightly smaller that those of the trunk CG. Note also that the trunk and body CG accelerate forward (into the -X direction) then decelerate (X direction acceleration) as they begin to accelerate in the Y direction. That is, the trunk accelerates forward horizontally first, then loses horizontal velocity as the body begins to pick up vertical velocity. One can hypothesize two functional sequences for this movement pattern. First, the body CG is moved forward off the seat and more nearly over the feet in preparation for shifting support from the seat and feet to the feet alone. Second, the trunk forward rotation establishes momentum which may then be used to provide impetus for the rise maneuver.

Figures 5.32 and 5.33 show the X- axis and Y-axis displacement data for a chair rise maneuver performed by a quadraplegic cerebral palsied subject. The flow of this maneuver is obviously less smooth and less well coordinated.

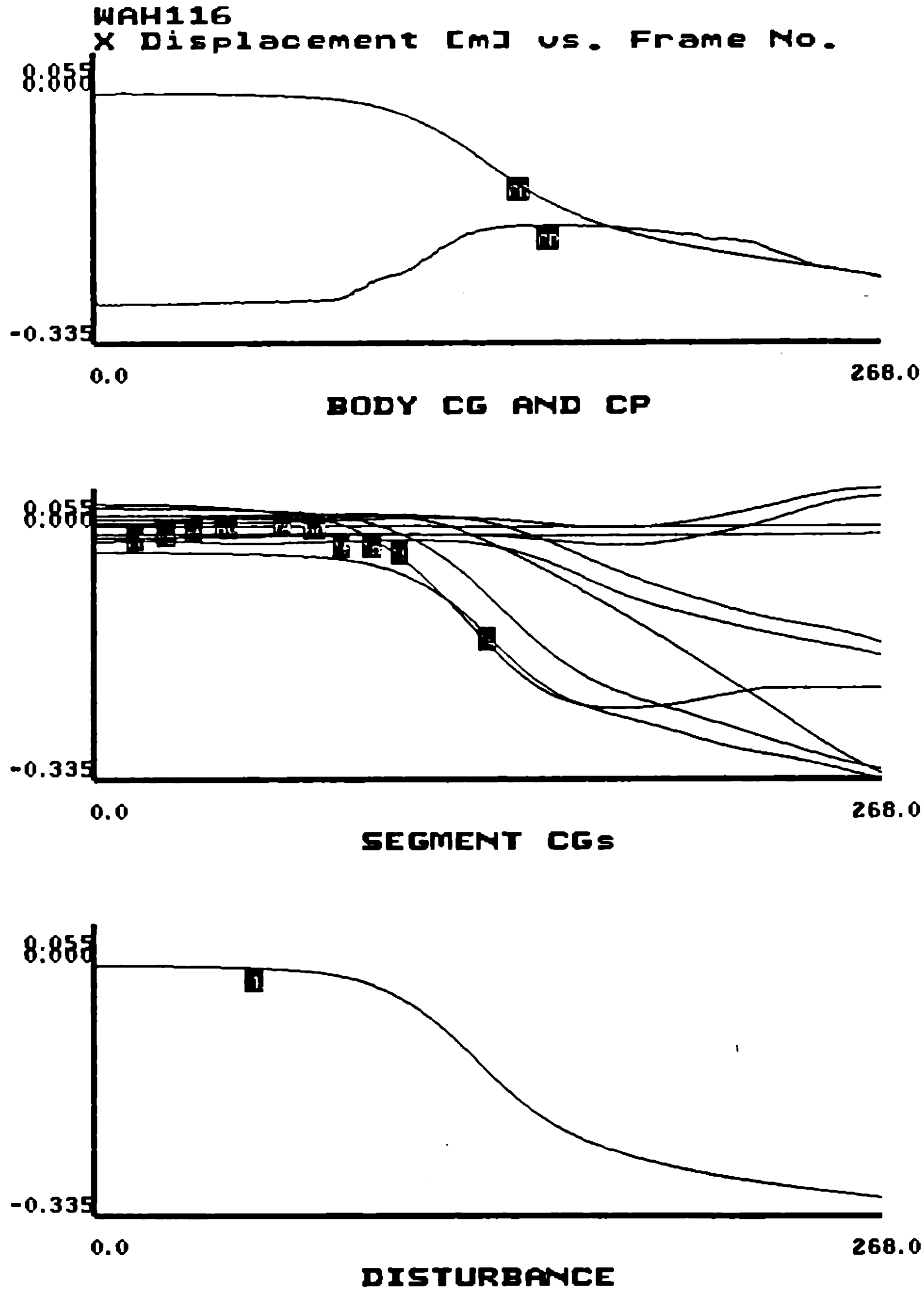

Figure 5.28 Control subject, chair rise X displacement

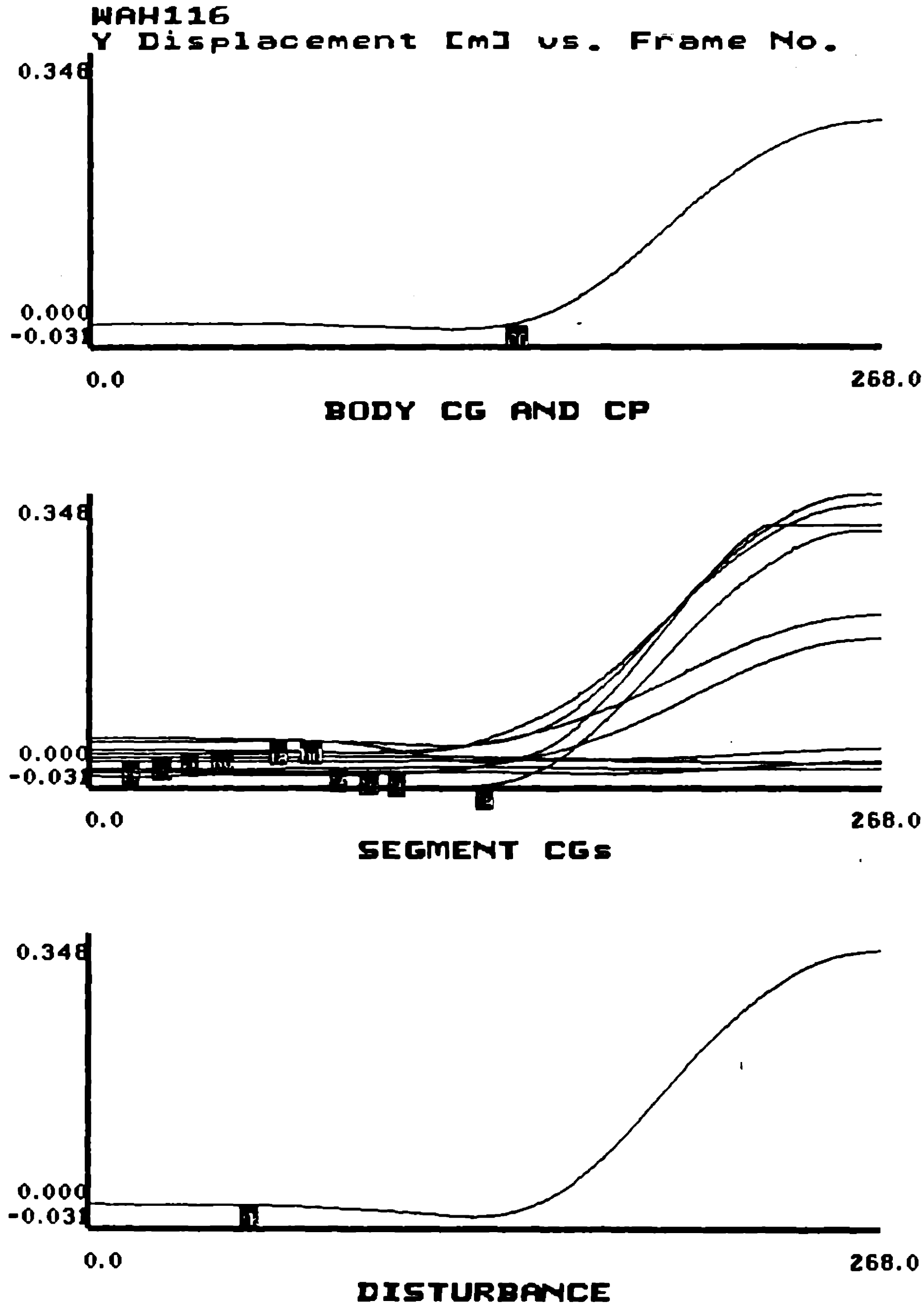

Figure 5.29 Control subject, chair rise Y displacement

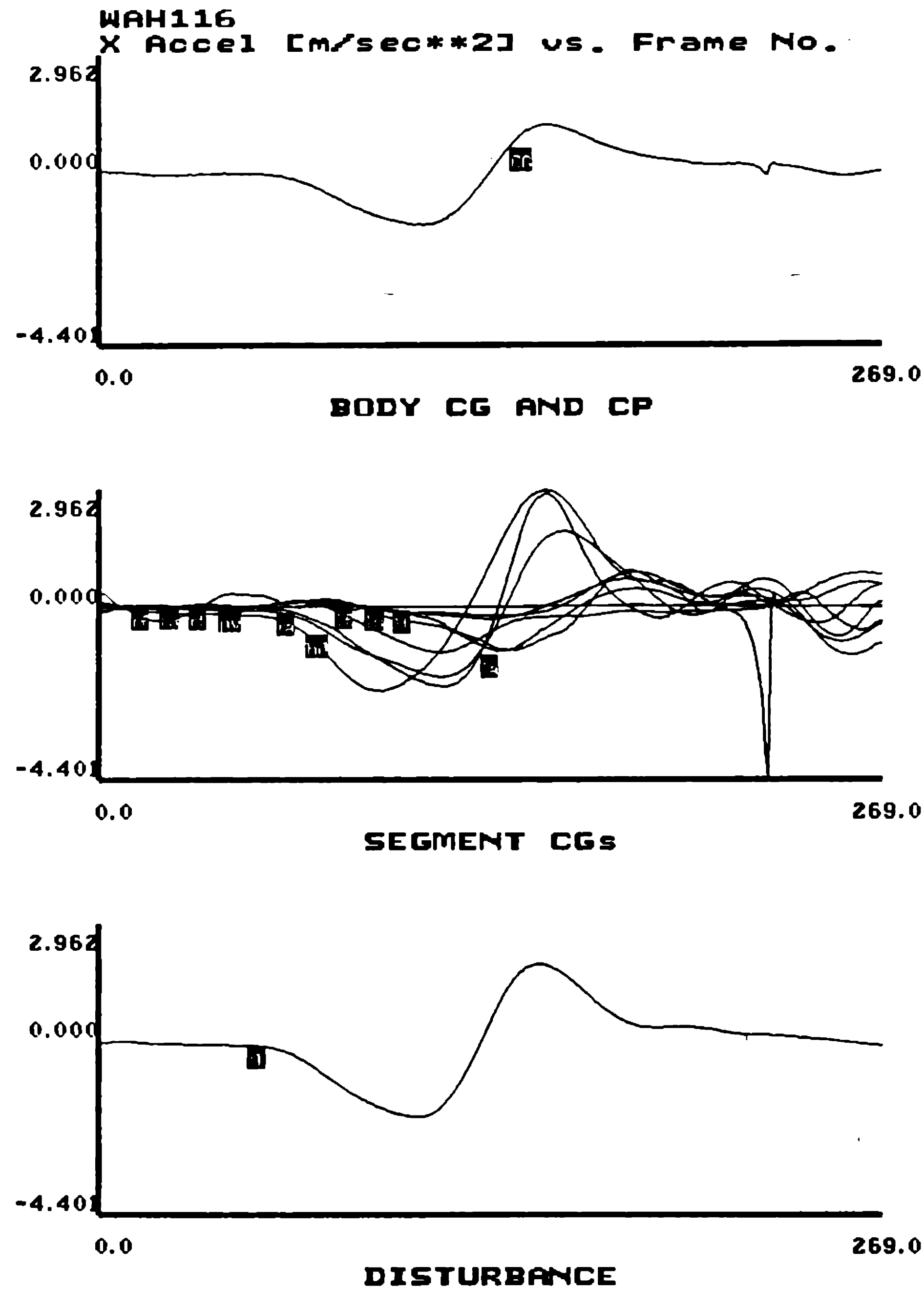

Figure 5.30 Control subject, chair rise X acceleration

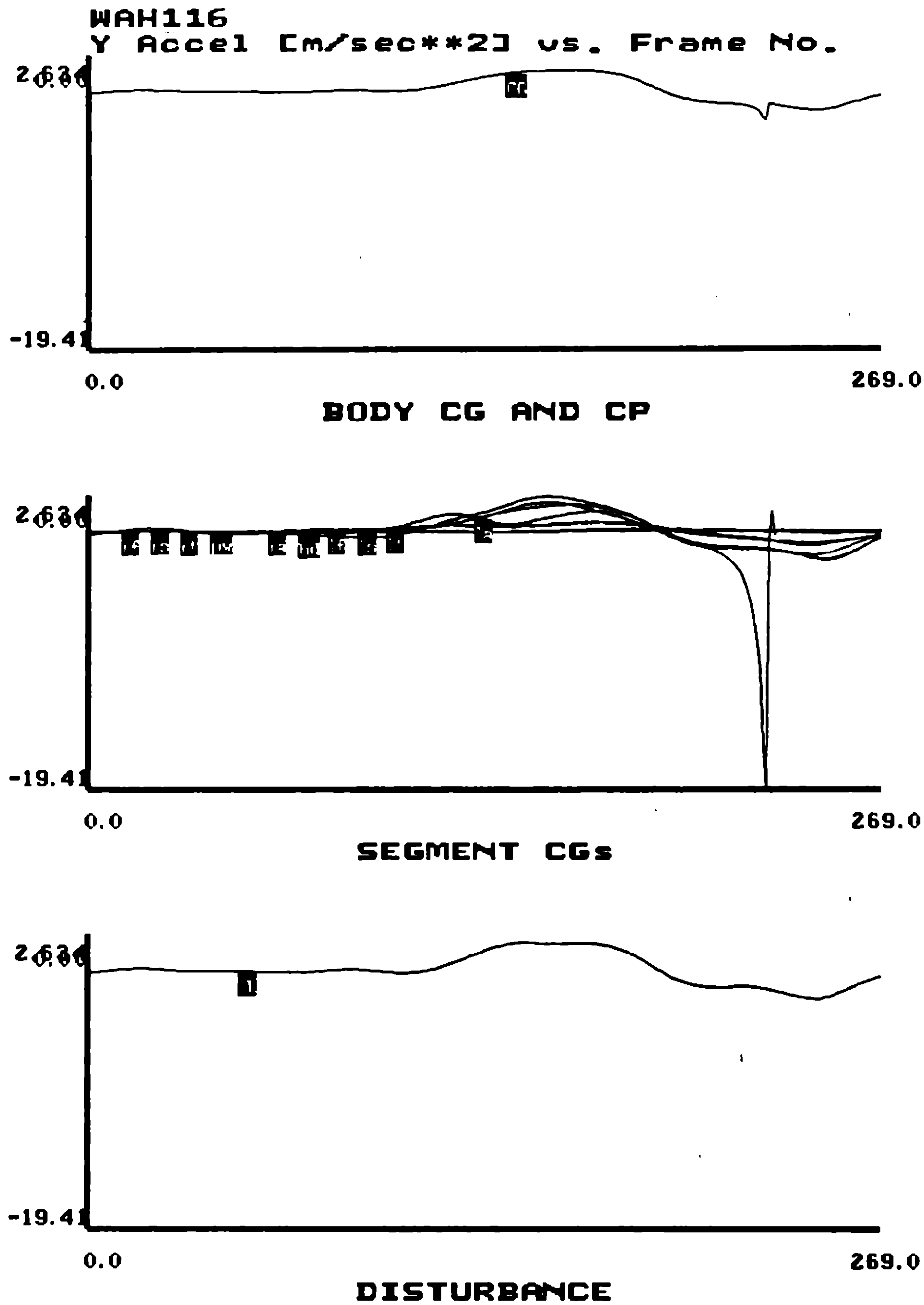

Figure 5.31 Control subject, chair rise Y acceleration

Note, the joint center were defined by the axes of rotation method for the gait and chair rise data.

Figures 5.34 through 5.37 show the X-axis and Y-axis data for a control subject doing the right arm raise maneuver. The displacement data indicate that there are small postural compensations associated with the right arm raise maneuver. The principle activity occurs in the trunk and head with some additional adjustment of the contralateral lower limb. The body CG displacement and accelerations are quite small. The acceleration data illustrate the coordination of compensation very well. With the exception of the contralateral arm, the acceleration patterns of the other body segments are scaled mirror images of the acceleration pattern of the right arm; i.e. they are exactly compensatory. The head CG appears to travel about 2 cm. back and 1 cm. up during the compensation process. The most normal case observed by Clement (12), preflight data for a non-space experienced cosmonaute, showed similar results.

Figures 5.38 through 5.41 show the same data for a quadraplegic cerebral palsied subject. In both cases anatomical landmarks were used to define joint centers. The body CG excursion is larger as would be expected from the earlier tests. There is evidence of adjustment of the position of all body segments during the maneuver. The body segment kinematics pattern is quit complex suggesting that

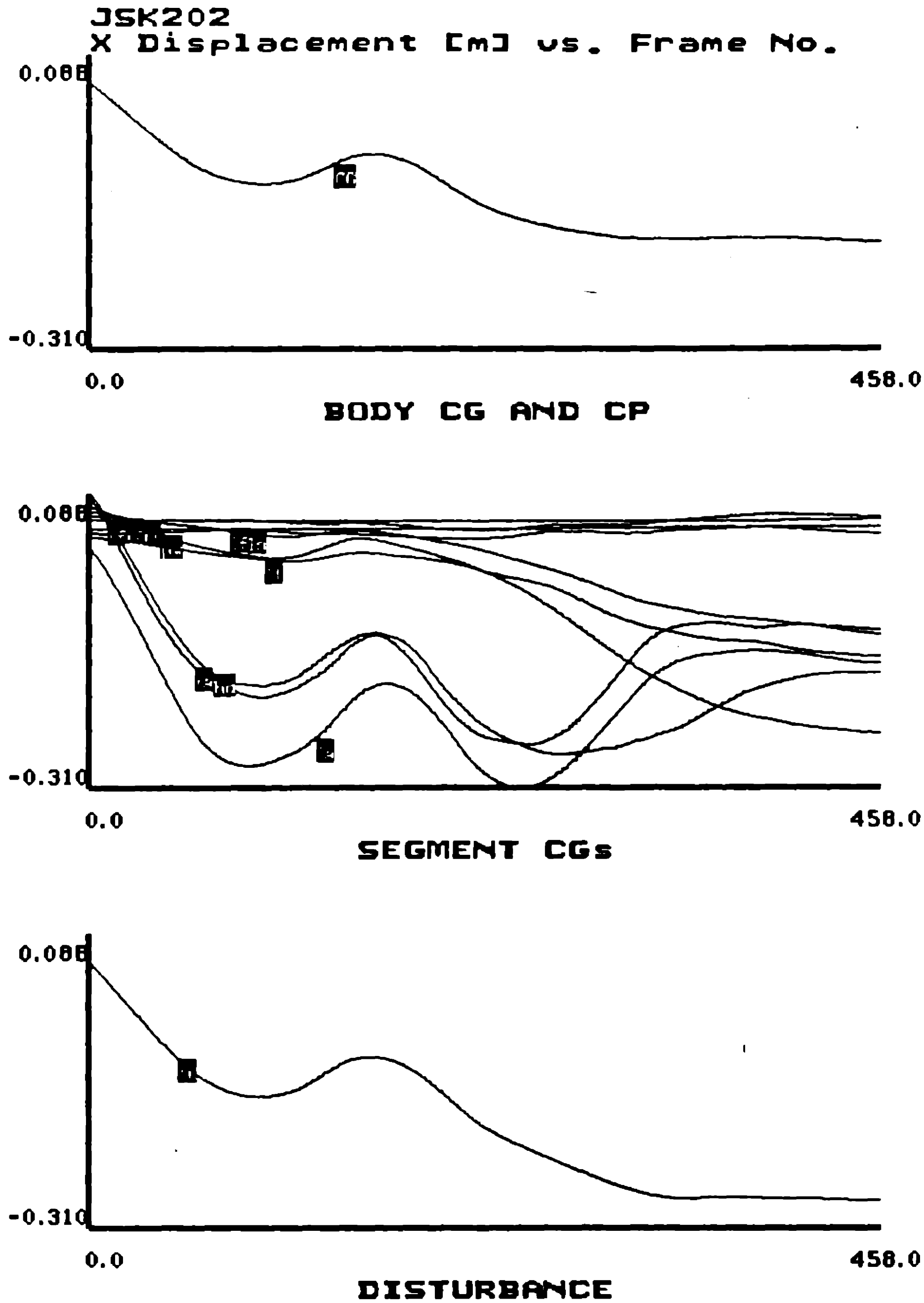

Figure 5.32 Cerebral Palsy, chair rise X displacement

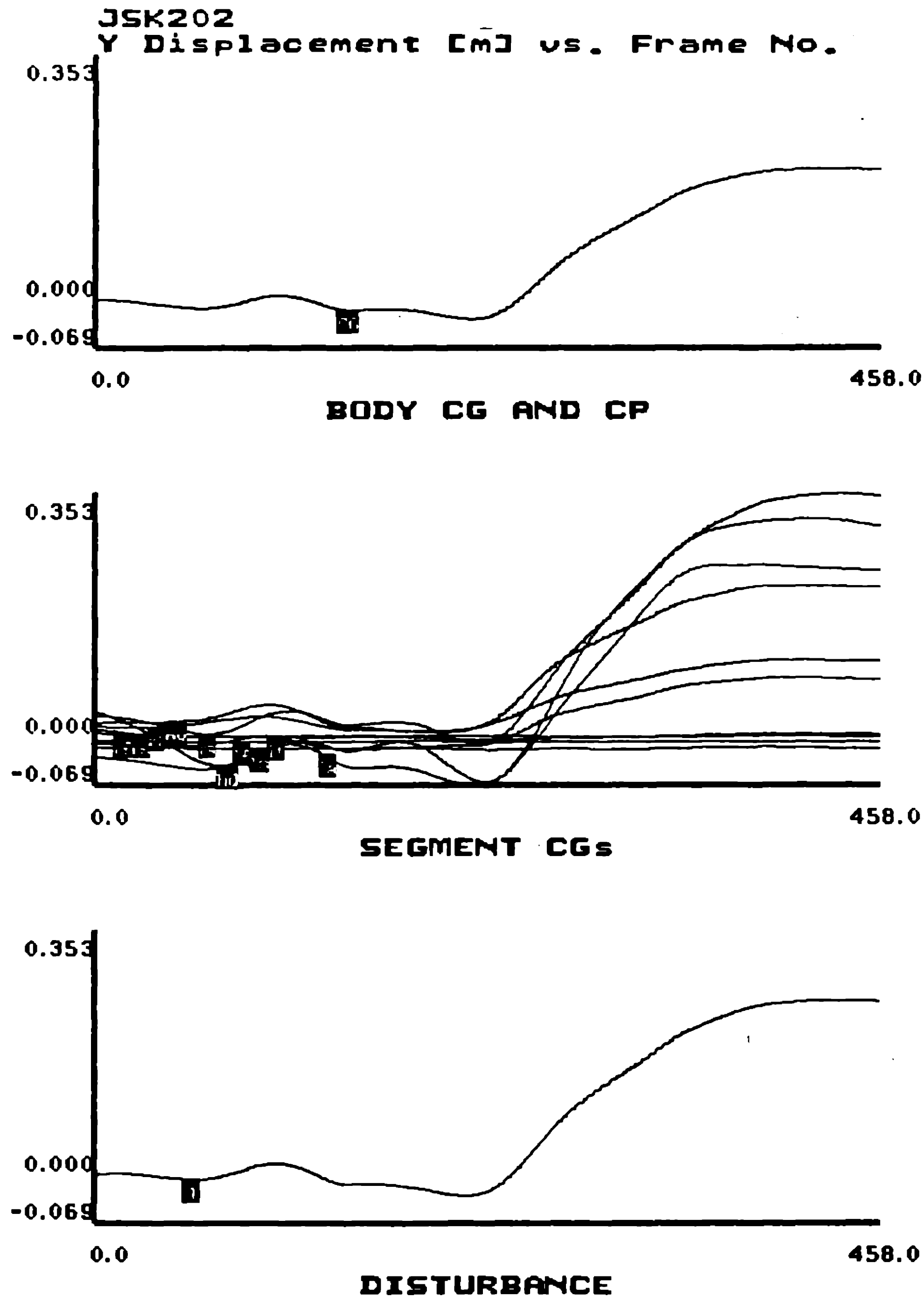

Figure 5.33 Cerebral Palsy subject, chair rise, Y displacement

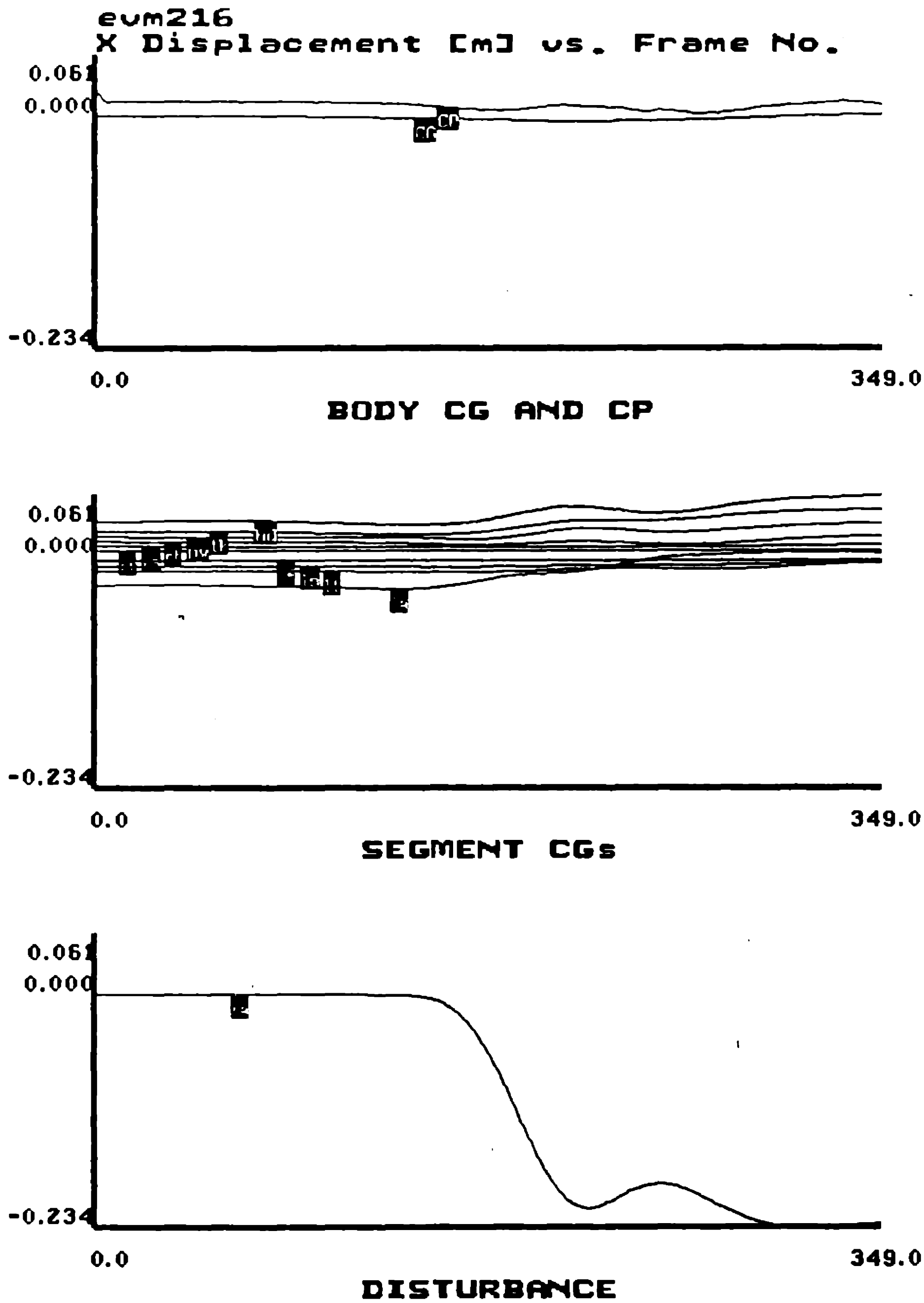

Figure 5.34 Control subject, right arm raise, X displacement

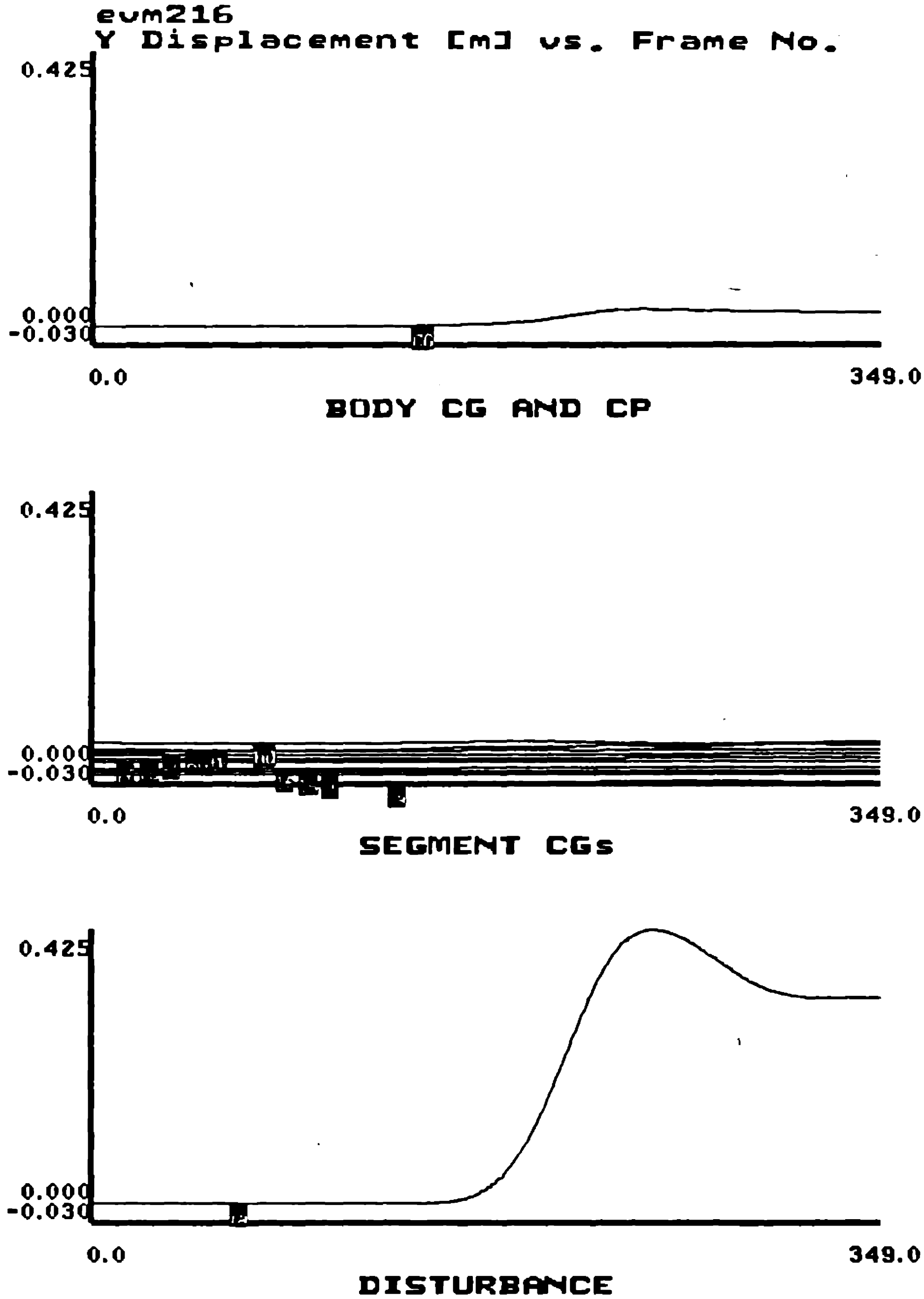

Figure 5.35 Control subject, right arm raise, Y displacement

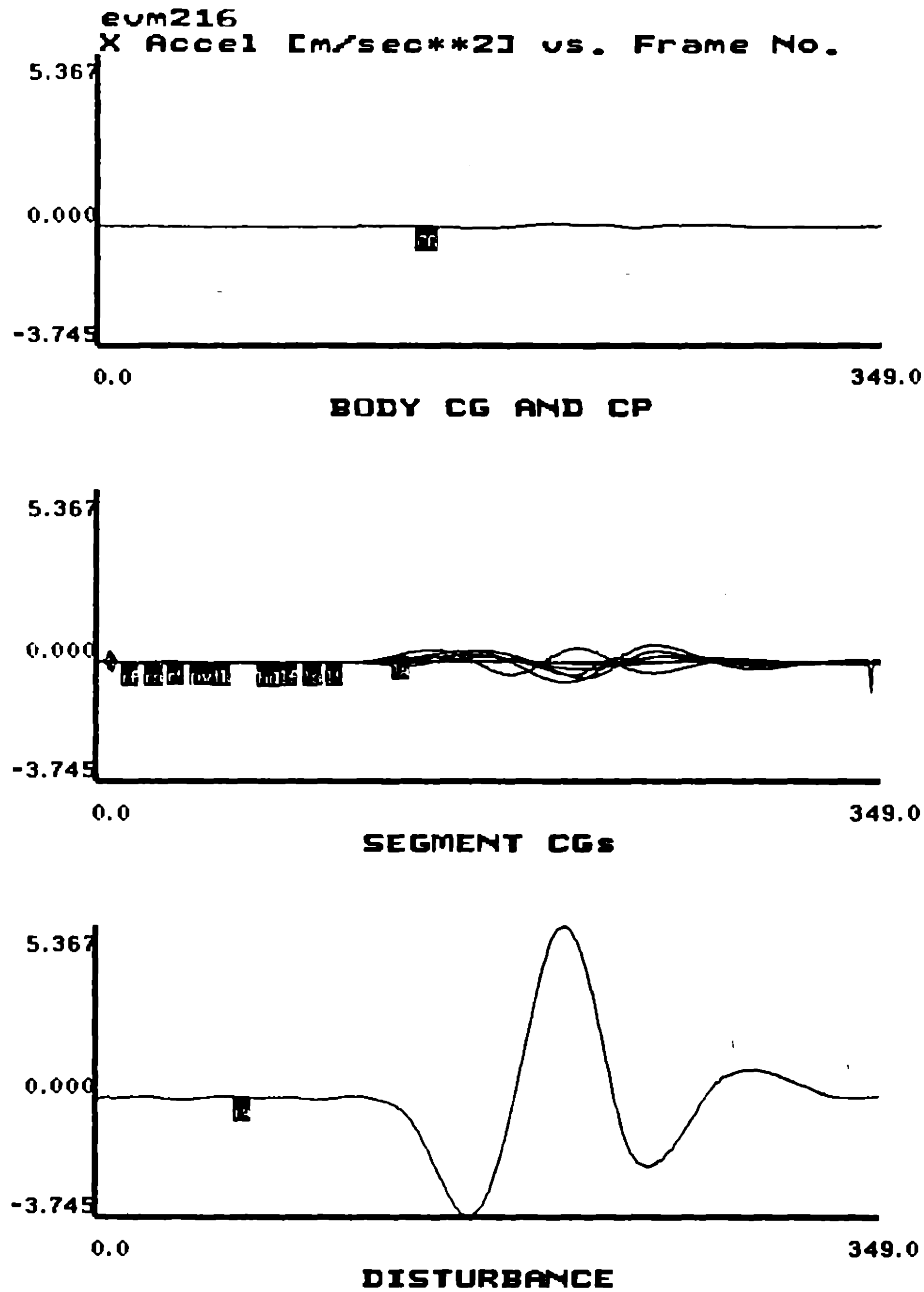

Figure 5.35 Control subject, right arm raise, X acceleration

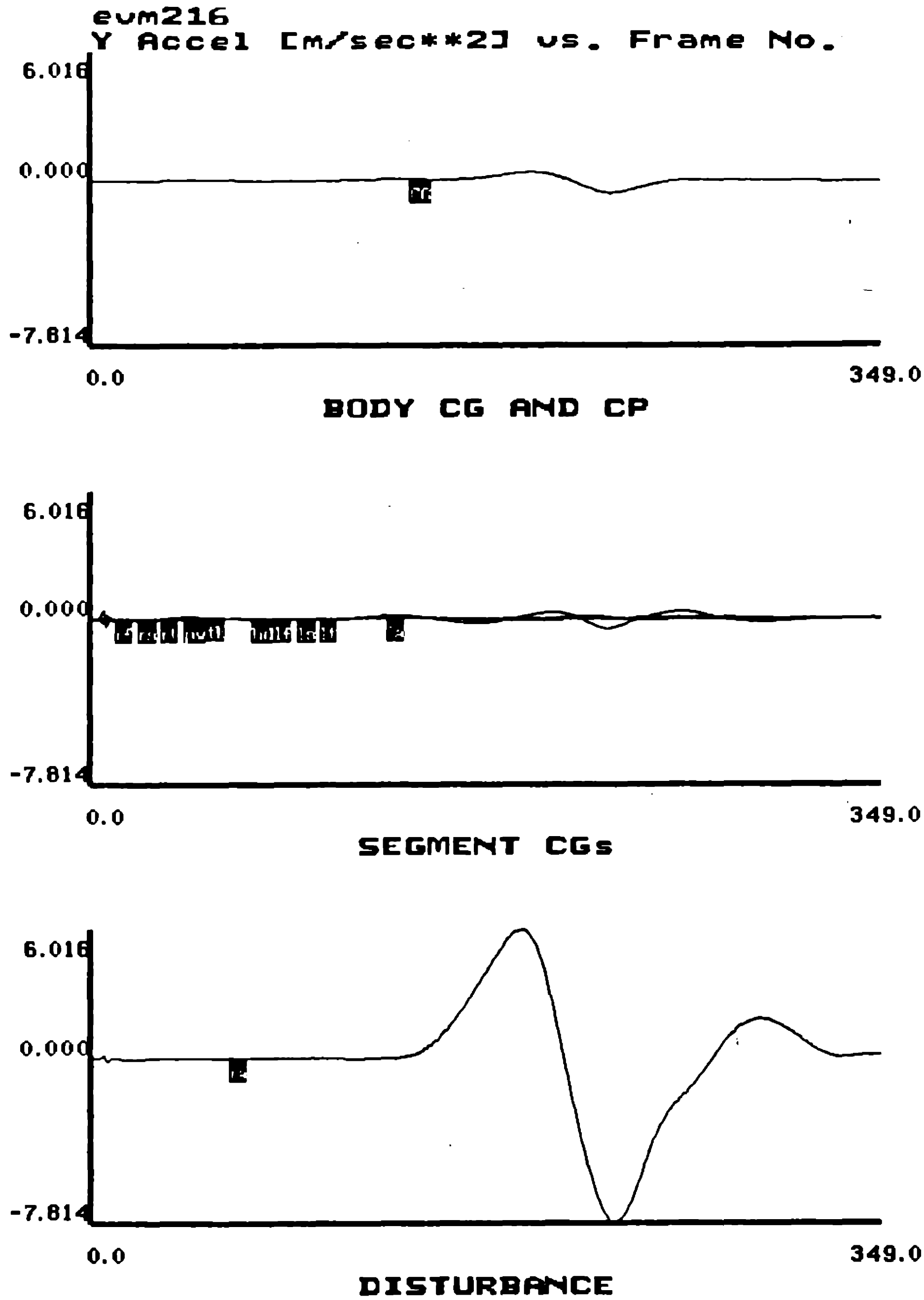

Figure 5.37 Control subject, right arm raise, Z acceleration

the response is not well coordinated. This is consistent with the less effective control of the body CG.

It is evident from the examples presented that the CG model provides a great deal of information on the biomechanics of postural adjustments. The detailed interpretation of this information is difficult since the processes themselves are quite complex. Fortunately this task is beyond the scope of this thesis. In general it will be possible to evaluate wholebody coordination using the technique developed herein. These methods should, therefore, prove extremely useful in a wide variety of research and clinical applications.

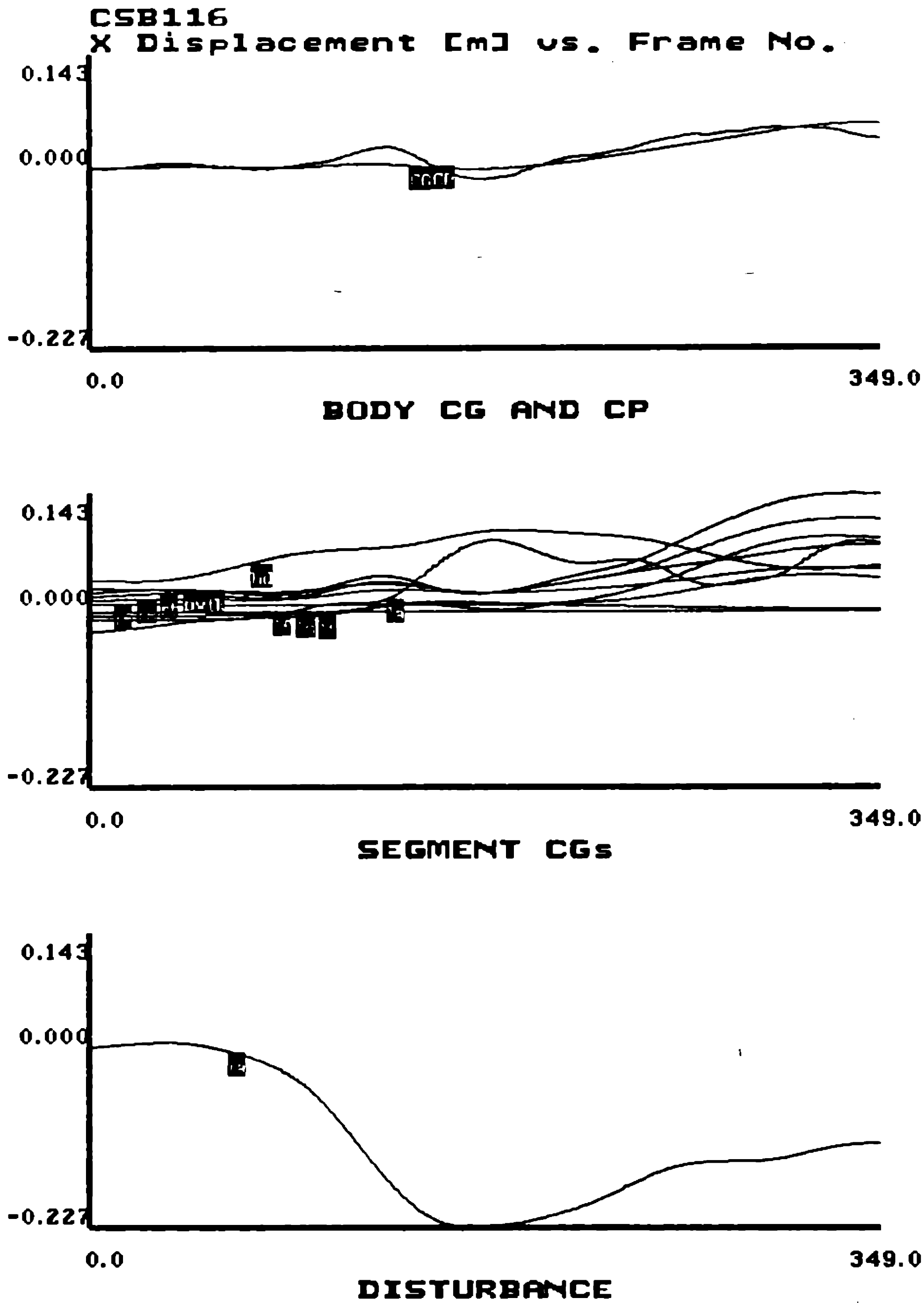

Figure 5.38 Cerebral Palsy subject, right arm raise, X displacement

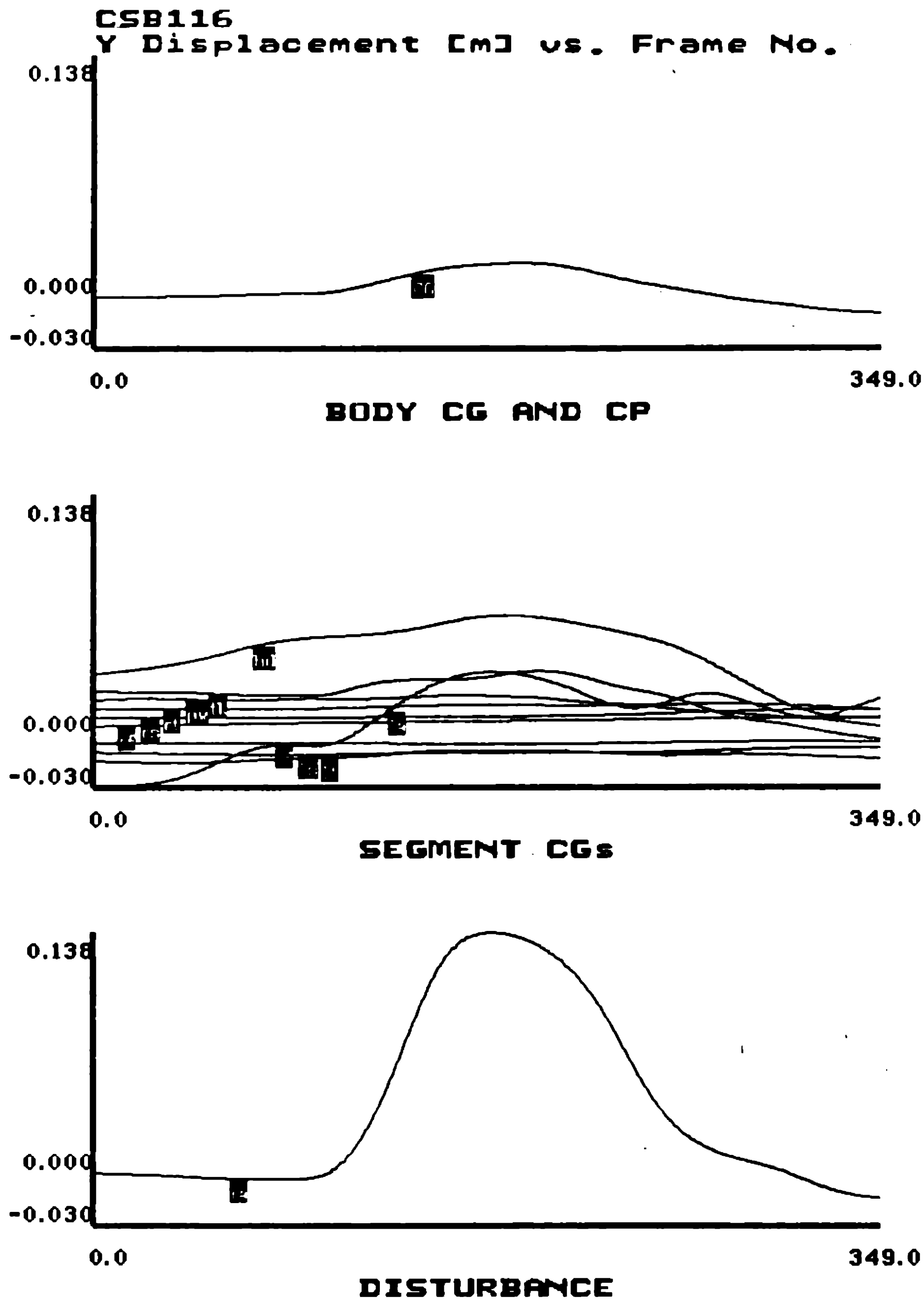

Figure 5.39 Cerebral Palsy subject, right arm raise, Y displacement

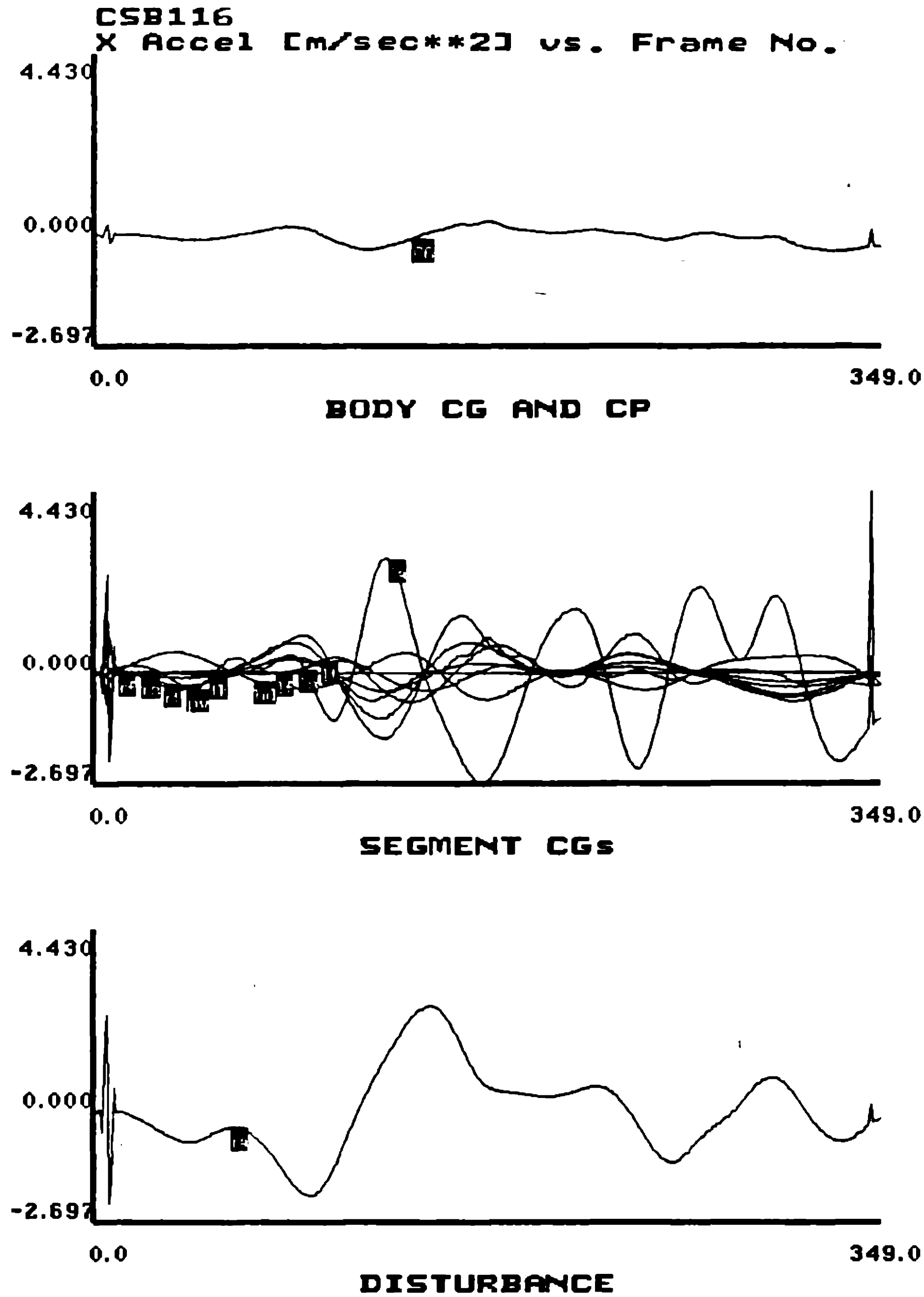

Figure 5.40 Cerebral Palsy subject, right arm raise, X acceleration

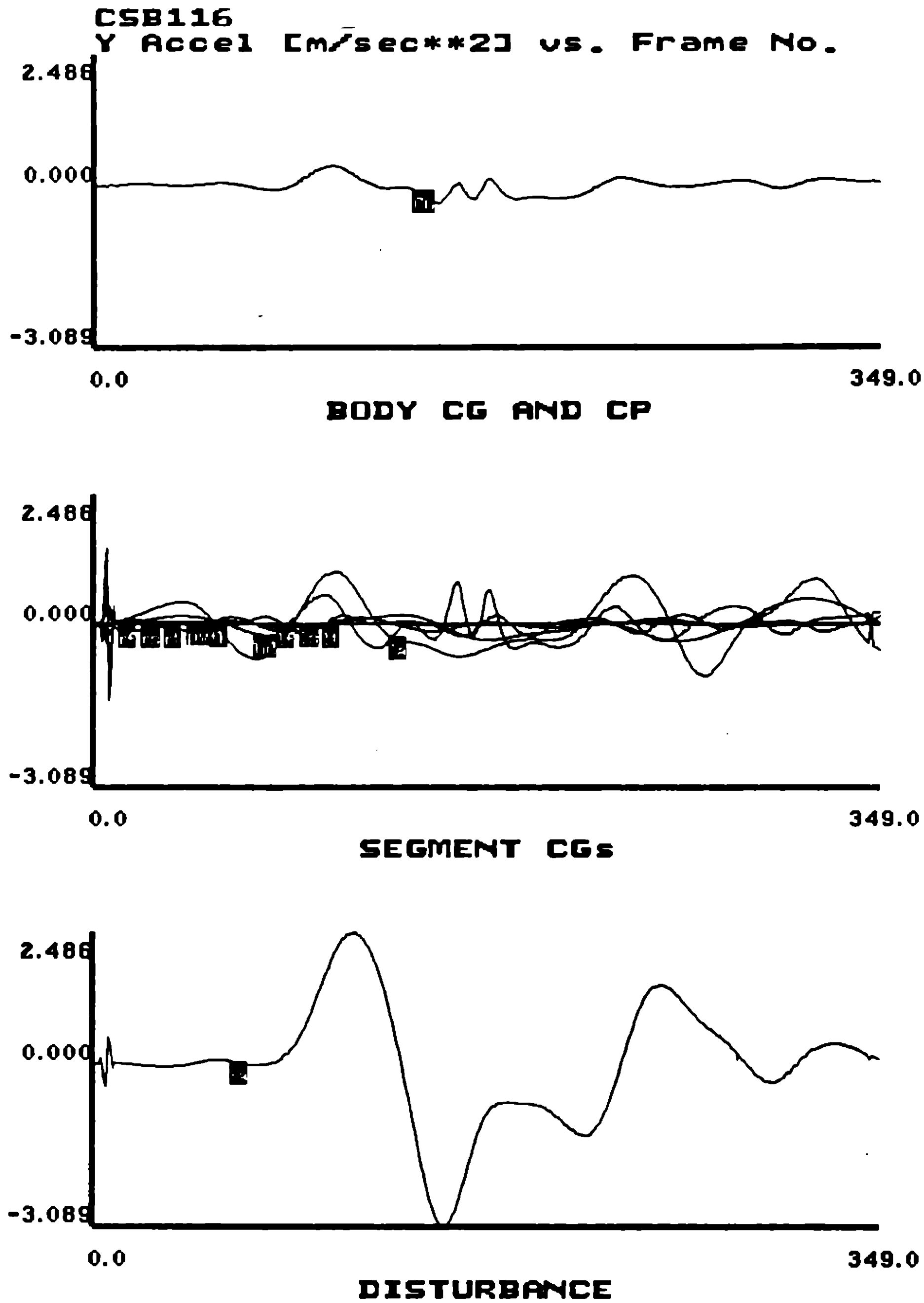

Figure 5.41 Cerebral Palsy subject, right arm raise, Z acceleration

CHAPTER VI

IMPLICATIONS

The goals of this research effort were to develop a model which quantifies the biomechanical relationship between posture and balance and to evaluate the model's potential usefulness in the clinical study of disorders of posture and balance. These goals have been met. Further, by paying careful attention to assessing the accuracy of each step in the development of the model, we know the overall accuracy of the model, the limitations on its application and the direction further improvement and development must take.

VI.1 DISCUSSION AND CONCLUSIONS

The tests of "static" posture and balance in normal and cerebral palsied children clearly show that the model is able to evaluate posture quantitatively and objectively. In modelling posture static and dynamic 3-D displays were developed which clearly capture the essential features of posture. This complete 3-D information may be expected to be useful to clinicians in evaluating their patients, in documenting the changes in posture brought about by progression of the disease and/or clinical intervention, and in understanding the biomechanics of postural deformity.

The kinematic indices and balance index, even in the simple forms used in these pilot studies, permitted the severity of postural deformity and undisturbed balance deficit to be measured in an objective quantitative manner. The joint angle index (ANG) provided the most definitive and easily interpretable measure of postural deformity, at least of the type common to the cerebral palsied population. The standard deviation of the center of pressure location (SDCP) is an adequate measure of the undisturbed balance control. The graphic representation of the time history of the CP position is useful for qualitatively evaluating static balance and disturbed balance.

The center of gravity model permits the interaction of posture control and balance to be studied and assessed. The wealth of information yielded by this model is indicated by the four examples provided in the results chapter. This tool permits wholebody coordination to be assessed in tasks as varied as "simple static" stance to one of the most complex and coordinated motor activity of all gait. If we conduct our experiments with sufficient care and if we ask the right questions, the CG model can be used to obtain very important results in the fields of biomechanics, movement control, and, of course, clinical assessment.

VI.2 FUTURE DEVELOPMENT

VI.2.1 Technical

We must do three things to improve the model technically: 1)improve the joint center determination, 2)improve the joint center determination, and 3) improve the joint center determination. First, the accuracy of the joint center determination and hence the array to body segment transformations is not comparable to the basic level of accuracy of the basic kinematic data acquisition system. The errors associated with the anatomical landmark method of joint center determination caused sufficient scatter in the ANG index to reduce it sensitivity and rendered the MOM index essentially useless. Use of the axes-of-rotation method for defining joint centers promises to reduce this source of error by a factor of about 2 making the ANG index even more useful and perhaps making it worthwhile to evaluate joint moments. However, if we look at the CG model, even when the joint centers are determined by the axes-of-rotation method, the array to body segment transformation is as important a source of error as the variability of the body segment parameter estimates. This is particularly disheartening when one considers the truly dismal quality of the body segment parameter data.

Improving the joint center estimations has not been, is not and will not be a simple task. A variety of mathematically techniques are available which work more or less well depending on the quality of the data and the nature of the kinematics. I would like to call attention to the fact that the 1° accuracy of the kinematic rotation data that we have accepted may be a critical factor limiting our ability to improve joint center determination. The model shown in Figure 5.5 is admittedly extremely simplified, but essentially correct. It shows that a 1cm error in the joint center can result in an error of about 1° in the segment angle. Interpreted another way, a 1° error in the rotation data may be expected to cause an error in the joint center location of on the order of 1cm.

The 1° accuracy of rotations is not a fundamental limit of the system. The fundamental limit of the system is the 3-D position accuracy of the individual LED positions. This imposes a limit on the accuracy of rotations only as a function of array geometry. That is, all other things being equal, a given LED position error produces a rotation error which is inversely proportional to the size of the array [2]. Producing larger rigid arrays is a solution but not a particularly attractive one. Large rigid arrays interfere with normal movement, obstruct each other's visibility and would be either very fragile or excessively heavy.

Perhaps non-rigid arrays might be attempted. The Biomotion Laboratory currently uses tight conforming stockings to mount arrays to the lower limb segments and pelvis. These could be modified to permit individual LEDS or LED clusters to be fixed on the stocking. Five of six LEDs distributed over the whole shank would constitute the shank array, similarly LEDs distributed over the lateral surface of the thigh would constitue the thigh array. Static data would define the array geometry (segment file in the TRACK system). It could be argued that muscle movement would affect the array geometry and degrade the data, but in this set up muscle movement would shift each LED individually and could perhaps be reduced as an effect by the inter-LED-length error check. When using rigid arrays muscle movements shift the whole array producing false movements and rotations which are difficult to detect and eliminate from the data. Efforts to develop array mountings which are immune to muscle artifacts have been only marginally successful.

Secondly, the use of an axis-of-rotation to define joint kinematics implies the assumption that the joint is a hinge joint. The knee and the elbow (which is not included in the model) are the only joints for which this assumption is approximately true. Even these are at best a sliding hinge joint and probably have even more degrees of freedom producing movements of sufficient magnitude to be detected by our system. The other joints certainly are capable of significant non-planar movements. It will be necessary to implement joint center determination methods which do not depend on the planar motion assumption before the level of error associated with this simplification of the model can be assessed.

Thirdly, the location of joint centers has been treated as an initial condition problem. This is justifiable for studies of "static" posture where joint rotations of a few degrees are typical. Under these conditions the changes in the joint center locations may be expected to be smaller than the error in the joint center location algorithm and, therefore, negligible. However, in moving into the realm of posture control or disturbance response studies, we are beginning to deal with larger ranges of motion. Here, the assumption of fixed location joint centers is not biomechanically reasonable [17, 37]. It must be looked upon as a necessary intermediate step while the necessary algorithms are developed to define instantaneous or quasi-instantaneous joint centers.

If we are to tackle the analysis of the full range of human motion this task is extremely difficult. We must find algorithms that accurately define the joint centers for suitable data. Several different algorithms may be needed because the joints are capable of operating in different modes, sometimes exhibiting planar rotation, sometimes very non-planar rotation. We must define algorithms which evaluate the data and determine what portions of the data contain kinematic information suitable for joint center determination and, perhaps, select the most appropriate technique to be applied to the data. Finally, we must define methods for dealing with those portions of the data where the kinematics does not permit joint center determination. For example, no method will ever accurately determine the knee joint center based on kinematic data during the stance phase of gait.

A great deal of work will be required to achieve the three improvements in the joint center determination that I have outlined. Certainly the mathematic of kinematics is up to the task and the computational power is available on computers of the type available in motion analysis laboratory (PC-based systems being excluded of course). Development effort seems to be all that is required. The effort will be significant, but necessary if we are to study the kinematics of the human body rather than the kinematics of LED arrays.

VI.2.2 Clinical

While cerebral palsy is not the only disease associated with postural abnormality, the clinical handling of this disorder is probably typical. The record of the treatment of this disorder is characterized by a time history of movement back and forth between various treatment modalities with little overall progress [27, 53]. The same basic surgical, medical, and physical therapy interventions have been in use for over a generation. One comes into vogue, is used for a time, falls into disfavor as the qualitative record of marginal results builds and is replaced by another pre -existing technique whose record of equally disappointing results has now been forgotten. This pattern speaks to the serious need for objective quantitative assessment of the basic disorder and of the interventions used to treat it.

A three step approach is needed in using the posture model for clinical studies. The first step is clinical studies to prove that the method can detect and quantify the postural and balance abnormalities associated with a particular pathology. These first studies will validate the model and, equally important, document the baseline posture and balance problems associated with the disease in question. There are a number of disorders in additions to cerebral palsy where this type of study would be appropriate.

In this regard, the study of normal and cerebral palsied children described in this work was a pilot study and not an adequate baseline clinical study of this disease. Also, the methods used to study "static" posture and balance are sufficiently developed to be ready to move into this clinical study phase. More experience with the body CG model is needed before we will know how to use it clinically. Finally, the term "clinical study" here refers to controlled scientific studies under specific protocols with specific clinical objectives; not to diagnostic testing. While there is a need for such diagnostic testing in posture and balance disorders, the model developed here is not directly suitable for that purpose. Research using the model will suggest appropriate parameters to be measured by simpler techniques for diagnostic testing.

The second step is to use the model to quantitatively and objectively assess the effectiveness of therapeutic interventions. This is a more difficult task for the method. In the first case it is required to measure the difference between normal posture and balance and the posture and balance resulting from pathology; in this second case the method is required to detect incremental changes in posture and balance. On the other hand, since the clinical interventions generally have specific goals or objectives, the measurements can be targeted to assess whether these have been achieved. However, the overall effect must also be determined.

The third step is to use what the model reveals about the biomechanics of posture, posture control and balance to suggest new or improved clinical techniques. This is the most difficult, but most promising step. In conversation with a clinician I was told that this step is not necessary; that "We already know what to do, we only need to document the effectiveness of our methods." Again I refer to the history of cerebral palsy treatment to prove that either this statement is false or clinicians are very foolish. I believe the first hypothesis holds. Clinicians are in general an extremely pragmatic lot; if one of the existing treatments had worked it would have come into general use and acceptance for the appropriate patient group whether or not we could put a number on how well it worked. Conversely, the fact that no treatment seems to have achieved wide long term general acceptance indicates that none is wholly satisfactory. This is in part due to the extreme difficulty of the problem being tackled; the consequences of major damage to the human brain are not going to be easily erased. One can hope that with quantative objective knowledge of the biomechanical and motor control issues involved, a rational for improved treatment will be forthcoming. This research was intended to be and, I believe, is a significant step in that direction.

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Appendix A

Laboratory Setup

The MGH Biomotion Laboratory is in a 30 x 40 foot carpeted, optically neutral room. The kinematic data acquisition system employs Selspot II hardware and TRACK software. The TRACK system was developed and evaluated at the MIT Newman Laboratory for Biomechanics and Human Rehabilitation. The system has been implemented in an expanded and enhanced form at the MGH. The MGH system is bilateral, capable of observing human movement from both sides, simultaneously. Two sets of two [a total of four] Selspot II infrared sensing cameras are arranged on two Klinger optical benches on either side of the study area. Each camera is mounted using a precision Klinger translator platform and rotator. This permits the distance between cameras to be controlled to 0.0001 meters and vertical axis rotation to be controlled to within 1/100°. Benchmark blocks are used to identify standard locations for the cameras on the optical bench. In the configuration installed at MGH, the Selspot II system is capable of tracking the position of 64 infrared LEDs at a sampling frequency of 153 hertz. The LEDs are generally configured in arrays of three or more for TRACK processing.

Two in-floor mounted piezoelectric force platforms (Kistler) are located in the approximate center of the test area. The force platforms are mounted on a granit subfoundation within the concrete floor. The system for anchoring the force platforms to this foundation provides an extremely rigid mounting while permitting the relative position of the platforms to be adjusted to the requirements of the experiment. The force platforms measure six components of the ground reaction force to an accuracy of 1% of full scale. The output consist of the 3-D ground reaction force vector, the center of pressure in 2-D of the force platform surface, and the free moment about the vertical.

Two computers are employed. A PDP 11/60 and a MicroVAX II Workstation are connected by an Eithernet local area net using DecNet software. In the current operating mode the PDP11/60 is used for data aquisition and storage and 3-D data display. The MicroVAX is used for data processing and 2-D data display. The 11/60 is equipped with an interactive 3-D Megatek vector graphics display which is used to display processed data in 3-D. 2-D plots of the data maybe developed using a graphics terminal on the 11/60 or the MicroVAX II Workstation. Hardcopy is available from a HP

7475 pen plotter or a DEC LN03PLUS laser printer. Data may be backed up using a ½ inch tape drive on the 11/60 or a CompacTape on the MicroVAX.

The TRACK software first calculates the 3-D position of each LED; then aggregates the LEDs in to arrays for calculation the array orientation and position. At each stage of processing the data is automatically tested to evaluate it quality. "Bad" data is rejected based on appropriate criterion. The position and orientation of each array is established to within 1 mm. and 1 degree of orientation. Velocities and accelerations are also estimated using either a fifth order Lagrangian filter or fifth order b-spline smoothing and differentiating algorithm. In the version of TRACK currently implemented at MGH, the data is smoothed and interpolated using b-spline smoothing of the three dimensions of the individual LED positions and of the array positions and orientation parameters.

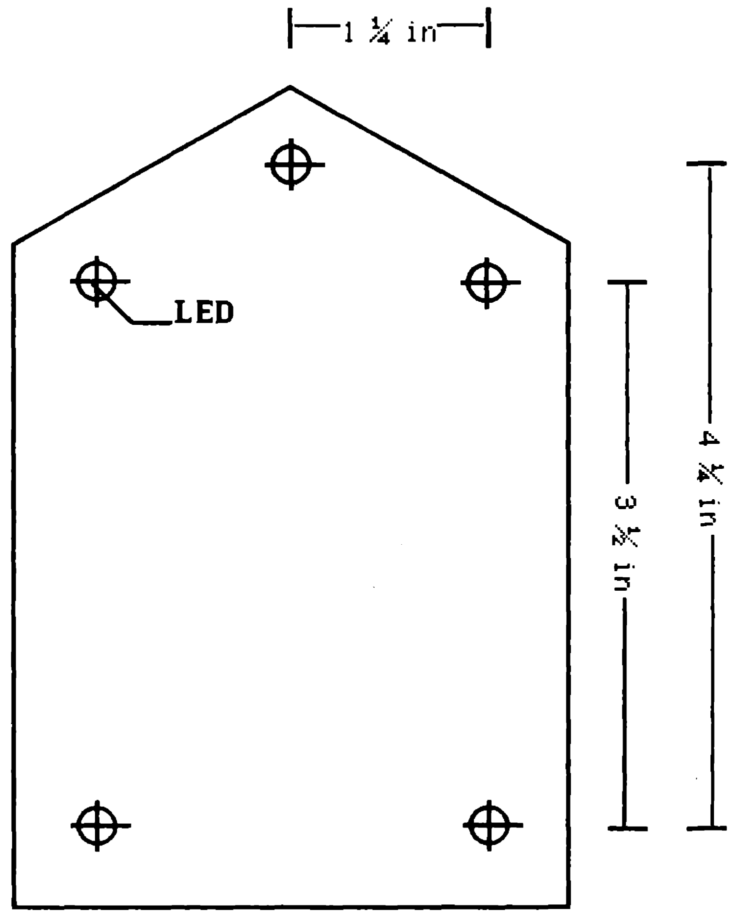
For estimation of the joint forces and torques software dubbed NEWTON is used. This software was also developed at MIT. NEWTON combines the measured ground reaction forces and array kinematics, with estimates of the body segment parameters and array to body segment coordinate transformations, to calculate the 3-D forces and torques at the lower limb joints, ankle, knee and hip.

Appendix B

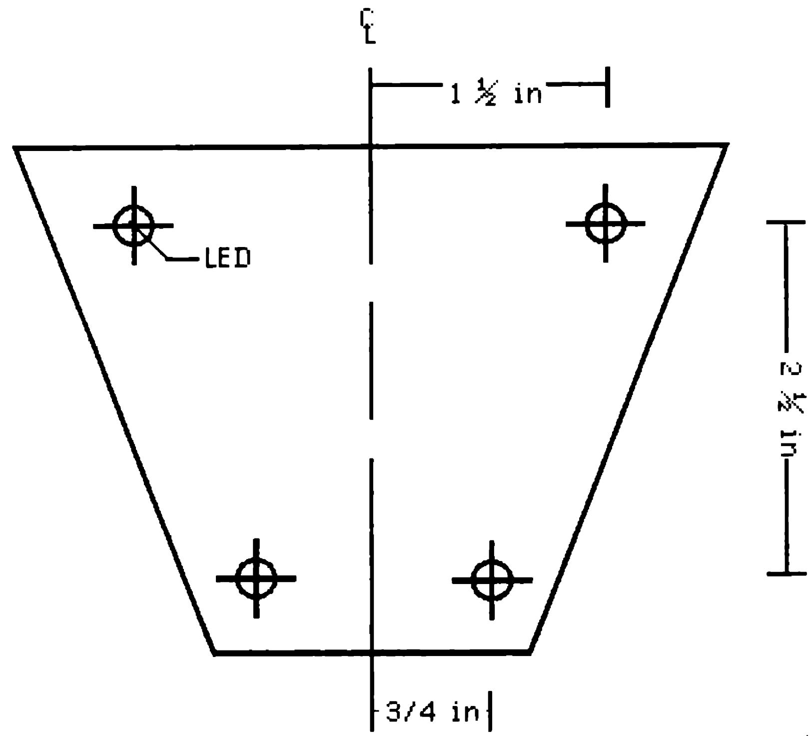
Arrays

The arrays of infrared light emitting diodes [LEDs] used in the kinematic studies of posture are illustrated in the following diagrams. All dimensions are in inches and indicate LED positions. The arrays are made of 1/8 inch thick plexiglass except for the pelvis and trunk arrays which are made of 1/16 inch thick plexiglass. The overall array dimensions are such that each led is approximate 0.5 inches, or 1.0 cm from the closest edge. All arrays are planar. The LEDs are mounted so that the lens are just flush with the planar camera facing surface. All camera facing surfaces are uniformly coated with a black low reflective paint. Mounting hardware which must penetrate to the camera facing surfaces is recessed; e.g. mounting screw (position not indicated) are counter-sunk so that the head is slightly below the plane of the array. The remaining mounting hardware and all wiring is on the hidden, non-camera-facing surface of the array. Wiring is light gage, AWG 22, ribbon cable to minimize weight and maximize flexibility.

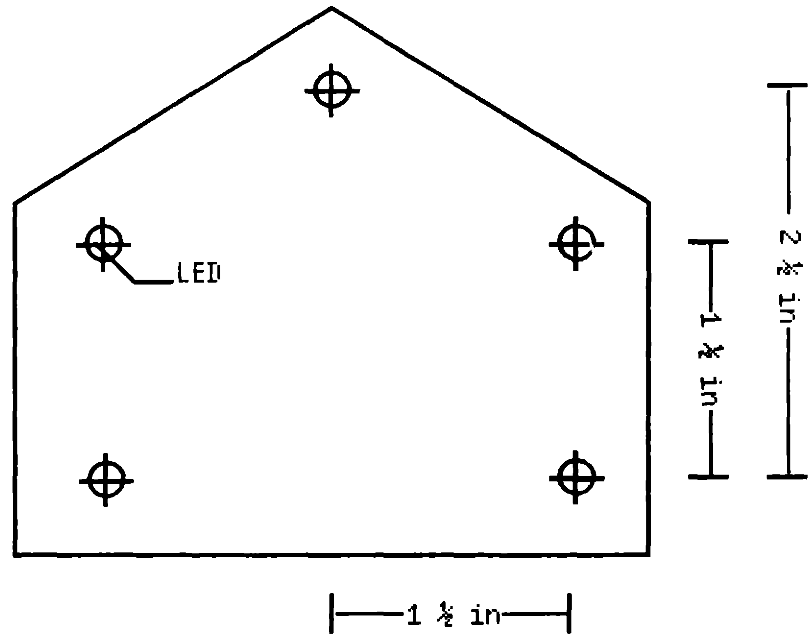
Multiple LED clusters were not used in these arrays. We found that given the combination of cameras, lenses and led control units used in the lab, it is possible to obtain satisfactory intensity levels using single LEDs. Since the effort required to manufacture arrays and the LED consumption is significantly greater when clusters are used, we elected to not add this feature to our already very complex array configuration.



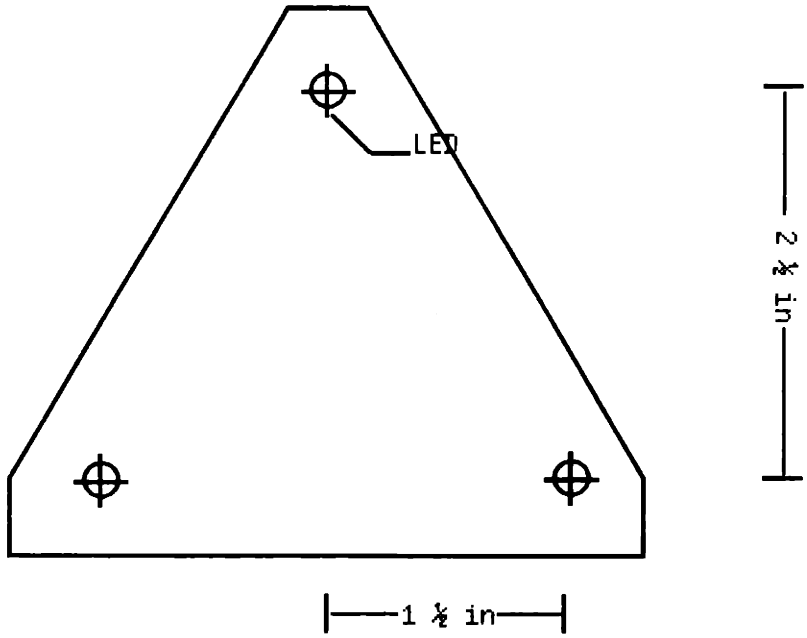
Shank and Thigh Array



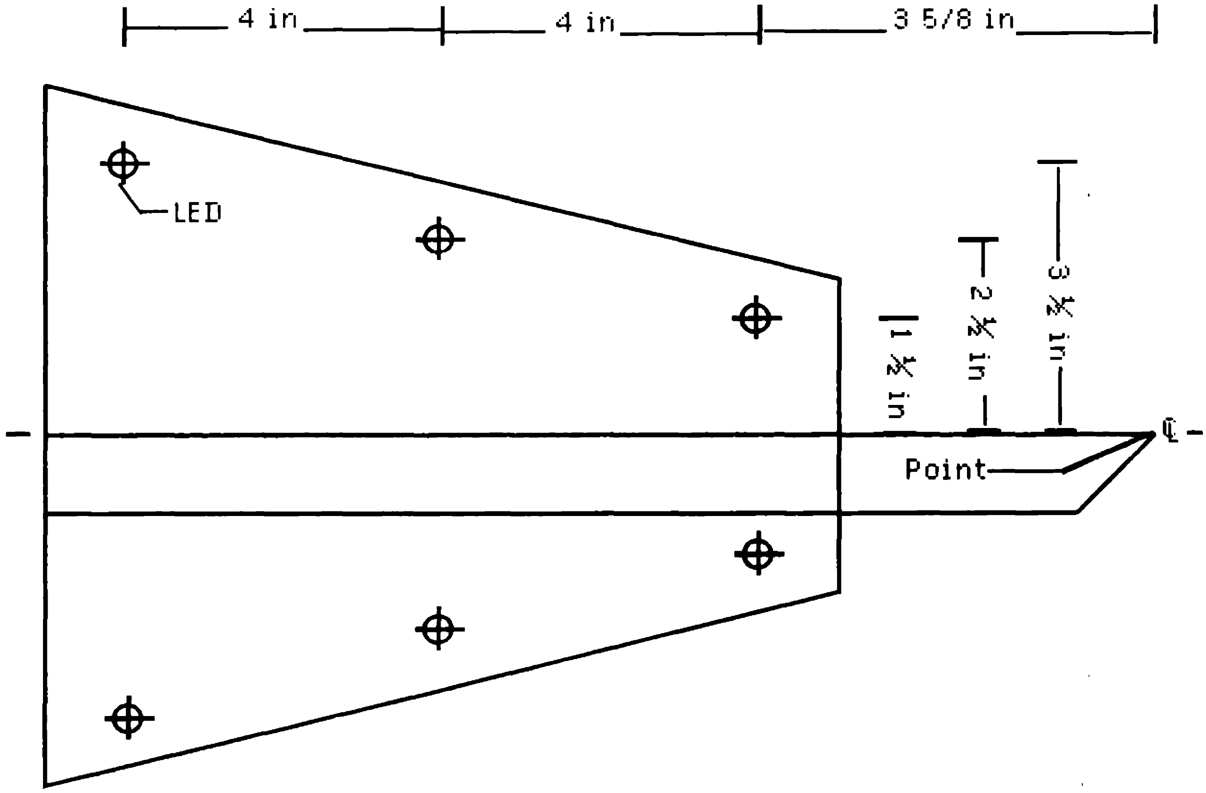
Head Array



Foot, Pelvis and Trunk Array



Arm Array



Anatomical landmark pointer array

Appendix C

Joint Centers

This Appendix contains program FJOINT.FOR which performs joint center determination and initial condition specification for the whole body posture data. This program uses an axis-of-rotation method to define the joint centers for the hip, knee and ankle. The axis-of- rotation subroutine T5SFLX.FOR is also included. Program FJOINT.FOR represents a very much expanded and modified version of program LJOINT.FTN originally developed for the Biomotion Laboratory by Fijan. The subroutine T5SFLX.FOR is also based on a program by Fijan, FLEX5.FTN.

```
C       FJOINT
C               NOV-86            PATRICK RILEY
C#################################################################################
C       THIS PROGRAM TAKES THE DATA FROM STATIC DATA FILES
C       INCLUDING SELECTED ARRAY POSITIONS AND ORIENTATIONS AS
C       WELL AS THE JOINT CENTER JIG POSITION AND ORIENTATION
C       AND DATA FROM A FILE WHERE LOWER LIMB JOINTS ARE
C       CONTAINS DATA RELATING THE ARRAYS TO THE ANATOMIC JOINT CENTERS
C       AND DEFINES THE INITIAL SEGMENT ANGLES.  THIS DATA IS USED BY
C       PROGRAM 3DSEG ALONG WITH A SET OF STATIC DATA TO DEFINE THE
C       SEGMENT FILES NEEDED FOR 3D DISPLAYS AND FOR OTHER PROCESSING
C       WHICH REQUIRES CONVERSION FROM ARRAY COORDINATES TO LIMB SEGMENT
C       COORDINATES.
C
C       BASED ON PROGRAM NJOINT BY BOB FIJAN
C
C       LINK FJOINT,T5SFLX,TRACK.OLB/LIB
C
C#################################################################################
C
        PROGRAM FJOINT

        CHARACTER*16 NLANKL,NLKNEE,NLHIP,NLFLX
        CHARACTER*16 NRANKL,NRKNEE,NRHIP,NRFLX
        CHARACTER*16 NLARM,NRARM,NLTRNK,NRTRNK,NHEAD
        CHARACTER*16 FPLNAM,NMROUT,NMLOUT
        CHARACTER*1 IBYTE,IJFOOT,IJKNEE,IJHIP
        BYTE BELL
        DIMENSION TEMP(3,3),VTEMP(3),RTEMP(3,3)
        COMMON /BLKROT/ ROTATE(3,3,7,15)
        COMMON /BLKANG/ ANGLS(2,16),RRVECT(3,12),RLVECT(3,10)
        COMMON /BLKRLG/ XYZRF(3,7),XYZRA(3,7),XYZRK(3,7),XYZRT(3,7),XYZRH(3,7)
        COMMON /BLKLLG/ XYZLF(3,7),XYZLA(3,7),XYZLK(3,7),XYZLT(3,7),XYZLH(3,7)
        COMMON /BLKTRK/ XYZLTK(3,7),XYZRTK(3,7),XYZLAM(3,7),XYZRAM(3,7)
        COMMON /BLKHED/ XYZHD(3,7),XYZEND(3,2),XYZHND(3,2)
        COMMON /BLKHIP/ VLHIP(3,2),VRHIP(3,2)
        COMMON /NAMFIL/ NLANKL,NLKNEE,NLHIP,NLFLX,
     1  NRANKL,NRKNEE,NRHIP,NRFLX,
     2  NLARM,NRARM,NLTRNK,NRTRNK,NHEAD,
     3  FPLNAM,NMROUT,NMLOUT
        COMMON /FLXOUT/UNIT4(3),POINT(3),ISTEP
        DATA BELL/"7/
        PARAMETER(LUI=5,LUO=6,PI = 3.1415927,BAD=-1000.)
C
        NLANKL ='TK4:NAMFIL.DTL;0'
        NLKNEE ='TK4:NAMFIL.DTL;0'
        NLHIP  ='TK4:NAMFIL.DTL;0'
        NLFLX  ='TK4:NAMFIL.DTL;0'
        NRANKL ='TK4:NAMFIL.DTR;0'
        NRKNEE ='TK4:NAMFIL.DTR;0'
        NRHIP  ='TK4:NAMFIL.DTR;0'
        NRFLX  ='TK4:NAMFIL.DTR;0'
        NLARM  ='TK4:NAMFIL.DTL;0'
        NRARM  ='TK4:NAMFIL.DTR;0'
        NLTRNK ='TK4:NAMFIL.DTL;0'
        NRTRNK ='TK4:NAMFIL.DTR;0'
        NHEAD  ='TK4:NAMFIL.DTR;0'
        NMLOUT ='TK4:NAMFIL.DTL;0'
        NMROUT ='TK4:NAMFIL.DTR;0'
```

```
 2        FORMAT(1A6)

          WRITE(LUO,3)BELL
 3        FORMAT('$Enter the name of the right ANKLE data file ',A1)
          READ(LUI,2) NRANKL(5:10)
          WRITE(LUO,4)
 4        FORMAT('$Enter the name of the right KNEE  data file ')
          READ(LUI,2) NRKNEE(5:10)
          WRITE(LUO,5)
 5        FORMAT('$Enter the name of the right HIP   data file ')
          READ(LUI,2) NRHIP(5:10)

          WRITE(LUO,7)
 7        FORMAT('$Enter the name of the right TRUNK data file ')
          READ(LUI,2) NRTRNK(5:10)
          WRITE(LUO,8)
 8        FORMAT('$Enter the name of the right ARM   data file ')
          READ(LUI,2) NRARM(5:10)
          WRITE(LUO,9)
 9        FORMAT('$Enter the name of the       HEAD  data file ')
          READ(LUI,2) NHEAD(5:10)
C
          WRITE(LUO,9786)
 9786     FORMAT(/,' Default left file name equals corresponding'
     *          ,' right file name.')

          WRITE(LUO,31)BELL
 31       FORMAT('$Enter the name of the left  ANKLE data file ',A1)
          READ(LUI,2) NLANKL(5:10)
          IF(NLANKL(5:5).EQ.' ')THEN
                NLANKL(5:10) = NRANKL(5:10)
          ENDIF

          WRITE(LUO,32)
 32       FORMAT('$Enter the name of the left  KNEE  data file ')
          READ(LUI,2) NLKNEE(5:10)
          IF(NLKNEE(5:5).EQ.' ')THEN
                NLKNEE(5:10) = NRKNEE(5:10)
          ENDIF

          WRITE(LUO,33)
 33       FORMAT('$Enter the name of the left  HIP   data file ')
          READ(LUI,2) NLHIP(5:10)
          IF(NLHIP(5:5).EQ.' ')THEN
                NLHIP(5:10) = NRHIP(5:10)
          ENDIF

          WRITE(LUO,35)
 35       FORMAT('$Enter the name of the left  TRUNK data file ')
          READ(LUI,2) NLTRNK(5:10)
          IF(NLTRNK(5:5).EQ.' ')THEN
                NLTRNK(5:10) = NRTRNK(5:10)
          ENDIF

          WRITE(LUO,36)
 36       FORMAT('$Enter the name of the left  ARM   data file ')
          READ(LUI,2) NLARM(5:10)
          IF(NLARM(5:5).EQ.' ')THEN
                NLARM(5:10) = NRARM(5:10)
          ENDIF
```

```
C  Obtain measurements

        WRITE(LUO,112)BELL
 112    FORMAT(//,'$Height of the rt foot during static ',
     *  'data[D:0.0]? ',A1)
        READ(LUI,89) RFLRHT
        IF (RFLRHT .LE. 0.0) RFLRHT=0.0
        WRITE(LUO,113)
 113    FORMAT('$Height of the left foot during ',
     *  'static data [D:0.0]? ')
        READ(LUI,89) SFLRHT
        IF (SFLRHT .LE. 0.0) SFLRHT=0.0

        WRITE(LUO,194)BELL
 194    FORMAT(1X,A1,//)

        WRITE(LUO,114)
 114    FORMAT('$Length of the Right foot (inches)?          ')
        READ(LUI,89) RFOOT
        WRITE(LUO,115)
 115    FORMAT('$Circumference of the Right ankle (inches)?')
        READ(LUI,89) RANKLE
C       RANKLE = RANKLE/39.37    !IN TO METERS
        WRITE(LUO,116)
 116    FORMAT('$Circumference of the Right knee (inches)? ')
        READ(LUI,89) RRKNEE
C       RRKNEE = RRKNEE/39.37    !IN TO METERS
        WRITE(LUO,117)
 117    FORMAT('$Circumference of the Right thigh (inches)?')
        READ(LUI,89) RTHIGH
C       RTHIGH = RTHIGH/39.37    !IN TO METERS
        WRITE(LUO,118)
 118    FORMAT('$Circumference of the Waste (inches)?       ')
        READ(LUI,89) CWASTE
C       CWASTE = CWASTE/39.37    !IN TO METERS
        WRITE(LUO,198)
 198    FORMAT('$Circumference of the Chest (inches)?       ')
        READ(LUI,89) CTRUNK
C       CTRUNK = CTRUNK/39.37    !IN TO METERS
        WRITE(LUO,119)
 119    FORMAT('$Circumference of the Right arm (inches)?  ')
        READ(LUI,89) CRARM
C       CRARM = CRARM/39.37      !IN TO METERS
        WRITE(LUO,120)
 120    FORMAT('$Length of the Right arm (inches)?          ')
        READ(LUI,89) RRARM
        WRITE(LUO,121)
 121    FORMAT('$Circumference of the NECK (inches)?        ')
        READ(LUI,89) CHEAD
        CHEAD = CHEAD/39.37      !IN TO METERS
        WRITE(LUO,1711)
1711    FORMAT('$Circumference of the HEAD (inches)?        ')
        READ(LUI,89) CTHEAD
C       CTHEAD = CTHEAD/39.37    !IN TO METERS

C LEFT SIDE
        WRITE(LUO,194)BELL
        WRITE(LUO,122)
 122    FORMAT('$Length of the Left foot (inches) [D:L=R]?          ')
```

```
      READ(LUI,89) SFOOT
      IF(SFOOT.LE.0.0)SFOOT = RFOOT
      WRITE(LUO,123)
123   FORMAT('$Circumference of the Left ankle (inches) [D:L=R]?')
      READ(LUI,89) SANKLE
      IF(SANKLE.LE.0.0)SANKLE = RANKLE
      WRITE(LUO,124)
124   FORMAT('$Circumference of the Left knee (inches) [D:L=R]? ')
      READ(LUI,89) SRKNEE
      IF(SRKNEE.LE.0.0)SRKNEE = RRKNEE
      WRITE(LUO,125)
125   FORMAT('$Circumference of the Left thigh (inches) [D:L=R]?')
      READ(LUI,89) STHIGH
      IF(STHIGH.LE.0.0)STHIGH =RTHIGH
      WRITE(LUO,126)
126   FORMAT('$Circumference of the Left arm (inches) [D:L=R]?  ')
      READ(LUI,89) CLARM
      IF(CLARM.LE.0.0)CLARM = CRARM
      WRITE(LUO,127)
127   FORMAT('$Length of the Left arm (inches) [D:L=R]?         ')
      READ(LUI,89) RLARM
      IF(RLARM.LE.0.0)RLARM=RRARM

89    FORMAT(F15.6)

C  CALL SUBROUTINE TO AVERAGE THE ROTATION MATRIX

      WRITE(LUO,194)BELL
      TYPE*,' Averaging the static data files.'

      CALL AVERAG(NLANKL,ROTATE(1,1,1,1),XYZLA)
      CALL AVERAG(NLKNEE,ROTATE(1,1,1,2),XYZLK)
      CALL AVERAG(NLHIP ,ROTATE(1,1,1,3),XYZLH)
      CALL AVERAG(NRANKL,ROTATE(1,1,1,4),XYZRA)
      CALL AVERAG(NRKNEE,ROTATE(1,1,1,5),XYZRK)
      CALL AVERAG(NRHIP,ROTATE(1,1,1,6),XYZRH)
      CALL AVERAG(NLTRNK,ROTATE(1,1,1,7),XYZLTK)
      CALL AVERAG(NRTRNK,ROTATE(1,1,1,8),XYZRTK)
      CALL AVERAG(NLARM ,ROTATE(1,1,1,9),XYZLAM)
      CALL AVERAG(NRARM ,ROTATE(1,1,1,10),XYZRAM)
      CALL AVERAG(NHEAD ,ROTATE(1,1,1,11),XYZHD)

C  CALCULATE SEGMENT ORIENTATION ANGLES

      WRITE(LUO,194)BELL
      TYPE*,' Finding segment orientation angles.'

1330  WRITE(LUO,1687)"7
1687  FORMAT(/,'$Were the feet flat on the floor? [Y/N] ',A1)
      READ(LUI,560,ERR=1330)IBYTE
      IF(IBYTE.EQ.'Y'.OR.IBYTE.EQ.'y')IBYTE = 'Y'
      IF(IBYTE.NE.'Y')IBYTE = 'N'

      CALL ANGLE(ROTATE(1,1,5,1),1,FLX1,ABD1,EXT1,'L')
      CALL ANGLE(ROTATE(1,1,5,2),1,FLX3,ABD3,EXT3,'L')
      CALL ANGLE(ROTATE(1,1,1,3),1,FLX4,ABD4,EXT4,'L')

      IF(IBYTE.EQ.'N')THEN
            ANGLS(1,1) = FLX1        !L FOOT
            ANGLS(1,2) = ABD1        !L FOOT
```

```
      ELSE
              ANGLS(1,1) = 0.0
              ANGLS(1,2) = 0.0
      END IF

      ANGLS(1,3) = EXT1          !L FOOT
      ANGLS(1,4) = EXT3          !L SHANK
      ANGLS(1,5) = FLX4          !L PELVIS

      CALL ANGLE(ROTATE(1,1,5,4),1,FLX1,ABD1,EXT1,'R')
      CALL ANGLE(ROTATE(1,1,5,5),1,FLX3,ABD3,EXT3,'R')
      CALL ANGLE(ROTATE(1,1,1,6),1,FLX4,ABD4,EXT4,'R')

      IF(IBYTE.EQ.'N')THEN
              ANGLS(2,1) = FLX1          !R FOOT
              ANGLS(2,2) = ABD1          !R FOOT
      ELSE
              ANGLS(2,1) = 0.0
              ANGLS(2,2) =  0.0
      END IF

      ANGLS(2,3) = EXT1          !R FOOT
      ANGLS(2,4) = EXT3          !R SHANK
      ANGLS(2,5) = FLX4          !R PELVIS
      ANGLS(1,5) = (ANGLS(1,5)+ANGLS(2,5))/2.0
      ANGLS(2,5) = ANGLS(1,5)

      CALL ANGLE(ROTATE(1,1,1,9),1,FLX3,ABD3,EXT3,'L')
      CALL ANGLE(ROTATE(1,1,1,10),1,FLX4,ABD4,EXT4,'R')
      CALL ANGLE(ROTATE(1,1,1,11),1,FLX5,ABD5,EXT5,'R')
      ANGLS(1,9) = FLX3          !L ARM FLX
      ANGLS(1,10) = ABD3         !L ARM ABD
      ANGLS(1,11) = EXT3         !L ARM ROTATION
      ANGLS(2,9) = FLX4          !R ARM FLX
      ANGLS(2,10) = ABD4         !R ARM ABD
      ANGLS(2,11) = EXT4         !R ARM ROTATION
      ANGLS(1,12) = FLX5         !HEAD FLX
      ANGLS(1,13) = ABD5         !HEAD TILT
      ANGLS(1,14) = EXT5         !HEAD ROTATION
      ANGLS(2,12) = FLX5         !HEAD FLX
      ANGLS(2,13) = ABD5         !HEAD TILT
      ANGLS(2,14) = EXT5         !HEAD ROTATION

C CALCULATE JOINT CENTERS UPPER BODY JOINT CENTERS FIRST IN GLOBAL
C COORDINATES

      WRITE(LUO,194)BELL
      TYPE*,' Finding the upper body joint centers in global coordinates'

      XYZLAM(1,1) = XYZLAM(1,1)        !+ ROTATE(3,1,1,9)*CLARM/(1.0*PI)
      XYZLAM(2,1) = XYZLAM(2,1)        !+ ROTATE(3,2,1,9)*CLARM/(1.0*PI)
      XYZLAM(3,1) = XYZLAM(3,1)        !+ ROTATE(3,3,1,9)*CLARM/(1.0*PI)

      XYZRAM(1,1) = XYZRAM(1,1)        !+ ROTATE(3,1,1,10)*CRARM/(1.0*PI)
      XYZRAM(2,1) = XYZRAM(2,1)        !+ ROTATE(3,2,1,10)*CRARM/(1.0*PI)
      XYZRAM(3,1) = XYZRAM(3,1)        !+ ROTATE(3,3,1,10)*CRARM/(1.0*PI)

C  PELVIS AND TRUNK ABDUCTION AND EXTERNAL ROTATION ANGLES
C  AND DIMENSIONS
```

```
      ANGLS(2,8)=ATAN((XYZLAM(1,1)-XYZRAM(1,1))/
     *                (XYZLAM(3,1)-XYZRAM(3,1)))
      ANGLS(2,8)=ANGLS(2,8)*180./PI
      ANGLS(1,8)=180.0+ANGLS(2,8)
      TWIDTH  = SQRT((XYZLAM(1,1)-XYZRAM(1,1))**2+
     *               (XYZLAM(2,1)-XYZRAM(2,1))**2+
     *               (XYZLAM(3,1)-XYZRAM(3,1))**2)

      ANGLS(2,6)=ATAN((XYZLTK(2,1)-XYZRTK(2,1))/
     *                (XYZLTK(3,1)-XYZRTK(3,1)))
      ANGLS(2,7)=ATAN((XYZLTK(1,1)-XYZRTK(1,1))/
     *                (XYZLTK(3,1)-XYZRTK(3,1)))
      ANGLS(2,6)=ANGLS(2,6)*180./PI
      ANGLS(2,7)=ANGLS(2,7)*180./PI
      ANGLS(1,6)=ANGLS(2,6)
      ANGLS(1,7)=180.0+ANGLS(2,7)
      PWIDTH  = SQRT((XYZLTK(1,1)-XYZRTK(1,1))**2+
     *               (XYZLTK(2,1)-XYZRTK(2,1))**2+
     *               (XYZLTK(3,1)-XYZRTK(3,1))**2)

C  LOWER BACK JOINT CENTER IS THE MIDDLE OF THE VECTOR
C  CONNECTING THE PELVIC CREST

      XYZLTK(1,1) = (XYZLTK(1,1) + XYZRTK(1,1))/2.0
      XYZLTK(2,1) = (XYZLTK(2,1) + XYZRTK(2,1))/2.0
      XYZLTK(3,1) = (XYZLTK(3,1) + XYZRTK(3,1))/2.0

      XYZRTK(1,1) = XYZLTK(1,1)
      XYZRTK(2,1) = XYZLTK(2,1)
      XYZRTK(3,1) = XYZLTK(3,1)

      XYZHD(1,1) = XYZHD(1,1) + ROTATE(3,1,1,11)*CHEAD/(2.0*PI)
      XYZHD(2,1) = XYZHD(2,1) + ROTATE(3,2,1,11)*CHEAD/(2.0*PI)
      XYZHD(3,1) = XYZHD(3,1) + ROTATE(3,3,1,11)*CHEAD/(2.0*PI)

C DETERMINE THE LOWER LIMB JOINT CENTERS USING AXIS OF ROTATION METHOD
C OPEN LEFT SIDE ROM DATA FILE

C JOINT CENTERS DETERMINED DIRECTLY IN ARRAY COORDINATES

      WRITE(LUO,194)BELL
      TYPE*,' Finding the lower limb joints and axes of rotation'

C  Find the floor height from the FPL file
      OPEN(UNIT=15,NAME=NLANKL,TYPE='OLD',ACCESS='DIRECT',
     *       READONLY)
      READ(15'4) FPLNAM(1:16)
      CALL CLOSE(15)
      OPEN(UNIT=15,NAME=FPLNAM,TYPE='OLD',READONLY)
             READ(15,1210) ITEMP6
             READ(15,1211) XFP1,YFP1,ZFP1
             READ(15,1210) ITEMP6
             READ(15,1210) ITEMP6
             READ(15,1210) ITEMP6
             READ(15,1211) XFP2,YFP2,ZFP2
1210         FORMAT(A2)
1211         FORMAT(3F15.7)
      CALL CLOSE(15)
      YFLOOR = (YFP1 + YFP2) / 2.0
```

```
C OUTPUT AXIS OF ROTATION DISPLAY FILES
        IJFOOT='Y'
        IJKNEE='Y'
        IJHIP='Y'

719     OPEN (UNIT=16,NAME='TK4:FJOINT.AXL',ACCESS='DIRECT',
     1          RECL=7,STATUS='UNKNOWN')
        OPEN (UNIT=17,NAME='TK4:FJOINT.AXR',ACCESS='DIRECT',
     1          RECL=7,STATUS='UNKNOWN')


        DO WHILE (IJFOOT.EQ.'Y')        !DO FOOT CALCULATIONS IF SELECTED

720     WRITE(LUO,1296)BELL
1296    FORMAT(/,'$Name of the LEFT ANKLE FLX-EXT data file '
     1          ,'[no default]:',A1)
        READ(LUI,2,ERR=720) NLFLX(5:10)

        OPEN (UNIT=15,NAME=NLFLX,ACCESS='DIRECT',READONLY,
     1          ERR=720,TYPE='OLD')

        READ(15'2)RNCHN,FRAMES
        NFRAME = IFIX(FRAMES)
        NINIT=1
721     WRITE(LUO,331)NINIT
331     FORMAT('$First frame for FLEX-EXT evaluation [I;D=',I5,']:')
        READ(LUI,332,ERR=721)NINIT1
332     FORMAT(I5)
        IF(NINIT1.GT.0)NINIT=NINIT1

5721    WRITE(LUO,5331)NFRAME
5331    FORMAT('$Last frame for FLEX-EXT evaluation [I;D=',I5,']:')
        READ(LUI,332,ERR=5721)NS
        IF(NS.LE.0)NS=NFRAME
        NSD=NS
        NEXT=NS-1
        NFINAL=NINIT

        ANGMIN = 7.5
722     WRITE(LUO,723)ANGMIN
723     FORMAT('$Angmin for the left ankle[DEGREES]? [D=',F5.1,']:')
        READ(LUI,9876,ERR=722)ANGMIN1
        IF(ANGMIN1.GT.0.0)ANGMIN=ANGMIN1
9876    FORMAT(F15.6)
        CALL T5SFLX(2,1,NS,NEXT,NFINAL,ANGMIN,15)!FIND ANKLE FROM FOOT

        WRITE(16'1)UNIT4,ANGMIN,POINT           ! COORDINATES

        IF(UNIT4(1).NE.BAD)THEN
         CALL VECPLN(UNIT4,POINT,ROTATE(1,1,1,1),XYZLA(1,1),
     1          XYZLA(1,5),ROTATE(1,1,5,1),RLVECT(1,1))!FIND ANKLE JOINT CENTER
         DO LK=1,3
          RLVECT(LK,1) = RLVECT(LK,1)*39.37!EXPRESS IN INCHES FOR CONSISTENCY
         ENDDO
        ELSE
         DO LK=1,3
          RLVECT(LK,1) = BAD
         ENDDO
         GOTO 720
        ENDIF
```

```
      NS=NSD
      NEXT=NS-1

      CALL T5SFLX(1,2,NS,NEXT,NFINAL,ANGMIN,15)       !FIND ANKLE FROM SHANK

      WRITE(16'2)UNIT4,ANGMIN,POINT

      IF(UNIT4(1).NE.BAD)THEN
       CALL VECPLN(UNIT4,POINT,ROTATE(1,1,2,1),XYZLA(1,2),
     1        XYZLA(1,5),ROTATE(1,1,5,1),RLVECT(1,2))!FIND ANKLE JOINT CENTEF
       DO LK=1,3
        RLVECT(LK,2) = RLVECT(LK,2)*39.37     !EXPRESS IN INCHES FOR CONSISTE
       ENDDO
      ELSE
       DO LK=1,3
        RLVECT(LK,2) = BAD
       ENDDO
       GOTO 720
      ENDIF

      CLOSE(UNIT=15)

C NOW RIGHT ANKLE
      TYPE*
750   WRITE(LUO,1396)NLFLX(5:10),BELL
1396  FORMAT('$Name of the RIGHT ANKLE FLX-EXT data file '
     1        ,'[DEFAULT-',1A6,']: ',A1)
      READ(LUI,2,ERR=750) NRFLX(5:10)
      IF(NRFLX(5:10).EQ.'      ')THEN
              NRFLX(5:10)=NLFLX(5:10)
      ELSE
              ISTEP=2
      ENDIF

      OPEN (UNIT=15,NAME=NRFLX,ACCESS='DIRECT',READONLY,
     1        ERR=750,TYPE='OLD')

      READ(15'2)RNCHN,FRAMES
      NFRAME = IFIX(FRAMES)
      IF(NSD.GT.NFRAME)NSD=NFRAME

751   WRITE(LUO,331)NINIT
      READ(LUI,332,ERR=751)NINIT1
      IF(NINIT1.GT.0)NINIT=NINIT1

5722  WRITE(LUO,5331)NSD
      READ(LUI,332,ERR=5722)NS
      IF(NS.LE.0)NS=NSD
      NSD=NS

      NEXT=NS-1
      NFINAL=NINIT
752   WRITE(LUO,753)ANGMIN
753   FORMAT('$Angmin for the right ankle[DEGREES]? [D=',F5.1,']:')
      READ(LUI,9876,ERR=752)ANGMIN1
      IF(ANGMIN1.GT.0.0)ANGMIN=ANGMIN1
      CALL T5SFLX(2,1,NS,NEXT,NFINAL,ANGMIN,15)       !FIND ANKLE FROM FOOT

      WRITE(17'1)UNIT4,ANGMIN,POINT
```

```
      IF(UNIT4(1).NE.BAD)THEN
       CALL VECPLN(UNIT4,POINT,ROTATE(1,1,1,4),XYZRA(1,1),
     1        XYZRA(1,5),ROTATE(1,1,5,4),RRVECT(1,1)) !FIND ANKLE JOINT CENTE
       DO LK=1,3
        RRVECT(LK,1) = RRVECT(LK,1)*39.37!EXPRESS IN INCHES FOR CONSISTENCY
       ENDDO
      ELSE
       DO LK=1,3
        RRVECT(LK,1) = BAD
       ENDDO
       GOTO 750
      ENDIF

      NS=NSD
      NEXT=NS-1

      CALL T5SFLX(1,2,NS,NEXT,NFINAL,ANGMIN,15)!FIND ANKLE FROM SHANK

      WRITE(17'2)UNIT4,ANGMIN,POINT

      IF(UNIT4(1).NE.BAD)THEN
       CALL VECPLN(UNIT4,POINT,ROTATE(1,1,2,4),XYZRA(1,2),
     1        XYZRA(1,5),ROTATE(1,1,5,4),RRVECT(1,2)) !FIND ANKLE JOINT CENTE
       DO LK=1,3
        RRVECT(LK,2) = RRVECT(LK,2)*39.37!EXPRESS IN INCHES FOR CONSISTENCY
       ENDDO
      ELSE
       DO LK=1,3
        RRVECT(LK,2) = BAD
       ENDDO
       GOTO 750
      ENDIF

      CLOSE(UNIT=15)

C SIZE STANDARD PROPORTIONED FEET TO TOUCH FLOOR
C Find the height of the ankle above the standing surface
C by subtracting floor height and platform height from ankle height.
C First get foot array to ankle vector in global coordinates
      CALL GMTRA(ROTATE(1,1,1,1),TEMP,3,3)
      CALL GMPRD(TEMP,RLVECT(1,1),VTEMP,3,3,1)
C Then perform subtractions with appropriate unit conversion
C to find the ankle height
      HLFOOT  =  (39.37*XYZLA(2,1)+VTEMP(2)-
     1               39.37*YFLOOR - SFLRHT)
C Similar operation for right ankle
      CALL GMTRA(ROTATE(1,1,1,4),TEMP,3,3)
      CALL GMPRD(TEMP,RRVECT(1,1),VTEMP,3,3,1)
      HRFOOT  =  (39.37*XYZRA(2,1)+VTEMP(2)-
     1               39.37*YFLOOR - RFLRHT)
C Match dimensions
      HFOOT = (HLFOOT+HRFOOT)/2.
      TYPE*,'FROM THE FOOT:HLFOOT     HRFOOT  HFOOT'
      TYPE*,HLFOOT,HRFOOT,HFOOT

C REPEAT BASED ON SHANK
C First get foot array to ankle vector in global coordinates
      CALL GMTRA(ROTATE(1,1,2,1),TEMP,3,3)
      CALL GMPRD(TEMP,RLVECT(1,2),VTEMP,3,3,1)
```

```
C Then perform subtractions with appropriate unit conversion
C to find the ankle height
        HLFOOT  =  (39.37*XYZLA(2,2)+VTEMP(2)-
     1                    39.37*YFLOOR - SFLRHT)
C Similar operation for right ankle
        CALL GMTRA(ROTATE(1,1,2,4),TEMP,3,3)
        CALL GMPRD(TEMP,RRVECT(1,2),VTEMP,3,3,1)
        HRFOOT  =  (39.37*XYZRA(2,2)+VTEMP(2)-
     1                    39.37*YFLOOR - RFLRHT)
C Match dimensions
        TYPE*,'FROM THE SHANK:HLFOOT    HRFOOT  '
        TYPE*,HLFOOT,HRFOOT

1331    WRITE(LUO,540)BELL
        READ(LUI,560,ERR=1331)IJFOOT
        IF(IJFOOT.EQ.'Y'.OR.IJFOOT.EQ.'y')IJFOOT='Y'
        ENDDO

C NOW THE KNEES
        DO WHILE (IJKNEE.EQ.'Y')          !DO KNEE CALCULATIONS IF
                                          !SELECTED
730     WRITE(LUO,1286)NRFLX(5:10),BELL
1286    FORMAT(/,'$Name of the LEFT KNEE FLX-EXT data file '
     1          ,'[DEFAULT-',1A6,']: ',A1)
        READ(LUI,2,ERR=730) NLFLX(5:10)
        IF(NLFLX(5:10).EQ.'      ')NLFLX(5:10)=NRFLX(5:10)

        OPEN (UNIT=15,NAME=NLFLX,ACCESS='DIRECT',READONLY,
     1          ERR=730,TYPE='OLD')


        READ(15'2)RNCHN,FRAMES
        NFRAME = IFIX(FRAMES)
        IF(NSD.GT.NFRAME)NSD=NFRAME

731     WRITE(LUO,331)NINIT
        READ(LUI,332,ERR=731)NINIT1
        IF(NINIT1.GT.0)NINIT=NINIT1

5723    WRITE(LUO,5331)NSD
        READ(LUI,332,ERR=5723)NS
        IF(NS.LE.0)NS=NSD
        NSD=NS

        NEXT=NS-1
        NFINAL=NINIT
        ANGMIN=15.

732     WRITE(LUO,733)ANGMIN
733     FORMAT('$Angmin for the left knee[DEGREES]? [D=',F5.1,']:')
        READ(LUI,9876,ERR=732)ANGMIN1
        IF(ANGMIN1.GT.0.0)ANGMIN=ANGMIN1
        CALL T5SFLX(3,2,NS,NEXT,NFINAL,ANGMIN,15)       !FIND KNEE FROM SHANK

        WRITE(16'3)UNIT4,ANGMIN,POINT

        IF(UNIT4(1).NE.BAD)THEN
         CALL VECPLN(UNIT4,POINT,ROTATE(1,1,2,2),XYZLK(1,2),
     1          XYZLK(1,5),ROTATE(1,1,5,2),RLVECT(1,3))!FIND KNEE JOINT CENTER
         DO LK=1,3
```

```
          RLVECT(LK,3) = RLVECT(LK,3)*39.37!EXPRESS IN INCHES FOR CONSISTENCY
         ENDDO
        ELSE
         DO LK=1,3
          RLVECT(LK,3) = BAD
         ENDDO
         GOTO 730
        ENDIF

        NS=NSD
        NEXT=NS-1

        CALL T5SFLX(2,3,NS,NEXT,NFINAL,ANGMIN,15)        !FIND KNEE FROM THIGH

        WRITE(16'4)UNIT4,ANGMIN,POINT

        IF(UNIT4(1).NE.BAD)THEN
         CALL VECPLN(UNIT4,POINT,ROTATE(1,1,3,2),XYZLK(1,3),
     1          XYZLK(1,5),ROTATE(1,1,5,2),RLVECT(1,4))!FIND KNEE JOINT CENTER
         DO LK=1,3
          RLVECT(LK,4) = RLVECT(LK,4)*39.37!EXPRESS IN INCHES FOR CONSISTENCY
         ENDDO
        ELSE
         DO LK=1,3
          RLVECT(LK,4) = BAD
         ENDDO
         GOTO 730
        ENDIF

        CLOSE(UNIT=15)

C NOW THE RIGHT SIDE
        TYPE*
760     WRITE(LUO,1386)NLFLX(5:10),BELL
1386    FORMAT('$Name of the RIGHT KNEE FLX-EXT data file '
     1          ,'[DEFAULT-',1A6,']: ',A1)
        READ(LUI,2,ERR=760)NRFLX(5:10)
        IF(NRFLX(5:10).EQ.'      ')THEN
                NRFLX(5:10)=NLFLX(5:10)
        ELSE
                ISTEP=2
        ENDIF
        OPEN (UNIT=15,NAME=NRFLX,ACCESS='DIRECT',READONLY,
     1          ERR=760,TYPE='OLD')


        READ(15'2)RNCHN,FRAMES
        NFRAME = IFIX(FRAMES)
        IF(NSD.GT.NFRAME)NSD=NFRAME


761     WRITE(LUO,331)NINIT
        READ(LUI,332,ERR=761)NINIT1
        IF(NINIT1.GT.0)NINIT=NINIT1

5724    WRITE(LUO,5331)NSD
        READ(LUI,332,ERR=5724)NS
        IF(NS.LE.0)NS=NSD
        NSD=NS
```

```
      NEXT=NS-1
      NFINAL=NINIT
762   WRITE(LUO,763)ANGMIN
763   FORMAT('$Angmin for the right knee[DEGREES]? [D=',F5.1,']:')
      READ(LUI,9876,ERR=762)ANGMIN1
      IF(ANGMIN1.GT.0.0)ANGMIN=ANGMIN1
      CALL T5SFLX(3,2,NS,NEXT,NFINAL,ANGMIN,15)        !FIND KNEE FROM SHANK

      WRITE(17'3)UNIT4,ANGMIN,POINT

      IF(UNIT4(1).NE.BAD)THEN
       CALL VECPLN(UNIT4,POINT,ROTATE(1,1,2,5),XYZRK(1,2),
     1        XYZRK(1,5),ROTATE(1,1,5,5),RRVECT(1,3))  !FIND KNEE JOINT CENTE
       DO LK=1,3
        RRVECT(LK,3) = RRVECT(LK,3)*39.37!EXPRESS IN INCHES FOR CONSISTENCY
       ENDDO
      ELSE
       DO LK=1,3
        RRVECT(LK,3) = BAD
       ENDDO
       GOTO 760
      ENDIF

      NS=NSD
      NEXT=NS-1
      CALL T5SFLX(2,3,NS,NEXT,NFINAL,ANGMIN,15)        !FIND KNEE FROM THIGH
      WRITE(17'4)UNIT4,ANGMIN,POINT

      IF(UNIT4(1).NE.BAD)THEN
       CALL VECPLN(UNIT4,POINT,ROTATE(1,1,3,5),XYZRK(1,3),
     1        XYZRK(1,5),ROTATE(1,1,5,5),RRVECT(1,4))  !FIND KNEE JOINT CENTER
       DO LK=1,3
        RRVECT(LK,4) = RRVECT(LK,4)*39.37!EXPRESS IN INCHES FOR CONSISTENCY
       ENDDO
      ELSE
       DO LK=1,3
        RRVECT(LK,4) = BAD
       ENDDO
       GOTO 760
      ENDIF

      CLOSE(UNIT=15)
C CHECK KNEE HEIGHTS
C CALCULATIONS THE KNEE HEIGHTS
C Find the height of the KNEE above the standing surface
C by subtracting floor height and platform height from ankle height.
C First get foot array to ankle vector in global coordinates
      CALL GMTRA(ROTATE(1,1,2,2),TEMP,3,3)
      CALL GMPRD(TEMP,RLVECT(1,3),VTEMP,3,3,1)
C Then perform subtractions with appropriate unit conversion
C to find the ankle height
      HLKNEE  =  (39.37*XYZLK(2,2)+VTEMP(2)-
     1                39.37*YFLOOR - SFLRHT)
C Similar operation for right ankle
      CALL GMTRA(ROTATE(1,1,2,5),TEMP,3,3)
      CALL GMPRD(TEMP,RRVECT(1,3),VTEMP,3,3,1)
      HRKNEE  =  (39.37*XYZRK(2,2)+VTEMP(2)-
     1                39.37*YFLOOR - RFLRHT)
C Match dimensions
      TYPE*,'REFERENCE THE SHANK: HLKNEE      HRKNEE'
```

```
      TYPE*,HLKNEE,HRKNEE

C REPEAT REFERENCE THE THIGH ARRAY
      CALL GMTRA(ROTATE(1,1,3,2),TEMP,3,3)
      CALL GMPRD(TEMP,RLVECT(1,4),VTEMP,3,3,1)
C Then perform subtractions with appropriate unit conversion
C to find the ankle height
      HLKNEE  =  (39.37*XYZLK(2,3)+VTEMP(2)-
     1                  39.37*YFLOOR - SFLRHT)
C Similar operation for right ankle
      CALL GMTRA(ROTATE(1,1,3,5),TEMP,3,3)
      CALL GMPRD(TEMP,RRVECT(1,4),VTEMP,3,3,1)
      HRKNEE  =  (39.37*XYZRK(2,3)+VTEMP(2)-
     1                  39.37*YFLOOR - RFLRHT)
C Match dimensions
      TYPE*,'REFERENCE THE THIGH :HLKNEE      HRKNEE'
      TYPE*,HLKNEE,HRKNEE
1332  WRITE(LUO,540)BELL
      READ(LUI,560,ERR=1332)IJKNEE
      IF(IJKNEE.EQ.'Y'.OR.IJKNEE.EQ.'y')IJKNEE='Y'
      ENDDO

C NOW FIND THE HIP JOINT CENTERS

      DO WHILE (IJHIP.EQ.'Y') !DO HIP CALCULATION IF SELECTED
740   WRITE(LUO,1276)NRFLX(5:10),BELL
1276  FORMAT(/,'$Name of the LEFT HIP FLX-EXT data file '
     1          ,'[DEFAULT-',1A6,']: ',A1)
      READ(LUI,2,ERR=740) NLFLX(5:10)
      IF(NLFLX(5:10).EQ.'      ')NLFLX(5:10)=NRFLX(5:10)

      OPEN (UNIT=15,NAME=NLFLX,ACCESS='DIRECT',READONLY,
     1          ERR=740,TYPE='OLD')

      READ(15'2)RNCHN,FRAMES
      NFRAME = IFIX(FRAMES)
      IF(NSD.GT.NFRAME)NSD=NFRAME

741   WRITE(LUO,331)NINIT
      READ(LUI,332,ERR=741)NINIT1
      IF(NINIT1.GT.0)NINIT=NINIT1

5726  WRITE(LUO,5331)NSD
      READ(LUI,332,ERR=5726)NS
      IF(NS.LE.0)NS=NSD
      NSD=NS

      NEXT=NS-1
      NFINAL=NINIT
      ANGMIN=15.0
742   WRITE(LUO,743)ANGMIN
743   FORMAT('$Angmin for the left hip[DEGREES]? [D=',F5.1,']:')
      READ(LUI,9876,ERR=742)ANGMIN1
      IF(ANGMIN1.GT.0.0)ANGMIN=ANGMIN1
      CALL T5SFLX(4,3,NS,NEXT,NFINAL,ANGMIN,15)      !FIND HIP FROM THIGH

      WRITE(16'5)UNIT4,ANGMIN,POINT

      IF(UNIT4(1).NE.BAD)THEN
       CALL VECPLN(UNIT4,POINT,ROTATE(1,1,3,3),XYZLH(1,3),
```

```
     1           XYZLH(1,1),ROTATE(1,1,1,3),RLVECT(1,5))!FIND HIP JOINT CENTER
         DO LK=1,3
          RLVECT(LK,5) = RLVECT(LK,5)*39.37!EXPRESS IN INCHES FOR CONSISTENCY
         ENDDO
        ELSE
         DO LK=1,3
          RLVECT(LK,5) = BAD
         ENDDO
         GOTO 740
        ENDIF

        NS=NSD
        NEXT=NS-1

        CALL T5SFLX(3,4,NS,NEXT,NFINAL,ANGMIN,15)       !FIND HIP FROM PELVIS

        WRITE(16'6)UNIT4,ANGMIN,POINT

        IF(UNIT4(1).NE.BAD)THEN
         CALL VECPLN(UNIT4,POINT,ROTATE(1,1,4,3),XYZLH(1,4),
     1           XYZLH(1,1),ROTATE(1,1,1,3),RLVECT(1,6))!FIND HIP JOINT CENTER
         DO LK=1,3
          RLVECT(LK,6) = RLVECT(LK,6)*39.37!EXPRESS IN INCHES FOR CONSISTENCY
         ENDDO
        ELSE
         DO LK=1,3
          RLVECT(LK,6) = BAD
         ENDDO
         GOTO 740
        ENDIF

        CALL CLOSE(15)            !CLOSE LEFT SIDE  ROM DATA FILE

C NOW THE RIGHT SIDE
        TYPE*
770     WRITE(LUO,1376)NLFLX(5:10),BELL
1376    FORMAT('$Name of the RIGHT HIP FLX-EXT data file '
     1          ,'[DEFAULT-',1A6,']: ',A1)
        READ(LUI,2,ERR=770) NRFLX(5:10)
        IF(NRFLX(5:10).EQ.'      ')THEN
                NRFLX(5:10)=NLFLX(5:10)
        ELSE
                ISTEP=2
        ENDIF

        OPEN (UNIT=15,NAME=NRFLX,ACCESS='DIRECT',READONLY,
     1          ERR=770,TYPE='OLD')

        READ(15'2)RNCHN,FRAMES
        NFRAME = IFIX(FRAMES)
        IF(NSD.GT.NFRAME)NSD=NFRAME

771     WRITE(LUO,331)NINIT
        READ(LUI,332,ERR=771)NINIT1
        IF(NINIT1.GT.0) NINIT=NINIT1

5725    WRITE(LUO,5331)NSD
        READ(LUI,332,ERR=5725)NS
        IF(NS.LE.0)NS=NSD
        NSD=NS
```

```
      NEXT=NS-1
      NFINAL=NINIT
772   WRITE(LUO,773)ANGMIN
773   FORMAT('$Angmin for the right hip[DEGREES]? [D=',F5.1,'] ')
      READ(LUI,9876,ERR=772)ANGMIN1
      IF(ANGMIN1.GT.0.0)ANGMIN=ANGMIN1
      CALL T5SFLX(4,3,NS,NEXT,NFINAL,ANGMIN,15)       !FIND HIP FROM THIGH

      WRITE(17'5)UNIT4,ANGMIN,POINT

      IF(UNIT4(1).NE.BAD)THEN
       CALL VECPLN(UNIT4,POINT,ROTATE(1,1,3,6),XYZRH(1,3),
     1        XYZRH(1,1),ROTATE(1,1,1,6),RRVECT(1,5))  !FIND HIP JOINT CENTER
       DO LK=1,3
        RRVECT(LK,5) = RRVECT(LK,5)*39.37!EXPRESS IN INCHES FOR CONSISTENCY
       ENDDO
      ELSE
       DO LK=1,3
        RRVECT(LK,5) = BAD
       ENDDO
       GOTO 770
      ENDIF

      NS=NSD
      NEXT=NS-1

      CALL T5SFLX(3,4,NS,NEXT,NFINAL,ANGMIN,15)       !FIND HIP FROM PELVIS

      WRITE(17'6)UNIT4,ANGMIN,POINT

      IF(UNIT4(1).NE.BAD)THEN
       CALL VECPLN(UNIT4,POINT,ROTATE(1,1,4,6),XYZRH(1,4),
     1        XYZRH(1,1),ROTATE(1,1,1,6),RRVECT(1,6))  !FIND HIP JOINT CENTER
       DO LK=1,3
        RRVECT(LK,6) = RRVECT(LK,6)*39.37!EXPRESS IN INCHES FOR CONSISTENCY
       ENDDO
      ELSE
       DO LK=1,3
        RRVECT(LK,6) = BAD
       ENDDO
       GOTO 770
      ENDIF

      CALL CLOSE(15)          !CLOSE RIGHT SIDE ROM DATA FILE

C CALCULATION HIP JOINT HEIGHTS
C Find the height of the HIP above the standing surface
C by subtracting floor height and platform height from ankle height.
C First get foot array to ankle vector in global coordinates
      CALL GMTRA(ROTATE(1,1,3,3),TEMP,3,3)
      CALL GMPRD(TEMP,RLVECT(1,5),VTEMP,3,3,1)
C Then perform subtractions with appropriate unit conversion
C to find the ankle height
      HLHIP  =  (39.37*XYZLH(2,3)+VTEMP(2)-
     1              39.37*YFLOOR - SFLRHT)
C Similar operation for right ankle
      CALL GMTRA(ROTATE(1,1,3,6),TEMP,3,3)
      CALL GMPRD(TEMP,RRVECT(1,5),VTEMP,3,3,1)
      HRHIP  =  (39.37*XYZRH(2,3)+VTEMP(2)-
```

```
     1                   39.37*YFLOOR - RFLRHT)
C Match dimensions
        TYPE*,'REFERENCE THE THIGH: HLHIP      HRHIP'
        TYPE*,HLHIP,HRHIP

C REPEAT REFERENCE THE PELVIS
        CALL GMTRA(ROTATE(1,1,4,3),TEMP,3,3)
        CALL GMPRD(TEMP,RLVECT(1,6),VTEMP,3,3,1)
C Then perform subtractions with appropriate unit conversion
C to find the ankle height
        HLHIP  =  (39.37*XYZLH(2,4)+VTEMP(2)-
     1                   39.37*YFLOOR - SFLRHT)
C Similar operation for right ankle
        CALL GMTRA(ROTATE(1,1,4,6),TEMP,3,3)
        CALL GMPRD(TEMP,RRVECT(1,6),VTEMP,3,3,1)
        HRHIP  =  (39.37*XYZRH(2,4)+VTEMP(2)-
     1                   39.37*YFLOOR - RFLRHT)
C Match dimensions
        TYPE*,'HLHIP    HRHIP'
        TYPE*,HLHIP,HRHIP
1333    WRITE(LUO,540)BELL
        READ(LUI,560,ERR=1333)IJHIP
        IF(IJHIP.EQ.'Y'.OR.IJHIP.EQ.'y')IJHIP='Y'
        ENDDO

        CALL CLOSE(16)          !CLOSE AXIS OF ROTATION FILE
        CALL CLOSE(17)
C CALCULATE ARRAY TO JOINT CENTER VECTORS FOR THE UPPER BODY

        WRITE(LUO,194)BELL
        TYPE*,' Calculate upper body array to joint center vectors.'

C  Left pelvic array to left sacral joint

        VTEMP(1) = (XYZLTK(1,1)-XYZLTK(1,4))*39.37
        VTEMP(2) = (XYZLTK(2,1)-XYZLTK(2,4))*39.37
        VTEMP(3) = (XYZLTK(3,1)-XYZLTK(3,4))*39.37
        CALL GMPRD(ROTATE(1,1,4,7),VTEMP,RLVECT(1,7),3,3,1)

C  Left trunk array to left sacral joint

        VTEMP(1) = (XYZLTK(1,1) - XYZLTK(1,5))*39.37
        VTEMP(2) = (XYZLTK(2,1) - XYZLTK(2,5))*39.37
        VTEMP(3) = (XYZLTK(3,1) - XYZLTK(3,5))*39.37
        CALL GMPRD(ROTATE(1,1,5,7),VTEMP,RLVECT(1,8),3,3,1)

C  Left trunk to left shoulder joint

        VTEMP(1) = (XYZLAM(1,1) - XYZLAM(1,5))*39.37
        VTEMP(2) = (XYZLAM(2,1) - XYZLAM(2,5))*39.37
        VTEMP(3) = (XYZLAM(3,1) - XYZLAM(3,5))*39.37
        CALL GMPRD(ROTATE(1,1,5,9),VTEMP,RLVECT(1,9),3,3,1)

C  Left arm to left shoulder

        VTEMP(1) = (XYZLAM(1,1) - XYZLAM(1,6))*39.37
        VTEMP(2) = (XYZLAM(2,1) - XYZLAM(2,6))*39.37
        VTEMP(3) = (XYZLAM(3,1) - XYZLAM(3,6))*39.37
        CALL GMPRD(ROTATE(1,1,6,9),VTEMP,RLVECT(1,10),3,3,1)
        CALL GMPRD(ROTATE(1,1,6,13),VTEMP,XYZHND(1,1),3,3,1)
```

```
C  Right pelvic array to Right sacral joint

      VTEMP(1) = (XYZRTK(1,1) - XYZRTK(1,4))*39.37
      VTEMP(2) = (XYZRTK(2,1) - XYZRTK(2,4))*39.37
      VTEMP(3) = (XYZRTK(3,1) - XYZRTK(3,4))*39.37
      CALL GMPRD(ROTATE(1,1,4,8),VTEMP,RRVECT(1,7),3,3,1)

C   Right trunk array to Right sacral joint

      VTEMP(1) = (XYZRTK(1,1) - XYZRTK(1,5))*39.37
      VTEMP(2) = (XYZRTK(2,1) - XYZRTK(2,5))*39.37
      VTEMP(3) = (XYZRTK(3,1) - XYZRTK(3,5))*39.37
      CALL GMPRD(ROTATE(1,1,5,8),VTEMP,RRVECT(1,8),3,3,1)

C  Right trunk to Right shoulder joint

      VTEMP(1) = (XYZRAM(1,1) - XYZRAM(1,5))*39.37
      VTEMP(2) = (XYZRAM(2,1) - XYZRAM(2,5))*39.37
      VTEMP(3) = (XYZRAM(3,1) - XYZRAM(3,5))*39.37
      CALL GMPRD(ROTATE(1,1,5,10),VTEMP,RRVECT(1,9),3,3,1)

C  Right arm to right shoulder

      VTEMP(1) = (XYZRAM(1,1) - XYZRAM(1,6))*39.37
      VTEMP(2) = (XYZRAM(2,1) - XYZRAM(2,6))*39.37
      VTEMP(3) = (XYZRAM(3,1) - XYZRAM(3,6))*39.37
      CALL GMPRD(ROTATE(1,1,6,10),VTEMP,RRVECT(1,10),3,3,1)

C  Right trunk to neck joint

      VTEMP(1) = (XYZHD(1,1) - XYZHD(1,5))*39.37
      VTEMP(2) = (XYZHD(2,1) - XYZHD(2,5))*39.37
      VTEMP(3) = (XYZHD(3,1) - XYZHD(3,5))*39.37
      CALL GMPRD(ROTATE(1,1,5,11),VTEMP,RRVECT(1,11),3,3,1)

C  Right head to neck joint

      VTEMP(1) = (XYZHD(1,1) - XYZHD(1,7))*39.37
      VTEMP(2) = (XYZHD(2,1) - XYZHD(2,7))*39.37
      VTEMP(3) = (XYZHD(3,1) - XYZHD(3,7))*39.37
      CALL GMPRD(ROTATE(1,1,7,11),VTEMP,RRVECT(1,12),3,3,1)

C       WRITE OUT GEOMETRY FILES

775     WRITE(LUO,500) "7
500     FORMAT('$Enter the name of the anatomical data file ',
     *          'to create ',A1)
        READ(LUI,501,ERR=775) NMLOUT(5:10)
501     FORMAT(1A6)

        NMROUT(5:10)=NMLOUT(5:10)

C       OUTPUT LEFT SIDE DATA
        OPEN(UNIT=17,NAME=NMLOUT,TYPE='NEW')
                WRITE(17,502)
                WRITE(LUO,502)
502             FORMAT('L SIDE')
                WRITE(17,503) ANGLS(1,3),ANGLS(1,1),ANGLS(1,2),
     1                  SFOOT,HFOOT
```

```
            WRITE(LUO,503) ANGLS(1,3),ANGLS(1,1),ANGLS(1,2),
     1             SFOOT,HFOOT
 503        FORMAT(F15.6,' LEFT FOOT EXTERNAL ROTATION ANGLE'/,
     *      F15.6,' LEFT FOOT DORSI-FLEXION ANGLE'/,
     *      F15.6,' LEFT FOOT INVERSION ANGLE',/,
     *      2F15.6,' LEFT FOOT LENGTH and LEFT FOOT HEIGHT')
            WRITE(17,504) ANGLS(1,4)
            WRITE(LUO,504) ANGLS(1,4)
 504        FORMAT(F15.6,' LEFT KNEE EXTERNAL ROTATION ANGLE')
            WRITE(17,505) ANGLS(1,7),ANGLS(1,6),ANGLS(1,5)
            WRITE(LUO,505) ANGLS(1,7),ANGLS(1,6),ANGLS(1,5)
 505        FORMAT(F15.6,' LEFT PELVIS EXTERNAL ROTATION ANGLE',/,
     *      F15.6,' LEFT PELVIS ABDUCTION ANGLE',/,
     *      F15.6,' LEFT PELVIS FLEXION ANGLE')
            WRITE(17,506) ANGLS(1,8),TWIDTH,PWIDTH
            WRITE(LUO,506) ANGLS(1,8),TWIDTH,PWIDTH
 506        FORMAT(F15.6,' LEFT TRUNK ROTATION ANGLE',/,
     *             F15.6,' TRUNK WIDTH',/,
     *             F15.6,' PELVIS WIDTH')
            WRITE(17,507) ANGLS(1,11),ANGLS(1,10),ANGLS(1,9)
            WRITE(LUO,507) ANGLS(1,11),ANGLS(1,10),ANGLS(1,9)
 507        FORMAT(F15.6,' LEFT ARM ROTATION ANGLE',/,
     *      F15.6,' LEFT ARM ABDUCTION ANGLE',/,
     *      F15.6,' LEFT ARM FLEXION ANGLE')
            WRITE(17,508)RLARM
            WRITE(LUO,508)RLARM
 508        FORMAT(F15.6,' LENGTH OF LEFT ARM')
C           WRITE(17,509) ANGLS(1,14),ANGLS(1,13),ANGLS(1,12)
 509        FORMAT(F15.6,' HEAD EXTERNAL ROTATION ANGLE',/,
     *      F15.6,' HEAD ABDUCTION ANGLE',/,
     *      F15.6,' HEAD FLEXION ANGLE')
            WRITE(17,510) ( (RLVECT(J,K),J=1,3), K=1,10)
            WRITE(LUO,510) ( (RLVECT(J,K),J=1,3), K=1,10)
 510        FORMAT(3F15.6)
            WRITE(17,512)SANKLE,SRKNEE,STHIGH
            WRITE(LUO,512)SANKLE,SRKNEE,STHIGH
 512        FORMAT(F15.6,' LEFT ANKLE CIRCUMFERENCE'/,
     1             F15.6,' LEFT KNEE CIRCUMFERENCE',/,
     2             F15.6,' LEFT THIGH CIRCUMFERENCE')
            WRITE(17,514)CWASTE,CTRUNK,CLARM
            WRITE(LUO,514)CWASTE,CTRUNK,CLARM
 514        FORMAT(F15.6,' WASTE CIRCUMFERENCE'/,
     1             F15.6,' CHEST CIRCUMFERENCE',/,
     2             F15.6,' LEFT ARM CIRCUMFERENCE')
      CALL CLOSE(17)
C
      OPEN(UNIT=17,NAME=NMROUT,TYPE='NEW')
            WRITE(17,602)
            WRITE(LUO,602)
 602        FORMAT('R SIDE')
            WRITE(17,603) ANGLS(2,3),ANGLS(2,1),ANGLS(2,2),
     1             RFOOT,HFOOT
            WRITE(LUO,603) ANGLS(2,3),ANGLS(2,1),ANGLS(2,2),
     1             RFOOT,HFOOT
 603        FORMAT(F15.6,' RIGHT FOOT EXTERNAL ROTATION ANGLE'/,
     *      F15.6,' RIGHT FOOT DORSI-FLEXION ANGLE'/,
     *      F15.6,' RIGHT FOOT INVERSION ANGLE',/,
     *      2F15.6,' RIGHT FOOT LENGTH AND HEIGHT')
            WRITE(17,604) ANGLS(2,4)
            WRITE(LUO,604) ANGLS(2,4)
```

```
604           FORMAT(F15.6,' RIGHT KNEE EXTERNAL ROTATION ANGLE')
              WRITE(17,605) ANGLS(2,7),ANGLS(2,6),ANGLS(2,5)
              WRITE(LUO,605) ANGLS(2,7),ANGLS(2,6),ANGLS(2,5)
605           FORMAT(F15.6,' RIGHT PELVIS EXTERNAL ROTATION ANGLE',/,
     *        F15.6,' RIGHT PELVIS ABDUCTION ANGLE',/,
     *        F15.6,' RIGHT PELVIS FLEXION ANGLE')
              WRITE(17,606) ANGLS(2,8),TWIDTH,PWIDTH
              WRITE(LUO,606) ANGLS(2,8),TWIDTH,PWIDTH
606           FORMAT(F15.6,' RIGHT TRUNK ROTATION ANGLE',/,
     *                F15.6,' TRUNK WIDTH',/,
     *                F15.6,' PELVIS WIDTH')
              WRITE(17,607) ANGLS(2,11),ANGLS(2,10),ANGLS(2,9)
              WRITE(LUO,607) ANGLS(2,11),ANGLS(2,10),ANGLS(2,9)
607           FORMAT(F15.6,' RIGHT ARM ROTATION ANGLE',/,
     *        F15.6,' RIGHT ARM ABDUCTION ANGLE',/,
     *        F15.6,' RIGHT ARM FLEXION ANGLE')
              WRITE(17,608)RRARM
              WRITE(LUO,608)RRARM
608           FORMAT(F15.6,' LENGTH OF RIGHT ARM')
              CCHEAD = CHEAD * 39.37
              WRITE(17,609) ANGLS(2,14),ANGLS(2,13),ANGLS(2,12),CCHEAD
              WRITE(LUO,609) ANGLS(2,14),ANGLS(2,13),ANGLS(2,12),CCHEAD
609           FORMAT(F15.6,' HEAD EXTERNAL ROTATION ANGLE',/,
     *        F15.6,' HEAD ABDUCTION ANGLE',/,
     *        F15.6,' HEAD FLEXION ANGLE',/,
     *        F15.6,' NECK CIRCUMFERENCE')
              WRITE(17,610) ( (RRVECT(J,K),J=1,3), K=1,12)
              WRITE(LUO,610) ( (RRVECT(J,K),J=1,3), K=1,12)
610           FORMAT(3F15.6)
              WRITE(17,612)RANKLE,RRKNEE,RTHIGH
              WRITE(LUO,612)RANKLE,RRKNEE,RTHIGH
612           FORMAT(F15.6,' RIGHT ANKLE CIRCUMFERENCE'/,
     1                F15.6,' RIGHT KNEE CIRCUMFERENCE',/,
     2                F15.6,' RIGHT THIGH CIRCUMFERENCE')
              WRITE(17,614)CWASTE,CTRUNK,CRARM
              WRITE(LUO,614)CWASTE,CTRUNK,CRARM
614           FORMAT(F15.6,' WASTE CIRCUMFERENCE'/,
     1                F15.6,' CHEST CIRCUMFERENCE',/,
     2                F15.6,' REFT ARM CIRCUMFERENCE')
              WRITE(17,616)CTHEAD
              WRITE(LUO,616)CTHEAD
616           FORMAT(F15.6,' HEAD CIRCUMFERENCE')
        CALL CLOSE(17)

        WRITE(LUO,520)
520     FORMAT(//,' ***********************************************')
1335    WRITE(LUO,540)BELL
540     FORMAT('$Repeat range of motion calculations? [Y/N]',A1)
        READ(LUI,560,ERR=1335)IBYTE
560     FORMAT(1A1)
        IF(IBYTE.EQ.'Y'.OR.IBYTE.EQ.'y')THEN
1336     WRITE(LUO,542)
542      FORMAT('$Repeat FOOT range of motion calculations? [Y/N;D=N]')
         READ(LUI,560,ERR=1336)IJFOOT
         IF(IJFOOT.EQ.'Y'.OR.IJFOOT.EQ.'y')IJFOOT='Y'
1337     WRITE(LUO,544)
544      FORMAT('$Repeat KNEE range of motion calculations? [Y/N;D=N]')
         READ(LUI,560,ERR=1337)IJKNEE
         IF(IJKNEE.EQ.'Y'.OR.IJKNEE.EQ.'y')IJKNEE='Y'
1338     WRITE(LUO,546)
```

```
546     FORMAT('$Repeat HIP range of motion calculations? [Y/N;D=N]')
        READ(LUI,560,ERR=1338)IJHIP
        IF(IJHIP.EQ.'Y'.OR.IJHIP.EQ.'y')IJHIP='Y'
        GOTO 719
       ENDIF

       CALL EXIT
       END
C***********************************************************
C
       SUBROUTINE AVERAG(NAME,ROT,XYZ)
       CHARACTER*16 NAME
       DIMENSION ROT(3,3,7),XYZ(3,7)
       DIMENSION SUM(16,7,1000),ICOUNT(7),X(7),Y(7),Z(7)
C
       BAD = -1000.0

       DO 22 K=1,7
               ICOUNT(K)=0
               X(K)=0.0
               Y(K)=0.0
               Z(K)=0.0
22     CONTINUE

       OPEN(UNIT=15,NAME=NAME,TYPE='OLD',ACCESS='DIRECT',
     *         READONLY)
       READ(15'2) RNCHN,FRAMES,FREQ,RNSEG
       NSEG=IFIX(RNSEG)
       NFRAME=IFIX(FRAMES)
C
       DO 21 K=1,NFRAME
               READ(15'K+10) ((SUM(I1,I2,K),I1=1,16),I2=1,NSEG)
21     CONTINUE
       CLOSE(UNIT=15)

       DO 1 K=1,NFRAME
               DO 2 ISEG = 1,NSEG
                       IF (SUM(1,ISEG,K).EQ.BAD) GOTO 2
                       IF (SUM(1,ISEG,K).EQ.0.0 .AND. SUM(2,ISEG,K).EQ.0.0
     *                 .AND. SUM(3,ISEG,K).EQ.0.0) GOTO 2
                       ICOUNT(ISEG)=ICOUNT(ISEG)+1
C
                       X(ISEG)=X(ISEG)+SUM(1,ISEG,K)
                       Y(ISEG)=Y(ISEG)+SUM(2,ISEG,K)
                       Z(ISEG)=Z(ISEG)+SUM(3,ISEG,K)
C
                       ROT(1,1,ISEG)=ROT(1,1,ISEG)+SUM(8,ISEG,K)
                       ROT(1,2,ISEG)=ROT(1,2,ISEG)+SUM(9,ISEG,K)
                       ROT(1,3,ISEG)=ROT(1,3,ISEG)+SUM(10,ISEG,K)
                       ROT(2,1,ISEG)=ROT(2,1,ISEG)+SUM(11,ISEG,K)
                       ROT(2,2,ISEG)=ROT(2,2,ISEG)+SUM(12,ISEG,K)
                       ROT(2,3,ISEG)=ROT(2,3,ISEG)+SUM(13,ISEG,K)
                       ROT(3,1,ISEG)=ROT(3,1,ISEG)+SUM(14,ISEG,K)
                       ROT(3,2,ISEG)=ROT(3,2,ISEG)+SUM(15,ISEG,K)
                       ROT(3,3,ISEG)=ROT(3,3,ISEG)+SUM(16,ISEG,K)

2              CONTINUE
1      CONTINUE

       DO 3 ISEG = 1,NSEG
```

```
              IF (ICOUNT(ISEG).EQ.0) GOTO 13

              UNITMG = SQRT(ROT(1,1,ISEG)**2+ROT(1,2,ISEG)**2+
     *        ROT(1,3,ISEG)**2)
              ROT(1,1,ISEG)=ROT(1,1,ISEG)/UNITMG
              ROT(1,2,ISEG)=ROT(1,2,ISEG)/UNITMG
              ROT(1,3,ISEG)=ROT(1,3,ISEG)/UNITMG

              UNITMG = SQRT(ROT(2,1,ISEG)**2+ROT(2,2,ISEG)**2+
     *        ROT(2,3,ISEG)**2)
              ROT(2,1,ISEG)=ROT(2,1,ISEG)/UNITMG
              ROT(2,2,ISEG)=ROT(2,2,ISEG)/UNITMG
              ROT(2,3,ISEG)=ROT(2,3,ISEG)/UNITMG

              ROT(3,1,ISEG) = ROT(1,2,ISEG)*ROT(2,3,ISEG)-
     *        ROT(2,2,ISEG)*ROT(1,3,ISEG)
              ROT(3,2,ISEG) = ROT(2,1,ISEG)*ROT(1,3,ISEG)-
     *        ROT(1,1,ISEG)*ROT(2,3,ISEG)
              ROT(3,3,ISEG) = ROT(1,1,ISEG)*ROT(2,2,ISEG)-
     *        ROT(2,1,ISEG)*ROT(1,2,ISEG)

              X(ISEG) = X(ISEG)/FLOAT(ICOUNT(ISEG))
              Y(ISEG) = Y(ISEG)/FLOAT(ICOUNT(ISEG))
              Z(ISEG) = Z(ISEG)/FLOAT(ICOUNT(ISEG))
              GOTO 14
13            X(ISEG)=BAD
              Y(ISEG)=BAD
              Z(ISEG)=BAD
14            XYZ(1,ISEG)=X(ISEG)
              XYZ(2,ISEG)=Y(ISEG)
              XYZ(3,ISEG)=Z(ISEG)
3       CONTINUE

        RETURN
        END
C*****************************************************************
C
C       SUBROUTINE TO FIND THE ANATOMICAL ANGLES
C
        SUBROUTINE ANGLE(R,JOINT,FLX,ABD,EXTROT,LEG)
        DIMENSION R(3,3)
        BYTE LEG
C
        PI = 3.1415927
        SCALE = 1.0
        IF (LEG.NE.'R') SCALE=-1.0
C
        FLX =  - ATAN2( R(2,1) , R(2,2) )*180./PI*SCALE
        ABD = ( PI/2.0 - ACOS( R(2,3) ) )*180./PI
        RNUMER = R(1,3)
        RDENOM = R(1,1)
        EXTROT = ATAN2(RNUMER,RDENOM) *180./PI*SCALE
C
        IF(JOINT.EQ.2) FLX = -FLX
C
        RETURN
        END
C***********************************************************************
C
```

```
C  VECPLN                DEC-1986          PATRICK O. RILEY
C
C        THIS SUBROUTINE DETERMINES THE INTERCEPT[JCENT] BETWEEN A VECTOR
C        [UNIT] WITH ORIGIN AT A POINT [POINT],  DEFINED IN AN ARRAY
C        COORDINATE SYSTEM [APNT],[AROT], AND A PLANE DEFINED IN THE GLOBAL
C        COORDINATE SYSTEM BY THE LOCATION AND ORIENTATION ANOTHER ARRAY.
C        THE SECOND ARRAY LOCATION IS [ORIGIN] AND ITS ORIENTATION IS
C        SPECIFIED BY THE ROTATION MATRIX [AXIS]
C***********************************************************************

        SUBROUTINE VECPLN(UNIT,POINT,AROT,APNT,ORIGIN,AXIS,JCENT)
        REAL JCENT(3)
        DIMENSION UNIT(3),POINT(3),AROT(3,3),ORIGIN(3),AXIS(3,3),APNT(3)
        DIMENSION TEMP1(3),TEMP2(3),TEMP3(3),RTEMP1(3,3),RTEMP2(3,3)

C FIND THE VECTOR FROM THE REFERENCE ARRAY [AROT,APNT]TO THE END OF [UNIT]
C  IN [AROT]COORDINATES

          TEMP3(1) = POINT(1) + UNIT(1)
          TEMP3(2) = POINT(2) + UNIT(2)
          TEMP3(3) = POINT(3) + UNIT(3)

C  ROTATE [POINT] NEW [TEMP3] INTO GLOBAL COORDINATE SYSTEM

        CALL GMTRA(AROT,RTEMP1,3,3)
        CALL GMPRD(RTEMP1,POINT,TEMP1,3,3,1)
        CALL GMPRD(RTEMP1,TEMP3,TEMP2,3,3,1)

C FIND THE VECTOR FROM THE [ORIGIN] TO [APNT] IN GLOBAL COORDINATES

        TEMP3(1)=APNT(1)-ORIGIN(1)        !VECTOR FROM ORIGIN TO
        TEMP3(2)=APNT(2)-ORIGIN(2)        !REFERENCE ARRAY
        TEMP3(3)=APNT(3)-ORIGIN(3)

C FIND THE VECTOR FROM [ORIGIN] TO [POINT] IN GLOBAL COORDINATES

        POINT(1) = TEMP1(1) + TEMP3(1)
        POINT(2) = TEMP1(2) + TEMP3(2)
        POINT(3) = TEMP1(3) + TEMP3(3)
C FIND THE VECTOR FROM [ORIGIN] TO THE END OF [UNIT] IN GLOBAL COORDINATES

        UNIT(1)  = TEMP2(1)  + TEMP3(1)
        UNIT(2)  = TEMP2(2)  + TEMP3(2)
        UNIT(3)  = TEMP2(3)  + TEMP3(3)
C ROTATE [POINT] AND [UNIT] INTO THE [AXIS] COORDINATE SYSTEM

        CALL GMPRD(AXIS,POINT,TEMP1,3,3,1)      !IN [AXIS]COORDINATES
        CALL GMPRD(AXIS,UNIT ,TEMP2,3,3,1)      !IN [AXIS]COORDINATES

C FIND THE VECTOR FROM [POINT] TO [UNIT] IN [AXIS] COORDINATES

        UNIT(1) = TEMP2(1) - TEMP1(1)
        UNIT(2) = TEMP2(2) - TEMP1(2)
        UNIT(3) = TEMP2(3) - TEMP1(3)

C FIND THE INTERCEPT OF UNIT WITH THE X-Y PLANE DEFINED BY [AXIS]
C PASSING THROUGH [ORIGIN]

          RATIO = -TEMP1(3)/UNIT(3)     !Z DISTANCE FROM [POINT] TO PLANE
```

```
        TEMP2(3) = 0.0                        !Z = ZERO PLANE IN [AXIS]
        TEMP2(1) = TEMP1(1) + UNIT(1)*RATIO     !TRAVEL IN [UNIT] DIRECTION
        TEMP2(2) = TEMP1(2) + UNIT(2)*RATIO     !ACROSS X-Y PLANE WHILE

C ROTATE VECTOR TO THE INTERCEPT POINT IN GLOBAL COORDINATES

        CALL GMTRA(AXIS,RTEMP1,3,3)
        CALL GMPRD(RTEMP1,TEMP2,POINT,3,3,1)

C FIND THE VECTOR FROM THE REF ARRAY
C TO THE INTERCEPT POINT IN GLOBAL COORDNATES

        POINT(1)= POINT(1) - TEMP3(1)
        POINT(2)= POINT(2) - TEMP3(2)
        POINT(3)= POINT(3) - TEMP3(3)

C ROTATE THE VECTOR INTO REFERENCE ARRAY COORDINATES

        CALL GMPRD(AROT,POINT,JCENT,3,3,1)
        END
C***************************************************************************
```

```
C       SUBROUTINE T5SFLX
C       JUL,1986         PATRICK O. RILEY
C       FROM    FLEX5             BY BOB FIJAN
C
C     FINITE DISPLACEMENT METHOD SUBROUTINE TO FIND
C     AN AVERAGE AXIS OF ROTATION
C
C    CALL
C       CALL T5SFLX(J,JR,NINIT,NEXT,NFINAL,ANGMIN,NLUN)
C       COMMON/FLXOUT/UNIT4(3),POINT(3),ISTEP
C
C    INPUT
C       J-SEGMENT AXIS OF ROTATION TO BE DETERMINED
C       WITH RESPECT TO SEGMENT-JR[0 FOR GLOBAL COORDINATES]
C       NINIT-FIRST FRAME
C       NEXT-LAST FRAME, INCREMENTED IF SEPERATION TOO SMALL
C       NFINAL-LAST ACCEPTABLE LAST FRAME
C       ANGMIN-MINIMUM ACCEPTABLE ROTATION ANGLE OF DATA POINTS
C       NLUN-LOGICAL UNIT NUMBER TO ASSIGNED INPUT FILE
C
C     OUTPUT
C       UNIT4-UNIT VECTOR AXIS OF ROTATION
C       POINT-VECTOR FROM ARRAY TO AXIS OF ROTATION
C       ISTEP-NUMBER OF FRAMES TO DISMIN
C**********************************************************************
C
        SUBROUTINE T5SFLX(J,JR,NINIT,NEXT,NFINAL,ANGMIN,NLUN)
        COMMON/FLXOUT/UNIT4(3),POINT(3),ISTEP
C
        REAL*8 X1,X2,X3,U1,U2,U3,THETA,COSTH,
     1         THETAM,AX,AY,AZ,ANGLE,DPOINT(3)
        DIMENSION UNIT8(3),VECT8(3)
        DIMENSION D(7,10),SEGRNT(9,10)
        DIMENSION TEMP(3),RTEMP(3,3)
        DIMENSION VECTOR(3,3),ROTATE(3,3,3)
        DIMENSION UNIT(3),UNIT0(3)
C
        PARAMETER(LUO=6,LUI=5,BAD=0.0,BAD1=-1000.)

        TYPE*,J,JR,NINIT,NEXT,NFINAL,ANGMIN
        THETAM = 0.0
        DO IR=1,3
                UNIT(IR)=0.0
                UNIT0(IR)=0.0
                UNIT4(IR)=0.0
                UNIT8(IR)=0.0
                POINT(IR)=0.0
                VECT8(IR)=0.0
                TEMP(IR)=0.0
                DO IJ=1,3
                  RTEMP(IR,IJ)=0.0
                  VECTOR(IR,IJ)=0.0
                  DO KL=1,3
                    ROTATE(IR,IJ,KL)=0.0
                  ENDDO
                ENDDO
        ENDDO
        UNIT(3)=1.0
```

```
      READ (NLUN'2) RNCHN,FRAMES,FREQ,RNSEG,FPLSCL
      NFRAME=IFIX(FRAMES)
      NSEG=IFIX(RNSEG)
      IFF=1
      ILF=NFRAME
      IF (NINIT.LT.IFF)NINIT=IFF+1
      IF (NINIT.LT.ILF)NINIT=ILF
      IF (NEXT.LT.IFF)NEXT=IFF

      JM=J
      IF(JR.GT.JM)JM=JR

      ICOUNT=0
      IMIN=1

C  Read Initial and Final Data and Rotate it, if necessary.
      IND=1
      K=NINIT

13    READ(NLUN'K+10)
     * (((D(J2,J3),J2=1,7),(SEGRNT(J2,J3),J2=1,9)),J3=1,JM)
C  Relative to another segment.

      IF (D(1,J).EQ.BAD .AND. D(2,J).EQ.BAD .AND.
     *                D(3,J).EQ.BAD) THEN
                      TYPE*,'BAD DATA'
                      K=K-1
                      GOTO 13
      ENDIF
      IF (D(1,JR).EQ.BAD .AND. D(2,JR).EQ.BAD .AND.
     *                D(3,JR).EQ.BAD) THEN
                      TYPE*,'BAD DATA'
                      K=K-1
                      GOTO 13
      ENDIF
      IF (D(1,J).EQ.BAD1 .AND. D(2,J).EQ.BAD1 .AND.
     *                D(3,J).EQ.BAD1) THEN
                      TYPE*,'BAD DATA'
                      K=K-1
                      GOTO 13
      ENDIF
      IF (D(1,JR).EQ.BAD1.AND.D(2,JR).EQ.BAD1 .AND.
     *                D(3,JR).EQ.BAD1) THEN
                      TYPE*,'BAD DATA'
                      K=K-1
                      GOTO 13
      ENDIF

      CALL GMTRA(SEGRNT(1,JR),RTEMP,3,3)

      DO 15 I1=1,3
              TEMP(I1)=D(I1,J)-D(I1,JR)
15    CONTINUE
      CALL GMPRD(RTEMP,TEMP,VECTOR(1,IND),3,3,1)
      CALL GMPRD(RTEMP,SEGRNT(1,J),ROTATE(1,1,IND),3,3,3)

C     ELIMINATE POINTS THAT ARE TOO CLOSE TOGETHER

      IF (IND.NE.1)THEN
              CALL GMTRA(ROTATE(1,1,1),RTEMP,3,3)
```

```
      CALL GMPRD(RTEMP,ROTATE(1,1,2),ROTATE(1,1,3),3,3,3)

      COSTH=0.50*(DBLE(ROTATE(1,1,3))+DBLE(ROTATE(2,2,3))
     1     +DBLE(ROTATE(3,3,3))-1.0)
      IF (COSTH.GE.1.0)COSTH=1.0
      IF (COSTH.LE.-1.0)COSTH=-1.0

      THETA=DACOS(COSTH)*180./3.1415927
C     TYPE*,'THETA=',THETA
      IF(THETA.GT.THETAM)THETAM = THETA

      IF(THETA.GE.DBLE(ANGMIN))GOTO 577
   ENDIF

      IF(IND.EQ.1)K = NEXT
      IF(IND.EQ.2)K = K - 1
      IF (K .LT. NFINAL) GOTO 9997
      IND = 2
      GOTO 13

C  Find the position and orientation of the final frame
C      with respect to the initial frame.

577   DO 20 I1=1,3
         TEMP(I1)=VECTOR(I1,2)-VECTOR(I1,1)
20    CONTINUE

      CALL GMPRD(RTEMP,TEMP,VECTOR(1,3),3,3,1)

C  Find the axis of rotation with respect to the initial frame's data.

      X1 = DBLE(VECTOR(1,3))  !relative  displacement vector
      X2 = DBLE(VECTOR(2,3))
      X3 = DBLE(VECTOR(3,3))

c axis vector from off diagional terms
      U1=DBLE(ROTATE(3,2,3))-DBLE(ROTATE(2,3,3))
      U2=DBLE(ROTATE(1,3,3))-DBLE(ROTATE(3,1,3))
      U3=DBLE(ROTATE(2,1,3))-DBLE(ROTATE(1,2,3))
2020  UNIMAG=DSQRT( U1**2 + U2**2 + U3**2 )
      U1 = U1/UNIMAG
      U2 = U2/UNIMAG
      U3 = U3/UNIMAG

c Rodrigues vector
      AX=THETA*U1
      AY=THETA*U2
      AZ=THETA*U3
C

      CALL AXIS( X1,X2,X3,AX,AY,AZ,ANGLE,DPOINT)

C  Find the axis of rotation with respect to the BCS of segment JR.

      DO LKJ =1,3
         POINT(LKJ)=SNGL(DPOINT(LKJ))
      ENDDO

      CALL GMPRD(ROTATE(1,1,1),POINT,TEMP,3,3,1)
      DO 21 I1=1,3
```

```
                POINT(I1)=VECTOR(I1,1)+TEMP(I1)
21      CONTINUE
        TEMP(1)=SNGL(AX)
        TEMP(2)=SNGL(AY)
        TEMP(3)=SNGL(AZ)
        CALL GMPRD(ROTATE(1,1,1),TEMP,UNIT4,3,3,1)
C
        IF (UNIT4(3).GT.0.0) GOTO 779
        UNIT4(1) = - UNIT4(1)
        UNIT4(2) = - UNIT4(2)
        UNIT4(3) = - UNIT4(3)
C AVERAGE THE VECTORS
779     DO IR=1,3
                UNIT8(IR)=UNIT8(IR)+UNIT4(IR)
                VECT8(IR)=VECT8(IR)+POINT(IR)
        ENDDO
        K=K-1
        ICOUNT=ICOUNT+1
        IF(K.GT.NFINAL.AND.ICOUNT.LE.20)GOTO 13
        TYPE*,'K FINAL',K
        UMAG = SQRT( UNIT8(1)**2 + UNIT8(2)**2 + UNIT8(3)**2 )

        DO 781 KK=1,3
                UNIT4(KK) = UNIT8(KK)/UMAG
                POINT(KK) = VECT8(KK)/FLOAT(ICOUNT)
781     CONTINUE

        ISTEP=IABS(K-NINIT)     !ADVANCE TO DISMIN THIS RUN

        GOTO 998                !successful determination

9997    TYPE*,'MINIMUM ANGLE NOT FOUND'
        TYPE*,'THETA MAX = ',THETAM
        POINT(1)=-1000.0
        POINT(2)=-1000.0
        POINT(3)=-1000.0
        UNIT4(1)=-1000.0
        UNIT4(2)=-1000.0
        UNIT4(3)=-1000.0
        ISTEP=INTVL

998     RETURN
        END
C**************************************************************
C       SUBROUTINE AXIS(X1,X2,X3,R1,R2,R3,ANGLED,VECT)
C
C       THIS SUBROUTINE FINDS THE VECTOR TO THE AXIS OF ROTATION
C       GIVEN:
C       X       THE TRANSLATION VECTOR
C       R       THE RODRIGUES VECTOR  (RETURNED AS A UNIT VECTOR)
C
        SUBROUTINE AXIS(X1,X2,X3,R1,R2,R3,ANGLED,VECT)
        IMPLICIT REAL*8(A-H,O-Z)
        DIMENSION X(3),R(3),VECT(3),U(3),ARRAY(3,3)
        X(1)=X1
        X(2)=X2
        X(3)=X3
        R(1)=R1
        R(2)=R2
        R(3)=R3
```

```
C
C       Find the unit vector and the rotation angle.
C
        PI=3.1415927

        ANGLED=DSQRT(R(1)**2+R(2)**2+R(3)**2)

        DO 1 I=1,3
                U(I)=R(I)/ANGLED
1       CONTINUE

        R1=U(1)
        R2=U(2)
        R3=U(3)

        ANGLE=ANGLED*PI/180.
C
C       Find the component of translation normal to the axis of rotation.
C
        XDOTU=(X(1)*U(1)+X(2)*U(2)+X(3)*U(3))
        DO 2 I=1,3
                X(I)=X(I)-XDOTU*U(I)
2       CONTINUE
C
C       Find the coefficients.
C
        A=1.0-DCOS(ANGLE)
        B=-1.0*DSIN(ANGLE)
        ARRAY(1,1)=A
        ARRAY(1,2)=-B*U(3)
        ARRAY(1,3)=B*U(2)
        ARRAY(2,1)=B*U(3)
        ARRAY(2,2)=A
        ARRAY(2,3)=-B*U(1)
        ARRAY(3,1)=-B*U(2)
        ARRAY(3,2)=B*U(1)
        ARRAY(3,3)=A
C
C       Solve the simultaneous equations to get the desired VECTOR.
C
        CALL SIMUL(ARRAY,X,VECT)

C       Find the components of VECTOR normal to the axis of rotation.

        VDOTU=(VECT(1)*U(1)+VECT(2)*U(2)+VECT(3)*U(3))

        DO 3 I=1,3
                VECT(I)=VECT(I)-VDOTU*U(I)
3       CONTINUE

        RETURN
        END

C*************************************************
        SUBROUTINE SIMUL(A,X,V)
        IMPLICIT REAL*8(A-H,O-Z)
        DIMENSION A(3,3),X(3),V(3)
C
C       DIAGONALIZE THE MATRIX
C
```

```
      XMAX=-1.0

      DO 1 I=1,3
            TMPMAX=DABS(A(I,1))
            IF (TMPMAX.LT.XMAX) GOTO 1
            XMAX=TMPMAX
            INDEX=I
1     CONTINUE
      IF (INDEX.EQ.1) GOTO 2
      S=X(1)
      X(1)=X(INDEX)
      X(INDEX)=S

      DO 3 J=1,3
            S=A(1,J)
            A(1,J)=A(INDEX,J)
            A(INDEX,J)=S
3     CONTINUE

2     DO 5 I=2,3
            RATIO=A(I,1)/A(1,1)
            X(I)=X(I)-X(1)*RATIO
            DO 6 J=1,3
                  A(I,J)=A(I,J)-A(1,J)*RATIO
6           CONTINUE
5     CONTINUE
      IF ( DABS( A(2,2) ) .GT. DABS( A(3,2) ) ) GOTO 7
      S=X(2)
      X(2)=X(3)
      X(3)=S
      DO 8 J=2,3
            S=A(2,J)
            A(2,J)=A(3,J)
            A(3,J)=S
8     CONTINUE

7     RATIO=A(3,2)/A(2,2)

      X(3)=X(3)-X(2)*RATIO
      A(3,3)=A(3,3)-A(2,3)*RATIO

C SOLVE THE DIAGONALIZED EQUATIONS
      V(3)=X(3)/A(3,3)
      V(2)=(X(2)-A(2,3)*V(3))/A(2,2)
      V(1)=(X(1)-A(1,3)*V(3)-A(1,2)*V(2))/A(1,1)

      RETURN
      END
```

Appendix D

Geometry Determination

This Appendix contains program CAMGEO.FTN which was developed to determine the position and orientation of the left hand camera coordinate system in terms of the right hand (master) camera coordinate system. This program finds the 4X4 rotation matrix which describes the translations and rotations necessary to produce the best fit of the array of positions described in the left side coordinate system into the corresponding array of positions defined in the right hand coordinate system. The arrays of positions represent a set of approximately 30 3-D positions of the center of a bilateral array; i.e. an array seen simultaneously by both camera systems. A modified version of the TRACK routine T3SOCA.SUB is used to fit the two arrays and is given here as T5SOCA.FTN.

```
C   CAMGEO -- TRACK III BATCH ORIENTATION CALCULATION
C   CAMGEO
C   CAMGEO         THIS PROGRAM FORMS DATA FROM ONE SET OF CAMERAS
C   CAMGEO         INTO A SEGMENT FILE CONSISTING OF ONE ARRAY WITH
C   CAMGEO         UP TO 64 ELEMENTS.  THE ARRAY ORIGIN IS THE ORIGIN
C   CAMGEO         OF THE SET OF CAMERAS.  THE ARRAY IS THEN ROTATED AND
C   CAMGEO         TRANSLATE TO FIT DATA FROM THE OPPOSITE SET OF
C   CAMGEO         CAMERAS USING T5SOCA.  THIS YIELDS THE POSITION
C   CAMGEO         OF THE ARRAY SIDE CAMERA ORIGIN IN THE OPPOSITE
C   CAMGEO         SIDE COORDINATE SYSTEM.  IT ALSO DETERMINES THE
C   CAMGEO         ROTATION MATRIX TO ROTATE THE ARRAY SIDE
C   CAMGEO         CAMERA COORDINATE SYSTEM INTO THE DATA SIDE
C   CAMGEO         CAMERA COORDINATE SYSTEM.
C
C       JUNE 1986                    PATRICK O. RILEY
C~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~
C
      PROGRAM CAMGEO
      COMMON /BLKCAL/RPOS(2,3),ROT(2,3)
      COMMON /BLK3DC/X(64),Y(64),Z(64)
      COMMON /BLKDAT/NCHN,NFRAME,IBUF(128)
      COMMON /BLKMO2/RSEGM(64),XHRAW(64),YHRAW(64),ZHRAW(64),RSEGMX
      COMMON /BLKRE1/XI(10),YI(10),ZI(10),AX(10),AY(10),AZ(10)
      COMMON /BLKRE2/DISCRD(64),ROTERR(10),SEGROT(10,3,3)
      COMMON /BLKDT1/SAVE,NPIC
      DIMENSION RLEDX(64),RLEDY(64),RLEDZ(64),RROT(3,3),TEMP(3)
      DIMENSION ICOMNT(16),ROTATE(3,3,2),TEMP1(3,3),TEMP2(3)
      BYTE NAMFIL(15),NAMSEG(15),NAMSTT(12)
      DATA NAMFIL/'T','K','4',':','T','E','S','T','0','0',
     * '.','D','T','R',"0/
      DATA NAMSEG/'T','K','4',':','T','E','S','T','0','0',
     * '.','D','T','L',"0/
C
      NAME10="60
      NAME9="60
      RSEGMX=1.0
      LUO = 5
      LUI = 5
C
C     WRITE(LUO,1000)
 1000 FORMAT(' Calculate BCS Result orientations (rotations) ')
C
C  Initialize the position input array to 0.0
C
      DO 300 I=1,64
      X(I)=0.
      Y(I)=0.
      Z(I)=0.
  300 CONTINUE
      R=1000.
C
C  Initialize the rotation array to 0.0
C
      DO 200 J=1,10
      XI(J)=0.
      YI(J)=0.
      ZI(J)=0.
      DO 400 I=1,3
      SEGROT(J,1,I)=0.
      SEGROT(J,1,I)=0.
```

```
      SEGROT(J,1,I)=0.
 400  CONTINUE
 200  CONTINUE
C
CCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCC
C
C Determine the number of files to be used.
      WRITE(LUO,3125)"7
3125  FORMAT('$Enter the number of files to be used
     *[I,1-64] ',A1)
      READ(LUI,3126)NFIL
3126  FORMAT(I4)
      IF(NFIL.GT.64)NFIL=64
      IF(NFIL.LE.0)NFIL=1
      WRITE(LUO,2) "7
2     FORMAT('$Enter the starting file name: ',A1)
      READ(LUI,3) (NAMSTT(KL),KL=1,6)
3     FORMAT(6A1)
      IF(NAMSTT(1).EQ.' ')GOTO 765
      DO 432 JU=1,6
              NAMFIL(JU+4)=NAMSTT(JU)
              NAMSEG(JU+4)=NAMSTT(JU)
432   CONTINUE
      NAME10=NAMFIL(10)-"1
      NAME9=NAMFIL(9)
C
C Read in the array center positions
765   DO 1 I=1,NFIL
      NAME10 = NAME10+"1
      IF(NAME10.GE."70)NAME9=NAME9+"1
      IF(NAME10.GE."70)NAME10="60
      NAMFIL(9) = NAME9
      NAMFIL(10) = NAME10
              DO 1789 LK=9,10
                      NAMSEG(LK)=NAMFIL(LK)
1789          CONTINUE
C
301   WRITE(LUO,302)(NAMFIL(IQ),IQ=5,10)
      PRINT 302,(NAMFIL(IQ),IQ=5,10)
302   FORMAT(1X,6A1)
C
C     GET THE RIGHT SIDE POINTS
              OPEN(UNIT=11,NAME=NAMFIL,TYPE='OLD',
     *        ACCESS='DIRECT',READONLY)
              READ(11'2) RNCHN,FRAMES,FREQ
              NFRAME=IFIX(FRAMES)
              ICOUNT=0
C
C     AVERAGE OVER NUMBER OF FRAMES
              DO 3456 IJK=1,NFRAME
              READ(11'IJK+10) DX,DY,DZ
              IF(DX.EQ.-1000.)GOTO 3456       !NO BAD PTS
              X(I)=X(I)+DX
              Y(I)=Y(I)+DY
              Z(I)=Z(I)+DZ
              ICOUNT=ICOUNT+1
3456          CONTINUE
              X(I)=X(I)/FLOAT(ICOUNT) !AVERAGE RESULT
              Y(I)=Y(I)/FLOAT(ICOUNT)
              Z(I)=Z(I)/FLOAT(ICOUNT)
```

```
C
                PRINT*,X(I),Y(I),Z(I)
                TYPE*,X(I),Y(I),Z(I)
                CLOSE(11)
C
C       NOW GET THE LEFT SIDE POINTS
                OPEN(UNIT=13,NAME=NAMSEG,TYPE='OLD',
     *          ACCESS='DIRECT',READONLY)
                READ(13'2) RNCHN,FRAMES,FREQ
                READ(13'3)((RPOS(J,IJ),IJ=1,3),J=1,2)
                NFRAME=IFIX(FRAMES)
                ICOUNT=0

                DO 4456 IJK=1,NFRAME
                READ(13'IJK+10) DX,DY,DZ
                IF(DX.EQ.-1000.)GOTO 4456        !NO BAD PTS
                XHRAW(I)=XHRAW(I)+DX
                YHRAW(I)=YHRAW(I)+DY
                ZHRAW(I)=ZHRAW(I)+DZ
                ICOUNT=ICOUNT+1
4456            CONTINUE

                XHRAW(I)=XHRAW(I)/FLOAT(ICOUNT) !COMPUTE
                YHRAW(I)=YHRAW(I)/FLOAT(ICOUNT) !AVERAGES
                ZHRAW(I)=ZHRAW(I)/FLOAT(ICOUNT)

                PRINT*,XHRAW(I),YHRAW(I),ZHRAW(I)
                TYPE*,XHRAW(I),YHRAW(I),ZHRAW(I)
                CALL CLOSE(13)
1       CONTINUE
C
        NCHN=NFIL
        NSEG=1
        DO 1367 I=1,NFIL
                RSEGM(I)=1.0
1367    CONTINUE
C
CCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCCC
C
        NFRAME=1
        OPEN(UNIT=14,TYPE='NEW',NAME='CAMGEO.DAT')
C
C  Calculate the motion and store in the output file.
C
        CALL T5SOCA
C
        PRINT *,' CAMGEO.DAT'
C       DO 100 K=1,NFRAME
        NSEGMX=IFIX(RSEGMX)
        WRITE(14,1987)(XI(J),YI(J),ZI(J),AX(J),AY(J),AZ(J),
     *SEGROT(J,1,1),SEGROT(J,1,2),SEGROT(J,1,3),
     *SEGROT(J,2,1),SEGROT(J,2,2),SEGROT(J,2,3),
     *SEGROT(J,3,1),SEGROT(J,3,2),SEGROT(J,3,3),J=1,NSEGMX)

        WRITE(14,1987)(ROTERR(J),J=1,NSEGMX)

        PRINT 1987,(XI(J),YI(J),ZI(J),AX(J),AY(J),AZ(J),
     *SEGROT(J,1,1),SEGROT(J,1,2),SEGROT(J,1,3),
     *SEGROT(J,2,1),SEGROT(J,2,2),SEGROT(J,2,3),
     *SEGROT(J,3,1),SEGROT(J,3,2),SEGROT(J,3,3),J=1,NSEGMX)
```

```
      PRINT 1987,(ROTERR(J),J=1,NSEGMX)

 100  CONTINUE
1987  FORMAT(3F15.6)
      CALL CLOSE(14)
C
C     USE TRANSPOSE OF SEGROT TO MODIFY CAMERA POSITIONS
      DO 567 KL=1,3
              DO 567 LL=1,3
                      RROT(LL,kL)=SEGROT(1,KL,LL)
567   CONTINUE
C
C     MULTIPLY THE LEFT SIDE CAMERA POSITION FILE TO
C     REFLECT THE DETERMINED GEOMETRY.
C
      OPEN(UNIT=19,NAME=RPOS,TYPE='OLD',READONLY)
              READ(19,125) ICOMNT
125           FORMAT(16A2)
              READ(19,126) RPOS(1,1),RPOS(1,2),RPOS(1,3)
              READ(19,126) ROTATE(1,1,1),ROTATE(1,2,1),
     *                ROTATE(1,3,1)
              READ(19,126) ROTATE(2,1,1),ROTATE(2,2,1),
     *                ROTATE(2,3,1)
              READ(19,126) ROTATE(3,1,1),ROTATE(3,2,1),
     *                ROTATE(3,3,1)
              READ(19,126) RPOS(2,1),RPOS(2,2),RPOS(2,3)
              READ(19,126) ROTATE(1,1,2),ROTATE(1,2,2),
     *                ROTATE(1,3,2)
              READ(19,126) ROTATE(2,1,2),ROTATE(2,2,2),
     *                ROTATE(2,3,2)
              READ(19,126) ROTATE(3,1,2),ROTATE(3,2,2),
     *                ROTATE(3,3,2)
126           FORMAT(3F15.7)
      CALL CLOSE(19)
C
C     ROTATE AND TRANSLATE TO LEFT CAMERA 1 POSITION VECTOR
              TEMP(1)=RPOS(1,1)
              TEMP(2)=RPOS(1,2)
              TEMP(3)=RPOS(1,3)
              CALL GMPRD(RROT,TEMP,TEMP2,3,3,1)
              RPOS(1,1) = TEMP2(1)+XI(1)
              RPOS(1,2) = TEMP2(2)+YI(1)
              RPOS(1,3) = TEMP2(3)+ZI(1)
C
C     ROTATE AND TRANSLATE TO LEFT CAMERA 2 POSITION VECTOR
              TEMP(1)=RPOS(2,1)
              TEMP(2)=RPOS(2,2)
              TEMP(3)=RPOS(2,3)
              CALL GMPRD(RROT,TEMP,TEMP2,3,3,1)
              RPOS(2,1) = TEMP2(1)+XI(1)
              RPOS(2,2) = TEMP2(2)+YI(1)
              RPOS(2,3) = TEMP2(3)+ZI(1)
C
C     ROTATE TO LEFT CAMERA 1 ROTATION MATRIX
              CALL GMPRD(RROT,ROTATE(1,1,1),TEMP1,3,3,3)
      DO 1644 KI=1,3
              DO 1644 KJ=1,3
                      ROTATE(KI,KJ,1)=TEMP1(KI,KJ)
1644  CONTINUE
```

```
C
C       ROTATE TO LEFT CAMERA 2 ROTATION MATRIX
                CALL GMPRD(RROT,ROTATE(1,1,2),TEMP1,3,3,3)
        DO 2644 KI=1,3
                DO 2644 KJ=1,3
                        ROTATE(KI,KJ,2)=TEMP1(KI,KJ)
2644    CONTINUE
C
C       WRITE OUT THE NEW LEFT SIDE ROTATION MATRIX
C
        OPEN(UNIT=19,NAME='NEWLFT.CAM',TYPE='NEW')
                WRITE(19,5125)
5125    FORMAT(' LEFT SIDE CAMERAS,RIGHT SIDE COORDINATES')
                WRITE(19,126) RPOS(1,1),RPOS(1,2),RPOS(1,3)
                WRITE(19,126) ROTATE(1,1,1),ROTATE(1,2,1),
     *                  ROTATE(1,3,1)
                WRITE(19,126) ROTATE(2,1,1),ROTATE(2,2,1),
     *                  ROTATE(2,3,1)
                WRITE(19,126) ROTATE(3,1,1),ROTATE(3,2,1),
     *                  ROTATE(3,3,1)
                WRITE(19,126) RPOS(2,1),RPOS(2,2),RPOS(2,3)
                WRITE(19,126) ROTATE(1,1,2),ROTATE(1,2,2),
     *                  ROTATE(1,3,2)
                WRITE(19,126) ROTATE(2,1,2),ROTATE(2,2,2),
     *                  ROTATE(2,3,2)
                WRITE(19,126) ROTATE(3,1,2),ROTATE(3,2,2),
     *                  ROTATE(3,3,2)
        CALL CLOSE(19)

C
C
        CALL EXST(1)
        END
```

```
C  T5SOCA -- TRACK III SUBROUTINE ORIENTATION CALCULATION
C  T5SOCA
C  T5SOCA
C  T5SOCA
C
C       26-SEP-80          by Erik Antonsson
C       25-AUG-81          modified by Dan Ottenheimer
C       21-NOV-81          modified by Erik Antonsson
C       10-JUL-86          Modified by Pat Riley
C
C~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~
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C~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~~
C
C  This routine calculates the orientative parameters for the number of
C  designated bodies and finds the new positional descriptions at body
C  coordinate system origins if the particular mean and the BCS origin
C  are not coincident.
C
C******* MODIFIED 3/22/78 TO WRITE THE RODRIGUES VECTOR INSTEAD ******
C******* OF THE ORTHOGONAL ROTATIONS IN THE OUTPUT FILE         ******
C
C******* MODIFIED on 8/25/81 by D. Ottenheimer to improve calculation of
C******* rotation matrix and to output the rotation vector instead of
C******* the Rodriguez vector.
C
C******* MODIFIED 21-NOV-81 by EKA to use the transpose of SEGROT(J,M,N)
C******* in the calculation of the correction vector from the average
C******* position to the origin of the Body Coordinate System.
C
C*************************************************************************
C
C       THIS SET OF SUBROUTINES MUST BE COMPILED AS:
C
C               F77 T5SOCA/-F77=T5SOCA
C
C       COMPILING USING THE FORTRAN 77 COMPILER WILL PRODUCE AN OBJECT
C       FILE WHICH WILL RUN WITH ERRORS WHEN TASK BUILT.  WILL GET A
C       FLOATING ZERO DIVIDE IN LINE 36 OF T3SGSC AND THE RESULT FILE
C       WILL CONTAIN MAJOR ERRORS.  THIS IS APPARENTLY A F77 COMPILER
C       ERROR.
C*************************************************************************

      SUBROUTINE T5SOCA
      COMMON /BLKDAT/NCHN,NFRAME,IBUF(128)
```

```
      COMMON /BLK3DC/ XD(64),YD(64),ZD(64)
      COMMON /BLKMO2/RSEGM(64),XARAW(64),YARAW(64),ZARAW(64),RSEGMX
      COMMON /BLKRE1/XI(10),YI(10),ZI(10),AX(10),AY(10),AZ(10)
      COMMON /BLKRE2/DISCRD(64),DISERR(10),SEGROT(10,3,3)
      COMMON/GAUDAT/S(4,4),A,B,C,D
      DIMENSION X(64),Y(64),Z(64)
      DIMENSION XHRAW(64),YHRAW(64),ZHRAW(64)
      DIMENSION VARDIS(10),ERRDIS(64)
      DIMENSION RMHATX(10),RMHATY(10),RMHATZ(10)
      DIMENSION XH(64),YH(64),ZH(64)
      INTEGER CHNSUM,CHNSKP(64),SEGSKP(10)
      DOUBLE PRECISION SR13,SR23,SR12,SCTH2
      REAL J1,J2,J3,L1,L2,L3,MAGD
      REAL J1S,J2S,J3S,J13,J23,J12
C
C  Determine tolerance
C
      write(5,2678)"7
2678  format('$Enter number of deviations of DISERR'
     *          ' to be accepted [R]',A1)
      READ(5,8762)RMUL
8762  FORMAT(F10.5)
      PRINT 6897,RMUL
6897  FORMAT(' STANDARD DEVIATION LIMIT:',F10.5)
C
C  Initialize arrays to zero.
C
      DO 255 I=1,NCHN
      CHNSKP(I)=0
  255 CONTINUE
      ISEGMX=IFIX(RSEGMX)
C
C  Find averages for the arbitrary and neutral data.
C
      IRUN=0
1997  DO 310 J=1,ISEGMX
      BD=-1000.0
      DO 250 JJ=1,10
      SEGSKP(JJ)=0.
      DISERR(JJ)=0.
      VARDIS(JJ)=0.
  250 CONTINUE
      CHNSUM=0.0
      SUMHX=0.0
      SUMHY=0.0
      SUMHZ=0.0
      SUMX=0.0
      SUMY=0.0
      SUMZ=0.0
      DO 305 I=1,NCHN
      IF(RSEGM(I).NE.FLOAT(J))GOTO 305
      IF(X(I).EQ.BD.AND.Y(I).EQ.BD.AND.Z(I).EQ.BD)CHNSKP(I)=1
      IF(CHNSKP(I).EQ.1) GOTO 305
      CHNSUM=CHNSUM+1
      SUMHX=SUMHX+XARAW(I)
      SUMHY=SUMHY+YARAW(I)
      SUMHZ=SUMHZ+ZARAW(I)
      SUMX=SUMX+XD(I)
      SUMY=SUMY+YD(I)
      SUMZ=SUMZ+ZD(I)
```

```
 305    CONTINUE
        IF(CHNSUM.LT.3) SEGSKP(J)=1
        IF(SEGSKP(J).EQ.1) GOTO 308
        RMHATX(J)=SUMHX/CHNSUM
        RMHATY(J)=SUMHY/CHNSUM
        RMHATZ(J)=SUMHZ/CHNSUM
        XI(J)=SUMX/CHNSUM
        YI(J)=SUMY/CHNSUM
        ZI(J)=SUMZ/CHNSUM
        GOTO 310
 308    XI(J)=0.0
        YI(J)=0.0
        ZI(J)=0.0
        RMHATX(J)=0.0
        RMHATY(J)=0.0
        RMHATZ(J)=0.0
 310    CONTINUE
        DO 350 J=1,ISEGMX
        IF(SEGSKP(J).EQ.1) GOTO 350
        DO 340 I=1,NCHN
        IF(RSEGM(I).NE.FLOAT(J)) GOTO 340
        IF(CHNSKP(I).EQ.1) GOTO 340
        XH(I)=XARAW(I)-RMHATX(J)
        YH(I)=YARAW(I)-RMHATY(J)
        ZH(I)=ZARAW(I)-RMHATZ(J)
        X(I)=XD(I)-XI(J)
        Y(I)=YD(I)-YI(J)
        Z(I)=ZD(I)-ZI(J)
 340    CONTINUE
 350    CONTINUE
        DO 360 I=1,NCHN
        IF(CHNSKP(I).EQ.1)GOTO 355
        RMPQH=SQRT(XH(I)**2+YH(I)**2+ZH(I)**2)
        RMPQ=SQRT(X(I)**2+Y(I)**2+Z(I)**2)
        IF(RMPQ.EQ.0.0) GOTO 355
        IF(RMPQH.EQ.0.0) GOTO 355
        XH(I)=XH(I)/RMPQH
        YH(I)=YH(I)/RMPQH
        ZH(I)=ZH(I)/RMPQH
        X(I)=X(I)/RMPQ
        Y(I)=Y(I)/RMPQ
        Z(I)=Z(I)/RMPQ
        GOTO 360
 355    X(I)=0.0
        Y(I)=0.0
        Z(I)=0.0
        XH(I)=0.0
        YH(I)=0.0
        ZH(I)=0.0
 360    CONTINUE
C
C  Use the Schut algorithm to statistically compute the Rodrigues
C  vector which gives the best least squares adjustment fit for those
C  data points which have not been eliminated by the ERRMAX criterion.
C
        DO 300 J=1,ISEGMX
        IF(SEGSKP(J).NE.1.)GO TO 200
        XI(J)=BD
        YI(J)=BD
        ZI(J)=BD
```

```
      AX(J)=BD
      AY(J)=BD
      AZ(J)=BD
      DISERR(J)=BD
      DO 600 M=1,3
      DO 700 N=1,3
      SEGROT(J,M,N)=BD
 700  CONTINUE
 600  CONTINUE
      GO TO 300
 200  CHNSUM=0.0
      DO 201 I=1,4
      DO 201 II=1,4
              S(I,II)=0.0
 201  CONTINUE
      G1=0.0
      G2=0.0
      G3=0.0
      G4=0.0
      DO 190 I=1,NCHN
      IF(RSEGM(I).NE.FLOAT(J)) GO TO 190
      IF(CHNSKP(I).EQ.1) GOTO 190
C
C  Contributions to the normal equations.
C
      CHNSUM=CHNSUM+1.
      G4=G4 + (XH(I)+X(I))**2 + (YH(I)+Y(I))**2 +
     *(ZH(I)+Z(I))**2
      S(1,1)=S(1,1) + XH(I)*X(I)
      S(2,2)=S(2,2) + YH(I)*Y(I)
      S(3,3)=S(3,3) + ZH(I)*Z(I)
      S(1,4)=S(1,4) + ZH(I)*Y(I)
      S(2,4)=S(2,4) + XH(I)*Z(I)
      S(3,4)=S(3,4)+ YH(I)*X(I)
      G1=G1 + Z(I)*YH(I)
      G2=G2 + X(I)*ZH(I)
      G3=G3 + Y(I)*XH(I)
 190  CONTINUE
C
C  Complete the normal equations.
C
      S(4,4)=G4 - 4.*(S(1,1)+S(2,2)+S(3,3))
      S(1,1)=G4 - 4.*S(1,1)
      S(2,2)=G4 - 4.*S(2,2)
      S(3,3)=G4 - 4.*S(3,3)
      S(1,2)=-2.*(G3+S(3,4))
      S(1,3)=-2.*(G2+S(2,4))
      S(2,3)=-2.*(G1+S(1,4))
      S(1,4)=2.*(G1-S(1,4))
      S(2,4)=2.*(G2-S(2,4))
      S(3,4)=2.*(G3-S(3,4))
C
      S(2,1)=S(1,2)
      S(3,1)=S(1,3)
      S(4,1)=S(1,4)
      S(3,2)=S(2,3)
      S(4,2)=S(2,4)
      S(4,3)=S(3,4)
C
C  Solve for the four parameters in the Schut algorithm (A,B,C,D)
```

```
C
      CALL T3SGSC
C
C  Solve for the equivalent orientational change parameters.
C
C  Compute the rotational matrix, SEGROT(J,3,3)
C  from the Schut parameters.
C
      AA=A*A
      BB=B*B
      CC=C*C
      DD=D*D
      AB2=2.*A*B
      AC2=2.*A*C
      AD2=2.*A*D
      BC2=2.*B*C
      BD2=2.*B*D
      CD2=2.*C*D
      ABC=AA+BB+CC
      ABCD=ABC+DD
      SEGROT(J,1,1)=(DD+AA-BB-CC)/ABCD
      SEGROT(J,2,2)=(DD-AA+BB-CC)/ABCD
      SEGROT(J,3,3)=(DD-AA-BB+CC)/ABCD
      SEGROT(J,1,2)=(AB2-CD2)/ABCD
      SEGROT(J,2,1)=(AB2+CD2)/ABCD
      SEGROT(J,1,3)=(AC2+BD2)/ABCD
      SEGROT(J,3,1)=(AC2-BD2)/ABCD
      SEGROT(J,2,3)=(BC2-AD2)/ABCD
      SEGROT(J,3,2)=(BC2+AD2)/ABCD
C
C  Calculate rotation vector
C
      IF(ABC)260,260,270
 260  AX(J)=0.
      AY(J)=0.
      AZ(J)=0.
      GOTO 290
 270  SABC=SQRT(ABC)
      IF(D.LE.0.)GOTO 275
      AX(J)=-A/SABC
      AY(J)=-B/SABC
      AZ(J)=-C/SABC
      GOTO 280
 275  AX(J)=A/SABC
      AY(J)=B/SABC
      AZ(J)=C/SABC
 280  COSTH=(DD-ABC)/(ABCD)
      THETA=ACOS(COSTH)*180./3.141593
      AX(J)=AX(J)*THETA
      AY(J)=AY(J)*THETA
      AZ(J)=AZ(J)*THETA

C  Find the position of the origin of the body coordinate
C  system based on the position of Q point coordinates.

C***************************************
C  MODIFIED 21-NOV-81 TO USE THE TRANSPOSE OF SEGROT(J,M,N) TO CALCULATE
C  THE CORRECTION VECTOR FROM THE AVERAGE LOCATION TO THE ORIGIN
C  OF THE ARBITRARY BODY COORDINATE SYSTEM.  (EKA)
C  This modification entailed reversing the indicies M and N in SEGROT(J,M,N)
```

```
C******************************************

  290   ROTX=SEGROT(J,1,1)*RMHATX(J)
     *+SEGROT(J,2,1)*RMHATY(J)+SEGROT(J,3,1)*RMHATZ(J)
        XI(J)=XI(J)-ROTX
        ROTY=SEGROT(J,1,2)*RMHATX(J)
     *+SEGROT(J,2,2)*RMHATY(J)+SEGROT(J,3,2)*RMHATZ(J)
        YI(J)=YI(J)-ROTY
        ROTZ=SEGROT(J,1,3)*RMHATX(J)
     *+SEGROT(J,2,3)*RMHATY(J)+SEGROT(J,3,3)*RMHATZ(J)
        ZI(J)=ZI(J)-ROTZ

C  Calculate the mean of the DISTANCE between data and calc.
C  also the varaince and record the DISTANCE error for each LED

        DO 240 I=1,NCHN
        IF(RSEGM(I).NE.FLOAT(J)) GO TO 240
        IF(X(I).EQ.0.AND.Y(I).EQ.0.AND.Z(I).EQ.0) GOTO 240
        XCALC=XH(I)*SEGROT(J,1,1)+YH(I)*SEGROT(J,1,2)+ZH(I)*SEGROT(J,1,3)
        YCALC=XH(I)*SEGROT(J,2,1)+YH(I)*SEGROT(J,2,2)+ZH(I)*SEGROT(J,2,3)
        ZCALC=XH(I)*SEGROT(J,3,1)+YH(I)*SEGROT(J,3,2)+ZH(I)*SEGROT(J,3,3)

C  Calculate the DISTANCE ERROR.

        DISPRD=(XCALC-X(I))**2 +( YCALC-Y(I))**2
     *           +( ZCALC-Z(I))**2
        VARDIS(J)=VARDIS(J)+DISPRD
        ERRDIS(I)=SQRT(DISPRD)
        DISERR(J)=DISERR(J)+ERRDIS(I)
  240   CONTINUE

C  CALCULATE MEAN ANGLE ERROR AND STANDARD DEVIATION

        DISERR(J)=(DISERR(J)/CHNSUM)
        VARDIS(J)=(VARDIS(J)-CHNSUM*DISERR(J)**2.)/
     *                           (CHNSUM-1)
        VARDIS(J)=SQRT(VARDIS(J))        !STANDARD DEVIATIONS
        WRITE(5,1632)J,DISERR(J),VARDIS(J)
 1632   FORMAT(/,' Segment number ',I2,/,
     *          ' Mean DISTANCE error ',F10.6,
     *          ' DISTANCE error standard deviation ',F10.6)

C  ELLIMINATE DATA MORE THAN A STANDARD DEVIATION FROM MEAN

        IF(IRUN.GE.1)GOTO 677
        DO 277 I=1,NCHN

                WRITE(5,4321)I,ERRDIS(I)
 4321   FORMAT(' LED #',I2,' ERRDIS=',F10.6)
                TEST=ABS(ERRDIS(I))

                IF(TEST.LT.RMUL*VARDIS(J))GOTO 277
                CHNSKP(I)=1
                WRITE(5,1227)J,I
 1227   FORMAT(' SEGMENT ',I2,' LED ',I2,' BEING ELLIMINATED')
                PRINT 1227,J,I
 277    CONTINUE


 677    CONTINUE
```

```
 300     CONTINUE
         IRUN=IRUN+1
         IF(IRUN.EQ.1)GOTO 1997   !SECOND RUN
 30      RETURN
         END
C*****************************************************************************
         SUBROUTINE T3SGSC
C
C
C        T3SGSC  --      TRACK III SUBROUTINE for GAUSSIAN SCHUT CALCULATION
C
C    T3SGSC does Gaussian Elimination with row and column pivoting on a 4 by
C    4 matrix.  The solution is found by setting the last variable to 1
C    and back-substituting to find the ratios between the variables.
C    The array and the solutions are passed through common.  The input array
C    is not saved.  Written 7/30/81 by Dan Ottenheimer.
C
         COMMON /GAUDAT/ S(4,4),ANS(4)
         DIMENSION IROW(4)
         DO 10 I=1,4
               IROW(I)=I
 10      CONTINUE
C
C  Loop through first two eliminations, selecting pivot element
C
         DO 100 ID=1,2
C
C  Find row number of largest diagonal element in array; pivot row & col
C
            IP=IROW(ID)
            DO 20 IR=ID+1,4
               J=IROW(IR)
               IF(ABS(S(J,J)).LE.ABS(S(IP,IP)))GOTO 20
               IROW(IR)=IP
               IP=J
 20         CONTINUE
C
C  Normalize pivot row
C
            IROW(ID)=IP
            FAC=1./S(IP,IP)
            DO 40 ICC=ID+1,4
               IC=IROW(ICC)
               S(IP,IC)=S(IP,IC)*FAC
 40         CONTINUE

C
C  Update S array, eliminating remaining rows
C
 45         DO 60 IR=ID+1,4
               J=IROW(IR)
               DO 50 ICC=ID+1,4
                  IC=IROW(ICC)
                  S(J,IC)=S(J,IC)-S(IP,IC)*S(J,IP)
 50            CONTINUE
 60         CONTINUE
 100     CONTINUE
C
C  Choose between last two rows for third pivot row
```

```
C
      IP=IROW(3)
      J=IROW(4)
      IF(ABS(S(J,J)).LE.ABS(S(IP,IP)))GOTO 120
      IROW(4)=IP
      IROW(3)=J
      IP=J
      J=IROW(4)
C
C  Normalize third pivot row and set fourth answer to 1.
C
 120  S(IP,J)=S(IP,J)/S(IP,IP)
      ANS(J)=1.
C
C  Back substitution:
C
      DO 150 IR=3,1,-1
          J=IROW(IR)
          FAC=0.
          DO 140 ICC=IR+1,4
              IC=IROW(ICC)
              FAC=FAC-S(J,IC)*ANS(IC)
 140      CONTINUE
          ANS(J)=FAC
 150  CONTINUE
C
      RETURN
      END
```

Appendix E

Segmant Parameter Estimation

This Appendix contains program CRGEOM.FOR which estimates the array to segment CG vectors for the body segment of the wholebody posture model. Program DINERT.FOR estimates the segment masses and inertial properties. Program CGPROG.FOR estimate the 3-D location of each segment center of mass and the combined center of mass based on the output of CRGEOM and DINERT and wholebody kinematic data. Programs CRGEOM and DINERT are base on similar program in the MIT NEWTON software package which perform similar functions for the lower limb segments only.

```
C       CRGEOM              PAT RILEY       APR-86  BILATERAL WHOLE BODY
C                           PAT RILEY       FEB-87  MODIFIED TO INCLUDE
C                                                   PARAMETERS FOR THE UPPER
C                                                   BODY.
C       THIS PROGRAM READS IN A .3DS DATA FILE AND CALCULATES AND STORES
C       THE GEOMETRY FILE FOR DYNAMIC CALCULATIONS. (USED IN PROGRAMS
C       "DINERT" AND "FORCE4".) CG LOCATIONS BASED ON "BODY SEGMENT MASS,
C       RADIUS AND RADIUS OF GYRATION PROPORTIONS OF CHILDREN", ROBERT K.
C       JENSEN, J. BIOMECHANICS VOL 19, NO. 5. PP 359-368,1986.
C
C
C   COMPILE WITH: FORT/NOI4 CRGEOM
C   BUILD WITH: LINK CRGEOM,TRACK.OLB/LIB

        CHARACTER*1 LEG,TURN,IANS
        CHARACTER*16 NAMFIL,FPNMFL
        DIMENSION X(7,10),Y(7,10),Z(7,10),RESULT(3,3,7),CENTER(3,7)
        DIMENSION RP(7,3),RD(7,3),RCG(7,3)
        DIMENSION FPROT(3,3)
        DIMENSION TEMP1(3),TEMP2(3),TEMP3(3),RTEMP(3,3)

        DATA NAMFIL/'TK4:NAMFIL.3DS;0'/

        PARAMETER(LUI = 5,LUO = 6)
        IANS = 'U'

        WRITE(LUO,51) "7
51      FORMAT('$Enter the name of the .3DS file to use ',A1)
        READ(LUI,52) NAMFIL(5:10)
52      FORMAT(1A6)
        ISIDE = 1
53      WRITE(LUO,55)
55      FORMAT('$Enter the subject age in years and months: [2I] ')
        READ(LUI,57)IYEAR,IMONTH
57      FORMAT(2I3)
        IF(IYEAR.EQ.0)GOTO 53
        AGE =  FLOAT(IYEAR) + (FLOAT(IMONTH)/12.)
        IF(AGE.GE.15.)AGE = 15. !DO NOT EXTRAPOLATE BEYOND DATA RANGE
43      IF(ISIDE.EQ.1)IANS='R'
        IF(ISIDE.EQ.2)IANS='L'
                IF(IANS.EQ.'L')NAMFIL(11:14)='.3DL'
                IF(IANS.EQ.'R')NAMFIL(11:14)='.3DR'

70      OPEN(UNIT=13,NAME=NAMFIL,TYPE='OLD',ACCESS='DIRECT',
     *          READONLY)

        if(ians.eq.'R')TYPE*,'Reading right side segment file'
        if(ians.eq.'L')Type*,'Reading left side segment file'
        IF(IANS.EQ.'R') OPEN(UNIT=14,NAME='TK3:F4NMFL.DTR',TYPE='NEW')
        IF(IANS.EQ.'L') OPEN(UNIT=14,NAME='TK3:F4NMFL.DTL',TYPE='NEW')

                NAMFIL(11:12)='.D'
                IF(IANS.EQ.'U')NAMFIL(13:14)='AT'
                IF(IANS.EQ.'L')NAMFIL(13:14)='TL'
                IF(IANS.EQ.'R')NAMFIL(13:14)='TR'
                WRITE(14,101) NAMFIL

                NAMFIL(11:12)='.3'
                IF(IANS.EQ.'U')NAMFIL(13:14)='DP'
                IF(IANS.EQ.'L')NAMFIL(13:14)='DL'
```

```
                IF(IANS.EQ.'R')NAMFIL(13:14)='DR'
                WRITE(14,101) NAMFIL
101             FORMAT(1A16)

        CALL CLOSE(14)

        if(ians.eq.'R')IDFP=1
        if(ians.eq.'L')IDFP=2

        WRITE(LUO,200)IDFP, "7
200     FORMAT('$Force plate [D=',I2,'] ',A1)
        READ(LUI,201) IFP
201     FORMAT(I6)
        IF (IFP.EQ.0) IFP=IDFP

        IREC = 1
        READ(13'IREC) MSCAL,LEG,TURN
        IF(LEG.EQ.'R')NSEGM=7
        IF(LEG.EQ.'L')NSEGM=6
        NJNT=NSEGM-1
        IREC = IREC + 1

        DO 1 J=1,NSEGM
                READ(13'IREC) (X(J,I),I=1,10)
                IREC = IREC + 1
                READ(13'IREC) (Y(J,I),I=1,10)
                IREC = IREC + 1
                READ(13'IREC) (Z(J,I),I=1,10)
                IREC = IREC + 1
1       CONTINUE


        DO 2 J=1,NSEGM
                READ(13'IREC) (CENTER(I,J),I=1,3)
                IREC = IREC + 1
2       CONTINUE


        DO 3 K=1,NSEGM
                READ(13'IREC) (   (RESULT(I,J,K),I=1,3),  J=1,3)
                IREC = IREC + 1
3       CONTINUE

        CALL CLOSE(13)

C       READ IN FORCE PLATE LOCATIONS

        NAMFIL(11:12)='.D'
                IF(IANS.EQ.'U')NAMFIL(13:14)='AT'
                IF(IANS.EQ.'L')NAMFIL(13:14)='TL'
                IF(IANS.EQ.'R')NAMFIL(13:14)='TR'


        if(ians.eq.'R')TYPE*,'Reading right side force plate location'
        if(ians.eq.'L')Type*,'Reading left side force plate location'
        OPEN(UNIT=17,NAME=NAMFIL,TYPE='OLD',ACCESS='DIRECT',
     *          READONLY)
        READ(17'4) FPNMFL(1:16)
        CALL CLOSE(17)
```

```
      OPEN(UNIT=17,NAME=FPNMFL,TYPE='OLD',READONLY)

      READ(17,1230) ITEMP6
      READ(17,1231) FPXX,FPYY,FPZZ
      READ(17,1231) FPROT(1,1),FPROT(1,2),FPROT(1,3)
      READ(17,1231) FPROT(2,1),FPROT(2,2),FPROT(2,3)
      READ(17,1231) FPROT(3,1),FPROT(3,2),FPROT(3,3)
      IF (IFP.EQ.1) GOTO 1232
      READ(17,1231) FPXX,FPYY,FPZZ
      READ(17,1231) FPROT(1,1),FPROT(1,2),FPROT(1,3)
      READ(17,1231) FPROT(2,1),FPROT(2,2),FPROT(2,3)
      READ(17,1231) FPROT(3,1),FPROT(3,2),FPROT(3,3)
1230  FORMAT(A2)
1231  FORMAT(3F15.7)
      CALL CLOSE(17)

1232  IF(IANS.EQ.'R')OPEN(UNIT=16,NAME='TK3:FPDATA.DTR',TYPE='NEW')
      IF(IANS.EQ.'L')OPEN(UNIT=16,NAME='TK3:FPDATA.DTL',TYPE='NEW')
              WRITE(16,202) FPXX,FPYY,FPZZ
              WRITE(16,202) FPROT(1,1),FPROT(1,2),FPROT(1,3)
              WRITE(16,202) FPROT(2,1),FPROT(2,2),FPROT(2,3)
              WRITE(16,202) FPROT(3,1),FPROT(3,2),FPROT(3,3)
              WRITE(16,203) IFP
202           FORMAT(3F14.7)
203           FORMAT(I6)

      CALL CLOSE(16)

C     CALCULATIONS FOR THE FOOT

      if(ians.eq.'R')TYPE*,'Writing Right side geometry file'
      if(ians.eq.'L')Type*,'Writing left side geometry file'

      FOOT = ABS(X(1,1)-X(1,4))/FLOAT(MSCAL)
      hfoot = abs(y(1,1)-y(1,2))/float(mscal)
      IFOOT = 1
      IF (LEG.NE.'R') IFOOT = -1
      FRAC    = 0.4351-.00186*AGE

C  PROXIMAL TO CG VECTOR IN LIMB COORDINATES
      TEMP1(1) = -FRAC * (0.8*FOOT) * IFOOT
      TEMP1(2) = -0.50 * hFOOT
      TEMP1(3) = 0.0
C  PROXIMAL TO CG VECTOR IN ARRAY COORDINATES
      CALL GMTRA(RESULT(1,1,1),RTEMP,3,3)
      CALL GMPRD(RTEMP,TEMP1,TEMP2,3,3,1)
C ARRAY TO CG VECTOR IN ARRAY COORDINATES
      RCG(1,1) = TEMP2(1) + CENTER(1,1)
      RCG(1,2) = TEMP2(2) + CENTER(2,1)
      RCG(1,3) = TEMP2(3) + CENTER(3,1)
C  PROXIMAL TO DISTAL VECTOR IN LIMB COORDINATES
      TEMP1(1) = -0.80 * FOOT * IFOOT
      TEMP1(2) = -0.50 * hFOOT
      TEMP1(3) = 0.0
C  PROXIMAL TO DISTAL VECTOR IN ARRAY COORDINATES
      CALL GMPRD(RTEMP,TEMP1,TEMP2,3,3,1)
C  CG TO DISTAL VECTOR IN ARRAY COORDINATES
      RD(1,1) = TEMP2(1) + CENTER(1,1) - RCG(1,1)
      RD(1,2) = TEMP2(2) + CENTER(2,1) - RCG(1,2)
      RD(1,3) = TEMP2(3) + CENTER(3,1) - RCG(1,3)
```

```
C  CG TO PROXIMAL VECTO IN ARRAY COORDINATES
        RP(1,1) = CENTER(1,1) - RCG(1,1)
        RP(1,2) = CENTER(2,1) - RCG(1,2)
        RP(1,3) = CENTER(3,1) - RCG(1,3)


C CALCULATIONS FOR THE SHANK
C SHANK LENGTH FRO SEG FILE DATA
        SHANK = ABS(Y(2,1)-Y(2,2))/FLOAT(MSCAL) !LENGTH OF SHANK
C DISTAL TO PROXIMAL VECTOR IN LIMB COORDINATES
        FRAC = 1.0 - (-.003 * AGE + .4526)
        TEMP1(1) = 0.0
        TEMP1(2) = SHANK
        TEMP1(3) = 0.0
C DISTAL TO PROXIMAL IN ARRAY COORDINATES
        CALL GMTRA(RESULT(1,1,2),RTEMP,3,3)
        CALL GMPRD(RTEMP,TEMP1,TEMP2,3,3,1)
C ARRAY TO CG VECTOR IN ARRAY COORDINATES
        RCG(2,1) = TEMP2(1)*FRAC + CENTER(1,2)
        RCG(2,2) = TEMP2(2)*FRAC + CENTER(2,2)
        RCG(2,3) = TEMP2(3)*FRAC + CENTER(3,2)
C CG TO DISTAL VECTOR IN ARRAY COORDINATES
        RD(2,1) = -TEMP2(1)*FRAC
        RD(2,2) = -TEMP2(2)*FRAC
        RD(2,3) = -TEMP2(3)*FRAC
C CG TO PROXIMAL VECTOR IN ARRAY COORDINATES
        RP(2,1) = TEMP2(1) + CENTER(1,2) - RCG(2,1)
        RP(2,2) = TEMP2(2) + CENTER(2,2) - RCG(2,2)
        RP(2,3) = TEMP2(3) + CENTER(3,2) - RCG(2,3)

C CALCULATIONS FOR THE THIGH
C THIGH LENGTH FROM SEGMENT FILE
        THIGH = ABS(Y(3,1)-Y(3,2))/FLOAT(MSCAL) !THIGH LENGTH
C DISTAL TO PROXIMAL VECTOR IN LIMB COORDINATES
        FRAC = 1.0 - (.4758 - .00115 * AGE)
        TEMP1(1) = 0.0                          !THIGH VECTOR
        TEMP1(2) = THIGH
        TEMP1(3) = 0.0
C DISTAL TO PROXIMAL VECTOR IN ARRAY COORDINATES
        CALL GMTRA(RESULT(1,1,3),RTEMP,3,3)     !ARRAY COORDINATED
        CALL GMPRD(RTEMP,TEMP1,TEMP2,3,3,1)     !THIGH VECTOR
C ARRAY TO CG VECTOR IN ARRAY COORDINATES
        RCG(3,1) = TEMP2(1)*FRAC + CENTER(1,3)  !DISTAL END OF
        RCG(3,2) = TEMP2(2)*FRAC + CENTER(2,3)  !THIGH
        RCG(3,3) = TEMP2(3)*FRAC + CENTER(3,3)
C CG TO DISTAL VECTOR IN ARRAY COORDINATES
        RD(3,1) = -TEMP2(1)*FRAC
        RD(3,2) = -TEMP2(2)*FRAC
        RD(3,3) = -TEMP2(3)*FRAC
C CG TO PROXIMAL VECTOR IN ARRAY COORDINATES
        RP(3,1) = TEMP2(1) + CENTER(1,3) - RCG(3,1)             !PROXIMAL TO
        RP(3,2) = TEMP2(2) + CENTER(2,3) - RCG(3,2)             !CG VECTOR
        RP(3,3) = TEMP2(3) + CENTER(3,3) - RCG(3,3)

C CALCULATIONS FOR THE PELVIS
C PELVIS SIZE FROM SEGMENT FILE DATA
        PELVIS = ABS(Y(4,1)-Y(4,2))/FLOAT(MSCAL)!HGT OF PELVIS
        PELWDT = ABS(Z(4,1)-Z(4,5))/FLOAT(MSCAL)!WIDTH OF PELVIS
C DISTAL CENTER TO PROXIMAL VECTOR IN LIMB COORDINATES
C Jenson combines trunk and pelvis.
```

```
      FRAC = .0018 * AGE + .5087      !COMBINED PELVIS TRUNK USED FOR EACH
      TEMP1(1) = 0.0                         !PELVIS AXIS VECTOR
      TEMP1(2) = PELVIS
      TEMP1(3) = 0.0
C DISTAL CENTER TO PROXIMAL VECTOR IN ARRAY COORDINATES
      CALL GMTRA(RESULT(1,1,4),RTEMP,3,3)    !PELVIS AXIS VECTOR
      CALL GMPRD(RTEMP,TEMP1,TEMP2,3,3,1)    !IN ARRAY COORDINATES
C DISTAL TO DISTAL CENTER VECTOR IN LIMB COORDINATES
      TEMP1(1) = 0.0                         !DISTAL END OF PELVIS AXIS
      TEMP1(2) = 0.0                         !DISPLACED IN FROM HIP
      TEMP1(3) = 0.25 * PELWDT               !JOINT
C DISTAL TO DISTAL CENTER VECTOR IN ARRAY COORDINATES
      CALL GMPRD(RTEMP,TEMP1,TEMP3,3,3,1)    !IN ARRAY COORDINATES
C ARRAY TO DISTAL CENTER VECTOR IN ARRAY COORDINATES
      TEMP3(1) = TEMP3(1) + CENTER(1,4)      !DISTAL END OF PELVIS
      TEMP3(2) = TEMP3(2) + CENTER(2,4)      !AXIS
      TEMP3(3) = TEMP3(3) + CENTER(3,4)
C ARRAY TO CG VECTOR IN ARRAY COORDINATES
      RCG(4,1) = TEMP2(1)*FRAC + TEMP3(1)
      RCG(4,2) = TEMP2(2)*FRAC + TEMP3(2)
      RCG(4,3) = TEMP2(3)*FRAC + TEMP3(3)
C CG TO DISTAL VECTOR IN ARRAY COORDINATES
      RD(4,1) = CENTER(1,4) - RCG(4,1)       !HIP JOINT CENTER TO
      RD(4,2) = CENTER(2,4) - RCG(4,2)       !CG VECTOR
      RD(4,3) = CENTER(3,4) - RCG(4,3)
C CG TO PROXIMAL VECTOR IN ARRAY COORDINATES
      RP(4,1) = TEMP2(1) + CENTER(1,4) - RCG(4,1)     !PROXIMAL TO CG
      RP(4,2) = TEMP2(2) + CENTER(2,4) - RCG(4,2)     !VECTOR
      RP(4,3) = TEMP2(3) + CENTER(3,4) - RCG(4,3)

C CALCULATIONS FOR THE TRUNK
C TRUNK SIZE FROM SEGMENT FILE
      TRUNK = ABS(Y(5,1)-Y(5,3))/FLOAT(MSCAL)!TRUNK HEIGHT
C DISTAL TO PROXIMAL VECTOR IN LIMB COORDINATES
C Jenson combines trunk and pelvis
      FRAC = .0018 * AGE +.5087       !COMBINED PELVIS TRUNK USED FOR
      TEMP1(1) = 0.0
      TEMP1(2) = TRUNK                       !TRUNK AXIS VECTOR
      TEMP1(3) = 0.0
C DISTAL TO PROXIMAL VECTOR IN ARRAY COORDINATES
      CALL GMTRA(RESULT(1,1,5),RTEMP,3,3)
      CALL GMPRD(RTEMP,TEMP1,TEMP2,3,3,1)
C ARRAY TO CG VECTOR IN ARRA COORDINATES
      RCG(5,1) = TEMP2(1)*FRAC + CENTER(1,5)
      RCG(5,2) = TEMP2(2)*FRAC + CENTER(2,5)
      RCG(5,3) = TEMP2(3)*FRAC + CENTER(3,5)
C CG TO DISTAL VECTOR IN ARRAY COORDINATES
      RD(5,1) = -TEMP2(1)
      RD(5,2) = -TEMP2(2)
      RD(5,3) = -TEMP2(3)
C CG TO PROXIMAL VECTOR IN ARRAY COORDINATES
      RP(5,1) = TEMP2(1) + CENTER(1,5) - RCG(5,1)
      RP(5,2) = TEMP2(2) + CENTER(2,5) - RCG(5,2)
      RP(5,3) = TEMP2(3) + CENTER(3,5) - RCG(5,3)

C CALCULATION FOR THE ARM
C ARM LENGTH FROM SEGMENT FILE
      ARM = (Y(6,1)-Y(6,2))/FLOAT(MSCAL)
      ARMW= ABS(Z(6,1)-Z(6,5))/FLOAT(MSCAL)
C PROXIMAL TO CG VECTOR IN LIMB COORDINATES
```

```
      FRAC = 1.0-0.56
      TEMP1(1) = 0.0
      TEMP1(2) = ARM * FRAC
      TEMP1(3) = -0.5 * ARMW
C PROXIMAL TO CG VECTOR IN ARRAY COORDINATES
      CALL GMTRA(RESULT(1,1,6),RTEMP,3,3)
      CALL GMPRD(RTEMP,TEMP1,TEMP2,3,3,1)
C ARRAY TO CG VECTOR IN ARRAY COORDINATES
      RCG(6,1) = TEMP2(1) + CENTER(1,6)
      RCG(6,2) = TEMP2(2) + CENTER(2,6)
      RCG(6,3) = TEMP2(3) + CENTER(3,6)
C PROXIMAL TO DISTAL VECTOR IN LIMB COORDINATES
      TEMP1(1) = 0.0
      TEMP1(2) = ARM
      TEMP1(3) = 0.0
C PROXIMAL TO DISTAL VECTOR IN ARRAY COORDINATES
      CALL GMPRD(RTEMP,TEMP1,TEMP2,3,3,1)
C CG TO DISTAL VECTOR IN ARRAY COORDINATES
      RD(6,1) = TEMP2(1) + CENTER(1,6) - RCG(6,1)
      RD(6,2) = TEMP2(2) + CENTER(2,6) - RCG(6,2)
      RD(6,3) = TEMP2(3) + CENTER(3,6) - RCG(6,3)
C CG TO PROXIMAL VECTOR IN ARRAY COORDINATES
      RP(6,1) = CENTER(1,6) - RCG(6,1)
      RP(6,2) = CENTER(2,6) - RCG(6,2)
      RP(6,3) = CENTER(3,6) - RCG(6,3)

C CALCULATIONS FOR THE HEAD
C HEAD SIZE FROM SEGMENT FILE
      HEAD = ABS(Y(7,1)-Y(7,2))/FLOAT(MSCAL)
C PROXIMAL TO CG VECTOR IN LIMB COORDINATES
      FRAC = .0025 * AGE + .4833
      TEMP1(1) = -0.20 * HEAD !SOMEWHAT ARBITRARY
      TEMP1(2) = FRAC * HEAD
      TEMP1(3) = 0.0
C PROXIMAL TO CG VECTOR IN ARRAY COORDINATES
      CALL GMTRA(RESULT(1,1,7),RTEMP,3,3)
      CALL GMPRD(RTEMP,TEMP1,TEMP2,3,3,1)
C ARRAY TO CG VECTOR IN ARRAY COORDINATES
      RCG(7,1) = TEMP2(1) + CENTER(1,7)
      RCG(7,2) = TEMP2(2) + CENTER(2,7)
      RCG(7,3) = TEMP2(3) + CENTER(3,7)
C CG TO PROXIMAL VECTOR IN ARRAY COORDINATES
      RP(7,1) =  CENTER(1,7) - RCG(7,1)
      RP(7,2) =  CENTER(2,7) - RCG(7,2)
      RP(7,3) =  CENTER(3,7) - RCG(7,3)
C PROXIMAL TO DISTAL VECTOR IN LIMB COORDINATES
      TEMP1(1) = 0.0
      TEMP1(2) = HEAD
      TEMP1(3) = 0.0
C PROXIMAL TO DISTAL VECTOR IN ARRAY COORDINATES
      CALL GMPRD(RTEMP,TEMP1,TEMP2,3,3,1)
C CG TO DISTAL VECTOR IN ARRAY COORDINATES
      RD(7,1) = CENTER(1,7) + TEMP2(1) - RCG(7,1)
      RD(7,2) = CENTER(2,7) + TEMP2(2) - RCG(7,2)
      RD(7,3) = CENTER(3,7) + TEMP2(3) - RCG(7,3)

C WRITE OUT THE RESULTS

      IRUN=1
```

```
      IF(IANS.EQ.'R') OPEN(UNIT=12,NAME='TK3:GEOMETRY.DTR',TYPE='NEW')
      IF(IANS.EQ.'L') OPEN(UNIT=12,NAME='TK3:GEOMETRY.DTL',TYPE='NEW')

C LOWER LIMB FIRST FORMAT COMPATIBLE WITH LOWER LIMB ONLY FILES
5432  WRITE(12,100) RP(1,1)
      WRITE(12,100) RP(1,2)
      WRITE(12,100) RP(1,3)
      WRITE(12,100) RP(2,1)
      WRITE(12,100) RP(2,2)
      WRITE(12,100) RP(2,3)
      WRITE(12,100) RP(3,1)
      WRITE(12,100) RP(3,2)
      WRITE(12,100) RP(3,3)
C
      WRITE(12,100) RD(1,1)
      WRITE(12,100) RD(1,2)
      WRITE(12,100) RD(1,3)
      WRITE(12,100) RD(2,1)
      WRITE(12,100) RD(2,2)
      WRITE(12,100) RD(2,3)
      WRITE(12,100) RD(3,1)
      WRITE(12,100) RD(3,2)
      WRITE(12,100) RD(3,3)
C
      WRITE(12,100) RCG(1,1)
      WRITE(12,100) RCG(1,2)
      WRITE(12,100) RCG(1,3)
      WRITE(12,100) RCG(2,1)
      WRITE(12,100) RCG(2,2)
      WRITE(12,100) RCG(2,3)
      WRITE(12,100) RCG(3,1)
      WRITE(12,100) RCG(3,2)
      WRITE(12,100) RCG(3,3)

C NOW UPPER BODY

      WRITE(12,100) RP(4,1)
      WRITE(12,100) RP(4,2)
      WRITE(12,100) RP(4,3)
      WRITE(12,100) RP(5,1)
      WRITE(12,100) RP(5,2)
      WRITE(12,100) RP(5,3)
      WRITE(12,100) RP(6,1)
      WRITE(12,100) RP(6,2)
      WRITE(12,100) RP(6,3)
C
      WRITE(12,100) RD(4,1)
      WRITE(12,100) RD(4,2)
      WRITE(12,100) RD(4,3)
      WRITE(12,100) RD(5,1)
      WRITE(12,100) RD(5,2)
      WRITE(12,100) RD(5,3)
      WRITE(12,100) RD(6,1)
      WRITE(12,100) RD(6,2)
      WRITE(12,100) RD(6,3)
C
      WRITE(12,100) RCG(4,1)
      WRITE(12,100) RCG(4,2)
      WRITE(12,100) RCG(4,3)
      WRITE(12,100) RCG(5,1)
```

```
      WRITE(12,100) RCG(5,2)
      WRITE(12,100) RCG(5,3)
      WRITE(12,100) RCG(6,1)
      WRITE(12,100) RCG(6,2)
      WRITE(12,100) RCG(6,3)

      IF (LEG.EQ.'R') THEN
              WRITE(12,100) RP(7,1)
              WRITE(12,100) RP(7,2)
              WRITE(12,100) RP(7,3)
C
              WRITE(12,100) RD(7,1)
              WRITE(12,100) RD(7,2)
              WRITE(12,100) RD(7,3)
C
              WRITE(12,100) RCG(7,1)
              WRITE(12,100) RCG(7,2)
              WRITE(12,100) RCG(7,3)
      END IF

      CALL CLOSE(12)

      IF(IRUN.EQ.1)THEN
       IF(IANS.EQ.'R')OPEN(UNIT=12,NAME='TK4:GEOMETRY.DTR',TYPE='NEW')
       IF(IANS.EQ.'L')OPEN(UNIT=12,NAME='TK4:GEOMETRY.DTL',TYPE='NEW')
              IRUN=IRUN+1
              GOTO 5432
      ENDIF

      if(ians.eq.'R')TYPE*,'Right side complete'
      if(ians.eq.'L')Type*,'Left side complete'

      IF(ISIDE.EQ.1)THEN
              ISIDE=2
              GOTO 43
      ENDIF

100   FORMAT(F14.7)
      CALL EXIT
      END
```

```
      PROGRAM DINERT
C
C This is a program to find segment masses
C and inertia tensor around the center of
C mass of each segment.  Right side data is presumed
C to consist of seven segments; left side data is
C presumed to consist of six segments.
C       The presumed segment ation is:
C               seg#    segment right   left
C               1       foot     x       x
C               2       shank    x       x
C               3       thight   x       x
C               4       pelvis   x       x
C               5       trunk    x       x
C               6       arm      x       x
C               7       head     x
C
C
C Dseg array=density array, Vseg= seg vol.
C SegM = segment mass, SEGL= segment length
C PRCG is BCS vector from proximal joint to CG, as in
C DYJGEO, DISCG is BCS vector from distal joint to CG
C CGBCS is vector from LED array origin to CG.
C GYRN is ratio of radius of gyration to segment
C length (from "Body segment mass, radius, and radius of
C gyration proportions of children,Robert K. Jensen,
C J. Biomechanics Vol 19, No 5, pp359-368,1086.)
C
C       SEGLGY is SEGL*GYRN, which is the radius of
C       gyration.
C
C       SGINRT is the inertia tensor about each
C       segment CG.
C
C
      DIMENSION DSEG(7),VSEG(7),SEGM(7)
      DIMENSION SEGL(7),PRCG(3,7),DISCG(3,7),GYRN(7)
      DIMENSION SEGLGY(7),CGL(7)
      DIMENSION RJOINT(9)
      DIMENSION SGINRT(3,7),CGBCS(3,7),SMRN(7)
      CHARACTER*1 SEX,FEM,MALE
      CHARACTER*2 METR(2),NBRIT(2),NUN(10)
      BYTE IANS

      DATA METR/'CM','KG'/
      DATA NBRIT/'IN','LB'/
      DATA FEM/'F'/
      DATA MALE/'M'/

      PARAMETER(LUO=5,LUI=5,LUP=8)

C     Initialize arrays

      DO 1 I=1,7
      DO 2 J=1,3
                PRCG(J,I)=0.0
                DISCG(J,I)=0.0
                CGBCS(J,I)=0.0
                SGINRT(J,I)=0.0
2         CONTINUE
```

```
                DSEG(I)=0.0
                VSEG(I)=0.0
                SEGM(I)=0.0
                SEGL(I)=0.0
                SEGLGY(I)=0.0
                CGL(I)=0.0
                RJOINT(I)=0.0
1       CONTINUE

C Ask user for HEIGHT and WEIGHT of
C               the subject.
C Change H and W to METRIC if necessary.

C   Also ask subject's sex, and circumferences
C of ankle, knee, upper thigh, trunk, upper arm, and neck.
C These are needed because the segment must be modelled as
C a truncated cone in order to find Iy.

        WRITE(LUO,1001)
1001    FORMAT(' What is the height of the subject?',/,
     *  ' (Please enter in same form as example (height,units))',//,
     *  ' EXAMPLE: 165.0,CM (Use only inches
     * (IN) or centimeters (CM)) ')
        READ(LUI,1002)HGT,NUN(1)
1002    FORMAT(F8.3,A2)

        IF(NUN(1).EQ.NBRIT(1))  HGT=HGT*2.54

110     WRITE(LUO,1003)
1003    FORMAT(/,' Now, enter the subject''s weight or mass.',/,
     *  ' ( Enter as in Example, (weight,units))',//,
     *  ' EXAMPLE: 54.2,KG  ( Use only pounds (LB)
     * or kilograms (KG)) ')
        READ(LUI,1004)W,NUN(2)
1004    FORMAT(F8.3,A2)
        IF(NUN(2).EQ.NBRIT(2))  W=W/2.2

C 111   WRITE(LUO,999)
 999    FORMAT(' What is the sex of the subject?',/,
     * ' (Please enter as F or M)')
C       READ(LUI,998)SEX
 998    FORMAT(A2)

        WRITE(LUO,1101)
1101    FORMAT(/,' Next, enter the appropriate circumferences.',/,
     *  ' ( In inches,(IN), or centimeters,(CM), like: 14.0,IN )')

        WRITE(LUO,1102)
1102    FORMAT($5X,' Ankle: ')
        READ(LUI,1103)RJOINT(2),NUN(3)
1103    FORMAT(F8.3,A2)

        WRITE(LUO,1104)
1104    FORMAT($5X,' Knee: ')
        READ(LUI,1103)RJOINT(3),NUN(4)

        WRITE(LUO,1105)
1105    FORMAT($5X,' Upper thigh: ')
        READ(LUI,1103)RJOINT(4),NUN(5)
```

```
      WRITE(LUO,1106)
1106  FORMAT($5X,' Waist: ')
      READ(LUI,1103)RJOINT(5),NUN(6)

      WRITE(LUO,1116)
1116  FORMAT($5X,' Chest: ')
      READ(LUI,1103)RJOINT(6),NUN(7)

      WRITE(LUO,1107)
1107  FORMAT($5X,' Upper arm: ')
      READ(LUI,1103)RJOINT(7),NUN(8)

      WRITE(LUO,1108)
1108  FORMAT($5X,' Neck: ')
      READ(LUI,1103)RJOINT(8),NUN(9)

      WRITE(LUO,1109)
1109  FORMAT($5X,' Head: ')
      READ(LUI,1103)RJOINT(9),NUN(10)

      DO I=3,10
              II=I-1
              IF(NUN(I).eq.NBRIT(1))then      !convert to cm
                      RJOINT(II)=RJOINT(II)*2.54
              end if
              RJOINT(II)=RJOINT(II)*.01/(2*3.14159)    !convert to radius
      ENDDO                                            !in meters
      rjoint(1) = rjoint(2)   !assume radius of foot constant

53    WRITE(LUO,55)
55    FORMAT('$Enter the subject age in years and months: [2I] ')
      READ(LUI,57)IYEAR,IMONTH
57    FORMAT(2I3)
      IF(IYEAR.EQ.0)GOTO 53
      AGE =  FLOAT(IYEAR) + (FLOAT(IMONTH)/12.)
      IF(AGE.GE.15.)AGE = 15. !DO NOT EXTRAPOLATE BEYOND DATA RANGE

C Get Geometrical information from
C latest GEOMETRY.DAT file.
      iside=1

9998  if(iside.eq.1)THEN
              IANS='R'
              TYPE*,'RIGHT SIDE PARAMETERS'
      ENDIF
      IF(ISIDE.EQ.2)THEN
              IANS='L'
              TYPE*,'LEFT SIDE PARAMETERS'
      ENDIF

      IF(IANS.EQ.'R')OPEN(UNIT=16,NAME='TK3:GEOMETRY.DTR;0',
     *        TYPE='OLD',READONLY)

      IF(IANS.EQ.'L')OPEN(UNIT=16,NAME='TK3:GEOMETRY.DTL;0',
     *        TYPE='OLD',READONLY)

      DO I=1,3
      READ(16,1000)(PRCG(J,I),J=1,3)
      ENDDO
      DO I=1,3
```

```
      READ(16,1000)(DISCG(J,I),J=1,3)
      ENDDO
      DO I=1,3
      READ(16,1000)(CGBCS(J,I),J=1,3)
      ENDDO

      DO I=4,6
      READ(16,1000)(PRCG(J,I),J=1,3)
      ENDDO
      DO I=4,6
      READ(16,1000)(DISCG(J,I),J=1,3)
      ENDDO
      DO I=4,6
      READ(16,1000)(CGBCS(J,I),J=1,3)
      ENDDO

      IF(IANS.EQ.'R')THEN
      READ(16,1000)(PRCG(J,7),J=1,3)
      READ(16,1000)(DISCG(J,7),J=1,3)
      READ(16,1000)(CGBCS(J,7),J=1,3)
      END IF

1000  FORMAT(F14.7/F14.7/F14.7)
      CLOSE(UNIT=16)


C Set the mass ratio and radius of gyration parameters
C Jensen looked at all male subjects, so females assumed
C to be similar.

      SMRN(1)=.0002*AGE + .0187
      SMRN(2)=.0012*AGE + .0381
      SMRN(3)=.0036*AGE + .0663
C Jensen presented data for combined head and trunk
C assumed weight distribution: 15% pelvis, 85%trunk.
      SMRN(4)=.15*(-.0006*AGE + .4246)
      SMRN(5)=.85*(-.0006*AGE + .4246)
      SMRN(6)=.0010*AGE + .0457
      SMRN(7)=-.0114*AGE + .2376

      GYRN(1)=sqrt((-.00203*AGE + .5022)**2
    1         -(-.00186*AGE + .4351)**2)
      GYRN(2)=sqrt((-.00224*AGE + .5307)**2
    1         -(-.003*AGE + .4526)**2)
      GYRN(3)=sqrt((-.00133*AGE + .5536)**2
    1         -(-.00115*AGE + .4758)**2)
      GYRN(4)=sqrt((.00136*AGE + .5872)**2
    1         -(.0018*AGE + .5087)**2)!COMBINED TRUNK PELVIS VALUE
      GYRN(5)=sqrt((.00136*AGE + .5872)**2
    1         -(.0018*AGE + .5087)**2)!USED FOR EACH INDIVIDUALLY
      GYRN(6)=sqrt((.001*AGE + .64)**2
    1         -(.00025*AGE + .56)**2) !UPPER ARM,LOWER ARM, HAND DATA
                                      !COMBINED
      GYRN(7)=sqrt((.0021*AGE + .5748)**2
    1         -(.0025*AGE + .4833)**2)

C FIND THE SEGMENT MASSES

      DO 116 I=1,7
              SEGM(I)=SMRN(I) * W
```

```
116     CONTINUE

C Find scalar segment length and length from
C proximal joint to CG.

        DO 117 I=1,7
        SL=0.0
        SLS=0.0
        CG=0.0
        DO 118 J=1,3
                SL=PRCG(J,I)-DISCG(J,I)
                IF(J.EQ.1)SL=SL/0.8
                IF(J.EQ.4)SL=2.0*DISCG(J,I)
                SLS=SL**2+SLS
                CG=PRCG(J,I)**2+CG
118     CONTINUE
        SEGL(I)=SQRT(SLS)
        CGL(I)=SQRT(CG)
117     CONTINUE

C Find radius of gyration

        DO 120 I=1,7
                SEGLGY(I)=SEGL(I)*GYRN(I)
120     CONTINUE

C Now, find segment principal moments of inertia, about the CG.
c The segments are treated as truncated cones with large radius A1 and
c small radius A2.

        DO 125 I=1,7     !for each segment
                II=I+1
                RMULT=SEGLGY(I)**2
                A1 = RJOINT(II) !proximal and distal joint radius
                A2 = RJOINT(I)  !for most segments

                IF(I.EQ.4)THEN  !pelvis
                        A1 = RJOINT(5)  !waist circumference
                        A2 = 2.*RJOINT(4)       !thigh circumference
                ENDIF

                IF(I.EQ.7)THEN  ! head
                        A1 = RJOINT(9)          !treated as cylinder
                        A1 = RJOINT(9)          !or use rjoint(8) for cone
                ENDIF
                IF(I.EQ.6)THEN  !arm upper circumference
                        A1 = RJOINT(7)          !measured, lower assumed.
                        A2 = RJOINT(7)*0.6
                ENDIF

                YMULT = 0.5*(A1**4 +A1**3*A2 +A1**2*A2**2
     1                  +A1*A2**3 +A2**4)/(A1**2 +A1*A2 +A2**2)

                SGINRT(1,I)=SEGM(I)*RMULT       !x axis moment of inertia
                SGINRT(3,I)=SEGM(I)*RMULT       !z axis moment of inertia
                SGINRT(2,I)=SEGM(I)*YMULT       !y [long]axis moment of
                                                ! inertia
125     CONTINUE

C       Write Results to File and to User
```

```
C               NOTE: For now, FORMAT masses and inertia
C               tensor as F14.7

        IF(IANS.EQ.'R') OPEN(UNIT=12,NAME='TK3:INERTIA.DTR;0',TYPE='NEW')
        IF(IANS.EQ.'L') OPEN(UNIT=12,NAME='TK3:INERTIA.DTL;0',TYPE='NEW')
        SWITCH=SGINRT(1,1)
        SGINRT(1,1)=SGINRT(2,1)
        SGINRT(2,1)=SWITCH
        DO 130 I=1,7
                WRITE(12,1010)SEGM(I),(SGINRT(J,I),J=1,3)
1010            FORMAT(F14.7/F14.7/F14.7/F14.7)
130     CONTINUE
        CALL CLOSE(12)

        LU1=LUO
140     WRITE(LU1,1016)HGT,METR(1)
1016    FORMAT(' Subject''s height is',F8.3,' ',A2)
        WRITE(LU1,1017)W,METR(2)
1017    FORMAT(' Subject''s total mass is',F8.3,' ',A2)
C       WRITE(LU1,1018)SEX(1)
1018    FORMAT(' Subject''s sex is ',A2)
 141    WRITE(LU1,1011)
1011    FORMAT(/,' The segment masses and lengths are:')
        DO 131 I=1,7
        WRITE(LU1,1012)I,SEGM(I),SEGL(I)
1012    FORMAT('     Segment',I3,' is',F14.7,' kg.',F14.7,' M')
131     CONTINUE
        WRITE(LU1,1013)
1013    FORMAT(/,' Moments of inertia about the CG are:')
        WRITE(LU1,1015)
1015    FORMAT(15X,' Ixx',12X,' Iyy',12X,' Izz')
        DO 132 I=1,7
        WRITE(LU1,1014)I,(SGINRT(J,I),J=1,3)
1014    FORMAT(' Segment',I3,3F14.7)
 132    CONTINUE
        IF(LU1.NE.LUP)GOTO 145
        CALL CLOSE(LUP)
        GOTO 150
145     CALL QUERY('Do you want a copy of results printed',IAR)
        IF(IAR.NE.1)GOTO 150
        LU1=LUP
        OPEN(UNIT=LUP,TYPE='NEW',DISPOSE='PRINT')
        GOTO 140
150     CONTINUE

        ISIDE=ISIDE+1
        IF(ISIDE.EQ.2)GOTO 9998


        CALL EXIT
        END
```

```
      PROGRAM CGPROG
C     This program reads in a complete set of TRACK data including the
C     DTR, DTL, 3DR, 3DL, and FPD, files.  It also reads in the
C     appropriate INERTIA and GEOMETRY files.

C     Using this data the program calculates for each frame of data the
C     location of the center of mass of each segment in 3D, the location
C     of the body center of mass in 3D, and the location of the center
C     of pressure in 2D.

C     The output consist of the CG and CP locations plotted against time.
C     Metafiles of each result may be stored as CGIMAGE.DAT files for
C     hardcopy reproduction or to be viewed later.  SPECIFIC FILE naming
C     will be done by the command file which runs this program and not by
C     the program itself.

C     The normal data locations will be:
C           DTR,DTL,FPD                   TK4: [PDP11/60,assigned by
c                                               SYS$system:[track]
c                                               track_run_assign.com]
C           INERTIA,GEOMETRY.DTR/DTL      TK3: [MicroVAX,assigned by
c                                               SYS$system:[track]
c                                               track_run_assign.com]
C           CGDATA.DAT                    PR4: [MicroVAX,assigned
C                                               by user]
C******************************************************************************
C           Patrick O. Riley        March, 1987
C     link CGPROG,TK7:track.olb/lib
C******************************************************************************
      CHARACTER*16    NMRFIL,NMLFIL
      CHARACTER*16    NAMOUT,NMFFIL,NMFFPL,NFPL
      CHARACTER*18    NMRGEO,NMLGEO
      CHARACTER*17    NMRINT,NMLINT
      CHARACTER*32    UNIT
      CHARACTER*1 AXIS
      BYTE LEG,TURN,IANS

      COMMON /INPUT/ NSEG,NSEGP,NFRAME,AXIS,ISCALE(3)
      COMMON /BLKINT/NMRGEO,NMLGEO,NMRINT,NMLINT
      COMMON /BLKCGV/CGCNTR(3,13),SEGMAS(13)

      COMMON /BLKFIL/NMFFPL,NMFFIL
      COMMON /SEGDAT/NRSEG,NLSEG,NSLS,NRJNT,NLJNT,NSLJ,NJNT

      DIMENSION DATA(15,1000),IPLOT(15)

      DATA IPLOT/2,2,2,2,2,2,2,2,2,2,0,0,2,2,2/
      DATA NMRFIL/'TK4:000000.DTR;0'/
      DATA NMLFIL/'TK4:000000.DTL;0'/
      DATA NMFFIL/'TK4:000000.FPD;0'/

      DATA NMRGEO/'TK3:GEOMETRY.DTR;0'/
      DATA NMLGEO/'TK3:GEOMETRY.DTL;0'/
      DATA NMRINT/'TK3:INERTIA.DTR;0'/
      DATA NMLINT/'TK3:INERTIA.DTL;0'/
      DATA NAMOUT/'PR4:CGDATA.XDT;0'/
      DATA ISCALE/1,1,1/
      DATA UNIT/'X Displacement [m] vs Frame No. '/

      PARAMETER(LUO=6,LUI=5,BAD=-1000.,LUP=8)
```

```
10      WRITE(LUO,1810)
1810    FORMAT(' Name of the file to be plotted: ')
        READ(LUI,1820)NMRFIL(5:10)
        WRITE(LUO,1822)NMRFIL(5:10)
1820    FORMAT(1A6)
1822    FORMAT(' Processing file: ',1A6,' to locate CGs')
        IF(NMRFIL(5:10).EQ.'      ')CALL EXIT

        NMLFIL(5:10)=NMRFIL(5:10)
        NMFFIL(5:10)=NMRFIL(5:10)
        NAMOUT(5:10)=NMRFIL(5:10)

        OPEN(UNIT=15,NAME=NMRFIL,TYPE='OLD',ACCESS='DIRECT',READONLY)
        OPEN(UNIT=16,NAME=NMLFIL,TYPE='OLD',ACCESS='DIRECT',READONLY)
        OPEN(UNIT=17,NAME=NMFFIL,TYPE='OLD',ACCESS='DIRECT',READONLY)

        READ(15'2)RRNCHN,FRAMES,FREQ,RRNSEG,FPLSCL,FPXSCL,FPZSCL
        READ(15'4) NMFFPL(1:16)
        IF(NMFFPL(11:14).NE.'.FPL')THEN
                NMFFPL(1:4)='TK6:'
                NMFFPL(11:16)='.FPL;0'
                WRITE(LUO,1865)
                READ(LUI,1861)NMFFPL(5:10)
        ENDIF
1861    FORMAT(1A6)
1865    FORMAT(' FORCE PLATE LOCATION FILE IN OLD FORMAT',/,
     1          ' READING NAME FROM SYS$INPUT')

        READ(16'2)RLNCHN,FRAMES,FREQ,RLNSEG
        NRSEG   =IFIX(RRNSEG)
        NLSEG   =IFIX(RLNSEG)
        NSLS    =NRSEG+1
        NSEG    =NRSEG+NLSEG
        NRJNT   =NRSEG
        NLJNT   =NLSEG
        NSLJ    =NRJNT+1
        NJNT    =NRJNT+NLJNT
        NFRAME  =IFIX(FRAMES)
        NFREQ   =IFIX(FREQ)
        WRITE(LUO,44)
44      FORMAT(' START FRAME, END FRAME:')
50      READ(LUI,1812)IS,IF
        IF(IS.LE.0)IS=1
        IF(IF.LE.0)IF=NFRAME
        IF(IF-IS.GT.1000) IF=IS+1000
        WRITE(LUO,1812)IS,IF
1812    FORMAT(2I4)
1814    FORMAT(' START FRAME, END FRAME:',2I4)
        NFRAME=IF-IS

        CALL DATARD(IF,IS)      !READ THE INPUT DATA INTO LARGE ARRAYS

        CLOSE(15)       !CLOSE THE INPUT DATA FILES
        CLOSE(16)
        CLOSE(17)

60      AXIS='X'
75      IF(AXIS.EQ.'X'.OR.AXIS.EQ.'Z')THEN
                NSEGP=NSEG+2
```

```
        ELSE
                NSEGP=NSEG+1
        ENDIF
        CALL GEOMETRY   !READ NECESSARY GEOMETRY AND INERTIA FILE DATA
        CALL CGFIND(DATA,IPLOT) !FIND THE SEGMENT AND BODY CG
        CALL CPFIND(DATA,IPLOT) !FIND THE CENTER OF PRESSURE
C WRITE OUT THE CG/CP POSITION ARRAY TO A DIRECT ACCESS FILE
        NAMOUT(12:12)=AXIS      !FOR VEL AND ACL CALCULATIONS
         OPEN(UNIT=13,NAME=NAMOUT,TYPE='NEW',ACCESS='DIRECT',
     1          RECL=NSEGP)
         WRITE(13'1)NMRFIL(1:16),FREQ,FLOAT(NFRAME)
         DO I=1,NFRAME
                WRITE(13'I+1)(DATA(NS,I),NS=1,NSEGP)
         ENDDO
         CLOSE(13)

        IF(AXIS.EQ.'X')THEN
                AXIS='Y'
                GOTO 75
        ENDIF
        IF(AXIS.EQ.'Y')THEN
                AXIS='Z'
                GOTO 75
        ENDIF

        END
C**********************************************************************
        SUBROUTINE GEOMETRY

        COMMON /BLKINT/NMRGEO,NMLGEO,NMRINT,NMLINT
        COMMON /BLKCGV/CGCNTR(3,13),SEGMAS(13)
        DIMENSION PRCG(3,13),DISCG(3,13)
        CHARACTER*18 NMRGEO,NMLGEO
        CHARACTER*17 NMRINT,NMLINT

C READ THE RIGHT SIDE GEOMETRY FILE
        OPEN(UNIT=13,NAME=NMRGEO,TYPE='OLD',READONLY)

        DO I=1,3
                READ(13,1000) (PRCG(J,I),J=1,3)
        ENDDO
        DO I=1,3
                READ(13,1000) (DISCG(J,I),J=1,3)
        ENDDO
        DO I=1,3
                READ(13,1000) (CGCNTR(J,I),J=1,3)
        ENDDO

        DO I=4,6
                READ(13,1000) (PRCG(J,I),J=1,3)
        ENDDO
        DO I=4,6
                READ(13,1000) (DISCG(J,I),J=1,3)
        ENDDO
        DO I=4,6
                READ(13,1000) (CGCNTR(J,I),J=1,3)
        ENDDO
        READ(13,1000) (PRCG(J,7),J=1,3)
        READ(13,1000) (DISCG(J,7),J=1,3)
        READ(13,1000) (CGCNTR(J,7),J=1,3)
```

```
      CLOSE(13)
C READ LEFT SIDE GEOMETRY FILE
      OPEN(UNIT=13,NAME=NMLGEO,TYPE='OLD',READONLY)

      DO I=8,10
              READ(13,1000) (PRCG(J,I),J=1,3)
      ENDDO
      DO I=8,10
              READ(13,1000) (DISCG(J,I),J=1,3)
      ENDDO
      DO I=8,10
              READ(13,1000) (CGCNTR(J,I),J=1,3)
      ENDDO

      DO I=11,13
              READ(13,1000) (PRCG(J,I),J=1,3)
      ENDDO
      DO I=11,13
              READ(13,1000) (DISCG(J,I),J=1,3)
      ENDDO
      DO I=11,13
              READ(13,1000) (CGCNTR(J,I),J=1,3)
      ENDDO
      CLOSE(13)

1000  FORMAT(F14.7)

C NOW GET THE SEGMENT MASSES
      OPEN(UNIT=13,NAME=NMRINT,TYPE='OLD',READONLY)
      DO I=1,7
              READ(13,1010) SEGMAS(I),XI,YI,ZI
      ENDDO
      CLOSE(13)

1010  FORMAT(F14.7)

      OPEN(UNIT=13,NAME=NMLINT,TYPE='OLD',READONLY)
      DO I=8,13
              READ(13,1010) SEGMAS(I),XI,YI,ZI
      ENDDO
      CLOSE(13)
      RETURN
      END
C*****************************************************************************
      SUBROUTINE DATARD(IF,IS)

      COMMON /INPUT/ NSEG,NSEGP,NFRAME,AXIS,ISCALE(3)
      COMMON /BLKCGV/CGCNTR(3,13),SEGMAS(13)

      COMMON /BLKFIL/NMFFPL,NMFFIL
      COMMON /SEGDAT/NRSEG,NLSEG,NSLS,NRJNT,NLJNT,NSLJ,NJNT

      COMMON /RDTBLK/DR(3,10,1000),SGRTRT(3,3,10,1000)
      COMMON /LDTBLK/DL(3,10,1000),SGRTLT(3,3,10,1000)
      COMMON /FPDBLK/FY(2,1000),AX(2,1000),AZ(2,1000),NFP
      BYTE AXIS

C  READ THE RIGHT SIDE KINEMATIC DATA INTO DATA ARRAYS
      DO J=1,NFRAME
```

```
                 READ(15'J+IS+10) (((DR(J2,J3,J),J2=1,3),A1,A2,A3,A4,
     1                          ((SGRTRT(J1,J2,J3,J),J2=1,3),J1=1,3))
     2                          ,J3=1,NRSEG)

        ENDDO
C   READ THE LEFT SIDE KINEMATIC DATA INTO DATA ARRAYS

        DO J=1,NFRAME

                 READ(16'J+IS+10) (((DL(J2,J3,J),J2=1,3),A1,A2,A3,A4,
     1                          ((SGRTLT(J1,J2,J3,J),J2=1,3),J1=1,3))
     2                          ,J3=1,NLSEG)

        ENDDO

C READ THE FORCEPLATE DATA INTO DATA ARRAYS
        READ(17'5)FPXSCL,FPZSCL
        NFP = IFIX(FPZSCL)

        DO J=1,NFRAME
              READ(17'J+IS+10) ((FX,FY(N,J),FZ,AX(N,J),MZ,AZ(N,J)),N=1,NFP)
        ENDDO

        RETURN
        END
C*************************************************************************
        SUBROUTINE CPFIND(DATA,IPLOT)

        COMMON /BLKFIL/NMFFPL,NMFFIL
        COMMON /INPUT/ NSEG,NSEGP,NFRAME,AXIS,ISCALE(3)
        COMMON /FPDBLK/FY(2,1000),AX(2,1000),AZ(2,1000),NFP
        CHARACTER*16    NMRFIL,NMLFIL,NMRSEG,NMLSEG,NMFFIL,NMFFPL,NFPL
        BYTE AXIS
        DIMENSION DATA(15,1000),IPLOT(15)
        DIMENSION ROTFP1(3,3),ROTFP2(3,3),RTEMP(3,3),TEMP(2,3)
        BAD=-1000.
        IF(AXIS.EQ.'X'.OR.AXIS.EQ.'Z')THEN

C read the force plate location file
          OPEN(UNIT=13,NAME=NMFFPL,STATUS='OLD',READONLY)
        READ(13,1230) ITEMP6
        READ(13,1231) XFP1,YFP1,ZFP1
        READ(13,1231) ROTFP1(1,1),ROTFP1(1,2),ROTFP1(1,3)
        READ(13,1231) ROTFP1(2,1),ROTFP1(2,2),ROTFP1(2,3)
        READ(13,1231) ROTFP1(3,1),ROTFP1(3,2),ROTFP1(3,3)
        READ(13,1231) XFP2,YFP2,ZFP2
        READ(13,1231) ROTFP2(1,1),ROTFP2(1,2),ROTFP2(1,3)
        READ(13,1231) ROTFP2(2,1),ROTFP2(2,2),ROTFP2(2,3)
        READ(13,1231) ROTFP2(3,1),ROTFP2(3,2),ROTFP2(3,3)
1230    FORMAT(A2)
1231    FORMAT(3F15.7)
          CLOSE(13)
C NEGLECT SMALL ROTATION OF THE FORCE PLATE
c find the combined CP in global coordinates
         IF(AXIS.EQ.'X')THEN
          DO I=1,NFRAME
                if(abs(fy(1,i)).le.2.5)fy(1,i)=0.0
                if(abs(fy(2,i)).le.2.5)fy(2,i)=0.0
                IF(FY(1,I).eq.0.0.and.FY(2,i).eq.0.0)THEN
                 DATA(NSEGP,I)=0.0
```

```
            ELSE
             DATA(NSEGP,I)=(FY(1,I)*(XFP1-AX(1,I))
     1                +FY(2,I)*(XFP2-AX(2,I)))/
     2                (FY(1,I)+FY(2,I))
            ENDIF.
       ENDDO
      ENDIF
      IF(AXIS.EQ.'Z')THEN
       DO I=1,NFRAME
            if(abs(fy(1,i)).le.2.5)fy(1,i)=0.0
            if(abs(fy(2,i)).le.2.5)fy(2,i)=0.0
            IF(FY(1,I).eq.0.0.and.FY(2,i).eq.0.0)THEN
             DATA(NSEGP,I)=0.0
            ELSE
             DATA(NSEGP,I)=(FY(1,I)*(ZFP1-AZ(1,I))
     1                +FY(2,I)*(ZFP2-AZ(2,I)))/
     2                (FY(1,I)+FY(2,I))
            ENDIF
       ENDDO
      ENDIF
     ENDIF
     RETURN
     END

C*****************************************************************************
     SUBROUTINE CGFIND(DATA,IPLOT)

     COMMON /INPUT/ NSEG,NSEGP,NFRAME,AXIS,ISCALE(3)
     COMMON /BLKCGV/CGCNTR(3,13),SEGMAS(13)

     COMMON /BLKFIL/NMFFPL,NMFFIL
     COMMON /SEGDAT/NRSEG,NLSEG,NSLS,NRJNT,NLJNT,NSLJ,NJNT

     COMMON /RDTBLK/DR(3,10,1000),SGRTRT(3,3,10,1000)
     COMMON /LDTBLK/DL(3,10,1000),SGRTLT(3,3,10,1000)

     DIMENSION DATA(15,1000),IPLOT(15)
     DIMENSION ROT(3,3),TEMP(3),CGVEC(3)
     BYTE AXIS

     IF(AXIS.EQ.'X')IAXIS=1
     IF(AXIS.EQ.'Y')IAXIS=2
     IF(AXIS.EQ.'Z')IAXIS=3
     TOTMAS =0.0

     DO I=    1,NSEG
             IF(I.NE.11.AND.I.NE.12)TOTMAS=TOTMAS+SEGMAS(I)
     ENDDO

     DO IFRM = 1,NFRAME
        DO ISEG = 1,NRSEG

             DO J=1,3
                DO K=1,3
                     ROT(J,K)=SGRTRT(K,J,ISEG,IFRM)
                ENDDO
             ENDDO

             DO I =1,3
                     TEMP(I)=CGCNTR(I,ISEG)
```

```
            ENDDO
            CALL GMPRD(ROT,TEMP,CGVEC,3,3,1)
            DATA(ISEG,IFRM)=DR(IAXIS,ISEG,IFRM)+CGVEC(IAXIS)
       ENDDO

       DO ISEG=1,NLSEG
            KSEG =ISEG+NRSEG

            DO J=1,3
               DO K=1,3
                    ROT(J,K)=SGRTLT(K,J,ISEG,IFRM)
               ENDDO
            ENDDO

            DO I =1,3
                    TEMP(I)=CGCNTR(I,KSEG)
            ENDDO

            CALL GMPRD(ROT,TEMP,CGVEC,3,3,1)

            DATA(KSEG,IFRM)=DL(IAXIS,ISEG,IFRM)+CGVEC(IAXIS)
       ENDDO

C       Average the left and right estimates of pelvis CG position
          DATA(4,IFRM) =(DATA(4,IFRM)+DATA(11,IFRM))/2.

C       Average the left and right estimates of trunk CG position
          DATA(5,IFRM) =(DATA(5,IFRM)+DATA(12,IFRM))/2.

          DATA(NSEG+1,IFRM)=0.0

          DO ISEG=1,NSEG
               IF(ISEG.NE.11.AND.ISEG.NE.12)THEN
                DATA(NSEG+1,IFRM)=DATA(NSEG+1,IFRM) +
     1                  DATA(ISEG,IFRM)*SEGMAS(ISEG)
               ENDIF
          ENDDO

          DATA(NSEG+1,IFRM)=DATA(NSEG+1,IFRM)/TOTMAS
       ENDDO
       RETURN
       END

C***********************************************************************
       SUBROUTINE DELTA(DATA,IPLOT,DELTY)
       COMMON /INPUT/ NSEG,NSEGP,NFRAME,AXIS,ISCALE(3)
       DIMENSION DATA(15,1000),IPLOT(15)

       DO I=2,NFRAME
               DO J=1,NSEGP
                       IF(J.LE.14)DATA(J,I)=DATA(J,I)-DATA(J,1)
                       IF(J.EQ.15)DATA(J,I)=DATA(J,I)-DATA(14,1)
               ENDDO
       ENDDO

       IF(NSEGP.EQ.15)DATA(15,1)=DATA(15,1)-DATA(14,1)

       DO J=1,NSEGP
               IF(J.LE.14)DATA(J,1)=0.0
       ENDDO
```

```
      DO NFRM =1,NFRAME
           DO J=1,NSEGP
              IF(IPLOT(J).EQ.2.AND.J.LE.7)
     1                DATA(J,NFRM)=DATA(J,NFRM)+DELTY*J
              IF(IPLOT(J).EQ.2.AND.J.GT.7)
     1                DATA(J,NFRM)=DATA(J,NFRM)-DELTY*J
           ENDDO
      ENDDO

      RETURN
      END
```

Appendix F

Test protocols

The protocol for the initial static posture and balance trials is given first. The expanded and modified protocol to be used for posture control trials is presented second. Each protocol provides for recording subject information, list certain conditions to be maintained during the trials and then list a series of tests to be performed. Subject/Guardian informed consent forms were obtained, but are seperate document from the protocol.

TEST PROTOCOL FOR STUDY OF POSTURE AND
BALANCE IN NORMAL AND CP CHILDREN

Patrick O. Riley

April, 1987
Rev 14

SUBJECT NAME____________________________________

RAW DATA DIRECTORY______________________________

PROCESSED DIRECTORY_____________________________

FORCE PLATE/EMG DATA____________________________

DATE____

SUBJECT INFORMATION

SUBJECT SEX______________________________

SUBJECT AGE______________YR_____________MO

SUBJECT HEIGHT___________________________IN

SUBJECT WEIGHT___________________________KG

FOOT LENGTH_____________________IN

ANKLE CIR_______________________IN

KNEE CIR________________________IN

THIGH CIR_______________________IN

WASTE CIR_______________________IN

CHEST CIRC______________________IN

ARM CIR_________________________IN

ARM LENGTH______________________IN

NECK CIR________________________IN

HEAD CIRC_______________________IN

GENERAL CONDITIONS

- 'usual stance' or 'standup straight'
  --best standing posture.
- feet together but not touching,
  one on each forceplate
- initial subjects- spastic quadraplegic CP 10,
  other CP 5, normals 10.
- forceplates located side-by-side in a
  standard configuration.
- verify the child can stand unsupported with eyes
  closed or blindfolded for the
  required length of time.
- if subject cannot remain relatively stationary
  shorten test.
- obtain EMG data during long forceplate data runs.
  Mounting EMG electrodes and posture arrays on
  subject simultaneously is impractical
- 'eyes closed' may require the use of a blindfold.
- In the case of CP subjects it may be necessary to
  obtain the kinematic data and the long sets
  of forceplate/EMG data in separate sessions.
  It may also be more convenient to obtain the
  normal subject data in two sessions if available.
- "audio feedback" shall consist of coaching by a
  physician and/or therapist
- passive disturbances will be by pushing the subject
  from behind at shoulder level being careful
  not to obstruct the upper trunk arrays
- subjects will be instructed as to how to perform
  the active arm raise disturbance and
  practice runs will be used to insure an
  appropriate maneuver
- when doing tactile reference tests insure that the subject
  is applying minimum force via hands.

JOINT CENTER DETERMINATION

- bilateral joint center determination
  including upper body using
  [5,1]JOINTP.CMD and/or
  [5,1]JOINTF.CMD

- measure heel to heel distance and great toe
  to great toe distance

- determine subject's overall height, i.e. the distance from
  the floor to the top of the head in the subject's
  normal posture, do not attempt to determine the
  true erect height.

STATIC POSTURE
FORCEPLATE AND KINEMATICS DATA
using [5,1]GAIT.CMD

Desired sampling rate 153hz
Length of each data record 5 sec minimum

1-normal stance eyes open

2-normal stance blindfold

3-tactile reference with eyes open

4-tactile reference with blindfold

5-audio feedback with eyes open

6-right arm raise eyes open

7-passive perturbation unalerted

8-passive perturbation alerted

9-normal stance eyes open with AFO

10-normal stance blindfold with AFO

11-normal stance eyes open [repeat]

POSTURE CONTROL STUDY
FORCEPLATE AND KINEMATICS DATA
using [5,1]GAIT.CMD

Desired sampling rate 153hz
Length of each data record 5 sec,
unless otherwise indicated.

1-normal stance eyes open

2-tactile reference with eyes open

3-audio feedback with eyes open

4-right arm raise eyes open

5-right arm raise eyes closed

6-passive perturbation unalerted eyes open

7-passive perturbation alerted eyes open

8-passive perturbation alerted eyes closed

9-normal stance eyes open with AFO

10-normal stance blindfold with AFO

11-normal stance eyes open [repeat]

12-gait-right foot on force plate [3 sec]

13-gait-left foot on force plate [3 sec]

14-chair rise 100% knee height+-10%

LONG FORCEPLATE AND EMG DATA TEST
using [5,1]EMG.CMD

Test length minimum 1.5 minute, unless otherwise indicated.
Sample rate 153hz

1-normal stance eyes open

2-normal stance blindfold

3-tactile reference with eyes open

4-tactile reference with blindfold

5-audio feedback with eyes open

6-audio feedback with blindfold

7-normal stance eyes open with AFO

8-normal stance blindfold with AFO

9-normal stance eyes open with earplugs

10-normal stance blindfold with earplugs

11-normal stance eyes open [repeat]

12-right arm raise, eyes open [5 sec]

13-right arm raise, blindfold [5,sec]

ADDITIONAL TEST FOR NORMAL SUBJECTS

-stand with simulated CP posture

-instruct subject to lean as far to right as possible and hold for 3 seconds

-instruct subject to lean as far to left as possible and hold for 3 seconds

-instruct subject to lean as far forward as possible and hold for 3 seconds

-instruct the subject to move head through the full range of motion while holding the rest of the body rigid

-instruct the subject to perform a lateral arm raise as well as a AP arm raise

-instruct subject to lean as far back as possible and hold for 3 seconds

-instruct subject to rise up on tip-toe and hold 3 sec

-instruct subject to balance on dominant leg for 3 sec

-instruct subject to balance on non-dominant leg for 3 seconds

-instruct subject to balance on a 2X4 for about 1 minute

-instruct subject to climb up steps arranged on the forceplates

TEST PROTOCOL FOR STUDY OF POSTURE CONTROL

Patrick O. Riley

MAY, 1987

SUBJECT NAME____

RAW DATA DIRECTORY__________________________________

PROCESSED DIRECTORY_________________________________

BALANCE DATA DIRECTORY______________________________

DATE____

SUBJECT INFORMATION

SUBJECT SEX________________________________

SUBJECT AGE_______________YR______________MO

SUBJECT HEIGHT_____________________________IN

SUBJECT WEIGHT__________________________LB/KG

SUBJECT KNEE HEIGHT_______________________IN

FOOT LENGTH_______________________IN

ANKLE CIR_________________________IN

KNEE CIR__________________________IN

THIGH CIR_________________________IN

WASTE CIR_________________________IN

CHEST CIRC________________________IN

ARM CIR___________________________IN

ARM LENGTH________________________IN

NECK CIR__________________________IN

HEAD CIRC_________________________IN

GENERAL CONDITIONS

- -When testing a minor a parent or gardian must be present during the entire testing period.
- -'usual stance' or 'standup straight' --best standing posture.
- -feet approximately 4 inches apart at heel centers
- -forceplates located side-by-side in a standard configuration.
- -'eyes closed' may require the use of a blindfold.
- -subjects will be instructed as to how to perform the active arm raise disturbance and practice runs will be used to insure an appropriate maneuver at acceptable speed.
- -subject will perform chair rise test from a chair height of 100±5% of knee height or 115±5% of kneeheight
- -metronome will be used to set cadence for the chair rise with the maneuver to be completed in 1 beat timing according to chair rise study protocol
  - 85%KH NORMAL SPEED--SETTING=63
  - 100%KH NORMAL SPEED--SETTING=66
  - 115%KH NORMAL SPEED--SETTING=69
- -metronome will be used to set cadence for the stair climb
  - NORMAL SPEED--SETTING=80
  - SLOW SPEED--Setting=60
- -when doing tactile reference tests insure that the subject is applying minimum force via hands.
- -initiation of gait test will begin with the subject at the near computer room edge of the viewing volume heading away from the computer room.

JOINT CENTER DETERMINATION

-bilateral joint center determination including upper body using [5,1]JOINTP.CMD

and/or

[5,1]JOINTF.CMD

-measure heel to heel distance and great toe to great toe distance

-determine subject's overall height, i.e. the distance from the floor to the top of the head in the subject's normal posture, do not attempt to determine the true erect height.

POSTURE CONTROL STUDY
FORCEPLATE AND KINEMATICS DATA
using [5,1]GAIT.CMD

Desired sampling rate 153hz
Length of each data record 5 sec,
unless otherwise indicated.

1-chair rise [3 sec] eyes open 115%Kh

2-chair rise [3 sec] eyes open 115%KH

3-chair rise [3 sec] eyes open 100%Kh

4-chair rise [3 sec] eyes open 100%KH

5-chair rise [3 sec] eyes open 85%KH

6-chair rise [3 sec] eyes open 85%KH

7-stair climb right first

8-stair climb left first

9-normal stance eyes open

10-normal stance blindfolded

11-corrected posture eyes open

12-right arm raise eyes open

13-right arm raise eyes open

14-gait-right foot on force plate

15-gait-left foot on force plate

16-initiation of gait right foot first

17-initiation of gait left foot first

18-normal stance eyes open [repeat]

LONG FORCEPLATE BALANCE TEST
using [5,1]EMG.CMD

Test length minimum 1.5 minute, unless otherwise indicated.
Sample rate 153hz

1-normal stance eyes open

2-normal stance blindfold

3-tactile reference with eyes open

4-tactile reference with blindfold

5-audio feedback with eyes open

6-audio feedback with blindfold

7-normal stance eyes open with AFO/CANE

8-normal stance blindfold with AFO/CANE

9-normal stance eyes open with earplugs

10-normal stance blindfold with earplugs

11-single leg stance

12-right arm raise, eyes open [5 sec]

13-right arm raise, eyes open [5,sec]

15-normal stance eyes open [repeat]

ADDITIONAL TEST FOR NORMAL SUBJECTS

- stand with simulated CP posture
- instruct subject to lean as far to right as possible and hold for 3 seconds
- instruct subject to lean as far to left as possible and hold for 3 seconds
- instruct subject to lean as far forward as possible and hold for 3 seconds
- instruct the subject to move head through the full range of motion while holding the rest of the body rigid
- instruct the subject to perform a lateral arm raise as well as a AP arm raise
- instruct subject to lean as far back as possible and hold for 3 seconds
- instruct subject to rise up on tip-toe and hold 3 sec
- instruct subject to balance on dominant leg for 3 sec
- instruct subject to balance on non-dominant leg for 3 seconds
- instruct subject to balance on a 2X4 for about 1 minute
- instruct subject to climb up steps arranged on the forceplates

Appendix G

Thesis Defense Text

MODELING THE BIOMECHANICS OF
POSTURE AND BALANCE

1. THIS MORNING LADIES AND GENTLEMEN, I AM GOING TO SUMMARIZE THE WORK DESCRIBED IN MY THESIS; "MODELING THE BIOMECHANICS OF POSTURE AND BALANCE." THIS PRESENTATION WILL COVER THE BASIC CONTENT OF THE THESIS WITH ONLY A SLIGHTY DIFFERENT ORDER OF PRESENTATION. IN THE WORK THAT I WILL DESCRIBE THE SELSPOT/TRACK KINEMATIC DATA AQUISITIONS SYSTEM IS USED TO SIMULTANEOUSLY STUDY WHOLEBODY POSTURE AND BALANCE. THE COORDINATION OF POSTURAL CONTROL TO MAINTAIN BALANCE IS, FOR THE NEURMUSCULAR OR MOTOR CONTROL SYSTEM, A PROBLEM GREAT COMPLEXITY WHICH RESEARCH TODATE HAS IGNORED OR MODELLED IN OVERLY SIMPLISTIC WAYS.

2. I WILL START BY DEFINING THE TERMS POSTURE AND BALANCE. I DEFINE POSTURE AS THE ALLIGNMENT OF THE BODY SEGMENTS.

3. I DEFINE BALANCE AS THE CONTROL OF THE POSITION OF THE CENTER OF GRAVITY OR OF THE CENTER OF PRESSURE.

4. THE CENTER OF PRESSURE IS A FAIRLY COMPLEX CONSTRUCTION, A PRESSURE DISTRIBUTION CAN FOR SOME PURPOSES BE REPRESENTATED BY A VECTOR FORCE WHOSE LINE OF ACTION AND MAGNITUDE ARE DETERMINED BY THE THE AREA INTERGRALS AS SHOWN HERE. IN GAIT STUDIES WE DEAL WITH THE GROUND REACTION FORCE VECTOR WHICH IS IN FACT THE RESULTANT VECTOR DUE TO THE PRESSURE DISTRIBUTION UNDER THE FOOT.

5. IN BALANCE STUDIES THE THE COMBINED CP IS USUALLY STUDIED-THE RESULTANT OF THE TWO RESULTANTS GROUND REACTION FORCE VECTORS, ONE AT EACH FOOT. THE PERSON MANINTAINS BALANCE IF THIS COMBINED CP REMAINS WITHIN THE BASE OF SUPPORT. THERE IS EVIDENCE THAT THE CP IS MUCH MORE TIGHTLY CONTROLLED.

6. IN BALANCE AND POSTURE CONTROL STUDIES THE TRAVEL OF THIS COMBINED CP IS MEASURED AND EVALUATED USUALLY USING SOME STATISTICAL TECHNIQUE WHICH PRESUMES THAT THE CP MOTION IS "RANDOM" AND THAT THE MOTION SHOULD BE MINIMIZED. FREQUENTLY IN POSTURE CONTROL STUDIES, A MODEL IS USED TO INFER BODY KINEMANTICS FROM THE MEASURED CP EXCURSION. BECAUSE THE INFORMATION AVAILABLE FROM THE FORCE PLATE DATA ALONE IS QUITE LIMITED, IT CAN BE USED AS INPUT TO ONLY QUITE SIMPLIFIED MODELS: EG SIMPLE OF SOMEWHAT CONSTRAINED COMPOUND INVERTED PENDULI. GIVEN THE NUMBER OF DEGREES OF FREEDOM OF THE HUMAN BODY ASSUMPTION OF SUCH SIMPLE MODELS IS NOT JUSTIFIABLE.

7. THE STUDY OF "STATIC" POSTURE HAS BEEN MOSTLY SUBJECTIVE AND QUALITATIVE WITH ONLY CRUDE ATTEMPTS AT QUANTITATIVE MEASUREMENT. SUBJECTS ARE OBSERVED AGAINST A POSTURE GRID AND SOMETIMES A SCORE IS OBTAINED TO ATTEMPT TO QUATNTIFY THE EVALUATION. THE KINEOSOLOGY LITERATURE

PRESENTS THIS TECHNIQUE FOR QUANTIFYING POSTURAL DEFORMITY. THE SUBJECT STANDS ON A BALANCE BEAM TO DETERMINE THE CENTER OF PRESSURE IN 1-D, A VERTICAL LINE IS DRAWN THROUGH THE CP AND HORIZONTAL LINES ARE DRAWN FROM CERTAIN ANATOMICAL LANDMARKS; THE COMBINED LENGTH OF THESE LINES BEING A MEASURE OF POSTURAL DEFORMITY.

8. I BEGAN WITH THE PRESUMPTION THAT THERE ARE BIOMECHANICAL LINKS BETWEEN POSTURE AND BALANCE CONTROL. SOME EARLY FORCE PLATE DATA THAT HINTED AT THIS BIOMECHANICAL LINK IS SEEN IN THE RESULTS OF SOME OF OUR FORCE PLATE STUDIED. HERE WE SEE A COMBINED CP DISTRIBUTION WHICH APPEARS TO BE PURELY RANDOM, THE TYPE OF DATA SEEN IN TYPICAL SINGLE FORCE PLATE BALANCE STUDIES. CERTAINLY LOOKING AT THIS DATA IT APPEARS TO INDICATE RANDOM MOVEMENT, AND STUDYING IT USING STATISTICAL METHODS PRESUMING RANDOM MOTION WOULD SEEM APPROPRIATE.

9. IF WE LOOK AT THE PATTERN OF THE CENTERS OF PRESSURE FOR EACH FOOT WE SEE THAT THE DISTRIBUTIONS ARE NOT RANDOM BUT HIGHTLY STRUCTURED WITH A MAJOR AXES WHICH SEEMS TO CORRESPOND TO THE MECHANICAL LINE OF ACTION OF THE CORRESPONDING LOWER LIMBS.

10. IN THIS CASE A SENSORY NEURAL, OR BIOMECHANICAL INTERVENTION HAS BEEN INCLUDED; THE SAME SUBJECT LIGHTLY TOUCHES A STABLE REFERENCE WITH HER LEFT HAND. THE COMBINED CP SEEMS LITTLE EFFECTED. OUR STATISTICAL ANALYSIS SAYS THAT THIS INTERVENTION HAS HAD NO EFFECT.

11. LOOKING AT THE INDIVIDUAL FOOT CP TRACES, WE SEE THAT THE INTERVENTION HAS EFFECTED THE LEFT SIDE DISTRIBUTION MARKEDLY.

12. SOMETIME THE BIOMECHANICAL CONNECTION TO CP CONTROL IS APPARENT EVEN ON THE COMBINED CP TRACES. HERE WE SEE THE TRACE PRODUCED BY A SUBJECT WITH A SEVERE AND ASSYMETRIC CROUTH DEFORMITY. I BELEIVE THAT AS OUR

ANALYSIS OF THE BIOMECHANIC OF POSTURE AND BALANCE ADVANCES WE WILL BE ABLE TO SHOW THAT THIS CP PATTERN IS A DIRECT CONSEQUENCE OT THE BIOMECHANICAL CONSTRAINTS IMPOSED BY THE SUBJECTS POSTURAL DEFORMITY.

13. THE FORCEPLATE DATA OF THE CROUCH DEFORMITY SUBJECTS SHOW ANOTHER BEHAVIOR WHICH IS I THINK A BIOMECHANICAL CONSEQUENCES OF THEIR POSTURE. THE TOTAL VERTICAL FORCE FOR STATIC DATA FOR MOST SUBJECTS SHOWS ALMOST NO VARIATION IMPLYING THAT VERTICAL ACCELERATIONS OF THE BODY CG ARE ALMOST NOEXISTENT. SUBJECTS WITH THE CROUTCH DEFORMITY TYPICALLY SHOW VARIATIONS IN THEIR COMBINED VERTICAL FORCE INDICATING THAT THEIR CG IS MOVING UP AND DOWM. I BELIEVE THAT THIS IS A CONSEQUENCE OF THE CROUCH DEFORMITY WHICH MAKE THE BODY INTO SOMETHING OF A VERTICLE SPRING; THESE SUBJECTS ARE CHARACTERIZED BY LARGE FELXION ANGLES AT THE ANKLE, KNEES, HIPS, AND LOWER BACK.

14. IN ORDER TO STUDY BALANCE AND POSTURE I FELT THAT IS WAS NECESSARY TO ACQUIRE WHOLEBODY KINEMATICS. THE KINEMATICS HAD TO DESCRIBE THE MOVEMENT OF A NUMBER OF BODY SEGMENTS SUFFICIENT TO BE REPRESENTATIVE OF THE WHOLE REPERTOIR OF MOVMENTS WHICH CAN EFFECT CG AND CP POSITION. FURTHER THE KINEMATICS HAD TO BE SUFFICIENTLY PRECISE TO PERMIT DETECTION OF THE RATHER SUBTILE MOVEMENTS WHICH FREQUENTLY CHARACTERIZE POSTURAL ADJUSTMENTS. I FELT THAT THE SELSPOT/TRACK SYSTEM HAD THE BASIC PRECISION AND FLEXIBILITY TO PERFORM THAT TASK. I SET OUT TO DEVELOP A SYSTEM FOR STUDING POSTURE AND BALANCE BASED TO THIS APPROACH. THIS IS THS PRIMARY GOAL OF MY RESEARCH; <u>TO DEVELOP A TECHNIQUE FOR THE STUDY THE BIOMECHANICS OF POSTURE AND BALANCE,</u> I AM USING THE TERM MODEL TO DESCRIBE A STRUCTURE WHICH I USE TO MAKE THE VAST AMOUNTS OF DATA COLLECTED IN A POSTURE STUDY UNDERSTANDABLE. I AM A EMPERICIS, I MODEL TO FACILITATE THE INTREPRETATION OF EXPERIMENTS, I DO NO

PERFORM EXPERIMENTS TO OBTAIN DATA FOR MY MODEL. THE MODEL IS AT THIS STAGE INTERPRETATIVE NOT PREDICTIVE. I ELECTED TO WORK AT THE MGH BIOMOTION LABORATORY FOR SEVERAL REASONS THE MOST BASIC OF WHICH WAS THAT THE BILATERAL CONFIGURATION OF THIS LAB, WHEN PROPERLY IMPLEMENTED WOULD FACILIATE ACQUISITION OF WHOLEBODY KINEMATIC DATA. THIS SLIDE SHOWS THE BASIC LAYOUT OUT OF THE LAB INCLUDING TWO FORCE PLATED AND TWO OPTICAL BENCHES EACH WITH FOUR CAMERAS.

15. THIS SLIDE SHOW A BLOCK DIAGRAM OF THE LABORATORY AND OUTLINES THE DATA FLOW.

16. THIS PICTURE SHOWS THE LABORATORY PHYSICALLY INCLUDING ONE OF THE TWO OPTICAL BENCHES AND THE FORCE PLATE LOCATION.

17. THIS PICTURE SHOWS THE MOUNTING OF A CAMERA ON THE OPTICAL BENCH, WITH EACH CAMERAS PRECISION TRANSLATOR AND ROTATOR.

18. THIS IS A DETAIL OF THE FORCE PLATE MOUNTING METHOD.

19. GETTING BACK TO THE MODELING OF POSTURE USING KINEMATIC DATA, I DEVELOPED AN ELEVEN SEGMENT 66-DOF MODEL OF THE BODY. IN DETAIL THERE ARE LIMITATIONS TO THIS MODEL, BUT IT IS CLEARLY SUPERIOR TO A SIMPLE INVERTED PENDULUM MODEL AND INCORPORATES THE ESSENTIAL FEATURES OF POSTURE; THE ALLIGNMENT OF THE UPPER AND LOWER LIMBS, THE CONTROL OF HEAD POSITION, AND THE CONTROL OF THE UPPER TRUNK RELATIVE TO THE PELVIS.

20. A NUMBER OF LED ARRAYS ARE USED TO ACQUIRE THE KINEMATIC INFORMATION NECESSARY TO DESCRIBE THIS MODEL, IN FACT BILATERL DATA IS ALMOST ESSENTIAL IF DATA FROM ALL SEGMENTS IS TO BE ACQUIRED SIMULTANEOUSLY. IN ORDER TO DO THIS THE LAB MUST BE TRULY BILATERAL; THE INFORMATION

FROM ALL FOUR CAMERAS AND THE FORCE PLATES MUST BE IN THE SAME COORDINATE SYSTEM.

21. LOOKING AT THE LABORATORY DIAGRAM INORDER TO ASSEMBLE ALL OF THE KINEMATIC DATA THE POSITION OF FOUR CAMERAS A MASTER AND SLAVE CAMERA ON THE RIGHT SIDE OPTICAL BENCH AND A MASTER AND SLAVE CAMERA ON THE LEFT SIDE OPTICAL BENCH MUST BE EXPRESSED ION A SINGLE COORDINATE SYSTEM. THE PROBLEM WAS DIVIDED INTO TWO PARTS, 1) THE GEOMETRY OF THE MASTER SLAVE COMBINATION ON EACH OPTICAL BENCH WAS DEFINED THEN 2) THE POSITION AND ORIENTATION OF THE LEFT HAND CAMERA ARRAY WAS FOUND WITH RESPECT TO THE RIGHT HAND CAMERA ARRAY.

22. THIS SLIDE SHOWS ONE OPTICAL BENCH AND SET OF CAMERAS, IN SEVERAL ATTEMPTS TO USE THE CAMERAS AS SURVEYING INSTRUMENTS TO DEFINE THE SYSTEM GEOMETRY BOB FIJAN AND I FOUND THAT IT WAS NOT VALID TO ASSUME THAT THE OPTICAL AXIS OF EACH CAMERA WAS PERPENDICULAR TO THE OPTICAL

BENCH OR THAT MECHANICAL SHIMMING OF THE CAMERA MOUNTS TO MAKE THIS TRUE AT ONE POSITION ON THE OPTICAL BENCH, NECESSARILY MADE IT TRUE AT ALL POSITIONS. BOB FIJAN DEVELOPED A TECHNIQUE FOR DEFINING THE GEOMETRIC RELATIONSHIP BETWEEN THE MASTER AND SLAVE CAMERAS. THE HEART OF THE TECHNIQUE WAS THE USE OF A GRADIENT SEARCH TECHNIQUE TO DEFINE A GEOMETRY WHICH PRODUCED THE MINIMUM AGGRAGATE SKEW RAY ERROR FOR A LARGE SET OF DATA WHERE THE LED LOCATION NEED NOT BE SPECIFIED.

23. THE DISTANCE BETWEEN CAMERAS AND THE CAMERA ROTATION ABOUT THE VERTICAL WERE TAKEN AS KNOWN, BEING PRECISELY MEASURABLE AND CONTROLLABLE WITH THE INSTALLED PRECISION ROTATOR AND TRANSLATORS.

24. THIS FIGURE ILLUSTRATES THE DEFINITION OF SKEW-RAY-ERROR AND INDICATES THE IMPORTANCE OF MINIMIZING THIS PARAMETER. MY TESTS OF THE RESULT SHOWED THAT THE METHOD CONSISTENTLY PRODUCED MEAN SKEW-RAY-ERRORS OF

1-2MM AND MAXIMUM SKEW RAY ERRORS OF ON THE ORDER OF 3-4MM FOR POSTURE STUDY DATA SETS. BOB USED A SIMILAR TECHNIQUE ON THE MIT SIDE WHERE THE PRECISION PLOTTER IS AVAILABLE TO PRECISELY POSITION LEDS AND SHOWED THAT THE METHOD YIELDED CORRECT POSITIONS TO 1MM IN THE CENTER OF THE VIEWING VOLUME.

25. THE SECOND PART OF THE PROBLEM WAS TO FIND THE POSITION OF THE TWO CAMERA SETS IN ONE COORDINATE SYSTEM. THE RIGHT HAND CAMERAS WERE DEFINED AS THE MASTER SET AND THE ROTATION MATRIX FOR CONVERTING LEFT SIDE TO RIGHT SIDE DATA WAS DEFINED. TO DO THIS I FABRICATED A LARGE BIDIRECTIONAL ARRAY, THAT IS AN ARRAY WITH TWO PLANES ONE FACING ONE WAY AND THE OTHER THE OPPOSITE. THE LEDS IN EACH PLANE WERE ARRANGED IN A CIRCLE ABOUT A SMALL CENTER HOLE. THE TWO PLANES WERE JOINED BY A BLOCK OF PRECISELY KNOW THICKNESS ALSO WITH A SMALL HOLE MACHINED THROUGHT THE CENTER. THE THREE COMPONENTS WERE JOINED WITH A TEMPORARY DOWELL THROUGHT THE ALLIGNING HOLES SO

THAT THE ARRAYS FACING EACH DIRECTION HAD A COMMON CENTER, THE CENTER OF THE ALIGNING HOLE. THE Z-POSITION OF THE COMMON ARRAY ORIGIN WAS THE CENTER OF THE ALIGNING WHOLE MIDWAY BETWEEN THE SURFACES. ONE THE SKEW RAY MINIMIZATION TECHNIQUE HAD BEEN USED TO DEFINE THE GEOMETRY OF EACH SIDE, THE POSITION OF THE BILATERAL ARRAY WAS DETERMINED IN EACH SIDES COORDINATE SYSTEM FOR A SET OF ARRAY POSITIONS SUFFICIENT TO REPRESENT THE COMBINED VIEWING VOLUME. THE SET OF RIGHT SIDE ARRAYS CENTERS WAS THEN TREATED AS A SET OF 3-D LED POSITIONS. THE SET OF LEFT SIDE CENTERS WAS TREATED AS A LARGE ARRAY SEGMENT FILE DISCRIPTION. SLIGHTLY MODIFIED VERSIONS OF THE STANDARD TRACK ROUTINES WERE THEN USED TO FINE THE ROTATION MATRIX [4X4] NECESSARY TO FIT THE LEFT SIDE SEGMENT FILE INTO THE RIGHT SIDE DATA. THIS IN TURN YIELDED THE CONVERSIONS NECESSARY TO DETERMINE THE POSITIONS AND ORIENTATION OF THE LEFT SIDE SET OF CAMERAS IN RIGHT SIDE COORDINATES. REPEATING THIS WHOLE

PROCESS INCLUDING THE SKEWRAY MINIMIZATION PORTION YIELDED RELATIVE CAMERA POSITIONS WHICH WERE CONSISTENT TO WITH 1MM POSITION AND 1/100 DEG ROTATION. WHEN THE BILATERAL ARRAY WAS USED AGAIN WITH THE POSITION SOLVED FOR USING THE COMBINED GEOMETRY THE LEFT AND RIGHT SIDE POSITIONS AGGREED TO WITHIN 1 TO 2 MM IN THE CENTER OF THE VIEWING VOLUME AND 3 TO 4 MM NEAR THE EDGE OF THE VOLUME. THUS THE SYSTEM WAS NOW TRUELY BILATERAL.

26. IN ORDER TO DESCRIBE POSTURE IT IS NECESSARY TO SPECIFY THE INITIAL CONDITIONS WITH ADEQUATE PRECISION. I WILL TALK ABOUT WHAT IS ADEQUATE PRECISION LATER. SPECIFING THE INITIAL CONDITIONS MEANS DEFINING THE TRANSFORMATION FROM ARRAY TO BODY SEGMENT COORDINATES, THIS IN TERN IS MAINLY A FUNCTION OF LOCATING JOINT CENTERS.

27. THIS FIGURE SHOWS ONE METHOD USED TO IDENTIFY JOINT CENTER WHICH IS ANALOGOUS TO THE ANATOMICAL LANDMARK METHOD USED BY MANY GAIT ANALYSIS CENTERS. THE

POINTER/ALIGNMENT ARRAY-A LARGE LED ARRAY WAS PLACE ON AN APPROPRIATE ANATOMICAL LANDMARK, THE JOINT CENTER WAS TAKEN TO BE SOME DISTANCE FROM THE POINTER IN THE MEDIAL DIRECTION.

28. THIS SHOW A MODIFICATION OF THE APPROACH USED TO DEFINE THE HIP JOINTS. BOB FIJAN INTRODUCED THESE TECHNIQUES AT THE MGHBL. EXACT ACCURACY IS DIFFICULT TO ASSESS, BASED ON SYMMETRY OF CONTROL SUBJECTS IN A GIVE SET OF TEST AND REPEATED TEST OF THE SAME SUBJECT, THE TYPICAL JOINT CENTER ESTIMATION ERROR USING THIS METHOD IS 2CM AND ERROR OF 4 CM SEEM POSSIBLE.

29. THIS FIGURE SHOWS THE ALIGNMENT POINTER ARRAY

30. INORDER TO ACHIEVE A MORE CONSISTENT JOINT CENTER DETERMINATION I IMPLEMENTED AN AXIS OF ROTATION METHOD OF THE TYPE BOB HAD SUGGESTED IN HIS SM THESIS. A FINITE DISPLACEMENT TECHNIQUE WAS USED TO DEFINE AXES OF ROTATION FOR THE ANKLE, KNEE AND HIP. THE KINEMATICS OF

A RISE FROM A CHAIR GENERALLY PROVIDE THE INPUT DATA. THE POSTION OF THE RISE AS THE SUBJECT APPROACHES UPRIGHT POSTURE WAS EXAMINED. THE AXES FOUND IN THIS WAY ARE THEN USED FOR THE STATIC POSTURE STUDIES. THE REPRODUCIBILITY OF THE JOINT CENTER DETERMINATION BY THIS METHOD IS ON THE ORDER OF 1CM. I WILL GET TO THE IMPLICATIONS OF THE ERRORS I AM DISCUSSING.

31. THIS RIG WAS USED TO TEST THE AXES OF ROTATIONS DETERMINATION ALGORITHM. ARRAYS ARE MOUNTED ON THE FIXED AND MOVABLE PARTS FACING BOTH THE LEFT AND RIGHT SIDE CAMERAS. THE LEFT SIDE RIGHT SIDE AGREEMENT OF THE AXES DEFINED OVER SEVERAL DIFFERENT TESTS WAS WITHIN 2 DEGREES INDICATING THE MAXIMUM ERROR OF THE AXIS DETERMINATION ALGORITHM AND OF THE LEFT SIDE RIGHT SIDE ALIGNMENT.

32. OUR INITIAL ATTEMPT TO APPLY A WHOLEBODY MODEL TO HUMAN POSTURE WAS A PILOT STUDY OF THE POSTURE AND BALANCE OF NORMAL AND CEREBRAL PALSIED CHILDREN. THE POSTURE PORTION OF THE PROTOCLOL CONSISTED OF A SERIES OF 5SEC TEST IN WHICH BILATERAL FORCEPLATE AND WHOLEBODY KINEMATIC DATA WERE ACQUIRED. "STATIC" STANCE WAS MAINLY EVALUATED WITH SOME VERY BASIC DISTURBANCE RESPONSE TRIALS THROWN IN.

33. THE FIRST QUESTION IS DOES THE MODEL ADEQUATELY REPRESENT THE IMPORTANT FEATURES OF POSTURE. THIS IS ONE OF THE FIRST SUBJECTS THAT WE OBTAINED WHOLEBODY KINEMATIC DATA ON. SHE IS A CEREBRAL PALSIED SUBJECT WITH A SEVERE CROUCH DEFORMITY.

34. THIS SIDE VIEW ILLUSTRATES THE FLEXED NATURE OF THE CROUTCH DEFORMITY. OBVIOUSLY WE RECOGINZE THAT ARRAY MOUNTING IS VERY IMPORTANT AND HAVE EVOLVED A SYSTEM WHICH IS MARKEDLY LESS ENCUMBERING, UNFORTUNATELY I

DON'T HAVE MORE UP TODATE SLIDES.

35. THIS IS THE 3-D GRAPHIC REPRESENTATION OF ONE FRAME OF THE SUBJECTS DATA. HER POSTURE IS WELL REPRESENTED IN THE AP AND LATERAL VIEWS.

36. THESE ARE SIMILAR VIEWS OF A HEMIPLEGIC CP SUBJECT FIRST FRONTAL

37. THEN LATERAL, NOTE THE HYPEREXTENDED LEFT LEG.

38. THIS IS THE 3D GRAPHIC OF THIS SUBJECT. AGAIN THE BASIC FEATURES OF HER POSTURE ARE WELL CHARACTERIZED. THE ADDITION OF FORCE PLATE INFORMATION PERMITS THE ASSYMETRY OF WEIGHT BEARING TO BE APPRECIATED.

39. THIS IS THE 3D GRAPHIC OF A QUADRAPLEGIC CEREBRAL PALSIED SUBJECT WITH A MARKED CROUCH DEFORMITY AND ASSYMETRY OF WEIGHT BEARING.

40. HERE IS ANOTHER SUBJECT WITH VERY SIMILAR POSTURE BUT QUITE SYMMETRIC WEIGHT BEARING.

41. THESE NEXT TWO SLIDE ILLUSTRATE HOW COMPARATIVELY SUBTILE EFFECTS OF A MECHANICAL INTERVENTIONS ON POSTURE CAN BE MEASURED. THIS IS A DIPLEGIC SUBJECT IN HIS NORMAL STANCE CONDITION. THE POSTURE APPEARS RATHER NORMAL WITH SLIGHT FLEXION AT THE ANKLE, HIP AND KNEE.

42. THIS IS THE SAME SUBJECT WEARING AFOS TO FORCE THE ANKLE TO 90° FLEXION. THE EFFECT IS TO BRING OUT THE EFFECT OF A SLIGHT FLEXION CONTRACTURE AT THE HIP WHICH WAS COMPENSATED FOR IN THE NORMAL STANCE. HERE BECAUSE THE ANKLES ARE FIXED AT 90° AND THE HIPS ARE CONSTRAINED TO BE FLEXED, HYPEREXTENSION OF THE KNEES RESULTS AND OVERALL POSTURE IS DEGRADED.

43. WE DEVELOPED NUMERICAL INDICES TO QUANTITATIVELY EXPRESSED THE SEVERITY OF POSTURAL DEFORMITY. ONE THE ANGLE INDEX WAS CALCULATED AS THE SUM OVER THE JOINTS OF INTEREST OF THE DIFFERENCE BETWEEN THE JOINT FLEXION ANGLE AND 0 (OUR ASSUMED NORMAL VALUE).

44. THIS SLIDES SHOW TYPICAL VALUES OF THIS INDICES FOR DIFFERENT LEVELS OF CEREBRAL PALSIED PATIENTS AND CONTROLS. THE THESIS PRESENTS THE COMPLETE SET OF DATA FOR THIS AND SOME OTHER INDICES FOR ALL SUBJECTS TESTED. THE ABILITY TO QUANTITATIVELY EVALUATE POSTURE IS DEMONSTRATED. THE SCATTER OF THE NORMAL DATA PROVIDES THE MOST INFORMATION ABOUT WHAT NEEDS TO BE DONE TO REFINE THE METHOD.

45. RECALL THAT I SAID THAT THE ERROR IN ESTIMATING JOINT CENTERS WAS TYPICALLY ON THE ORDER OF 2CM FOR THE ANATOMICAL LANDMARK METHOD, WITH 4 CM ERRORS BEING POSSIBLE. THIS SLIDE SHOWS A CRUDE BUT INFORMATIVE

MODEL OF HOW THIS ERROR EFFECTS JOINT ANGLE MEASUREMENTS. FOR A SEGMENT OF ABOUT 0.45M A 4 CM JOINT CENTER LOCATION ERROR CAN PRODUCE A 5° ANGLE ERROR, A 2CM JOINT CENTER ERROR PRODUCES ABOUT A 2.5° ANGLE ERROR AND A 1CM JOINT CENTER ERROR PRODUCES AN ANGLE ERROR OF A LITTLE OVER 1°. ALL OF THE DATA FOR THE INITIIAL SET OF POSTURE TRAILS WAS TAKEN USING THE ANATOMICAL LANDMARK METHOD OF JOINT CENTER DETERMINATION. AN ERROR OF $\pm 2°$ IN THE JOINT ANGLES FACTORED INTO THE ANGLE INDEX WOULD ACCOUNT FOR MOST OF THE VARIABILITY OF THIS INDEX AMONG CONTROLS. THUS THE METHOD IS USEFUL AS IS; 1 CM JOINT CENTER ACCURACY CAN BE CONSIDERED ADEQUATE FOR JOINT ANGLE MEASUREMENT IF NOT FOR MOMENT CALCULATIONS. HOWEVER, IMPROVING THE JOINT CENTER LOCATION DETERMINATION WILL GREATLY STRENGTHEN IT.

46. BY WAY OF REVIEW BEFORE GOING ON TO THE NEXT TOPIC THIS IS THE 3D REPRESENTATION OF THE POSTURE OF A TYPICAL CONTROL

47. A TYPCIAL DIPLEGIC CEREBRAL PALSIED SUBJECT

48. A TYPICAL HEMIPLEGIC CEREBRAL PALSIED SUBJECT

49. AND A TYPICAL QUADRAPLEGIC CEREBRAL PALSIED SUBJECT.

50. IN THIS INITIAL SERIE OF TEST WE ALSO DID MORE TRADITIONAL EVALUATIONS OF BALANCE USING 90 SECOND FORCE PLATE TEST

51. THE PARAMETER MEASURED WAS THE STANDARD DEVIATION OF THE COMBINED CP LOCATION.

52. I HAVE INDICATED SOME OF THE LIMITATIONS OF THIS MEASURE, NEVERTHELESS IT DOES PERMIT QUANTIFYING BALANCE IMPAIRMENT.

53. SOME DISTURBANCE RESPONSE TRAILS WERE ALSO ATTEMPTED. THE RAPID RIGHT ARM RAISE MANEUVER WAS THE MOST REPEATIBLE. THE SUBJECT ON SIGNAL DID A STRAIGHT ARM RAISE "AS FAST AS POSSIBLE" TO STRIKE A TARGET AT EYE LEVEL THEN HELD THE POSITION TO THE END OF THE TRIAL. THIS SLIDE SHOWS THE SEQUENCE FROM INITIATION TO APOGEE FOR A CONTROL SUBJECT. GROSSLY THE ONLY POSTURAL ADJUSTMENT OBSERVED IS A TRUNK ROTATION.

54. THIS SLIDE SHOWS THE CP DEVIATION DURING THE MANEUVER.

55. THIS SLIDE IS A CEREBRAL PALSIED SUBJECT PERFORMING THE SAME MANEUVER. A SIMILAR TRUNK ROTATION IS PRESENT, THERE IS ALSO AN APPARENTLY UNCOORDINATED MOVEMENT OF THE CONTRALATERAL ARM.

56. THIS SHOW THE CP DEVIATION DURING THIS MANEUVER.

57. THIS IS ANOTHER CP SUBJECT, AGAIN THEIR IS SLIGHT TRUNK ROTATION.

58. THIS SLIDE SHOWS THE CP DEVIATION DURING THIS MANEUVER. THE CP EXCURSION IS OF ABOUT THE SAME MAGNITUDE AS THE CONTROL'S, HOWEVER, ABOUT TWICE AS MANY FRAMES OF DATA ARE REQUIRED TO CAPTURE THE WHOLE ARM RAISE MANEUVER.

59. I SOUGHT TO DEVELOP A TECHNIQUE WHICH WOULD PERMIT THE COORDINATION OF SUCH MOVEMENTS TO BE BETTER EVALUATED. I FELT THAT IF WE COULD LOOK AT THE MOVEMENT OF THE CENTER OF MASS OF EACH OF THE DEFINED BODY SEGMENTS AND HOW THE MOVEMENT OF THAT SEGMENT EFFECTED THE LOCATION OF THE BODY CENTER OF GRAVITY WE WOULD GAIN SOME INSIGHT INTO COORDINATION.

60. THIS IS A BLOCK DIAGRAM FLOW CHART OF HOW THIS PORTION OF THE MODEL WORKS.

61. TO TEST THE CG ESTIMATION ALGORITHM I USED THIS TWO SEGMENT BODY MODEL. I FIRST DETERMINED THE MASS OF EACH SEGMENT AND THE ARRAY TO CG VECTOR IN ARRAY COORDINATES, THE VECTORS WERE DETERMINED TO ±1MM WHICH IS NOT AS TRIVIAL AS IT SOUNDS. THEN I TOOK STATIC DATA WITH THE SMALLER SEGMENT MOUNTED IN VARIOUS POSITIONS AND ORIENTATIONS ON THE LARGER SEGMENT. THE PREDICTED POSITION OF THE CG GENERALLY AGREED WITH THE CP POSITION TO WITHIN A FEW MILLIMETERS, I ONE CASE THE DISPARITY WAS OVER A CM DUE I BELIEVE TO THE UPPER ARRAY REFLECTING OFF OF THE UPPER SURFACE OF THE LOWER SEGMENT.

62. THIS SLIDE SHOWS SOME OF THE DATA FOR THE TEST I JUST DESCRIBE AND IS MAINLY HERE TO GIVE ME A CHANCE TO EXPLAIN THE GRAPHIC.

63. THIS IS THE X DISPLACEMENT DATA FOR A CONTROL SUBJECT DURING NORMAL STANCE. NOTE THAT THE SEGMENTS ARE MOVING TOGETHER WITH LARGER MAGNITUDE FURTHER FROM THE ANKLE JOINT, SOMETHING LIKE AN INVERTED PENDULUM BEHAVIOR MAY BE OCCURING HERE.

64. THIS IS THE CORRESPONDING ACCELERATION DATA. AGAIN IN THIS PARTICULAR CASE ALL SEGMENTS SEEM TO BE MOVING IN SYNCHRONY. PLEASE NOTE THE SCALE OF THE ACCELERATION PLOTS AS THEY WILL CHANGE MARKEDLY AND ARE IMPORTANT TO UNDERSTANDING THE IMPLICATIONS OF THIS DATA.

65. THIS IS DATA A CEREBRAL PALSIED SUBJECT DURING NORMAL STANCE.

66. THIS IS THE CORRESPONDING ACCELERATION PLOT. DESPITE AUTO SCALINGS ATTEMPT TO HIDE THE FACT, THE CEREBRAL PALSIED SUBJECT IS DISPLAYING A LESS WELL COORDINATED ACTIVITY WITH MUCH GREATER ACCELERATIONS OF THE SEGMENT AND COMBINED CGS.

67. THIS IS A 3D GRAPHIC OF THE CEREBRAL PALSIED SUBJECTS STANCE AT ABOUT FRAME 300.

68. THIS IS THE DISPLACEMENT DATA FOR A CONTROL SUBJECT PERFORMING THE RIGHT ARM RAISE MANEUVER, THE COMPENSATORY NATURE OF THE OTHER SEGMENT MOVEMENTS AND THE COORDINATION OF MOVEMENT IS EVIDENT

69. THE CORRESPONDING ACCELERATION PLOTS EVEN MORE CLEARLY HIGHLIGHT THE COORDINATION OF THE RESPONSE. THE FEET DO NOT OF COURSE MOVE, THE ACCELERATION PATTERN OF THE OTHER BODY SEGMENTS IS ALMOST EXACTLY 180° OUT FROM THAT OF THE ARM. THE ONE EXCEPTION IS THE CONTRALATERAL ARM WHICH IS WEEKLY FOLLOWING THE ACTIVE ARM. NOTE THAT THESE TWO SLIDE INDICATE THAT ALMOST ALL BODY SEGMENTS PARTICIPATE IN COUNTERING THE EFFECT OF THE ARM RAISE.

70. HERE A CEREBRAL PALSIED SUBJECT PERFORMS THE RIGHT ARM RAISE MANEUVER

71. THE CORRESPONDING ACCELERATION PLOT.

72. RECALL THE TOTAL VERTICLE FORCE OSCILLATIONS I OBSERVED IN THE CROUCH DEFORMITY SUBJECTS. THIS IS THE FORCE TRACE FOR THE FIRST SUCH SUBJECT WE LOOKED AT. ATTEMPT TO FIND THE KINEMATIC CORRELATE USING THE DATA PROCESSING AVAILABLE AT THE TIME AND THE MODEL AS IT EXISTED THEN; E.G. LOOKING CHANGES IN JOINT ANGLES AND SEGMENT ACCELERTIONS FAILED.

73. THIS HOWEVER IS THE OUTPUT OF THE CG ESTIMATION MODEL FOR THIS DATA SET. RECALLING THAT THE CG ACCELERATION AND GROUND REACTION FORCE SHOULD BE 180° OUT OF PHASE, THE PLOTS ARE SIMILAR. AGREEMENT IS QUITE GOOD CONSIDERING THE FORCE OSCILLATIONS ARE A PERCENT OR SO OF BODY WEIGHT AND THE AREA UNDER THE CG ACCELERATION CURVES REPRESENTS ONLY A FEW MM DISPLACEMENT.

74. IN CONCLUSION MY GOAL WAS TO DEVELOP A TECHNIQUE FOR STUDING POSTURE AND BALANCE. WHILE THERE IS CLEARLY ROOM FOR IMPROVEMENT, I HAVE SUCCEEDED IN DEVELOPING A TECHNIQUE WHICH NOT ONLY PERMITS THE EVALUATION OF STATIC POSTURE AND BALANCE, BUT PERMITS THE COORDINATION OF POSTURAL CONTROL MOVEMENTS TO BE ASSESSED.

75. THANK YOU. AND NOW ARE THEIR ANY QUESTIONS