A BIOMECHANICAL RATIONALE FOR PREFERRED RUNNING STYLE

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B.Sc.(Honours) (Kinesiology) , Simon Fraser University , 1983

A THESIS SUBMITTED IN PARTIAL FULFILLMENT OF THE REQUIREMENTS

FOR THE DEGREE OF

MASTER OF SCIENCE (KINESIOLOGY)

in the School

of

Kinesiology

Simon Fraser University

May 1988

APPROVAL

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DEGREE: Master of Science (Kinesiology)

TITLE OF THESIS: A Biomechanical Rationale For Preferred Running Style

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ABSTRACT

A person, when asked to run, must choose a pattern of motion, a style, in order to satisfy the request. In this investigation the hypothesis that the subject would prefer a style that minimized a specific criterion quantity (**e**), defined in mechanical units, was tested.

Previous researchers have reported that walkers, runners, and race-walkers preferred, at a fixed speed, a stride length associated with near minimal oxygen consumption. The assumption was made that these results reflect a general physiological rationale for the preference for style. The criterion quantity **e** was derived , for the present study, as an analogy to physiological effort. It was defined as the sum of the separate flexor and extensor impulses produced by hip, knee, and ankle torque generators over the period of the running cycle. The torque generators were, in turn, defined with respect to a rigid segment model of the lower limb.

Cine-film and force-platform data were collected from a subject running with his preferred style and with non-preferred styles and submitted for inverse dynamic analysis. At least one non-preferred style required as low a level of **e** as did the preferred style at each of three speeds tested (2.47, 2.86, and 3.52 ms^{-1}). As such the hypothesis was not supported.

Three possible explanations for the results of this experiment are presented. The first is that criteria such as balance, mechanical load, and aesthetics are relevant to the preference for style. The second is that, in the conditions of this experiment, the assumption that physiological effort is minimized is incorrect. The empirical data on which the assumption was based are not inconsistent with alternative explanations. The third is that the derived quantity **e** was an inappropriate model of physiological effort.

Over-simplification of the models of the physiological and mechanical systems is implicated in the third explanation. The influence of factors like the action of two joint

muscles, co-contraction, and the length and velocity dependencies of muscular force on the results could not be assessed directly.

ACKNOWLEDGEMENTS

The author wishes to thank his family, his friends, and his teachers for their support and their tolerance while he led two lives.

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CHAPTER 1
THE BIOMECHANICS OF RUNNING

Introduction

The following is a discussion of the theoretical and empirical motivations for, and the methods, results, and conclusions of, a biomechanical investigation of human running.

A person, when asked to run, must choose a pattern of motion, a style, in order to satisfy the request. On what basis would an individual choose his particular style from the large number which could satisfy the request? Is there a biomechanical rationale for the preferred style? The investigation documented here was undertaken in an attempt to answer these questions.

A criterion quantity, defined in mechanical units, was derived as a simplified model of the physiological resources required for motion. It was hypothesised that the criterion quantity would be minimized in the preferred running style. Mechanical data were collected from a single subject who ran with freely chosen, preferred style and with non-preferred styles. The findings have been discussed with respect to the hypothesis and the limitations of the experimental process.

Biomechanical Research on Running

The number of mechanical and physiological variables which may be of interest to the biomechanist is large. The relationship of kinematic, kinetic, and energetic variables to each other, to physiological measures, and to time have all been investigated under many experimental conditions. Research on the biomechanics of running has been reviewed extensively by Winter (1984), McMahon (1984), and Williams (1985a).

A common research strategy is to compare characteristic functions of the kinematics of running under different conditions. Examples of characteristic functions include stride length, stride frequency, joint and segment angles at specific points in the cycle, and the duration of stance and recovery. For example Kunz and Kaufmann (1981) characterized some systematic and individual differences in several kinematic characteristic functions between elite and less able sprinters. While this investigation quantified style in the two groups studied it neither revealed the reasons for the differences nor identified the characteristics critical to sprint performance.

Newton's laws link the motion of a body to a set of forces which can be said to have caused that motion. It is not surprising therefore to find that running style has been investigated in terms of the kinetic quantities associated with the style itself. Direct measurement of mechanical variables in the study of running style has been limited to kinematic variables and to a few kinetic variables such as external forces and torques. Of the internal forces acting on the body, muscular forces and compressive forces on bone and cartilage have not been measured in situ during locomotion. Komi, Salonen, Jarvinen, and Kokko (1987) and Gregor, Komi, and Jarvinen (1987) have measured achilles tendon forces in human subjects during walking and cycle ergometry. They used surgically implanted force transducers. There are obvious technical and ethical problems inherent in invasive measurement techniques (Andrews, 1982; Crowninshield and Brand, 1982) which preclude extensive use.

Kinetic variables are generally evaluated by performing an inverse dynamic analysis. The laws of motion for a model of the body are written and kinematic data, usually measured directly from a subject, and modelled inertial parameters are used to calculate a set of applied forces or torques. Kinetic variables quantified in this manner are only as valid as the models used. The use of models in biomechanics has been reviewed by Miller (1979) and King (1984).

The similarities of joint torque patterns among sprinters (Mann,1981) and of joint torques and powers among joggers (Winter, 1983) were identified using inverse dynamic analysis. These investigations illustrate some of the subtleties of the laws of motion of the models used for the analysis. The common kinetic characteristics identified may represent strategy for determining style. They may also represent some of the necessary conditions for achieving a running motion. What hasn't been explained by this type of investigation is why some of the differences seen in kinetic patterns within and across speeds, subjects, or other conditions can and do exist. Without complete representations of the laws of motion and the necessary and sufficient conditions for running it isn't possible to separate what must be done in order to satisfy the objective to run from what can be manipulated in subservience to a preference for style.

The relationship between physiological quantities and mechanical variables during running has been investigated directly and indirectly. An example of an indirect approach has been reported by Chapman and Medhurst (1981). They measured changes in several kinematic characteristic functions of the running stride of subjects just subsequent to the start and just prior to the finish of a 400 metre sprint. The assumption was that differences reflected the effects of fatigue. Sprague and Mann (1983) measured changes in both kinematic and kinetic quantities using a similar experimental paradigm. Both of these investigations quantified the idiosyncratic running styles of several subjects and the changes associated with fatigue. Neither explained why the subjects used the particular styles observed in the fatigued or unfatigued condition. Interpretation was further complicated by the fact that running speed changed with fatigue. The changes in style required by the onset of fatigue and its effect on the ability to produce force could not be separated from the changes due solely to the differences in running speed.

The empirical relationship between oxygen consumption and kinematic characteristics of locomotion have been obtained under a number of conditions. Zarrugh, Todd, and Ralston (1974) measured the oxygen cost of walking at various combinations of stride frequency and stride length. At each speed there was a unique stride frequency at which the oxygen cost per unit distance was minimized. Cavanagh and Williams (1982), found that while running with a fixed speed, the time rate of oxygen consumption varied as the stride length was changed. Further, the subjects tended to prefer a stride length which was near the one associated with minimal oxygen consumption. Morgan and Martin (1986) found similar results using trained race-walkers.

The tendency to prefer a stride length near the one associated with minimal oxygen consumption may be indicative of a physiological rationale for preferred style. However, stride length is insuffient to completely determine the motion, the style, of a subject. One can imagine a subject running with the same stride length and stride frequency as in the preferred case but with an exaggerated high knee motion. Another style with the same stride length and stride frequency may require a lower rate of oxygen consumption. It is also clear that while empirical relationships between stride length and oxygen consumption were determined in these investigations the relationship between a complete description of style and oxygen consumption remains unknown.

Andrews (1983) addressed the theoretical relationship between kinetic and physiological quantities. He argued that net joint torques are a reasonable instantaneous measure of muscular effort and, indirectly, of metabolic cost for a wide range of human activities. Andrews' argument can be extended. It could be hypothesized that the total metabolic or physiological cost of a running stride might be some function of the time integral of the applied forces or torques over the period of the stride.

In a series of articles Williams and Cavanagh (1983), Cavanagh and Kram (1985), and Williams (1985b) took an alternate approach to the assessment of the metabolic cost of running. They attempted to quantify the transduction of metabolic energy to mechanical energy. An implication of their work is that the efficacy of movement patterns in locomotion could be quantified in terms of the efficiency of the transduction of physiological to mechanical energy. A further implication is that the transduction process and the motion of the body are inter-dependent. These authors identified a number of problems in their attempt to calculate mechanical work, mechanical power, and efficiency from the fluctuation of mechanical energy levels. Their calculations required certain assumptions concerning the mechanical laws for linked rigid bodies, the metabolic cost of muscular work, the influence of mechanical elements not included directly in their model, and the signal-to-noise ratio of their data. A number of conceptual and experimental problems need to be solved before this method can be used to analyze and rationalize human running style.

Some recent discussion of the methodology of biomechanical research is relevant to the study of running style. Williams (1985a) identified several factors which may tend to limit the validity of biomechanical investigations of running. He suggested that restricting the number of subjects or the range of speeds covered can reduce the validity. Repeating an experiment at different speeds would certainly be useful. However, the comparison of mechanical quantities across speeds can be difficult to interpret. Changes which result from the change in speed would have to be separated from those which are volitional or which result from a change in force producing ability. Collapsing or comparing mechanical data across subjects suffers from a similar problem. How are the differences which must exist because the subjects are physically different to be separated from volitional differences or differential degradation of force producing ability?

Cappozzo (1983) and Winter (1987) have discussed the use of speculation and hypothesis testing in biomechanical research. Cappozzo advocated more use of the classical method of scientific investigation while Winter stated that much could be learned without the testing of specific hypotheses. Winter argued that specific findings could be applied to the general case because the biomechanical system is deterministic. In fact different subjects or even a single subject can achieve the same objective, to run at a specific speed for example, in many different ways. This suggests that there is some room for variability despite the deterministic laws of motion. The use of models in biomechanical research by Winter and others carries with it the hypothesis that those models are valid representations of the human body. A rational scientific process might be to test the hypothesis that a specific finding applies in other conditions.

Several authors have advocated general goals for the biomechanical research of locomotion of which the study of running is a sub-discipline. Invariably these generalized goals for research are motivated by general theories of locomotion.

Biological systems have been characterized by Hatze (1984) as having teleological tendencies;

> "...these (bio) systems attempt to achieve the envisaged goal
> in the best possible way, according to some criterion and
> under given constraints."

The value of considering biological behaviour in this manner is that a scientific objective is immediately apparent, the identification of the criterion, the measure of the 'best possible way'. The identification of a quantity that can serve as a criterion would provide evidence for a teleological theory of human movement.

Cappozzo (1983) discussed the evaluation of clinical gait and advocated a similar approach. He felt that the identification of strategies for movement rather than of patterns of biomechanical variables was more relevant to the understanding of locomotion;

> "...basic biomechanics research must overcome the stage
> where it supplies information about how man walks and
> begins to answer the relevant whys."

The biomechanical investigation of running style documented in this thesis was motivated, in part, by the research discussed above. An outline for the present investigation is given below along with a guide to the rest of this document.

Hatze's teleological view of locomotion was accepted. In Chapter 2 the argument for this view is stated in a symbolic, mathematical form. It is derived on the basis of a simple thought experiment and with the formalisms of mathematical optimization.

The possibility that there is a physiological rationale for style is used as a basis for deriving the criterion quantity. The physiological and metabolic systems were dealt with in a symbolic manner only. This investigation did not test directly the assertion that the criterion quantity is a valid representation of the physiological and metabolic processes underlying running.

In Chapter 3 a biomechanical investigation of running style is documented. The hypothesis that a specific criterion quantity, defined in mechanical units, would be minimized in the subject's freely chosen style was tested. The criterion quantity was derived from kinetic rather than kinematic quantities. This was done to take advantage of the complete generality of Newton's laws; the motion of the body must be consistent with these laws. The criterion quantity was defined as a function of the integrals, with respect to time, of the applied torques, over the period of the running cycle. The details of the model used, of the derivation of the criterion quantity, of the methods used, and of the results and discussion of the investigation are also presented.

A single subject was used. In this way the constraints on the movement were held constant in all conditions. Different subjects would have had different abilities to produce force, different inertias, and different ranges of movement.

The procedure was repeated and the hypothesis tested separately at each of three different speeds. At each speed stride length and stride frequency were held constant. The results of the test of the hypothesis rather than the biomechanical quantities themselves were compared across speeds.

Chapter 4 contains a discussion of the relevance of the conclusions of the investigation in light of the limits of the scientific process used to obtain them.

CHAPTER 2

OPTIMIZATION THEORY AND PREFERRED RUNNING STYLE

The Locomotion Problem

Accepting Hatze's teleological theory of locomotion is useful for two reasons. The first is that an immediate scientific objective, the identification for the criterion for movement, is apparent. The second is that, as shall be argued shortly, satisfying an instruction to run is mathematically equivalent to teleological behaviour.

Consider a simple thought experiment. A subject is asked to run down a path with a specified stride length and stride frequency. Within these objectives the subject has complete freedom to choose his preferred style. In fact, the request presents the subject with a problem. He must somehow choose or determine his style from the infinite number of possibilities which could satisfy the request. The request to run and the problem it presents will be referred to as the locomotion problem (L).

The locomotion problem can be restated in the form of a mathematical optimization problem. The body is a system, the configuration of which has n degrees of freedom q_i

(i=1,...,n). The configuration of the body can be represented with the vector X where

$$X = (q_1, q_2, ..., q_n)^T .$$

The motion of the body over time can be designated as X(t) where

$$X(t) = (q_1(t) , q_2(t) , ..., q_n(t))^T .$$

The request to run, L, is analogous to the objective defined for an optimization problem. It is, in principle, defined in terms of the motion of the body;

$$L = L(X(t)) .$$

L can be thought of as a system of equations defining the necessary and sufficient conditions for satisfying the request to run. As long as the system L has fewer than n equations the request is insufficient to determine fully the running style.

To define quantitatively a request to run is more complex than might appear. As such the system of equations L, has not been written explicity. A few properties of L can be deduced however. The first and most obvious is that running is a cyclic activity. It has undoubtedly been observed that a subject does not run with exactly uniform motion cycle after cycle. For the purposes of this discussion however, only the motion of the subject over the period of a single typical cycle shall be considered. As well, style shall be defined as the motion, X(t), of the subject over the period of a typical running cycle. The second important property is that L is insufficient to determine completely the style of the subject and therefore consists of less than n equations.

A set of constraints (C) are also defined in an optimization problem. The geometry of the musculoskeletal system acts to limit the motions of which a subject is capable. There are also limits on the ability of the subject to produce force. The constraints can be represented as a system of equalities and inequalities and defined in the units of X and time;

$$C = C(X,t) \geq 0.0 \ .$$

As with L, the constraints are presented only in symbolic form. Equality constraints and the locomotion problem act together to reduce the order of system X. Despite this there is still room, within the inequality constraints, for subservience to a preference for style.

For optimization problems involving dynamic systems a control function is defined. The behaviour of the system is said to be caused by the action of the control function. There are two ways to view the control of human running style. The first is embodied in a mathematical system called the equations of motion which are, in turn, derived from Newton's second law. This law states that the motion of any body, animate or inanimate, can be described as having been caused by the action of a set of forces (F). The second view is that human motion is caused by the action of physiological, neurophysiological, and metabolic processes within the body. For the purposes of this discussion the sum of these processes will be referred to as the physiological system. As will be discussed shortly, combining mechanical and physiological causal models of human motion is an important step in deriving a relevant hypothetical rationale for preferred running style.

The subject's preferred running style, his motion over the period of the running cycle, is analogous to the solution of the optimization problem. From the investigator's perspective it is not the solution which is of interest. Obtaining the solution is a matter of observing the subject's preferred style. It is the criterion quantity, the variable or function whose value is to be optimized in determining the solution, which is the scientific objective.

The goal of this investigation was to identify a criterion function which could explain the subject's preference and which is relevant to what is already understood about human running. The hypothetical criterion quantity, referred to as effort and designated as **e** , was to be stated in the units of X and time. It was defined a priori as that quantity whose minimization could explain the subject's preference for style.

The Criterion Function

The optimization framework calls for a criterion quantity defined in the mechanical units of the motion of the body. In contrast to this there is the strong intuitive and empirical relationship between human motion and the action of the physiological system. For example the empirical relationship between stride length and oxygen consumption provides indirect evidence for a physiological rationale for style.

There is an apparent conflict between a possible physiological rationale and a theoretical mechanical rationale for a preference for style. The first rationale is, in principle, defined in the units of the physiological system while the second is, in principle, defined in mechanical units.. This conflict can be resolved by accepting that the subject, a biological system, consists of separate but interacting mechanical and physiological subsystems. The corollary to this is that some transformation exists between the coordinates of the two subsystems; action by one is evidence for the action of the other. A rationale defined for one subsystem will have an equivalent representation in the other.

The empirical evidence for a physiological rationale for preferred style should, in principle, be helpful in finding the criterion quantity **e** . In the following sections of this chapter the transformation between the physiological and mechanical subsystems is considered and a theoretical general form of the criterion quantity is derived. The first step in the process is to consider the mechanical system in isolation. A completely general teleological law of motion, Newton's second law, provides a causal model of human running. The second step is to develop a teleological model of running incorporating the physiological system. It is drawn in analogy to the mechanical laws. The third step is to combine the two causal models to derive a hypothetical form of the criterion quantity.

All motion, including human running, is consistent with Newton's Laws of which there are many equivalent formulations. Newton's second law for a body with n degrees of freedom can, in principle, be written as a system of n second order differential equations. Further, these equations can be written in such a way that the forces of constraint can be ignored. The general form of the laws of motion for the human body can be written as

$$F_q = \mathbf{M}X'' \ .$$

The term X'' is the second derivative, with respect to time, of the position of the body. The inertia of the body, with dependencies on position and on constant inertial parameters is represented by the matrix **M**. The term F_q represents a set of applied forces which act in parallel to the coordinates q_i defined for X.

Any motion in the coordinate system X can be said to be the solution of the differential system described above. As such that motion can be said to have been caused by the application of the forces F_q. The solution of a differential system is also dependent on the initial conditions $X'(t_0)$ and $X(t_0)$. Thus the motion of the body during the running cycle can be described as a function of the applied forces F_q and of the initial conditions;

$$X(t) = X(F_q(t), X'(t_0), X(t_0)) \ .$$

The next step is to consider the role of the physiological system in determining the preference for running style. The empirically derived relationships between oxygen consumption ($V'O_2$) and stride length (λ) provide a starting point. For a particular subject this can be represented symbolically as

$$V'O_2 = V'O_2(\lambda) \ .$$

Stride length in turn can be represented as a function of the coordinates X;

$$\lambda = \lambda(X) \ .$$

Knowledge of the stride length alone is not enough to determine completely the motion X(t) even at a fixed speed. For this reason (X) cannot be used to transform $V'O_2(\lambda)$ and the relationship between oxygen consumption and style,

$$V'O_2 = V'O_2(X(t)) \ ,$$

remains unknown.

There are two consequences of this lack of understanding. The first is that the existence of styles other than the preferred which have the same stride frequency and stride length but which require lower levels of oxygen consumption cannot be ruled out. This possibility leads to the second consequence. Physiological quantities other than $V'O_2$ may be relevant to the subject's preference for style.

It is apparent that the minimization of oxygen consumption may or may not account for the preferred style. The empirical evidence cited earlier does, however, provide a compelling argument for the existence of a physiological minimization principle for the determination of locomotion patterns. As a basis for deriving a hypothetical criterion quantity it has been assumed that a subject chooses his running style so as to minimize some as yet undetermined physiological quantity. The unknown quantity can be referred to as physiological effort (E).

There are many physiological quantities which could be relevant to human movement in general. Consider the system P;

$$P = (p_1 , p_2 , ..., p_v)^T$$

where v is the number of relevant physiological variables. The action of the physiological system over time can be designated P(t). Hypothetical physiological effort (E) can be represented as an as yet unknown function of the relevant physiological variables;

$$E = E(P) .$$

The next step is to consider the relationship between the action of the P system and the motion of the body. Assume on the basis of the undeniable link between physiological processes and human motion that running is <u>caused</u> by the action of the P system. Assume that the relationship is dynamic analogous to the mechanical laws of motion. The relationship of the motion of the body to the action of the P system will be considered to be a second order differential system analogous to the mechanical laws of motion;

$$X''(t) = X''(P(t)) .$$

The motion of the body is then the solution of this differential system. The motion can therefore be described as a function of P(t) and of the initial conditions $X'(t_0)$ and $X(t_0)$;

$$X(t) = X (P(t) , X'(t_0) , X(t_0)) .$$

The forms of two causal models of human motion have been derived, the first is mechanical;

$$X(t) = X(F_q(t), X'(t_0), X(t_0)) ;$$

the second is physiological;

$$X(t) = X(P(t), X'(t_0), X(t_0)) .$$

Assume that these relationships are such that the variables $X'(t_0)$ and $X(t_0)$ can be eliminated. It would then be possible, in principle, to write an equation describing the forces as a function of the action of the physiological system;

$$F_q(t) = F_q(P(t)) ..$$

The relationship

$$E = E(P)$$

can, in principle, be transformed using the relationship describing the forces F_q as a function of P;

$$E(t) = E(F_q(t)) .$$

It is possible to deduce the existence of this relationship on the basis of assumptions and laws already mentioned. If E, the physiological criterion for style, exists it must be describable as a function of motion X(t) . Every motion, including running style, can be said to have been caused by the action of a set of forces $F_q(t)$ in compliance with Newton's second law. It must be true , therefore, that some relationship exists between the physiological effort required for motion and the set of forces that cause it.

The value E can be re-written as a definite integral over the period of the running cycle;

$$E = \int_{t_0}^{t_f} E'(\,F_q(t)\,)\, dt \ .$$

This relationship defines the form of physiological effort. The hypothetical physiological rationale for style is that the motion will be such that this definite integral is minimized over the period of the running cycle. Because the transformation relating the physiological system to the mechanical system is unknown this integral cannot be evaluated directly. While the validity of the integral E is not tested directly in this investigation its form, and the assumptions inherent in it, provide a perspective by which the relevance of the findings can be discussed.

In the next chapter the derivation of the quantity, **e** , an approximation of physiological effort defined in mechanical units , is described. An investigation designed to test the hypothesis that **e** is minimized in the preferred style is documented.

CHAPTER 3

A BIOMECHANICAL INVESTIGATION OF RUNNING

Introduction

The theoretical and empirical analyses of running style discussed in the first two chapters provided the motive for the investigation described in this chapter. A number of the terms and concepts which are described below had to be defined prior to performing the experiment. In each case the definition reflects a trade-off between validity and experimental tractability. In this chapter the results are discussed within the limitations of those definitions . A more general discussion of the validity of the investigative process and its impact on the conclusions is presented in chapter 4.

The remainder of this chapter is organized as follows. A mechanical model of human running is defined. Included in this are definitions of the configuration of the body (X), the inertia of the body ($\mathbf{M}$), and the generalized applied forces acting on the body (F_q). The laws of motion for the body are outlined. A locomotion problem, L, is defined in words and presented symbolically. A procedure for testing the hypothesis that a criterion quantity is minimized in the subject's preferred style is outlined. A second locomotion problem, one which results in the subject performing non-preferred styles is also defined. The derivation of the criterion quantity **e** is presented and the hypothesis stated explicitly. Sections detailing the methods used and the results of the experiment are then presented. The final section includes a discussion of the results and the conclusions that have been drawn.

The Model

A coordinate system was needed to define the configuration of the body and its position in the laboratory frame of reference. Two factors bearing on the choice were mathematical tractability and biological significance. The coordinate system had to be algebraically manageable yet some relationship to the musculoskeletal geometry was desired. The trade-off is best explained with an example. The length and orientation of a muscle are more directly reflected in the angle of the joint it crosses than in the angle in space of the segments of origin or insertion. Designation of the degrees of freedom in terms of joint angles rather than segment angles is preferable.

Several different, equivalent coordinate systems were used for the technical execution of this experiment. For the purpose of posing the hypothesis and for discussing the results the coordinate system shown in figure 3-1 can be used.

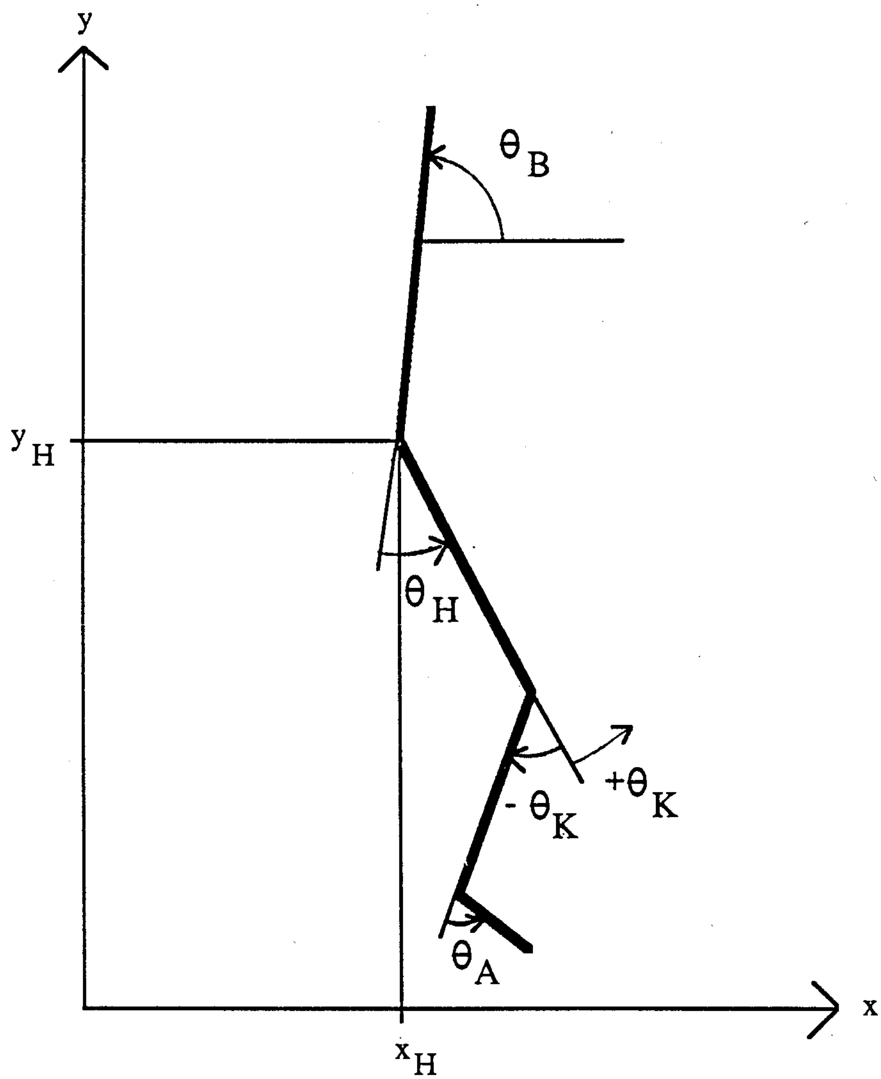
The model in figure 3-1 comprises the foot, the shank, and the thigh of one of the lower limbs and a single segment representing the upper body. The reference vector H,

$$H = (x_H, y_H)^T ,$$

refers to the position of the hip joint in the plane of progression. This is the plane in the laboratory frame of reference (LAB F-O-R) within which the limb is considered to move. The angle of the upper body (θ_B) is defined with respect to the horizontal (x) axis of the

Figure 3-1

Coordinate System (X)



LAB F-O-R. The angle of the thigh with respect to the upper body (θ_H) is representative of the hip joint angle. The angle of shank with respect to the thigh (θ_K) is representative of the knee joint angle. The angle of the foot with respect to the shank (θ_A) is representative of the ankle joint angle. The configuration of the model and its position in the LAB F-O-R can be specified with this set of six coordinates;

$$X = (\theta_B , x_H , y_H , \theta_H , \theta_K , \theta_A)^T \ .$$

There are several important assumptions implicit in the definition of X. The segments are assumed to be rigid; they can neither bend nor change length. This model is more than simply structural; it is a functional definition of the degrees of freedom within which running occurs. This is not to deny that the other parts of the body are necessary for running. An assumption of the investigation is that running style can be characterized and understood in terms of the sagittal plane motion of the four segments and three joints in the model. The implications of the narrow definition of the mechanical degrees of freedom of the body in running are discussed in the fourth chapter.

The inertia of the body, represented as the matrix **M**, is dependent on the position of the body (X) and on a set of constant inertial parameters characteristic of each segment. The segmental inertial parameters are obtained from a set of regression equations in which the segmental parameters are defined as a function of some easily measured parameters, total body weight and segment length. The conclusions of this investigation are subject to the errors of the dissection experiments from which the regression equations were obtained.

The final term to be defined in the laws of motion for the model is the set of forces F_q. A generalized force is associated with each coordinate;

$$F_q = (\tau_B , F_x , F_y , \tau_H , \tau_K , \tau_A)^T .$$

Torques (τ) act parallel to the angular coordinates and linear forces (F) parallel the linear coordinates.

With a complete set of independent forces F_q arbitrary motion is possible (ignoring any constraints on the forces). It is clear, however, that humans aren't capable of arbitrary motion. A person could not move across a room under his own power without flexing and extending at least one joint. This is because humans are, in a sense, mechanically underdetermined. Humans do not have muscles attached to the LAB F-O-R and therefore will have at least one degree of freedom within which no force can be produced. In the case of the model in figure 3-1 there are no muscles directly controlling the angle of the upper body with respect to the horizontal. As such

$$\tau_B = 0.0$$

The linear forces F_x and F_y, while defined with respect to the LAB F-O-R, do not present quite the same case. The contralateral limb, the one not included in the model, can produce forces at the hip. The ipsilateral leg has a similar action on the motion of the contralateral leg. The other independent generalized forces acting on the model are torque generators acting at the hip (τ_H), knee (τ_K), and ankle (τ_A) joints.

If the action of generalized forces in the model are to be representative of muscular function then two important functional characteristics are absent. The first is co-contraction. Where two or more muscles may cross a single joint only a single torque generator is represented. Antagonist muscular contraction and its potential influence on style cannot be accounted for. Within a single degree of freedom human motion is overdetermined. The second phenomenon that is unaccounted for is the action of muscles that cross more than one joint, several of which are present in the lower limb. These muscles tend to make the torque at adjacent joints dependent on each other, a dependency that can be hidden by antagonistic and agonistic co-contraction. The impact of this inadequacy is presented in the discussion of the results further in this chapter.

Given the terms F_q, M, and X, it should be possible to write the equations of motion for the model in figure 3-1 in the form,

$$F_q = MX'' ,$$

without an explicit representation of the ground reaction force. For the purposes of this investigation writing the equations of motion as a function of the forces F_q alone was disadvantageous. While specifying the forces F_q and the accelerations X'' is straightforward the specification of the matrix **M** is not. The inertia term has a complex dependence on X, the position of the system. A second disadvantage is that modelling the upper body as a rigid segment is probably less appropriate than doing so for the limb segments. To use a more realistic model of the upper body would greatly increase the complexity of the equations of motion.

An inverse dynamic analysis was required to calculate the joint torques. It was derived by modelling the segments as separate but interacting bodies and using Newton's third law, the so-called law of action and reaction. The torques are represented in the equations of motion as functions of the movement of the segments in the LAB F-O-R and, during stance, of the ground reaction forces. There are several advantages to this method.

It is computationally simple. The assumption of rigidity is limited to the segments of the leg where it is most appropriate. The hip joint torque can be calculated without measuring the hip joint angle directly.

The Locomotion Problem

The purpose of this investigation was to rationalize a subject's preference for style in response to a request to run. A request to run that fails to specify completely the motion of the body presents the subject with a problem. The locomotion problem presented to the subject in this investigation consisted of a set of specific objectives to attain. The subject was asked to run with a specific stride length along a path in the laboratory. This instruction can be broken down into two subsets of objectives; the instruction to run and the instruction to use a specific stride length. The instructions were given to the subject verbally. Tape was used to mark the desired stride length on the floor . The instruction to run can be represented symbolically, and in words, as

$$R = (\text{'run'}) \ .$$

The requested stride length (λ_r) can be represented in a similar manner;

$$\lambda_r = (\text{'use the stride length marked on the laboratory floor'}) \ .$$

The two components combined together represent the locomotion problem;

$$L = (R , \lambda_r)^T \ .$$

It was with respect to the subject's running style in response to this request in the experimental setting that the hypothesis was posed. The hypothesis that a criterion quantity **e**, the form of which is described in a subsequent section, is minimized in the preferred style was tested at three stride lengths, each resulting in a different running speed. In each case the subject was free to use any running style. The subject was free to choose his stride frequency and therefore his running speed. Styles with different combinations of stride length and stride frequency but at the same speed were not tested. As such the levels of **e** in those styles are unknown. The satisfaction of the hypothesis would explain the subject's preference for style in comparison to the possible styles at the specific combination of stride length and stride frequency tested. The possibility that styles with other combinations of stride length and stride frequency may require lower levels of **e** cannot be ruled out.

A Test of the Hypothesis

Given a locomotion problem and a measure of effort, how was the hypothesis that **e** is minimized in preferred style to be tested? This question was answered in two stages. The first was to develop an experimental strategy for assessing the level of **e** in the preferred style. The second was to devise a method for assessing whether or not there were other styles with lower values of **e** .

Human motion is inevitably variable in nature. This is true in well learned and in novel movements. A measure of the level of effort in preferred style would have to reflect the possible variation. No particular distribution was assumed beforehand; relatively simple measures of central tendency and dispersion were used. The subject was presented with the locomotion problem a number of times. The effort levels (**e**) evaluated for each of these preferred style trials were summarized as a sample mean ($\underline{\mathbf{e}}^{\underline{PS}}$) and sample standard deviations (S^{PS}). A confidence interval ranging from 2.0 standard deviations below the mean **e** to 2.0 standard deviations above the mean **e** was established;

$$[\, \underline{\mathbf{e}}^{\underline{PS}} - 2.0\, S^{PS} \,] \;\leq\; \mathbf{e}^{PS} \;\leq\; [\, \underline{\mathbf{e}}^{\underline{PS}} + 2.0\, S^{PS} \,] \;.$$

A level of **e** inside this region was considered equal to that of the preferred style. If the levels of **e** in the preferred style condition were found to be distributed normally the confidence interval would include 95% of the PS trial results.

In principle it should be possible to derive analytically the style that minimizes **e** and compare it to the subject's preferred style. The advanatage of an analytical method is that the style that minimizes a particular objective function can be identified directly. The solution of an optimization problem for a non-linear dynamic system is a large, complex computational problem. The numerical analysis required is subject to error. Perhaps the greatest disadvantage is that the system dynamic equations

$$F_q = MX'' ,$$

the locomotion problem

$$L = (R , \lambda_r)^T ,$$

and the constraints

$$C = C(X , t) \geq 0.0 ,$$

all need to be written explicitly.

The hypothesis that **e** is minimized in the preferred style implies that there are no other styles which require levels of **e** within or below the confidence interval. Empirical methods involve a search for such a style in an attempt to prove the null hypothesis. As such empirical methods are less definitive.

A set of non-preferred styles could conceivably be generated using some algorithm. A computer based synthesis could produce large numbers of examples for which the levels of e could be calculated and compared. However, the equations for F_q, L, and C would be needed in order to ensure that the derived styles satisfied the objectives and did not violate the constraints.

The method chosen for this investigation was to have the subject generate data for both his preferred style (X^{PS}) and a selection of non-preferred styles (X^{NP}).

Non-preferred style is a behaviour which cannot be defined before the fact. A strategy for inducing the subject to use a non-preferred style was devised. In the non-preferred trials the subject was presented with new locomotion problems (L^{NP}) that consisted of the original locomotion problem (L) and two additional conditions. A request to run with the same stride frequency (f_r) as in the preferred condition and an inequality condition. The inequality condition asked the subject to perform aspects of the movement (γ) to a greater or a lesser extent than in the preferred style. The case where γ is to be greater than normal can be represented symbolically as

$$\gamma^{NP} > \gamma^{PS} \ .$$

In both the preferred style and the non-preferred style trials the investigator's judgement was to be used to decide whether or not the subject had run as oppossed to walking or hopping or some other motor behaviour.. The degree to which the subject had achieved the stride length was assessed quantitatively from the position data collected during the experiment. The degree of difference with which the subject had performed the inequality objective was assessed subjectively from the joint angular position data.

The major disadvantage of the experimental procedure was that only a few examples of non-preferred styles were generated for comparison with the preferred style. It was very difficult to vary systematically any particular aspect of the running style.

The Derivation of Effort (e)

A hypothetical physiological rationale for preferred running style is the minimization of a physiological quantity. This quantity is defined as a scalar valued definite integral;

$$E = \int_{t_0}^{t_f} E'(F_q(t))\,dt \ .$$

Consider the quantity E' at a time t_d in the interval between t_o and t_f;

$$E'(t_d) = E'(F_q(t_d)) \ .$$

In words this equation states that, at the time t_d, the rate at which physiological effort is being consumed can be described as some function of the forces being applied to cause the motion of the body.

In a mathematical and mechanical sense there are many equivalent systems of forces that are capable of causing a particular motion of a particular body. In a biological sense, however, there is a particular set of forces that can be varied to control the motion of the body, the set of muscular forces (F_m). Each muscle, in turn, is a consumer of physiological effort. The rate at which effort is being consumed by the j^{th} muscle is some function of the force being produced by that muscle;

$$E'_j(t_d) = E_j'(F_j(t_d)) \ .$$

The amount consumed over the period of the cycle by the j^{th} muscle is equal to the integral of E_j over the period;

$$E_j = \int_{t_0}^{t_f} E_j'(F_j(t))\, dt \ .$$

The total amount of effort consumed to cause the particular running style is the sum of the amounts consumed by the set of muscles;

$$E = \sum_{j=1}^{m} E_j \ ,$$

where m is the total number of muscles.

Reconsider the relationship between the rate at which a muscle consumes physiological effort and the force which it produces. At the time t_d there exists some factor g_j such that

$$F_j(t_d) = g_j(t_d)\ E_j'(t_d) \ .$$

The unknown factor g_j quantifies the transformation of physiological effort to muscular force. Because it is representative of the coupling of the physiological to the mechanical system it may have dependencies on both mechanical and physiological variables. The first and simplest approximation of the factor g is assume that g_j is constant over the period of the cycle. As such the muscular force is assumed to be directly proportional to the rate of consumption of physiological effort;

$$E_j'(t) \propto F_j(t) \ .$$

A model ($\dot{e}_j$) of the physiological effort required to produce a muscular force can then be defined as

$$\dot{e}_j(t) = F_j(t) \ .$$

This equation preserves some intuitively reasonable properties of the relationship between physiological effort and motion. The rate of consumption of modeled effort rises as the muscular force rises. If the force drops to zero the consumption also drops to zero. As such basal physiological processes are excluded from the assessment of effort required for movement. Isometric contractions which require physiological effort but with which no displacement is associated are accounted for.

The effort required by the muscle over the period of the cycle is equal to the integral of e' with respect to time. The muscular force F_j can be substituted for $\mathbf{e}'$;

$$e_j = \int_{t0}^{tf} F_j \, dt \ .$$

The total effort consumed in causing the motion of the body is equal to the sum of the effort consumed by each muscle;

$$\mathbf{e} = \sum_{j=1}^{m} \mathbf{e}_j \ .$$

As defined above the quantity **e** qualifies as a model of physiological effort defined in mechanical units. However, it is stated in the coordinates of muscle rather than in the joint angle coordinates defined previously for the model of the body. As such it cannot, in its present form, be used with that model for the analysis of running style. A straight forward mathematical transformation between the coordinates in which muscles act and the joint angles defined for the model is not available. The only recourse is to consider each generalized force generator as the model's equivalent of muscle. This assumption places a limitation on the validity of the investigation. It also illustrates the price that must be paid for using an inverse dynamic analysis. This issue is discussed more fully in chapter 4. The adoption of the muscular effort model to the mechanical model is discussed next.

The generalized forces associated with each coordinate q in X include the torques τ_H, τ_K, τ_A, and τ_B, and the linear forces F_X and F_Y. The torque associated with the non-muscular coordinate θ_B was determined to be zero at all times;

$$\tau_B = 0.0 \ .$$

As such no effort is consumed;

$$E_B' = 0.0 \ .$$

The linear forces acting at the hip joint, F_X and F_Y, are reaction forces caused by the motion and inertia of the contralateral leg. As they are not direct reflections of muscular force the rate of consumption of **e** associated with them is set to be zero;

$$E_X' = 0.0 \ ;$$

$$E_Y' = 0.0 \ .$$

The ipsilateral leg has a similar effect on the contralateral. In accounting for the physiological effort associated with the torque generators in the muscular coordinates the effort required to produce F_X and F_Y is included implicitly.

The individual torque generators acting at the hip, knee and ankle joints are capable, as presently defined, of producing both flexion and extension torques. Human muscles are only capable of producing tension in one direction and therefore separate sets of muscles are needed at each joint. The model being used in the investigation was given that property simply by considering the extensor and flexor torques as having been created by separate torque generators. At the time t_d the joint torque τ_j is either flexor or extensor; the model does not have the property of co-contraction.

Six separate sources of torque, a flexor and extensor (k = 1,2) at each of the three joints (j = 1,3), each a separate consumer of effort, were designated;

$$e_{jk}' = e_{jk}'(\tau_{jk}) \ .$$

e_{j1}' is the rate at which **e** is consumed by the torque producer acting in the negative direction at the j^{th} joint and e_{j2}' is the rate of consumption of **e** by the torque producer acting in the positive direction. Over the period of a stride cycle the total effort consummed by each torque generator is;

$$e_{jk} = \int_{t_0}^{t_f} e_{jk}' \, dt$$

The physiological cost of a flexor contraction and of an extensor contraction both represent a positive physiological effort. In analogy, the total mechanical effort over the period of the cycle was defined as the sum of the moduli of the efforts e_{jk};

$$e = \sum_{j=1}^{3} \sum_{k=1}^{2} | e_{jk} | \ .$$

The function **e** is not a valid mechanical variable <u>per se</u>, but rather a function of mechanical variables meant to reflect the physiological resources required for motion.

Before continuing with the discussion it is of value to digress slightly and consider mechanical work, a possible form of hypothetical effort which was not investigated. The relationship between mechanical work and motion is, in the classical mechanical sense, a fundamental consequence of Newton's second law. The coupling of the physiological to the mechanical system and the efficacy of that coupling in producing motion can be considered in terms of the transduction of physiological energy stores to mechanical work. Despite this there are drawbacks to the use of mechanical work as a reflection of effort.

A model of physiological effort was defined previously as the integral of muscular force with respect to time. This is, in mechanical terms, the mechanical impulse produced by the muscle. The mechanical work performed by the muscle is, on the other hand, the integral of muscular force with respect to displacement;

$$W_j = \int_{q_0}^{q_f} F_j \, dq_j ,$$

where q_j is the coordinate within which the muscle acts. The total work done by the muscles in causing running motion would be the sum of the work done by each of the n muscles acting;

$$W = \sum_{j=1}^{m} W_j .$$

A problem is immediately apparent. In uniform cyclic motion the mechanical energy of the body is equal at the beginning of each cycle. The net work done over the period of the running cycle is zero by definition. Because this is equally true for preferred and non-preferred styles the amount of mechanical work cannot be used to distinguish between them.

Muscles increase the mechanical energy of the body with concentric contractions and decrease it with eccentric contractions. Both types of contraction require the consumption of physiological effort. A reasonable model of effort might, therefore, be the sum of the absolute value of the differential work done over the cycle as a model of effort;

$$W_j = \int_{t_0}^{t_f} \left| F_j(t)\, dq_j(t) \right| ;$$

$$W = \sum_{j=1}^{m} \left| W_j \right| .$$

This value would not, in general, be zero in uniform cyclic motion.

Another conceptual problem with the mechanical work model of effort becomes apparent if the differential work term is re-written;

$$F_j\, dq_j = F_j\, q_j'\, dt \ .$$

The term

$$F_j\, q_j'$$

is the mechanical power produced by the muscle and is, according to the mechanical work model, equal to the rate of consumption of effort ($\mathbf{e}'$). If the velocity within which the muscle acts (q_j') is zero then the rate of consumption of **e** drops to zero. This would be true even if a force was being produced. Isometric contractions would be deemed not to require effort, clearly an unreasonable finding.

The exclusion of isometric effort is not limited to the simple work model of effort. It is a property of any model which is dependent on the force produced and is integrated over displacement. This can be illustrated by writing the definite integral for physiological effort for the j^{th} muscle;

$$E_j = \int_{t0}^{tf} E_j'(F_j)\, q_j'\, dt$$

It is the dependence on q_j' rather than the form of $E_j'(F_j)$ that restricts the validity of the model of effort. It is for this reason that mechanical impulse rather than mechanical work was chosen as the basic mechanical model of physiological effort.

The Hypothesis

This experiment was designed to test the suitability of a hypothetical criterion quantity as the basis by which preferred style could be determined. To qualify as a suitable criterion that quantity must be minimized in the preferred style. To find that a particular non-preferred style requires levels of the hypothetical quantity equal to or less than that found in the preferred style would render it unsuitable.

The subject was asked to run repeated trials with both his preferred and non-preferred styles at three different combinations of stride length and stride frequency. The hypothesis (H_1) to be tested separately at each combination of stride length and stride frequency, was that effort (**e**) is minimized in the preferred style;

$$H_1: \ \mathbf{e}^{PS} < \mathbf{e}^{NP} \quad \text{for } \underline{\text{all}} \text{ NP} .$$

The alternative, null, hypothesis (H_0) is that at least one NP style requires a level of **e** less than or equal to that required in the preferred style;

$$H_0: \ \mathbf{e}^{PS} \geq \mathbf{e}^{NP} \quad \text{for } \underline{\text{any}} \text{ NP} .$$

Methods

Subject

The subject was a 26 year old male. He was a competitive, international class distance runner. He was 1.74m in stature and had mass of 62.5 Kg.

Inertial Parameters

The regression equations used to evaluate the segmental inertial parameters were obtained from Winter (1979) who had collated them from several sources. The segments themselves were defined by the proximal and distal anatomical landmarks listed in table 3-1. The various parameters were defined as fractions of either the total body mass or the length of the segment in question as shown in table 3-2.

Laboratory Frame of Reference

A schematic diagram of the laboratory is shown in figure 3-2. The path along which the subject ran was approximately 13m long. It was possible for the subject to complete 5 or 6 (depending on the running speed) stride cycles on the path. For an investigation at running speeds faster than those reached in this investigation a longer path would be necessary.

A force-platform was situated approximately 8m down the running path. A cine-film camera was mounted on a tripod with the lens centered 1.0m above the ground and 5.50m from the center of the force-platform on a line normal to the running path.

The force-platform was a KIAG SWISS type 9261A (Kistler Instrumente A.G., Winterthur, Switzerland). Charge amplifiers and summing amplifiers yielded six channels

Table 3-1

Segment Endpoints

SEGMENT	PROXIMAL ENDPOINT	DISTAL ENDPOINT
foot	lateral malleolus (ankle)	2.0 cm above sole at distal end of foot (toe)
shank	lateral femoral epicondyle (knee)	ankle
thigh	greater trochanter (hip)	knee

Table 3-2

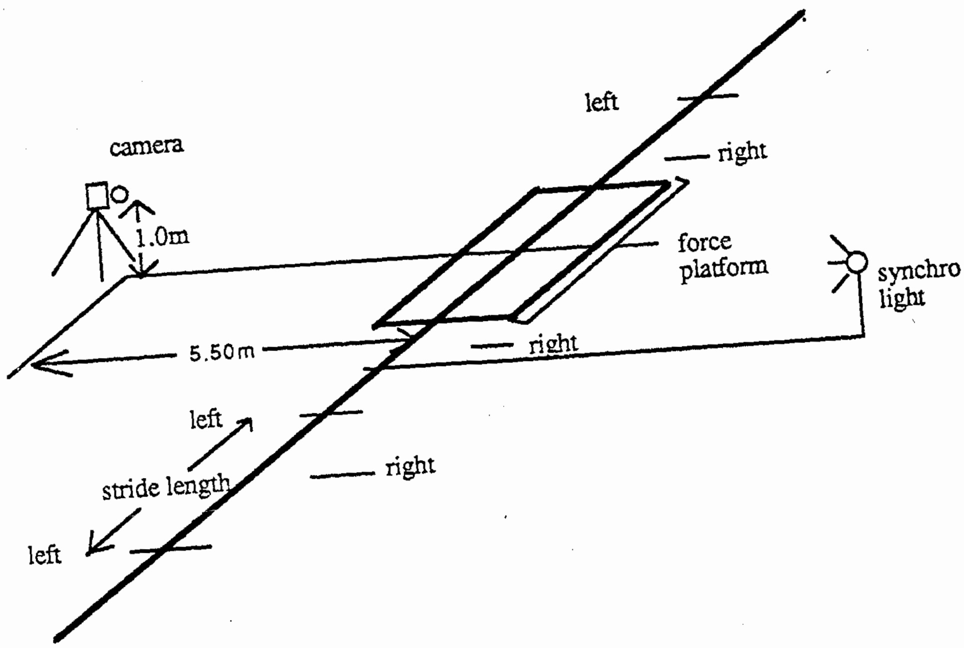
Segment Inertial Parameters

SEGMENT	$M_s:M_b$	$L_c:L_s$	$K_c:L_s$
foot	0.0145:1.0	0.500:1.0	0.457:1.0
shank	0.0465:1.0	0.433:1.0	0.302:1.0
thigh	0.1000:1.0	0.433:1.0	0.323:1.0

M_s = Segment Mass
M_b = Body Mass
L_s = Segment Length
L_c = Distance From Proximal Endpoint to Segment Centre of Mass
K_c = Radius of Gyration About Centre of Mass

Figure 3-2

Laboratory Set-up



of analogue data corresponding to the three components of ground reaction force and three components of torque resolved about the geometric center of the force-platform. The six components allowed for the calculation of the location, direction, and magnitude of the reaction force vector acting on the foot during stance. Calibration factors supplied by the manufacturer were available for the conversion of voltage data to the mechanical units of force (N) and torque (Nm).

A Locam model 51 (Redlake Corp., Santa Clara, Ca., U.S.A.) 16 mm high-speed cine-film camera was used. The filming was done with shutter factor of 1/3 (equivalent to a 60° opening in the shutter disc). A Tokina f1.8, 12.5-75 mm zoom T.V. lens was used. It was set at a focal length of 12.5 mm and an f-stop opening of 1.8. Four hundred ASA black and white cine film (Kodak 4-X reversal type 7277, 16mm) was used. Filming was done indoors under a set of overhead mercury vapour floodlights. A supplementary set of floodlights were used to sidelight the subject.

The laboratory frame of reference was defined to be a plane parallel to the running path and bisecting the force-platform. The subject was instructed to run with his left (ipsilateral) leg in this plane. To aid the subject in attaining the objectives of the locomotion problem in the proper plane the desired stride-lengths for both the ipsilateral and contralateral leg were marked with adhesive tape on the laboratory floor. Three separate sets of trials, each with a different stride length, were performed by the subject in this experiment. The sets of trials designated as 'slow', 'medium', and 'fast' were performed with arbitrarily chosen stride lengths of 2.0m, 2.6m, and 3.2m respectively.

A grid with markings of known distance was placed in the plane bisecting the force-platform and filmed. The calibration factors obtained from this film allowed for the determination of position in the laboratory frame of reference.

Time in the laboratory frame of reference was defined by two sources. An inverted pendular metronome was used to define the desired stride frequency during the running trials. The camera's calibrated frame rate of 179.0 frames per second was also used as a measure of time in the laboratory frame of reference. It was equipped with a shutter correlator generating a discrete voltage as an event-time signal. This signal was used to drive the computer based sampling of the force-platform's analogue output.

A hand held multiplexing switch was used to turn on a light-emitting diode in the field of view of the camera while simultaneously initiating the force platform data collection. This allowed for synchronization of the force-platform and cine-film data records. An analogue-to-digital converter interfaced to an Apple II+ microcomputer (Apple Computer Inc., Cupertino, Ca., U.S.A.) was used to collect data from the force-platform.

Determination of $X(t_i)$

The motion ($X(t)$) was quantified as a digital signal ($X(t_i)$) obtained from cine-film records of the subject running down the path. The equipment used to digitize the film included a microprocessor-based Numonics digitizer (Numonics Corp., Lansdale, Pa., U.S.A.) interfaced to an Apple II+ microcomputer. The film image was projected (using a Photo Optical Data Analyzer Model 224A (Mark VI), LW International, Woodland Hills, Ca., U.S.A.) onto a matte white table top. The digitizer was fixed to the surface.

Before digitizing each trial the stride cycle was defined. The timing of events such as footstrike and toe-off and the magnitude of joint and segment angles were used as guides. The final choice of frames defining the start and end of the stride cycle depended, ultimately, on the judgement of the investigator. The beginning and end the running cycle were defined as the frames in which the configuration of the body were most nearly equal. As a precaution extra frames prior to and subsequent to the defined cycle were digitized although this was not possible for some of the 'fast' speed trials. Prior to the acquisition of each trial film the calibration grid was digitized in order to allow for conversion of data to the units of length in the laboratory frame of reference. Although filming was done at 179.0 frames per second (fps) only every third frame was digitized making the effective frame rate 59.66 fps.

The x and y coordinates of the four landmarks described in table 3-1 and of a fixed reference marker were obtained from each frame. A simple presentation of the data in graphic form allowed for a preliminary data check. Frames were re-digitized if gross errors were apparent. The uncalibrated digital position data was transferred to an HP 1000 minicomputer (Hewlett-Packard Co., Cupertino, Ca., U.S.A.) for further analysis.

The raw coordinate data were converted to units of length, using the calibration factors obtained from the digitization of the calibration grid, and time, using the sampling period (t = 1/59.66 = 0.01676s). All position data were converted to the LAB F-O-R using the reference marker location.

Filtering of human motion data is nearly as much art as it is science. In the end an essentially arbitrary and indefensible choice must be made by the investigator. Because filtering is arbitrary the validity of the results of any investigation are called into question. A robust result might be one which doesn't change arbitrarily when the filtering is changed. Filtering was considered twice in this investigation. The first step was to choose a set of frequency cutoffs, the point after which a fourier series representation of the data was truncated to yield low-pass filtered data. The manner in which the cutoffs were chosen is described below. The sensitivity of the results of this investigation to changes in filtering was tested and is described later in this chapter.

The fourier series representation of a signal x(t) and its derivatives x′(t) and x′′(t) are as follows;

$$x(t) = \sum_{n=1}^{\infty} a_0 + (a_n \sin(n\omega t) + b_n \cos(n\omega t)) \ ;$$

$$x'(t) = \sum_{n=1}^{\infty} (a_n n\omega \cos(n\omega t) - b_n n\omega \sin(n\omega t)) \ ;$$

$$x''(t) = \sum_{n=1}^{\infty} (- a_n n^2\omega^2 \sin(n\omega t) - b_n n^2\omega^2 \cos(n\omega t)) \ .$$

The power (P_n) of the n^{th} harmonic of the acceleration series (x′′) can be defined by

$$P_n = ((-a_n n^2\omega^2)^2 + (-b_n n^2\omega^2)^2)^{1/2} .$$

The percentage of the total power of the infinte series in the n^{th} harmonic can be defined by

$$\%P_n = P_n \left(\sum_{i=1}^{\infty} P_i \right)^{-1} \times 100 \ .$$

Each signal can be reconstructed by the superposition of each of the individual terms in the series. A low-pass filtered version of each signal can be reconstructed by using a finite number of the sine and cosine terms.

The movement of the body was defined by eight digital position signals, an x-position and a y-position signal for each of the four anatomical landmarks. The fourier coefficients of each signal were obtained using a digital fourier transform algorithm. The coefficients for 26 harmonics were calculated for each signal.

The inverse dynamic analysis required the calculation of acceleration ($x''(t)$ and $y''(t)$). Because of the dependence on the quantity n^2 in the power of the n^{th} harmonic in the acceleration signal it is more sensitive to the presence of noise than the lesser derivatives. Because of this the choice of the number of harmonics to include in the filtered version of each signal was based on the power in the harmonics of the accelerations signals.

Note that the percent power of the c^{th} and last harmonic in the truncated version of the signal would be

$$\%P_c = P_c \left(\sum_{i=1}^{\infty} P_i \right)^{-1} \times 100 \ .$$

The summation represents the power accumulated in superimposing the first C harmonics. The term $^{\%}P_c$ can be called the percent of the accumulated power in the C^{th} and last harmonic.

Data from all trials in all conditions were considered in choosing a cut-off harmonic, the harmonic after which the series representation would be truncated. A cut-off was chosen for each coordinate of each of the four anatomical landmarks. The cut-off for each coordinate was then used in the reconstruction of data for all trials in all conditions. The same truncation point was used to reconstruct the position, velocity, and acceleration signals. The process used to choose the cut-off is described below.

Truncated Fourier series representations of the eight acceleration signals (x'' and $y''$for each of four landmarks) were derived in a systematic manner for each and every trial. The value of C was increased in increments of 1 from 1. For each value of C the percent of accumulated power in the C^{th} and last harmonic was calculated.

For the hip and knee data the first harmonic whose $\%P_c$ value dropped below 5% was identified. For the ankle and foot data the first C harmonic in the region of the 9th, 10th, and 11th harmonic for which the percent of accumulated power rose over 10% was identified (inclusion of a powerful harmonic in this region was found to help retain the discontinuous characteristics near foot strike and toe-off).

The highest identified C harmonic for each landmark seen in all the trials was chosen as the cutoff for all trials for that landmark.

Inverse Dynamic Analysis

A computer-based inverse dynamic analysis was used to solve for the applied torques. The algorithm, written in FORTRAN and run on an HP 1000 minicomputer, was developed by G. E. Caldwell and A. E. Chapman in the biomechanics laboratory at Simon Fraser University, Burnaby, B.C., Canada.

The inputs to the inverse dynamic analysis include the segmental inertial parameters, the filtered coordinate data (position, velocity, and acceleration), and the digital ground reaction force and torque data. The outputs of the analysis are the digital signals $\tau_H(t_i)$, $\tau_K(t_i)$,and $\tau_A(t_i)$.

Calculation of e

The separate components e_{jk} were evaluated by digital integration of the torque histories. These were done using an algorithm based on the trapezoidal rule. The integrations were performed separately on the positive and negative sections of each digital joint torque history;

$$e_{j1} = \int_{t0}^{tf} \tau_{j1}\, dt \quad , \tau_j < 0.0 \ ;$$

$$e_{j2} = \int_{t0}^{tf} \tau_{j2}\, dt \quad , \tau_j > 0.0 \ .$$

where the subscript j = 1 pertains to the ankle, j = 2 pertains to the knee, and j = 3 pertains to the hip. Total mechanical effort (e) was calculated as the sum of the six components;

$$e = \sum_{j=1}^{3} \sum_{k=1}^{2} | e_{jk} | \ .$$

Experimental Protocol

Three separate sets of trials were performed on two separate days aproximately three weeks apart.. All 'medium' speed trials ($\lambda_r = 2.6$m) and the PS trials for the 'fast' condition ($\lambda_r = 2.0$m) were performed on the first test date. The NP fast trials and all the 'slow' trials ($\lambda_r = 2.0$m) were performed on the second.

Prior to filming, the subject warmed up with a 15 minute jog, stretching exercises, and wind sprints. The same routine was used on each of the filming dates. On each of the two days the lighting and camera were set and the calibration grid was filmed. The anatomical landmarks were highlighted with adhesive markers prior to filming.

Table 3-3

Nonpreferred Style Instructions

STYLE	INSTRUCTION
HF	run with greater than normal hip flexion
KF	run with greater than normal knee flexion
SL	run with knee joint motion minimized and held in an extended position
OJ	run with knee and hip motion minimized and held in an flexed position

At each of the three speeds the preferred styles (PS) were performed first. The subject was told to run along the path and across the force-platform while hitting each mark on the floor with the appropriate foot. Practice trials were performed until the investigator was satisfied that the subject was hitting the marks reliably. During the practice trials the frequency of a metronome was matched, by trial and error, to the subject's freely chosen stride frequency. The subject was then filmed performing the run until, in the subjective opinion of the investigator, five trials with reliable stride length and frequency had been recorded.

The non-preferred (NP) styles were then performed. The subject was told to run with the same stride length, as marked on the floor, and the same stride frequency, as indicated by the metronome, as in the previous PS trials. Further, an NP instruction was also given to the subject. Four different instructions, as described in table 3-3, were used. Each NP style was practiced until the investigator was satisfied that the requested strde length and stride frequency were being matched. Two subjectively reliable trials of each NP style were recorded. Force-platform data were collected with each trial.

Results

Discarded Trials

Three of the 39 trials were excluded from consideration after serious errors in their data records were discovered. The force platform data record for the fifth preferred style trial (PS_5) in the medium speed condition was incomplete. The landmark coordinate data for two non-preferred trials, OJ_1 in the medium speed condition and KF_1 in the slow condition were found to be very erratic. The reasons for this are unknown although digitizer hardware error is suspected.

Filter Effects

Varying the filter cutoff frequency between the third and tenth haarmonic data produced virtually no visible effect on the position signals of the hip and knee joint markers.. The importance of filtering can be seen with the acceleration signal. Choosing the point where the acceleration cumulative power dropped below 5% was felt to retain most of the true signal. All trials tended to show this drop between the third and fifth harmonic. Within a few harmonics the power then rose dramatically as the $n^2\omega^2$ term began to dominate the power calculation. The high power at high harmonics was considered to be noise.

The case was slightly different for the toe and ankle marker data. Filtering produced visible changes in the position data especially during footstrike, stance, and toe-off. It was observed that by retaining a peak in the cumulative power spectrum in the region of the 10th harmonic two specific characteristics of position signal were controlled. First, the oscillation of the markers while the foot is on the ground and stationary was minimized. Also, during the periods in time when the foot's motion was arrested at footstrike and initiated near toe-off the signal better retained a corner-like appearance with the inclusion of the 10th or 11th harmonic. The set of harmonic cutoffs thus chosen are shown in table 3-4 in the row marked 'original'.

Running Style ($X(t_i)$)

The comparison of requested stride length (λ_r), stride frequency (f_r), and running speed ($\lambda_r \times f_r$) to that which the subject actually performed is given in tables 3-5.1, 3-5.2, and 3-5.3. Stride frequency was calculated as the inverse of the period of the stride cycle. Stride length was determined from the net horizontal displacement of the toe marker over

Table 3-4

Filter Cutoffs

FILTER	MARKER							
	toe		ankle		knee		high	
	x	y	x	y	x	y	x	y
low	9	9	9	9	3	4	3	3
original	11	11	11	11	5	6	5	5
high	13	13	13	13	7	8	7	7

the period of the cycle. Running speed was calculated in two ways. The first was simply to multiply the performed stride length (λ) by the the performed stride frequency (f). Speed was calculated a second time as the grand average ($\underline{V}$) of the average velocities of the hip, knee, ankle, and toe markers over the period of the stride cycle.

As might be expected the subject's stride length, stride frequency, and running speed varied between trials in all three conditions. The findings can be summarized as follows.

Slow Condition. Stride length in the PS trials tended to be long averaging about 11 cm greater than requested. The stride frequency was greater in all PS and NP trials. This suggests that the preferred f, as recorded during the practice trials, may have been incorrect. As a result running speed tended to be higher than requested. Both measures of running speed varied similarly from trial to trial.

Medium Condition. Stride lengths in the PS trials were tightly distributed about 5 cm short of the requested λ_r of 2.6 m. The NP trials also tended to be short (in 5 of 7 trials) with the worst case being the second OJ trial. The shortfall was 17 cm, an error of 6.5%. With one exception the performed f were higher than or equal to that reqested, the greatest error being 0.05 HZ (approximately 4.3%). The running speed in the PS trials tended to be high, the average error being about +3%. The worst case in the NP trials was a running speed of 3.06 m/s, about 7% higher than the requested speed of 2.86 ms^{-1}.

Fast Condition. The subject reproduced the requested λ_r more reliably in the PS trials than in the NP trials. The worst case was a λ of 2.76 m in the second KF trial, a shortfall of almost 14%. The story for the f is very similar. The PS trials were performed reliably and with little error ($S^{PS} \approx 2\%$) while greater error was seen in the NP trials.

Table 3-5.1

Stride Length,Stride Frequency,Speed

Condition: Slow

	PS	S^{PS}	HF_1	HF_2	KF_1	KF_2	SL_1	SL_2	OJ_1	OJ_2
λ_r	2.0		2.0	2.0		2.0	2.0	2.0	2.0	2.0
λ	2.11	0.03	2.00	1.99		1.99	2.10	2.08	1.93	2.11
f_r	1.23		1.23	1.23		1.23	1.23	1.23	1.23	1.23
f	1.29	0.03	1.36	1.39		1.30	1.27	1.33	1.36	1.27
$\lambda_r x f_r$	2.47		2.47	2.47		2.47	2.47	2.47	2.47	2.47
$\lambda x f$	2.72	0.06	2.71	2.76		2.58	2.67	2.76	2.62	2.67
$\underline{v}$	2.74	0.05	2.75	2.77		2.60	2.67	2.76	2.64	2.71

λ_r = Requested Stride Length (m)
λ = Performed Stride Length (m)

f_r = Requested Stride Frequency (s^{-1})
f = Performed Stride Frequency (s^{-1})

$\lambda_r x f_r$ = Requested Speed (ms^{-1})
$\lambda\ xf$ = Performed Speed (ms^{-1})
$\underline{v}$ = Performed Speed (average of markers) (ms^{-1})

KF_1 Discarded Trial

Table 3-5.2

Stride Length,Stride Frequency,Speed

Condition:Medium

	PS	S^{PS}	HF_1	HF_2	KF_1	KF_2	SL_1	SL_2	OJ_1	OJ_2
λ_r	2.6		2.6	2.6	2.6	2.6	2.6	2.6		2.6
λ	2.55	0.09	2.68	2.54	2.46	2.61	2.59	2.58		2.43
f_r	1.10		1.10	1.10	1.10	1.10	1.10	1.10		1.10
f	1.15	0.03	1.10	1.05	1.13	1.13	1.15	1.10		1.13
$\lambda_r x f_r$	2.86		2.86	2.86	2.86	2.86	2.86	2.86		2.86
$\lambda\ xf$	2.93	0.15	2.96	2.66	2.77	2.94	2.97	2.85		2.74
$\underline{v}$	2.95	0.15	2.98	2.77	2.81	2.95	3.06	2.92		2.80

λ_r = Requested Stride Length (m)
λ = Performed Stride Length (m)

f_r = Requested Stride Frequency (s^{-1})
f = Performed Stride Frequency (s^{-1})

$\lambda_r x f_r$ = Requested Speed (ms^{-1})
$\lambda\ xf$ = Performed Speed (ms^{-1})
$\underline{v}$ = Performed Speed (average of markers) (ms^{-1})

OJ_1 Discarded Trial

Table 3-5.3

Stride Length,Stride Frequency,Speed

Condition: Fast

	PS	S^{PS}	HF_1	HF_2	KF_1	KF_2	SL_1	SL_2	OJ_1	OJ_2
λ_r	3.2		3.2	3.2	3.2	3.2	3.2	3.2	3.2	3.2
λ	3.22	0.10	3.01	3.25	3.28	2.76	3.06	3.45	3.18	3.02
f_r	1.10		1.10	1.10	1.10	1.10	1.10	1.10	1.10	1.10
f	1.12	0.03	1.13	1.17	1.13	1.15	1.10	1.08	1.22	1.15
$\lambda_r x f_r$	3.52		3.52	3.52	3.52	3.52	3.52	3.52	3.52	3.52
$\lambda x f$	3.60	0.15	3.39	3.80	3.70	3.17	3.38	3.74	3.88	3.47
$\underline{v}$	3.58	012	3.46	3.84	3.70	3.42	3.46	3.62	3.94	3.54

λ_r = Requested Stride Length (m)
λ = Performed Stride Length (m)

f_r = Requested Stride Frequency (s^{-1})
f = Performed Stride Frequency (s^{-1})

$\lambda_r x f_r$ = Requested Speed (ms^{-1})
λ xf = Performed Speed (ms^{-1})
$\underline{v}$ = Performed Speed (average of markers) (ms^{-1})

The greatest error was seen in the first OJ trial where f was high by approximately 11%. The running speed for this trial, 3.94 ms^{-1}, represented an 11% error in speed, the largest seen.

Variations in speed can be expected with deviations from the λ_r and f_r objectives. The greatest deviations from the requested speed ($\lambda_r \times f_r$) were approximately 11%. Errors of this magnitude warranted an attempt to assess their impact on the test of the hypothesis. Just such an attempt is included with the results of the effort analysis.

While stride length and stride frequency were to be held constant in both PS and NP trials other aspects were not. The contrasts between the PS and NP styles ,as requested by the NP instructions, are exemplified in the joint angle patterns.

HF vs. PS. The examples are taken from the fast condition trials. The trial HF_1 is contrasted to the PS trials in Figure 3-3.1. The instruction given in each HF trial was to run with greater hip flexion than normal. The thigh angle in space is used as an indicator of the hip angle which was not measured directly. During ipsilateral stance the profiles are similar. In recovery the angles diverge sharply. Peak hip flexion angle, seen during contralateral support, is 1.48 radians in the HF_1 trial in contrast to the lesser hip flexion peak of approximately 0.75 radians in the PS trials.

KF vs. PS. Data from the PS trials and the second KF trial (KF_2) in the slow condition shown in Figure 3-3.2, illustrate the contrast of the knee joint angle patterns in the PS and the KF styles. The subject was instructed to run with greater knee flexion than normal. As with the HF style the differences are minimal during ipsilateral stance and are largest during contralateral stance. Peak knee flexion angle in KF_2 was -2.46 radians in contrast to the lesser peak flexion angle of approximately -1.5 radians in PS condition. The peak-to-peak range in knee joint angle histories was about 0.8 radians larger in KF_2 than in the PS trials.

SL vs. PS. Knee joint angle patterns for the trials SL_2 and the four PS trials in the medium speed trials are displayed in figure 3-3.3. The subject was told to run with `straight legs' thus minimizing knee joint motion and holding the knee joint in an extended position. It is apparent from the figure that the subject performed as instructed. The peak-to-peak range of motion is much less in the SL_2 trial. The maximum angle of knee flexion in the PS condition is much greater than in the SL trial. The peak extensor angle was slightly greater in the PS case though in the SL case the profile was much flatter near the peak values and over the entire cycle.

OJ vs. PS. For the OJ trials the subject was told to run like an `old jogger' with greater hip flexion and knee flexion than normal. This results in the subject looking as though he were seated in a chair or saddle while running. On Figure 3-3.4 thigh and knee joint angle data from trials OJ_1 and the PS trials in the medium speed condition are graphed.

The thigh angle profiles are similar between ipsilateral toe-off (ITO) and contralateral footstrike (CFS). Throughout the rest of the cycle the thigh angle reflects greater hip flexion in the OJ style. A peak flexor angle of 0.82 radians can be contrasted to the peak of approximately 0.50 radians in the PS trials.

The knee joint angle histories are similar in shape. The knee joint remained more flexed during ipsilateral support in the OJ trial but reached lesser peak flexion angle during recovery.

Figure 3-3.1

<u>HF vs. PS : Hip Angle</u>

Condition: Fast

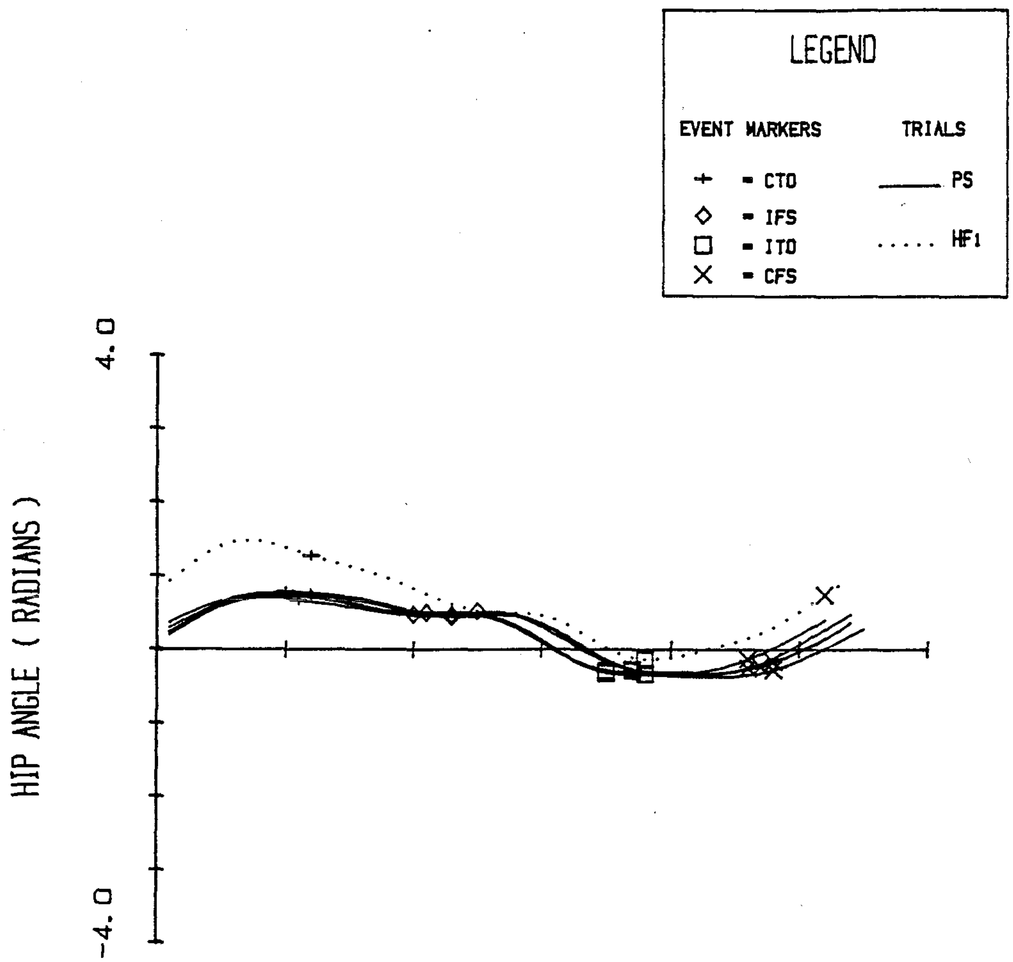


Figure 3-3.2

<u>KF vs. PS : Knee Angle</u>

Condition: Slow

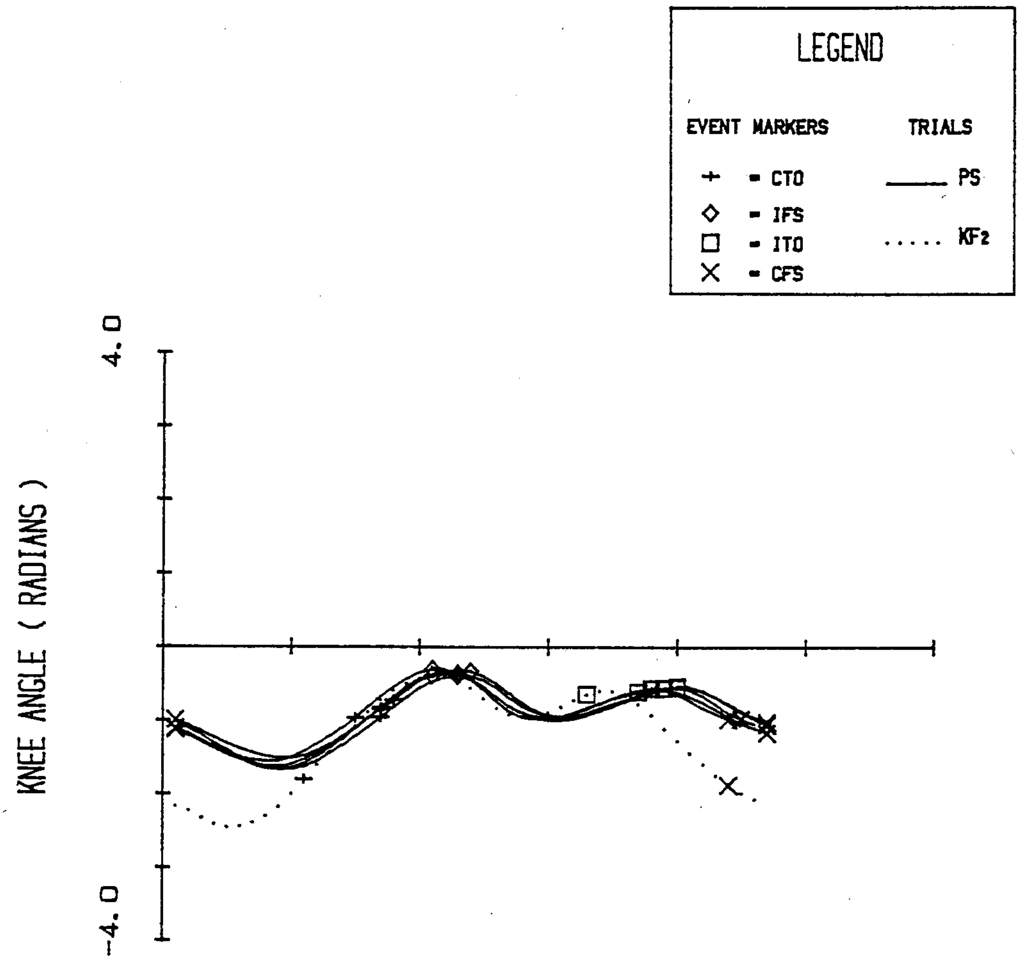


Figure 3-3.3

SL vs. PS : Knee Angle

Condition: Medium

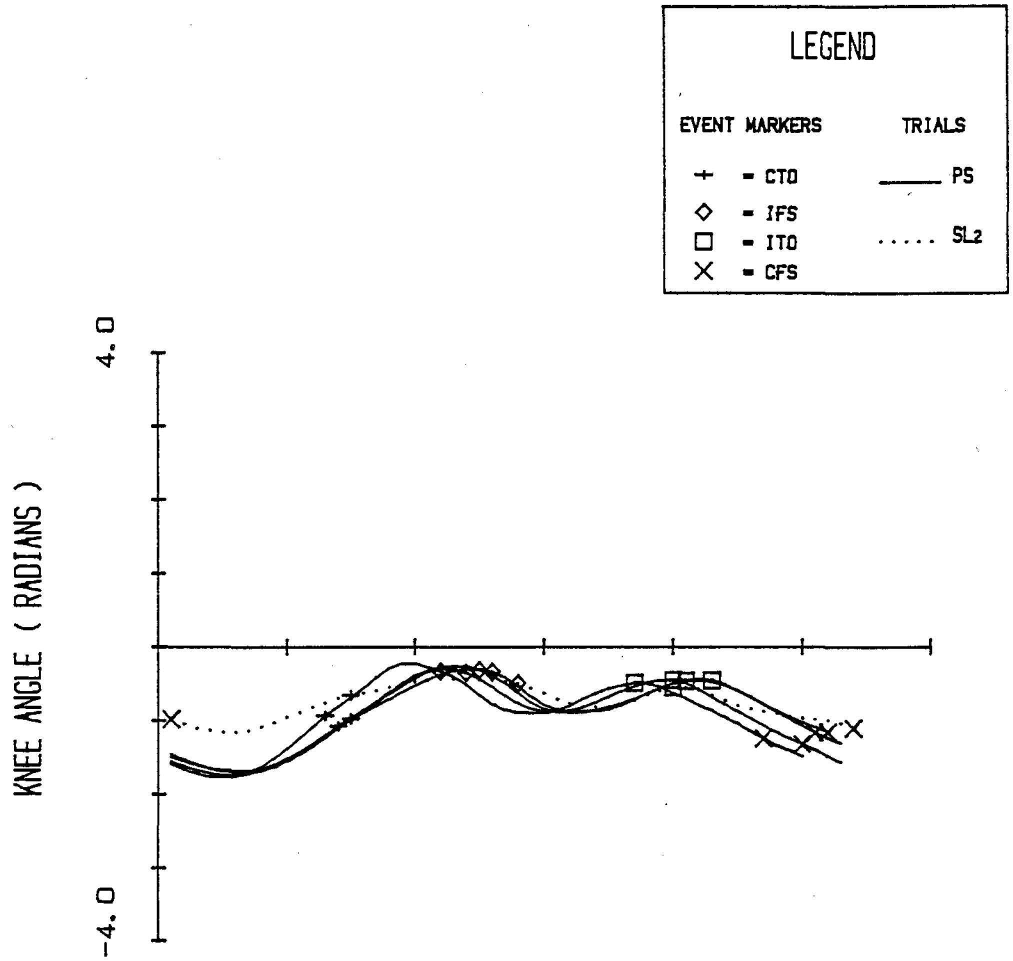
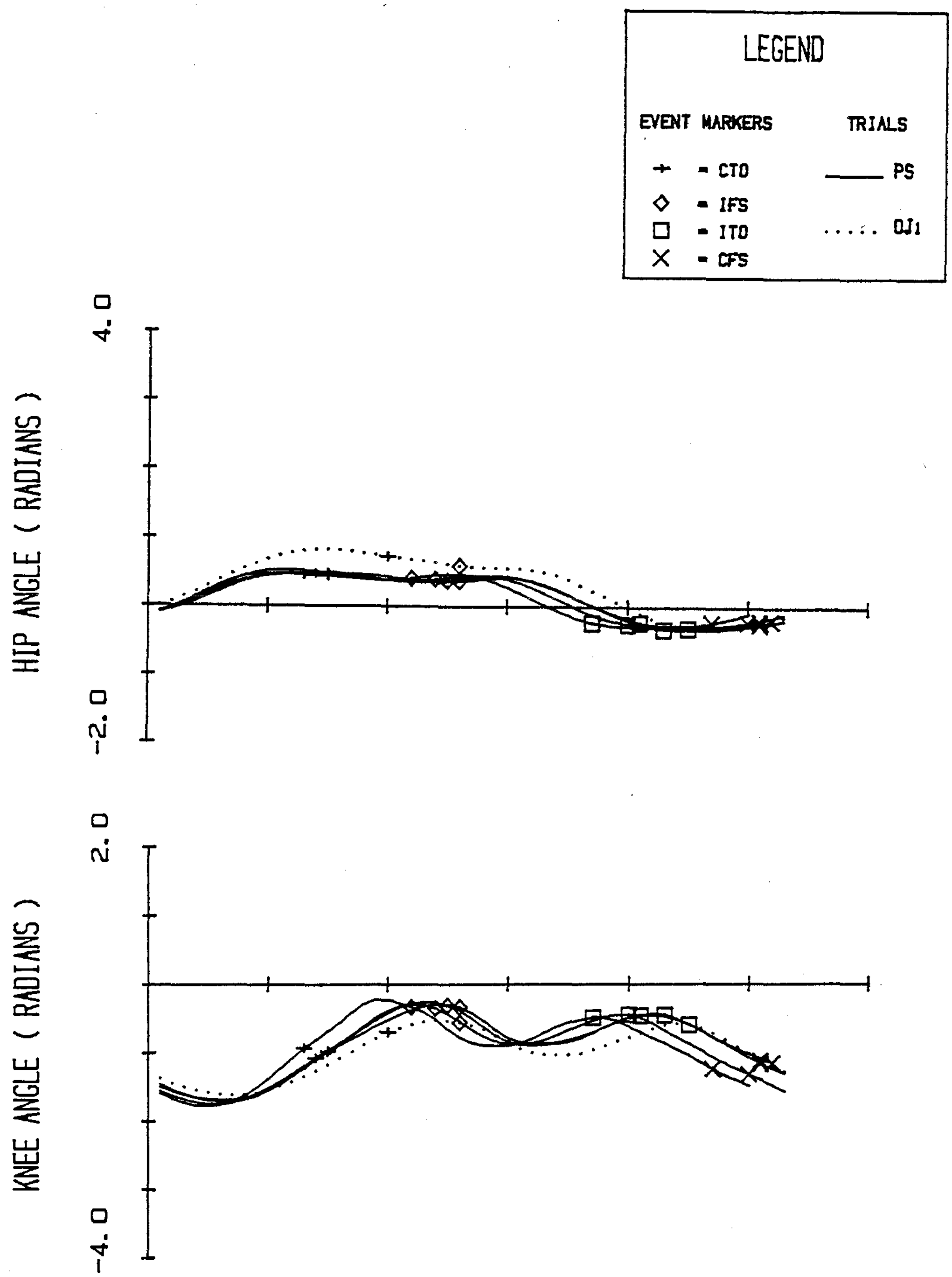


Figure 3-3.4

OJ vs. PS : Hip Angle and Knee Angle

Condition: Medium



Torque Data ($\tau_{jk}(t)$)

The segment inertial data and the kinematic data were submitted to a computerized inverse dynamic analysis algorithm. The algorithm yielded ankle, knee, and hip joint torque histories in digital form. Graphs of some typical torque data are shown in Figures 3-4.1 through 3-4.4. Note that the data has not been normalized or adjusted in time. The data from different trials tend to be out of phase.

Figure 3-4.1 shows the ankle torque patterns ($\tau_A(t_i)$) of the four PS trials in the medium speed condition. Small dorsi flexor torques, applied just after IFS, preceded larger plantar flexor torque during ipsilateral stance. During the balance of the cycle ankle torque activity was virtually nil. The shape of these curves was similar for ankle torque data in the PS and NP trials at all three speeds.

Knee torque patterns ($\tau_K(t_i)$) from the five PS trials of the fast speed condition are graphed in Figure 3-4.2. As was the case for τ_A these curves were subjectively similar in form across styles and speeds. Low levels of knee flexor torque are seen prior to IFS. High knee extensor torques are seen during ipsilateral stance. The torque drops to zero then becomes slightly extensor again in early recovery A small speed dependent difference was seen in the slow condition PS trials (not shown in Figures 3-4). Larger magnitudes of flexor torque were seen in the phase prior to IFS with smaller extensor torques seen subsequently.

Hip torque profiles ($\tau_H(t_i)$) for the PS trials in the fast condition are graphed in Figure 3-4.3. Prior to IFS the hip torque is mainly extensor. Shortly after IFS the torque becomes flexor and remains that way through stance and early recovery. These profiles can be contrasted to the data for hip torque in slow condition PS trials (Figure 3-4.4).

Figure 3-4.1

Ankle Torque : Preferred Style

Condition: Medium

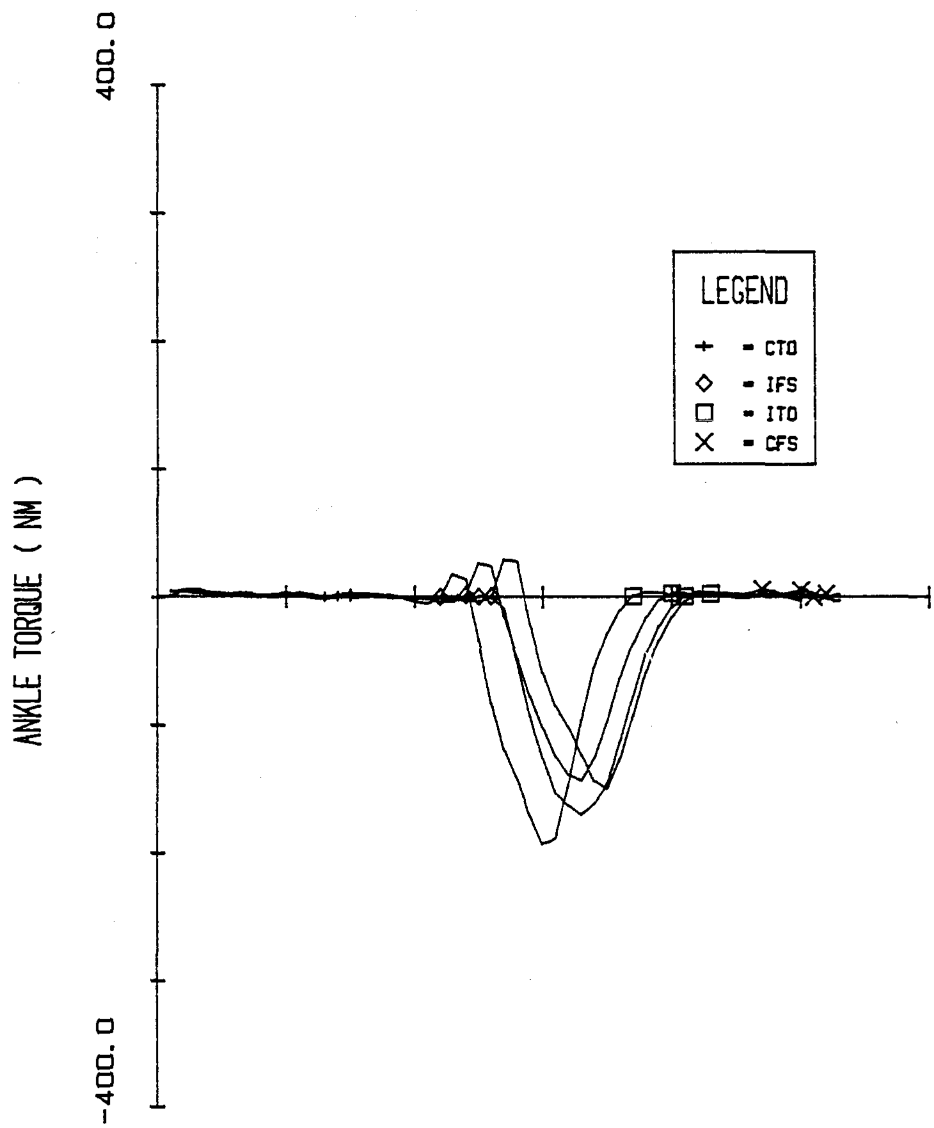


Figure 3-4.2

Knee Torque : Preferred Style

Condition: Fast

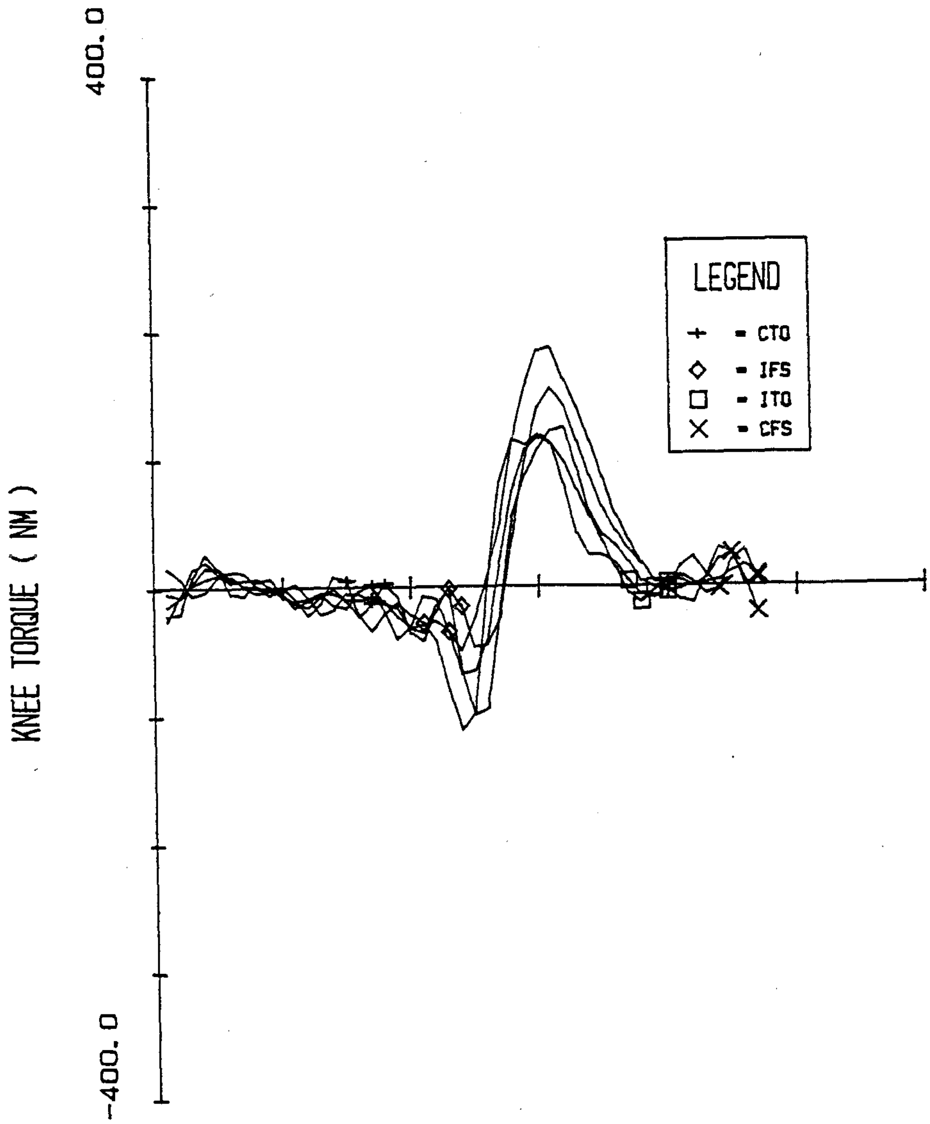


Figure 3-4.3

Hip Torque : Preferred Style

Condition: Fast

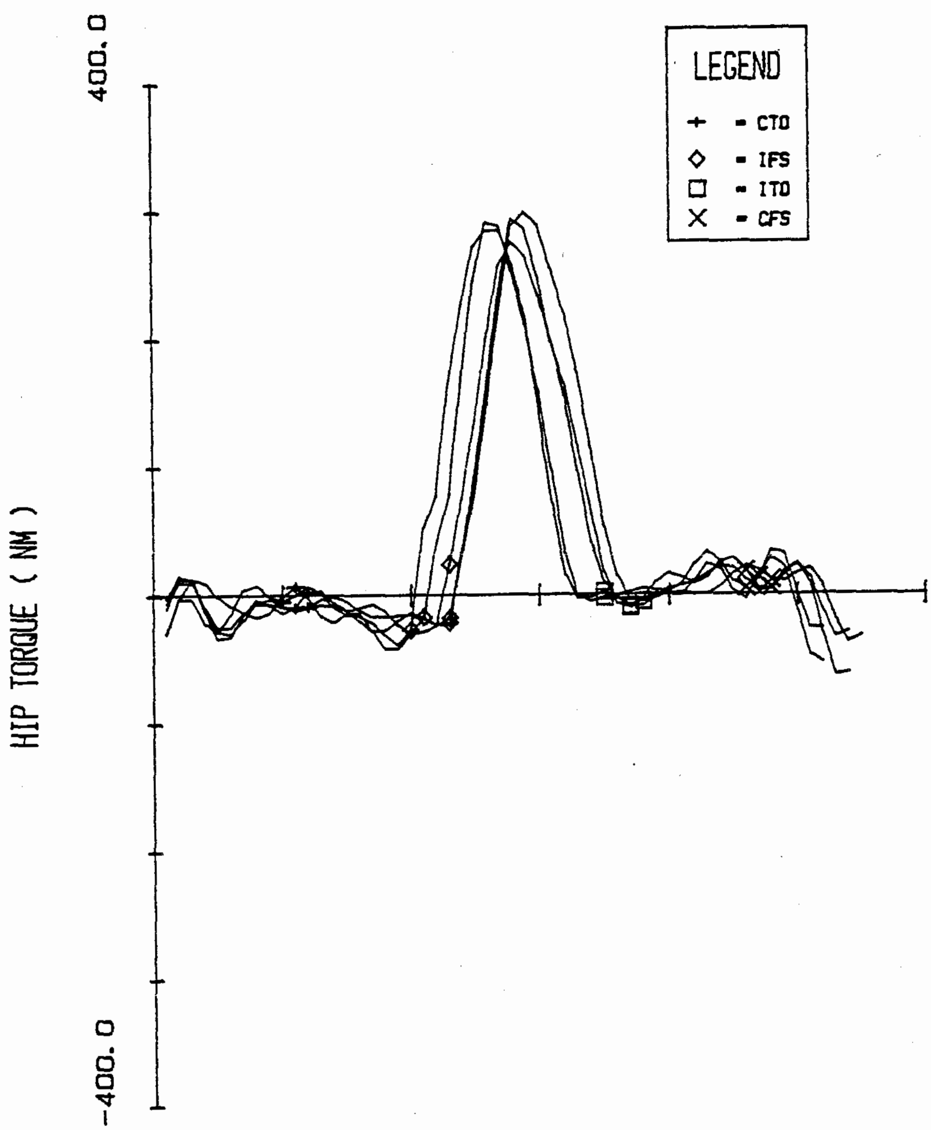
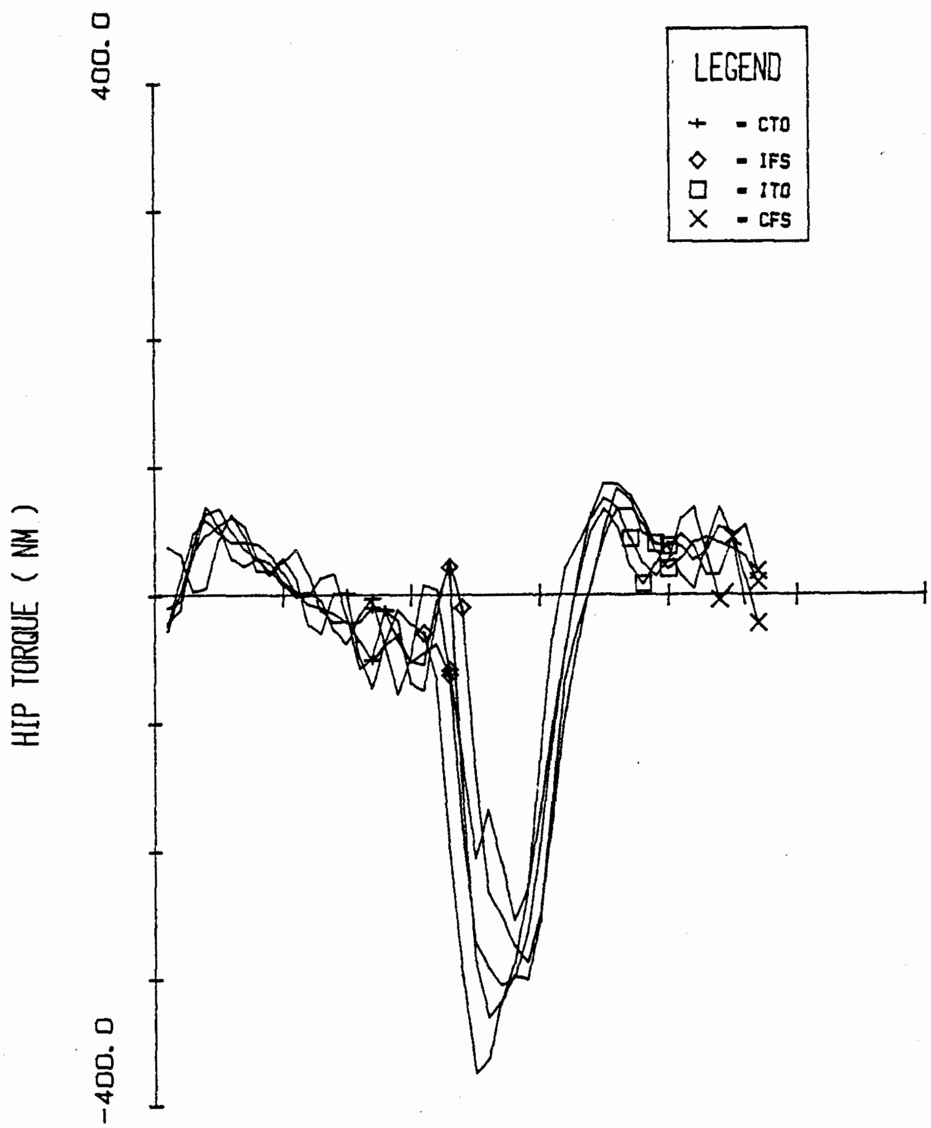


Figure 3-4.4

Hip Torque : Preferred Style

Condition: Slow



From ITO until just before CTO the torque is flexor. It becomes slightly extensor until CFS. During most of the stance phase the hip torque is extensor. This is true, in general, in all the PS and NP trials at the slow speed and may reflect a speed dependent characteristic.

A subjective view of the repeatability of the torque data in the PS trials can be obtained from these graphs. It is not, however, the shape of these curves which is of interest. The parameter of interest in this investigation, the characteristic whose difference explains the preference for the PS style is, by hypothesis, the integral of these curves. It is by comparison of mechanical effort (**e**) that the similarity and differences in the PS and NP torque data will be tested.

Mechanical Effort (**e**).

Graphic examples of the integration of PS and HF style torque data in the medium speed condition are shown in figure 3-5.1 and 3-5.2. The shaded area above the zero torque line represents the positive (hip flexor) effort while the shaded area below the zero line represents the negative (hip extensor) effort. Differences between the PS and the HF trial are diagrammed in the adjoining bar graph (figure 3-5.3). The flexor efforts were nearly equal with the HF trial being about 2.5% greater than the PS trial. The extensor effort required in the HF trial was nearly 36% larger than that required for the PS trial.

Results of the integration for all effort sources for all trials in the slow, medium, and fast condition are tabulated (tables A-1, A-2, A-3) and graphed (figures A-1.1 through A-3.3) in Appendix A. An example, the graph of knee effort ($\mathbf{e}_{2K}$) required for the PS and NP trials in the medium speed condition, is reproduced in figure 3-6. To the left of the zero line is the knee flexor effort and to the right, the knee extensor effort. Data from

Figure 3-5.1

Hip Torque vs. Time

Trial: HF_1 Condition: Medium

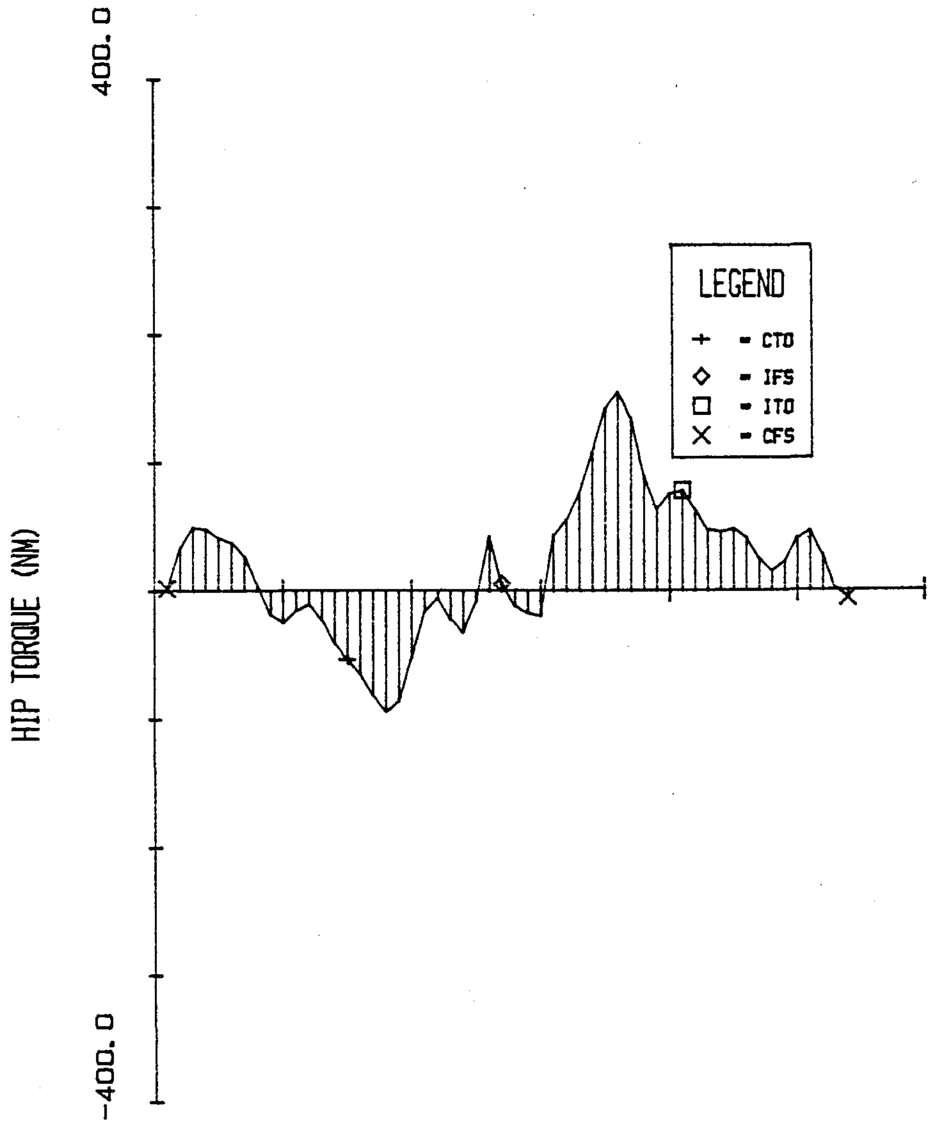


Figure 3-5.2

Hip Torque vs. Time

Trial: PS4 Condition: Medium

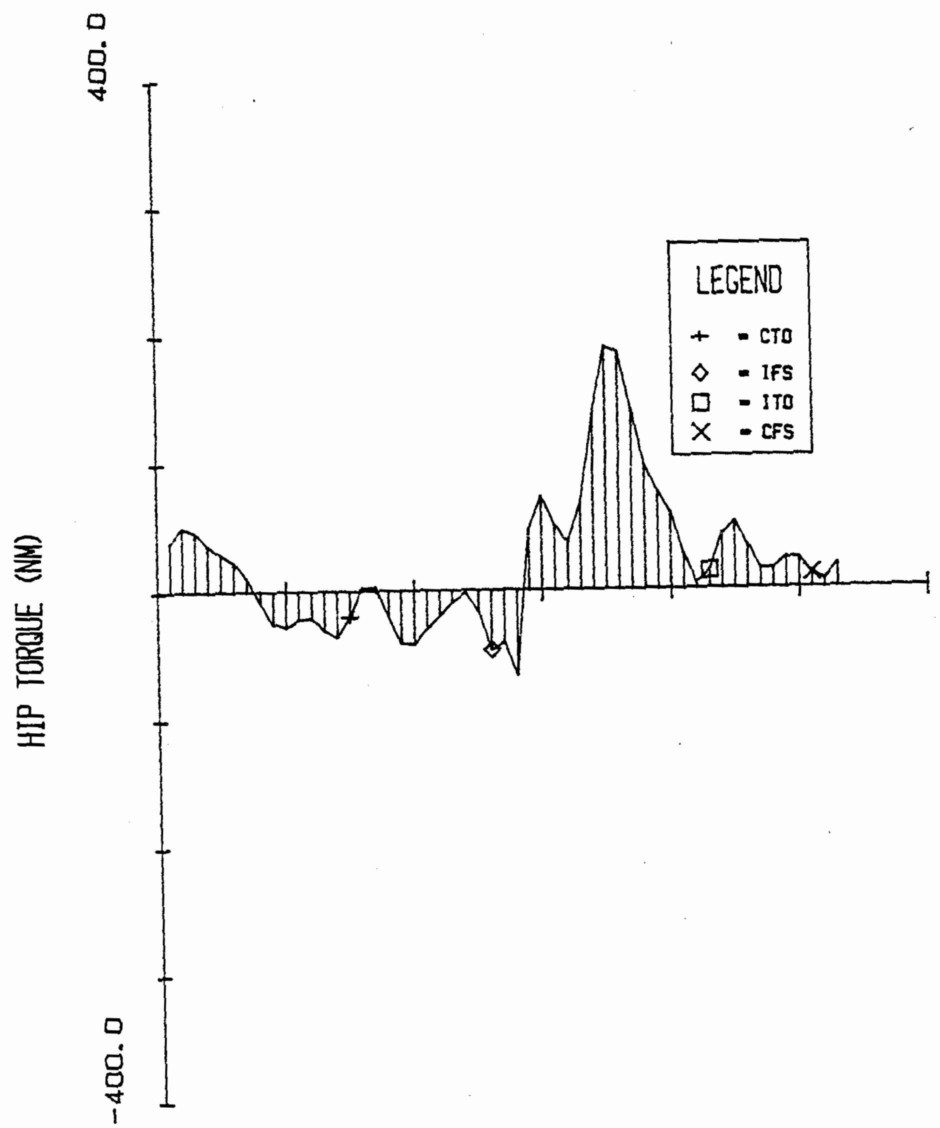


Figure 3-5.3

Hip Effort

Trial: HF_1,PS_4 Condition: Medium

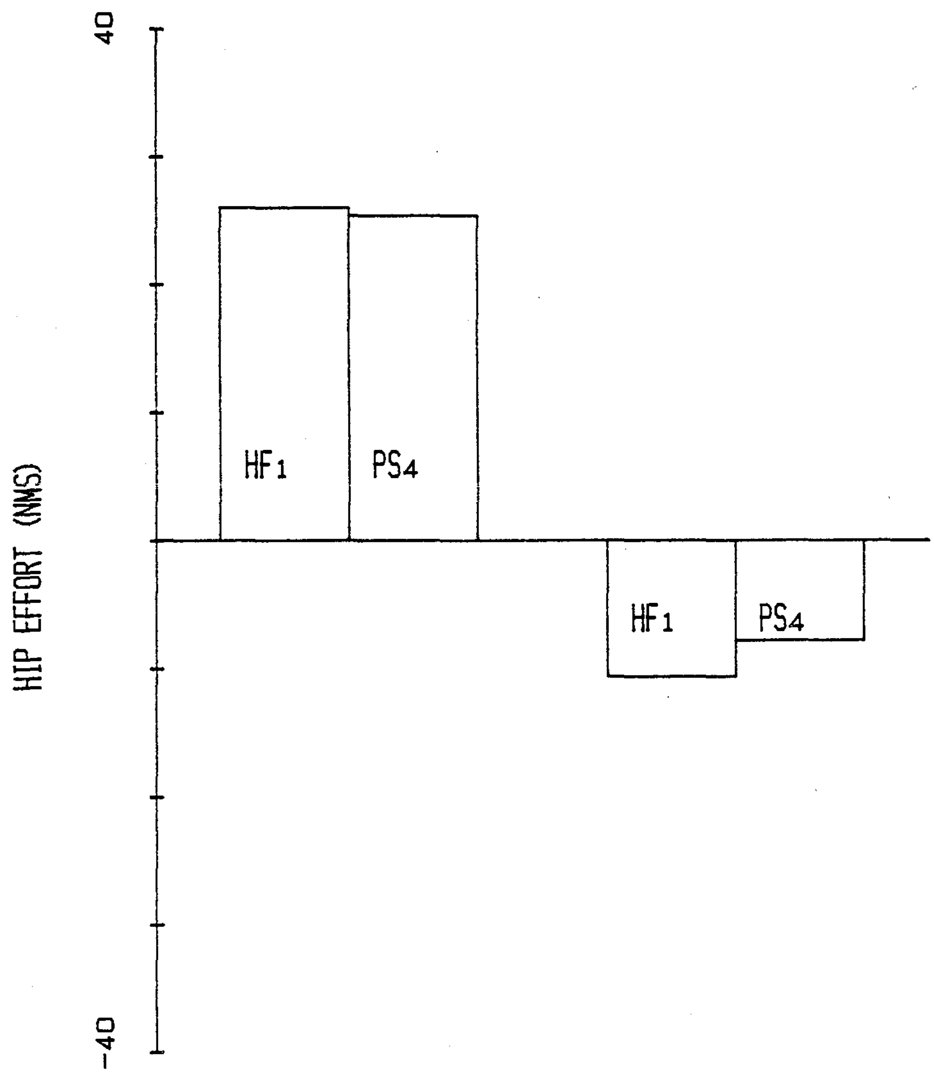
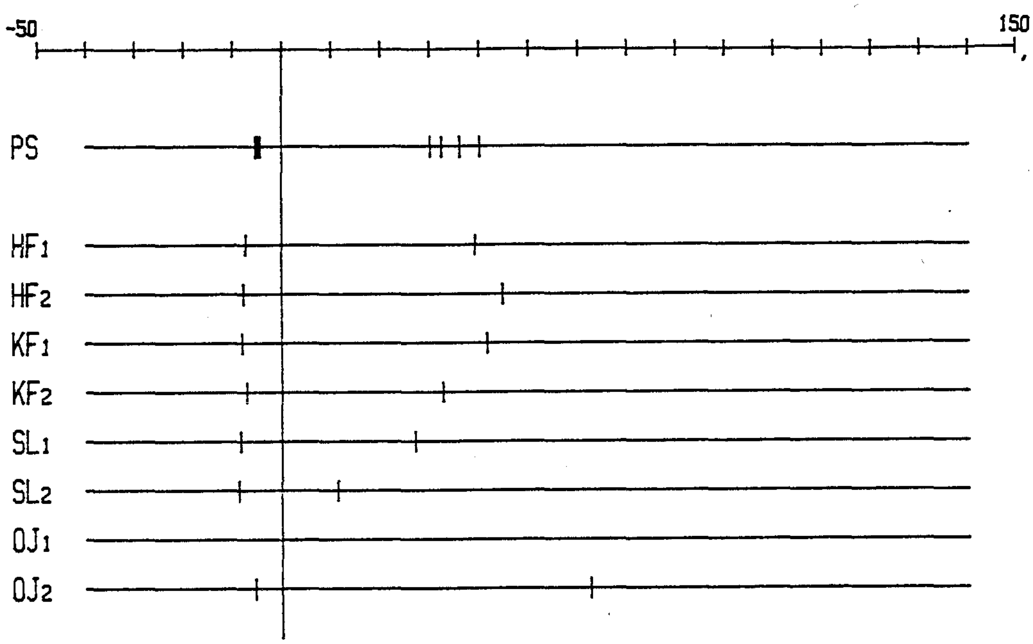


Figure 3-6

(e_{2k}) Knee Effort (Nms)

Condition: Medium



the PS trials are plotted on the uppermost line with the eight NP trials shown on the lower lines. It is apparent that all the NP trials required greater (more negative) knee flexor effort than was seen in any of the PS trials although the excess was slight in the case of the OJ_2 trial. On the extensor side the trials HF_1 and KF_2 fell within the range of the PS data; HF_2 and KF_1 were slightly greater than PS. The trial OJ_2 required a still larger extensor effort. The trials SL_1 and SL_2, on the other hand, required less knee extensor effort than the PS trials.

Some of the highlights of the effort data as shown in the tables and figures of Appendix A are described below. In the following discussion the range within which the PS data falls is used as a qualitative guide for judging the closeness of the levels e in NP trials to the PS condition. Equivalence to PS can be defined, more precisely, in terms of sample mean and sample standard deviation. These values are reported in tables A-1.1 through A-1.3.

Slow condition. The magnitude of hip extensor efforts were generally greater than the hip flexor efforts. All NP trials required greater extensor effort than was required in the PS trials. NP hip flexor effort tended to fall in or near the range of the PS data. The exception was the SL_2 trial which was about 35% higher than the mean of the PS data. Knee flexor effort was, with the exceptions of the two OJ trials, greater in the NP trials. The OJ knee extensor efforts were, again, within the PS range while the other NP trials required less extensor effort than in the PS condition. The ankle plantar flexor efforts in the NP trials all varied within 25% of the mean of the PS trials. KF_2 and SL_1 required larger efforts while the OJ trials required slightly less. The dorsi flexor efforts, smaller in magnitude than the plantar flexor efforts, were tightly distributed in the PS trials with little difference visible in the NP conditions. Subjectively assessed the OJ trials tended to remain the closest to PS in terms of effort while the SL trials deviated the greatest.

Medium condition. The SL style required the greatest hip extensor effort at the medium speed condition with the SL_2 trial requiring an effort approximately three times that of the PS mean value. All NP trials except OJ_2 required a higher than normal knee flexor effort. Knee extensor effort varied widely across the NP trials. On the high end the OJ_2 trial required an extensor effort 180% of the PS mean. At the other extreme was the SL_2 trial requiring only 30%. Ankle plantar flexor effort results are highlighted by a high requirement in the KF_2 trial, twice the PS mean, and the low value for the OJ_2 trial at about 15%. The OJ style, unlike the other NP styles, required a higher than normal ankle dorsi-flexor effort.

Fast condition. The standard deviation for the hip extensor data is small in contrast to the other speed conditions. All the NP trials required greater magnitudes of hip extensor effort with the SL_1 trial the largest at approximately eight times the PS mean value. Both SL trials also required hip flexor efforts about 1.5 times the PS mean. The knee efforts present an interesting pattern. All the NP knee flexor efforts were above the PS range in magnitude with the SL_1 trial showing the greatest deviation. All the NP knee extensor efforts were, with the exception of HF_1, below PS range with the SL_1 trial requiring the least of all. Ankle plantar flexor effort was above the PS range in all NP trials while all ankle dorsi-flexor efforts, with the exception of KF_2, were below or within the range of the PS trials.

The hypothesis to be tested in this experiment was that total effort (**e**) was less in the preferred (PS) style than in any of a set of non-preferred (NP) styles. Total mechanical effort was defined as

$$\mathbf{e} = \sum_{j=1}^{3} \sum_{k=1}^{2} | e_{jk} |$$

The results of this summation performed on the PS and NP trials can be found in table 3-6.1 and, in graphical form, in figures 3-7.1 through 3-7.3. The total mechanical effort data for the PS trials is reported as sample means and sample standard deviations.

Brief descriptions of the results in the slow, medium and fast conditions are given below.

Slow condition. The PS data can be summarized with a sample mean of 101.4 Nms and a sample standard deviation of 5.0 Nms. Four of seven NP trials were more than 4 standard deviations above the mean and well above the observed range of PS data. The other three were less than 3 standard deviations above the mean. The first `hip flexion' trial, $HF_1$, fell within the range of PS data at 106.3 Nms. The greatest total effort was seen for the second `straight-leg' trial, SL_2, at 133.4 Nms.

Medium condition. The mean and the standard deviation of the PS data at this speed were 93.4 ± 5.6 Nms. The second knee flexion trial (KF_2) fell within the observed range of PS data with a total effort of 99.3 Nms. The trial SL_1 was just above the range of PS data and less than 2.0 standard deviations above the PS mean. All other NP trials were more than 3.0 standard deviations above the mean. The greatest total was for the HF_2 trial with 126.4 Nms (+5.89 S).

Fast condition. The mean and standard deviation of the five PS trials in the fast

Table 3-6

(e) Effort (Nms)

TRIAL	CONDITION		
	SLOW	MEDIUM	FAST
PS_1	100.13	91.62	88.73
PS_2	99.58	91.68	102.93
PS_3	94.67	89.21	100.74
PS_4	107.15	100.00	94.63
PS_5	105.47		94.89
$\underline{e}^{PS}$	101.4	93.4	96.4
S^{PS}	5.0	5.6	5.2
HF_1	106.34	114.42	121.97
HF_2	130.40	126.38	111.55
KF_1		113.24	137.85
KF_2	123.32	99.31	127.67
SL_1	126.21	106.15	185.57
SL_2	133.42	119.47	173.73
OJ_1	114.74		129.84
OJ_2	107.54	122.67	130.13

Figure 3-7.1

(e) Effort (Nms)

Condition: Slow

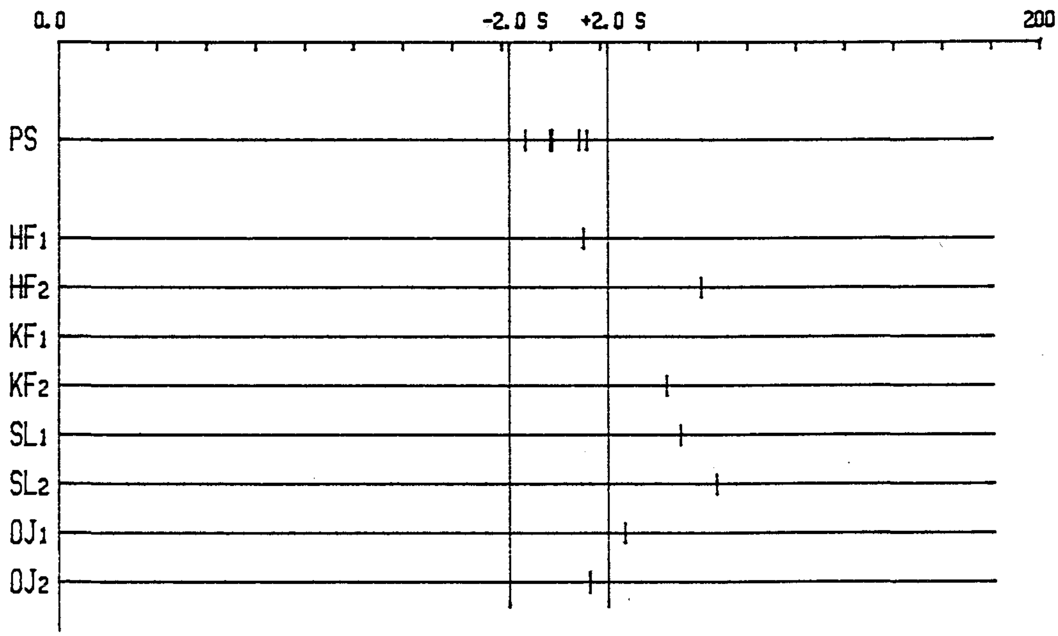


Figure 3-7.2

(e) Effort (Nms)

Condition: Medium

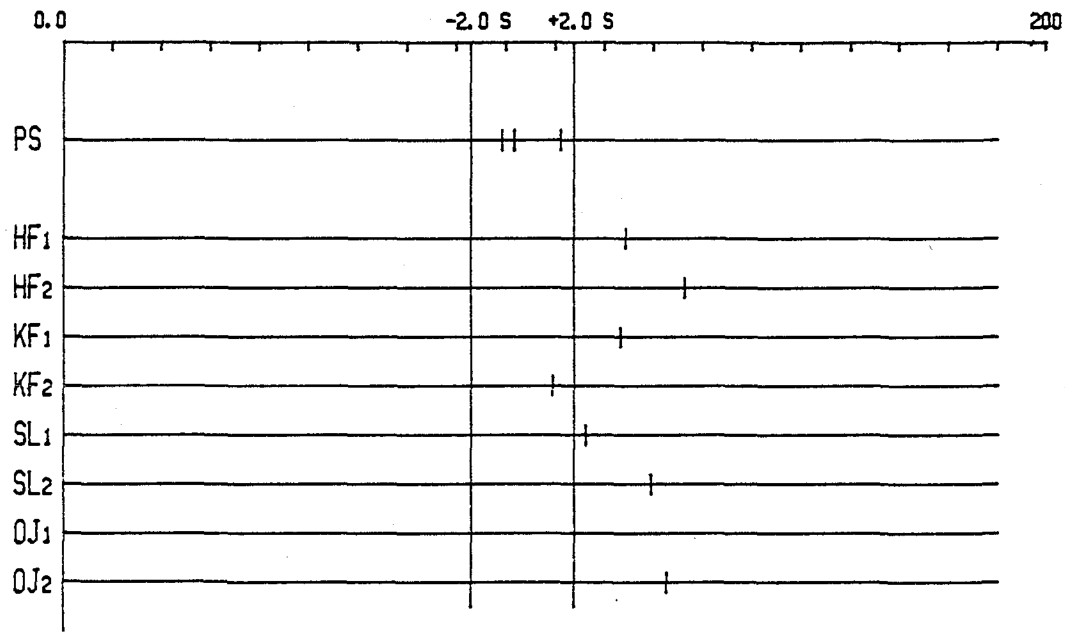
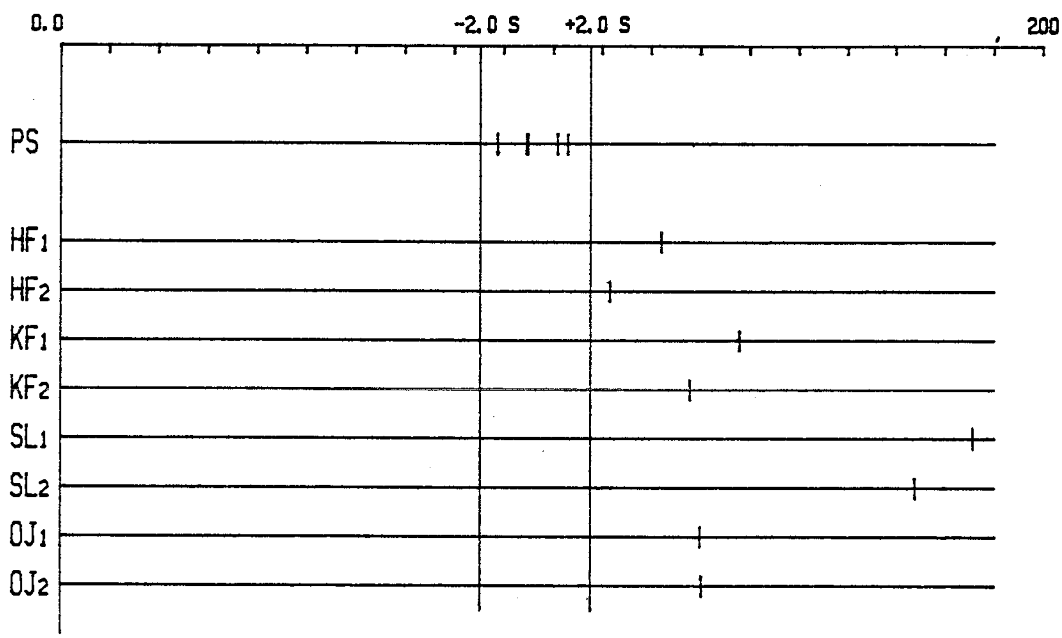


Figure 3-7.3

(e) Effort (Nms)

Condition: Medium



condition were 96.4 ± 5.2 Nms. All the NP trials fell outside the observed range of PS data. The trial HF_2 was nearest with a total effort of 111.6 Nms (+2.91 S). The greatest total was for the SL_1 trial at 185.6 Nms.

The Test of the Hypothesis

The confidence interval was defined as being 2.0 standard deviations above and below the PS mean value. All the PS trials fell within this interval at all three speed conditions. If the PS data were found to be distributed normally the confidence interval would encompass 95.44% of the PS data. It must be remembered that the choice is essentially arbitrary. The few NP trials which fall close to the confidence interval limits will be discussed specifically.

The confidence interval at each of the three speed conditions are given in table 3-7 in the row marked 'original'. Of the 22 NP trials in the three speed conditions, the total mechanical effort data for which are found in table 3-6, there are three trials which fall within the confidence interval and which therefore do not satisy the hypothesis. Further, there are three NP trials which lie within one standard deviation of the confidence interval. Because these trials lie close to the arbitrary confidence interval boundary it is only with caution that they are considered to have required greater than PS effort. None of the NP trials fall below the PS mean for total mechanical effort.

The six exceptions or near exceptions to the hypothesis are as follows.

Slow condition. The first hip flexor trial (HF_1) was, at 106.3 Nms, 1.0 standard deviation above the mean and well within the observed range of PS data. The trial OJ_2 also fell within the normal bounds with a total effort of 107.5 Nms (+1.23 S). This value fell just outside the observed range of PS data. The trial OJ_1, at 114.7 Nms (+2.66 S) fell close to the $\mathbf{e}^{PS}$ interval.

Medium condition. The trial KF_2, at 99.3 Nms (+1.1 S) was within the observed range of PS data and the confidence interval. The trial SL_1 was close at +2.46 S (106.2 Nms).

Fast condition. All the NP trials in this condition required greater than PS total effort. The trial HF_2, with a total effort of 111.6 Nms (+2.70 S) fell the closest.

Two factors which influence, in a quantitative manner, the calculation of total effort were mentioned previously. The first was the filtering strategy used. The second was the subject error in reproducing the requested stride length (λ_r) and stride frequency (f) . Changing the filtering process or accounting for the λ and f error would certainly change the magnitudes of $\mathbf{e}$. This is not really a concern. More critical to the investigation is how the changes in magnitude alter the test of the hypothesis. Will a change in the filtering strategy, or correcting for the λ and f error alter the relationship of any or all the NP trials to $\mathbf{e}^{PS}$ interval? In this section the results of an effort to address this question will be presented. To simplify discussion the effects of correcting λ and f error and of changing the filtering strategy will be discussed separately. Also in the interests of simplicity, only the PS mean and standard deviation, and the NP trials which were clear or near exceptions to the hypothesis will be discussed. Complete sets of the recalculated data can be found in appendix B and will be cited as they are discussed.

Errors in λ and f result in errors in running speed. A speed correction factor was devised to try and account for it. The simple model used is not intended to represent the ultimate, correct method for accounting for the errors. The speed-corrected total effort

(e^c) was calculated as:

$$e^c = e \times v_r \times \underline{v}^{-1}$$

where **e** is the original measure of total effort, V_r is the requested speed ($\lambda_r \times f_r$) and $\underline{V}$ is the average speed of the trial in question;

$$\underline{v} = (v_{toe} + v_{ankle} + v_{knee} + v_{hip}) \times 1/4.0$$

The results of the speed correction for all trials can be found in tables B-1 through B-3 in Appendix B.

The means, standard deviations and the upper limits of the e^{PS} confidence interval and the `close' regions (+2.0 S and +3.0 S respectively) for the original and speed corrected PS data are shown in table 3-7. The effect of the speed correction on the PS means is negligible. The influence is more pronounced on the standard deviations. In the slow condition the distribution was tightened. The reduction of variability might be expected of a manipulation designed to offset error. On the other hand the standard deviations increased in magnitude in the medium and fast conditions.

Discussion of some of the individual NP trials at each of the three speed conditions follows. Data for these trials can be found in table 3-8.

Slow condition. HF_1 became smaller in magnitude and closer to the mean in the units of total effort but greater in terms of standard deviations as a result of the speed correction. OJ_2 increased slightly in the units of effort and moved from +1.2 S to +1.84 S. In both cases the conclusion that total effort was within the PS range was not changed by the correction. The trial OJ_1, at +2.66 S (114.7 Nms) with the original analysis, rose to +4.50 S (119.3 Nms). The large rise was due to the reduction in the standard deviation with the speed correction.

Table 3-7

Original and Speed Corrections $\underline{e}^{PS}$ Regions

CONDITION		$\underline{e}^{PS}$	S^{PS}	$\underline{e}^{PS}+2.0S^{PS}$	$\underline{e}^{PS}+3.0S^{PS}$
slow	original	101.4	5.0	111.4	116.4
slow	corrected	101.3	4.0	109.3	113.3
medium	original	93.4	5.2	104.6	110.2
medium	corrected	93.6	8.6	110.8	119.4
fast	original	96.4	5.6	106.8	112.0
fast	corrected	96.8	7.6	112.0	119.6

Table 3-8

($\underline{e}$) Original and Speed Corrected Effort

CONDITION	TRIAL	ORIGINAL		SPEED CORRECTED	
		e	$_se$	e	$_se$
slow	HF_1	106.3	0.98	105.9	1.15
	OJ_2	107.5	1.23	108.6	1.84
	OJ_1	114.7	2.66	119.3	4.50
medium	KF_2	99.3	1.13	99.1	0.66
	SL_1	106.2	2.46	102.2	1.00
	KF_1	113.2	3.81	118.9	2.94
	HF_1	114.4	4.04	113.2	2.28
fast	HF_2	111.6	2.70	104.2	0.98

$$_se = (e - \underline{e}^{PS}) \times (S^{PS})^{-1}$$

Medium condition. The total effort for KF_2 changed very little while the standard score dropped from +1.13 S to +0.66 S. This trial stayed well within the bounds of the PS confidence interval. SL_1, initially classified as close (+2.45 S), fell to within the bounds of the PS interval at +1.00 S. The trials KF_1 and HF_1 both moved from well above PS, +3.81 S and +4.04 S respectively, to the close range (+2.94 S and +2.28 S). The conclusion remains that both required greater than PS effort.

Fast condition. The trial HF_2 moved from the close category well into the PS region (+0.98 S).

Only two of 22 trials in the three speed conditions moved into the PS range as a result of the speed correction and thus affected the test of the hypothesis. The results of the test of the hypothesis were relatively insulated from the errrors in stride length, stride frequency, and running speed.

As described previously the filtering strategy, the truncation of a fourier series representation of the x-y position data of anatomical landmarks, required an essentially arbitrary decision as to where the series was to be truncated. To change the truncation point, the filter cut-off, in an arbitrary way produces quantitative changes to the data. The important question is whether the change in the filter cut-off will change the results of the test of the hypothesis.

The truncation points used in the original analysis are given in table 3-4 beside the designation 'original'. Two other sets also appear in that table. The set designated 'low' are all 2 harmonics below the original set. The set designated 'high' are all 2 harmonics above the original set. The entire analysis was repeated using the low and high filter cutoffs. The total effort (e) results, along with the e_{jk} data, are given in tables C-1.1

through C-2.3 in Appendix C. Only the PS means and standard deviations, which are given in table 3-9, and the NP trials that are clear or near exceptions to the hypothesis will be discussed.

As the filter cutoffs changed from low to original to high the PS means for **e** rose gradually. This occurred at all three speeds. In the slow condition the standard deviation changed very little as the cutoff changed. In the medium speed condition the standard deviation dropped as the cutoffs changed from low to high. In the fast condition the standard deviation dropped with the high cutoffs.

The influence of the change in filter cutoff on individual NP trials in each of the three speed conditions is discussed next. The data for these trials can be found in table 3-10.

Slow condition. With original data there were two exceptions to the hypothesis, HF_1 and OJ_2, while OJ_1 was close. Using the low cutoffs the close trial, OJ_1, moved just into the PS region (+1.99 S). The other two remained inside the PS boundaries. Using the high cutoffs OJ_2 moved from within the PS region to the close region (from +1.23 S to +2.26 S) leaving HF_1 within the PS region and OJ_1 above PS in the close region. A large change was seen with the trial KF_2. Well above the $\mathbf{e}^{PS}$ range with the low and original cutoffs it moved into the close region (+2.84 S) with the high cutoffs. Despite the large change the conclusion that KF_2 required effort above the PS level didn't change.

Medium condition. With the original cutoffs KF_2 (+1.14 S) and SL_1 (+2.45 S) were the clear and near exceptions to the hypothesis. Going to the low cutoffs produced only small changes to the **e** values and standard deviation scores. Using the high cutoffs KF_2 remained within the PS region while SL_1 moved out of the close category with a

Table 3-9

Variation of Mean Effort ($\underline{e}^{PS}$) With Filtering

CONDITION	FILTER	$\underline{e}^{PS}$ (Nms)	S^{PS} (Nms)	$\underline{e}^{PS}+2.0S^{PS}$ (Nms)	$\underline{e}^{PS}+3.0S^{PS}$ (Nms)
slow	low	99.8	5.0	109.8	114.8
	original	101.4	5.0	111.4	119.6
	high	103.7	5.3	114.3	119.6
medium	low	91.9	6.0	103.9	109.9
	original	93.4	5.2	103.8	109.0
	high	95.8	4.2	104.3	108.5
fast	low	95.3	5.5	106.3	111.8
	original	96.4	5.6	107.6	113.2
	high	101.6	5.1	111.8	116.9

Table 3-10

Original and Refiltered Effort (e)

CONDITION	TRIAL	LOW		ORIGINAL		HIGH	
		e	$_s$**e**	**e**	$_s$**e**	**e**	$_s$**e**
slow	HF_1	106.6	1.35	106.3	0.99	108.3	0.85
	OJ_2	104.7	0.98	107.5	1.23	115.7	2.26
	OJ_1	109.8	1.99	114.7	2.67	119.0	2.88
	KF_2	118.8	3.79	123.3	4.38	118.8	2.84
medium	KF_2	98.8	1.16	99.3	1.13	101.7	1.37
	SL_1	105.8	2.32	106.2	2.45	110.0	3.35
fast	HF_2	106.1	1.96	111.6	2.70	117.2	3.06

$$_s\mathbf{e} = (e-\underline{e}^{PS}) \times (S^{PS})^{-1}$$

higher score of +3.35 S. SL_1 appears to be close but not within the PS region and therefore not an exception to the hypothesis.

Fast condition. With the original set of cutoffs HF_2 was designated as close to PS at +2.70 S. Using the low set HF_2 fell just inside the PS confidence interval at +1.96 S and with the high set of cutoffs it moved out of the clcse region to +3.06 S. HF_2 appears to be a possible but not clear exception to the hypothesis.

By changing the cutoffs to the high and to the low sets the hypothesis was tested twice more on each of the 22 NP trials. Of these 44 cases there were three cases where the conclusion differed from that made using the original set.

Summary

Exceptions to the hypothesis using two criteria are listed in table 3-11. The first column lists those tials that fell below +2.0 S using the original filter only. The second column lists additional trials that fell below +2.0 standard deviations with any of the three filters used. Using the other filters adds two trials to the list of exceptions, one in the slow speed and one in the fast speed. There appears to be at least one exception to the hypothesis at each of the three speeds. Accepting the speed correction as valid would add SL_1 in the medium speed condition. Applying the speed correction to the original filtered data moves the trial HF_2 in the fast condition, onto the list of exceptions.

Table 3-11

Exceptions to the Hypothesis

CONDITION	EXCEPTIONS WITH ORIGINAL FILTER	ADDITIONS WITH LOW AND HIGH FILTERS
slow	HF_1 OJ_2	OJ_1
medium	KF_2	
fast		HF_2

Discussion

None of the NP styles in any of the three speed conditions required levels of **e** less than the PS mean. However, at least one of the NP styles at each speed fell within the $\mathbf{e}^{PS}$ confidence interval for effort. Levels of **e** inside the confidence interval were regarded as being equal to the amount required in the preferred style. As such the preferred style cannot be rationalized on the basis that it required less **e** than any other style and the initial hypothesis (H_1) was not supported.

Several possible explanations for the failure are discussed below. Experimental uncertainty may have acted to obscure a true difference between the preferred and non-preferred styles. The failure to account for co-contraction and the action of two-joint muscles is considered as is the possibility that factors such as balance, mechanical load, and aesthetic criteria may be relevant to the preference for style. Finally, the physiological rationale, the validity of which was assumed, may have been incorrect. In chapter 4 a general discussion of the validity of the investigation is presented. The impact of the approximations and simplifications required to define the mechanical and physiological models on the analysis are assessed qualitatively. The limits of a teleological theory of locomotion to explain preferred style are also discussed.

Experimental Uncertainty

Several types of experimental uncertainty could have contributed to the failure to support the hypothesis. The extent to which the subject failed to satisfy the objectives, the uncertainty caused by the digital filtration and the arbitrary definition of the $\mathbf{e}^{PS}$ confidence interval all could have acted to lessen the resolution of investigation. This in turn could have prevented support for the hypothesis.

Deviations by the subject from the requested stride length (λ_r) and the requested stride frequency (f_r) would likely have been associated with erroneous measures of e in both the PS and NP trials. These errors in turn may have obscured the difference between e^{PS} and e^{NP}.

A speed correction factor, described in the Results section, was used to assess the impact of the stride length and stride frequency error on the test of the hypothesis. The correction factor itself provided only a first approximation of the influence of the error.

The application of the correction factor scaled total effort values by as much as 11%. While the changes could be large in both the amount of **e** and in the number of standard deviations the conclusion that an NP trial was either within or above the $\mathbf{e}^{PS}$ interval changed in only a few cases. To a first approximation the differences in e caused by stride length and stride frequency error was of a lesser order than the differences due to the contrasting PS and NP style objectives. With this in mind and with the understanding that the correction factor itself was only an approximation the errors in stride length and stride frequency will be ignored. The lone exception to this will be to accept that the trial HF_2 in the fast condition may have been an exception to the hypothesis. The errors in stride length and stride frequency remain a source of uncontrolled but, to the extent that the speed correction factor approximates the its influence, bounded error.

If the subject had not heeded the request to perform differently in the NP trials than in the PS trials any difference in **e** would certainly have been obscured. The extent to which the subject performed differently in the PS than in the NP trials was assessed by a subjective analysis of the joint angle data. Examples of contrasting joint angle patterns are given in figures 3-3.1, 3-3.2, 3-3.3, and 3-3.4. In each and every NP trial a clear deviation from the PS pattern is evident. Greater peak hip flexion was seen in the HF trials and greater peak knee flexion was seen in the KF trials. Knee joint range of motion was much less in the SL trials and fell near the peak knee extensor angles seen in the PS trials. The OJ trials can be characterized by greater hip flexion and knee flexion than the PS style throughout the stride cycle. Deviations from PS patterns were not necessarily limited to the particular aspect specified in the NP request. For example, knee joint angles varied from the PS pattern in the 'hip flexion' (HF) trials.

Subjectively assessed, the differences in **e** between NP and PS styles were not obscured by the possibility that the subject may have used his preferred style in an NP trial.

An unstated objective, implied in the model used, was that the subject's motion be restricted to the sagittal plane. Movements out of the sagittal plane and deviations from the path centered on the platform would result in error. Restricting the analysis to the lower limb helped to minimize the cumulative impact of non-coplanar motion. It remains, however, a source of uncontrolled error.

The truncation of a Fourier series representation of the digital position data was used to minimize the high frequency error introduced in the digitization process. There is no reliable means of assessing the spectrum of the true position signal or, alternatively, of the noise. Any filtering strategy must therefore be essentially arbitrary. Inappropriate low-pass filtering could conceivably contribute to a failure to support the hypothesis. Retaining too many high harmonics could allow noise to obscure the differences between NP and PS styles. Using too low a cutoff could eliminate the higher frequencies of the true signal. The difference between NP and PS styles could lie in these higher harmonic regions.

The cutoff for the coordinate data was varied over a four harmonic range. Within this range the conclusion that the hypothesis was not satisfied did not vary in the slow and medium speed conditions. As such it appears unlikely that either digitization error or the choice of filtering cutoff obscured the difference between NP and PS trials in these conditions.

The filtering did influence the test of the hypothesis in the fast condition. With the high filter cutoffs one NP trial, HF_2, moved from above to within the $\mathbf{e}^{PS}$ confidence interval. The conservative interpretation is that the hypothesis was not satisfied. If the set of high filter cutoffs included an inappropriate amount of high frequency noise the hypothesis may well have been satisfied.

The arbitrarily defined $\mathbf{e}^{PS}$ confidence interval may have been too wide. As such it could have obscured a true difference between PS and NP effort. In the slow and medium speed conditions this seems unlikely to have happened. The slow-speed trial HF_1 (+0.99 S) and the medium-speed trial KF_2(+1.13 S) required levels of $\mathbf{e}$ less than that required for several of the PS trials at the same speed. To have excluded these trials from the $\mathbf{e}^{PS}$ region would have meant excluding the several PS trials as well.

The conclusion that the hypothesis failed in the fast speed condition was based on the findings for trial HF_2 whose effort levels fell within the confidence interval with speed correction (+0.97 S) and with the low filter (+1.96 S). The speed correction is not considered comprehensive The $\mathbf{e}$ value calculated with the low filter, however, was close to, and is therefore dependent on, the arbitrarily defined confidence boundary. A better recommendation for the definition of the $\mathbf{e}^{PS}$ confidence interval would arise from a better established $\mathbf{e}^{PS}$ distribution. Repeating the experiment with more PS trials is advised.

The conservative interpretation is that none of the experimental factors mentioned appear capable of obscuring a true difference between $\mathbf{e}^{PS}$ and $\mathbf{e}^{NP}$. It is therefore unlikely that by somehow accounting for experimental uncertainty the failure of the hypothesis to explain preferred style can be reversed. Accepting that there exists some quantity **e** whose optimization accounts for the preferred style gives leave to only one conclusion; the hypothetical form of **e** was incorrect.

Another type of uncertainty results from the lack of quantitative representation of the constraints (C), the locomotion problem (L), and the equations of motion ($F_q = MX''$). This in turn adds a degree of uncertainty to the understanding of the reasons for the failure of hypothesis.

Explicit identification of the constraints was avoided by using only one subject, effectively holding them constant. The investigator's judgement and measures of stride length and stride frequency were used to assess whether or not the subject had achieved the objectives of the locomotion problem. Writing L explicitly was thus avoided. The inverse dynamic analysis, while allowing for the solution of the torques, was not derived from a system of equations representing motion as a function of those torques alone. The ground reaction force, a variable that was measured during the experimental trials, was required as input for the inverse dynamic analysis. Even with a characterization of the equations of motion without an explicit dependence on the ground reaction force the understanding would be incomplete. Solving a set of coupled, non-linear, second order differential equations is a very complex computational problem. Without quantitative representations of these concepts it is impossible to assess, with any certainty, the reasons for, or the limits on, the torque patterns observed in each condition.

As an example of how the lack of quantitative representations of the equations for L, C, and F_q hinders the understanding of running style consider the variation of $\mathbf{e}^{PS}$ across speed. The effort rose as the speed changed from medium to fast speed, an intuitively reasonable result. The mean effort was highest, however, at the lowest speed. Much of the excess magnitude can be attributed to the high levels of hip extensor effort (see table A-1.1). Because this was true in the NP and PS trials the effect appears to be speed rather than style dependent.

Hip torque profiles for the PS trials in the fast and slow speed conditions are graphed in figures 3-4.3 and 3-4.4 respectively. At both speeds the torques are extensor in the region of foot strike. At the high speed the hip torque becomes flexor in early to mid-stance. In contrast to this the hip torque at the slow speed goes through a strong extensor phase before becoming flexor late in stance. A possible interpretation is that the hip extensors are needed to pull the center of mass of the body across the base of support. Perhaps, at the slow speed, the body did not have enough momentum at foot strike to maintain its forward progression.

Other interpretations are possible. The subject could be using a large hip extensor effort to reduce the load on another torque generator, a load it may not be capable of sustaining. The subject could be shifting his emphasis towards some other as yet undetermined criterion. Without the quantitative representations of the equations of motion $F_{\dot{q}}$, the locomotion problem L, and the constraints C, only conjecture is possible.

The single trial in the fast condition that fell within the PS confidence interval did so only with the high filter and with the speed correction. There is a possibility that that trial did in fact require greater than preferred levels of effort. This, coupled along with the unequivocal failure of the hypothesis at the slow and medium speeds, could be interpreted as a weak tendency to satisfy the hypothesis as speed increases. The next question to ask is why this might be so. Did the criterion vary with speed to conform with the constraints acting on the motion? Within what range of speed is the subject free to choose his criterion for movement? A more complete representation of the laws of motion, the locomotion problem, and the constraints could help to answer or, at least, to put bounds on the range of possible answers to these hypothetical questions.In the following sections two other possible explanations for the failure are discussed. The uncertainty involved in the interpretation of the torque data does, of course, impact on them as well.

Co-contraction and Two-joint Muscle Action

Assume for the purposes of this discussion that the impulse model of physiological effort is correct. The co-contraction of antagonist and agonist muscles and the contraction of two-joint muscles could not be identified explicitly with the model used in this investigation. As such the effort associated with two-joint muscle contraction was over-estimated and the effort required for any co-contraction was under-estimated. If in fact these factors obscured a true difference between the PS and NP trials which were exceptions to the hypothesis one of the following must be true. Either the co-contractile effort is greater in the NP styles or the savings afforded by two-joint muscle contraction is less. A third option is that both are true.

A single muscular force produced by a two-joint muscle will produce torque at the two joints it crosses. This could result in a saving of physiological effort over the use of separate muscles at each joint. As such the use of two-joint muscles is consistent with the theory of minimization of effort. Synergistic co-contraction may occur at one or both joints as well.

A two-joint muscle may produce an inappropriate torque at one of the joints it crosses and an off-setting, physiologically demanding, antagonististic co-contraction may be required. Timing and coordination are obviously critical to the extent to which the action of a two-joint muscle can spare physiological effort.

The possibility of any co-contraction in the preferred style presents, in terms of the single joint torque generators defined for model used in this investigation, something of a contradiction. A prediction of a theory of minimization of physiological effort is that, with only single joint muscles acting and without any constraints on the ability of those muscles to produce torque , no antagonist co-contraction would occur. As is demonstrated with the data, the preferred style can be can be accounted for without any co-contraction. Any antagonist muscular contraction would incur an additional physiological cost.

If the theory of minimization of effort is accepted as true and if co-contraction does occur in the preferred style there appears to be only one explanation; the model of the body incorporating only the action of single-joint muscles is inappropriate. The action of two-joint muscles, themselves an effort saving mechanism, must somehow force the use of co-contraction in satisfying the objectives of the locomotion problem. The interplay of the action of two-joint muscles with co-contraction, an effect that is indiscernable with the model used in this investigation, might well account for the failure to support the hypothesis.

Balance, Load, and Aesthetic Criteria

Cappozzo (1983) proposed assessing the "...reliability of the locomotor act..." in terms of several criteria. He mentioned the maintenance of balance, the mechanical load on tissues, aesthetic criteria, and energy criteria. Effort is a concept not unlike energy expenditure. The others were not assessed directly in this investigation. However, the argument that they are relevant to locomotion in general, and therefore to running style, is compelling.

The NP styles certainly looked unusual, even humourous. If aesthetic standards play a role in determining style their origins are of interest. What is pleasing to the eye may reflect a subconscious understanding of what is best.

In a crude sense the mechanical load on the tissues was equivalent in the PS and NP styles. The subject was able to perform all styles without injury or gross discomfort. Musculoskeletal stress and strain almost certainly varied to some degree between styles and could well be relevant to the choice of style.

The PS and NP styles might also be considered equivalent in terms of balance. The subject was able to complete all trials without falling down. However, some styles may have been more susceptible to perturbations than others. As such balance could be relevant to the preference for style.

The consideration of balance and mechanical loading may have an indirect physiological rationale. Large strains on musculoskeletal tissue can result in damage which would require physiological action to repair. Mechanical perturbations could also cause large strains.

If there are a number of criteria relevant to the preference for style the subject must somehow account for each in solving a locomotion problem. Mathematical techniques for solving more complex optimization problems can shed light on how a subject may determine his preference for style. Nelson (1983) described two general methods for solving optimization problems with more than one relevant criterion. The first is to optimize the behaviour for one criterion while setting the others as constraints. An example might be to choose a running style such that the effort is minimized and for which a specified maximum amount of mechanical strain will be tolerated. The disadvantage of this, as Nelson points out, is that the solution may end up at the extremes of one or more of the relevant criteria. Operating at a maximum level of strain, for example, makes the tissues more susceptible to damage from control errors and perturbations.

The second approach is to strike a compromise between the relevant criteria. A comprehensive function with terms for each of the criteria, a weighted sum for example, is optimized to yield the solution. None of the individual criteria, however, need be optimized. The next step in the study of running style may be to identify, and derive hypothetical forms for, the criteria relevant to running style. The optimization of the individual terms and of a comprehensive criterion function could be tested.

Nelson's examples serve to illustrate how potentially relevant criteria such as those mentioned by Cappozzo could act to shift the preference for style away from the one which minimizes effort. It is interesting to note that the occurence of any co-contraction in the preferred style might also be explained by the existence of other relevant criteria. Stability and balance, for example, might be enhanced by co-contraction.

While the existence of other relevant criteria may serve to explain the results of this experiment it does so at the expense of the hypothesis that the preferred style is chosen so as to minimize effort.

Incorrect Physiological Rationale

An assumption of this investigation was that running style was chosen so as to minimize physiological effort. In the previous section an explanation for the failure to support the hypothesis which requires abandoning that assumption is given. Several questions arise. Is it possible to reject the original physiological rationale in view of the empirical data on which it was based? Are the empirical results consistent with any other rationale for style? Are the results of this investigation consistent with any other rationale for style?

The empirical evidence (Zarrugh, Todd, and Ralston, 1974; Cavanagh and Williams, 1982; Morgan and Martin, 1986) that gave rise to the hypothetical physiological rationale was, in fact, inconclusive. Style was not varied within fixed combinations of stride length and stride frequency. Styles other than those which were used by the subjects and which reaquire lower levels of oxygen consumption may exist. The subjects preferred stride lengths were close to but not necessarily equal to the one associated with minimal oxygen consumption. These results are sufficiently vague to allow interpretations other than the original physiological rationale. It is possible that the preferred stride length is chosen on a basis other than the minimization of oxygen consumption or that other criteria are also relevant to the choice.

While oxygen consumption may not be the criterion which is minimized in the determination of the preferred stride length preferred running style it is not necessarily irrelevant either. From a physiological standpoint the relationship between oxygen consumption and human movement is more than just relevant, it is fundamental. Oxygen consumption is a direct reflection of the total energy metabolism occurring in the body and is acutely dependent on the amount of muscular contraction.

While the hypothesized form of **e** did not provide for a definitive explanation for the preferred style, there is some evidence that it too may be relevant. The most compelling was that no NP trial in any condition required less **e** than the PS mean ($\underline{\mathbf{e}}^{\underline{PS}}$). This was true even at the slow speed where $\underline{\mathbf{e}}^{\underline{PS}}$ was greater than at the two faster speeds. Any opportunity for an NP style to require less **e** than the PS mean was probably greatest at the slow speed. The variability (S^{PS}) around the PS mean was similar at all three speeds.

This suggests that the higher $\underline{e}^{\underline{PS}}$ in the slow condition was not due to carelessness by the subject. The influence of the constraints (C) and the objectives of the locomotion problem (L) may have interacted to force the subject to run with a style requiring relatively high levels of effort at the slow speed. Despite this the tendency to keep the levels of **e** low relative to the NP styles was maintained. This trend argues for the relevance of **e** to style in analogy to the fundamental relationship between stride length and oxygen consumption.

A reconsideration of the physiological rationale and of **e** is in order. In accepting that there exists some quantity whose optimization accounts for the preference for style there are two possible routes. The first is to derive an entirely new form of **e**(X(t)) and test for its minimization. To do this would be to ignore the relevance that the original form of **e** had to the preference for style. It would also mean ignoring the similarity in the relationships between **e** and style and oxygen consumption and stride length. It would be unreasonable to assert that oxygen consumption is irrelevant to the preference for stride length just because it is not quite minimized.

A second possibility exists. Consider the possibility that effort is not a quantity to be spared but as a resource whose effect is to be maximized. No longer would **e** be considered as a <u>criterion</u> measure but as a <u>constraint</u> on motion. From the investigator's perspective the scientific objective would still be the identification of the criterion. The goal of any locomotor activity is, in general, to move in the environment. Two obvious effects to maximize are the distance moved and the time spent moving. At a fixed combination of stride length and stride frequency speed is constant and therefore the maximization of distance is equivalent to the maximization of time. The distance that can be run or the time that can be spent running can be referred to, in common, simply as endurance. The second possibility can be restated as the hypothesis that preferred style is chosen so as to maximize endurance.

Recall that the term E could be considered a sum;

$$E' = \sum_{j=1}^{m} E_j'(F_j)$$

The mechanical analogy **e** was considered to be the sum of six separate measures $\mathbf{e}_{jk}$;

$$\mathbf{e} = \sum_{j=1}^{3} \sum_{k=1}^{2} | \mathbf{e}_{jk} |$$

In either case the implication of the summation is that an excess consumption of effort at one force (or torque) generator can be offset by a savings at another. A further implication is that there is a quantity of effort which can be distributed to each force generator as it is needed without restriction. With a given quantity of effort the style that required the least per stride is the one that would yield the most strides. Effort would be minimized and endurance maximized with one specific style.

It seems unlikely that the physiological quantities required for locomotion are as universally available as is implied. For example, while liver glycogen can be mobilized and carried in the blood stream muscle glycogen cannot. It is available, in a direct way, only to the muscle in which it is stored. There are undoubtedly other physiological and metabolic resources whose distribution is similarly restricted.

Consider a system whose resources are localized. Each force generator would have an exclusive and finite quantity of effort at its disposal. Any particular style would require the consumption of a certain amount of effort with each stride. Each style would have a critical force generator, the one which would deplete its source of effort first. The style that maximizes endurance is the one whose critical force generator uses the smallest fraction of its resources per stride as compared to the critical force generator (not necessarily the same one) for all other styles. Most interestingly, the style that maximizes endurance need not be the one that minimizes the total effort consumed per stride.

This observation may be relevant to the interpretation of physiological data. Oxygen consumption is, in a sense, analogous to the derived quantity **e** . It reflects a summation of energy metabolism and, as has been demonstrated empirically, it is acutely dependent on running style. The availability of the reactants of aerobic metabolism are, to a certain extent, restricted to the muscles in which they are stored. To the extent that this is true oxygen consumption can be misleading as a criterion measure of the efficacy of style. The stride length that minimizes oxygen consumption may not be the one that maximizes endurance. Clearly, the extent to which physiological resources are locally or globally available can impact on the interpretation of mechanical and physiological data.

The capacity of each of the six torque generators represented in the model could not be determined from the results of this experiment. As such the endurance hypothesis could not be tested directly. The fact that a new hypothesis arose from the results of the experiment and from a reconsideration of the physiological system is important. The physiological properties that may be relevant to the preference for style are emerging. Further speculation and hypothesis testing using a model with more properties like the real physiological system is called for.

Summary

The hypothesis that the derived quantity **e** is minimized in preferred style was not supported by the results of this investigation. The failure in the fast speed condition may have resulted from inappropriate filtering or by an inappropriate definition of the $\mathbf{e}^{PS}$ confidence interval.

The potential relevance of the derived quantity was inferred from two results. The first was that none of the NP styles required a level of **e** less than the preferred style mean value. The second was the similarity of the relationship between **e** and style obtained in this investigation to empirically derived relationships between stride length and oxygen consumption.

Several explanations for the results obtained in this investigation were presented. The first was that criteria such as balance, mechanical load, and aesthetics are also relevant to the preference for style. The failure to account for co-contraction and the action of two-joint muscles may have contributed to the failure to support the hypothesis. Another explanation was that physiological effort may act as a constraint to, rather than a criterion of, running style. The possibility that endurance is maximized was presented as an alternative hypothetical criterion.

One possible explanation for the failure to support the hypothesis has not been discussed. Implied in the discussion thus far is that the derived quantity **e** is the correct mechanical model of the physiological resources required for movement. The quantity **e** was derived in two steps. The first was to derive a relationship between physiological effort and muscular force. The second step was to transform that relationship from the muscular frame of reference to the angular degrees of freedom of the model used. The approximations and assumptions in both steps are considered in the next chapter in the context of a discussion of the validity of the investigation.

CHAPTER 4

GENERALITY

Introduction

This investigation was undertaken in an attempt to establish the relevance of a mechanical phenomenon, preferred running style, to a biological system, the subject. A number of the definitions and algorithms used in this investigation were subject to error. Errors tend to reduce the generality of findings, the range in which the findings are valid. The following is a discussion of the generality of this investigation. It is organized into the separate, arbitrarily defined categories of mathematical, mechanical, biological, and paradigmatic generality.

Mathematical Generality

While the laws of mathematics are accepted as being completely general the use of approximations, as in this investigation, introduces error. Analytical differentiation of the fourier series representation of the anatomical landmark position data yielded their velocities and accelerations. Joint angular velocities and accelerations were obtained using a finite difference method. The error in the differentiation technique wasn't assessed and no effort was made to control for it.. The bounds of the error for digital methods are dependent on the nature of the signal and the range in which they operate. The numerical method used to calculate the coefficients of the fourier series also introduced bounded, uncontrolled error.

Mechanical Generality

An idealized model of the human body provided the basis for the analysis used in this investigation. Motion was assumed to occur only in the sagittal plane. The foot, shank, and thigh were assumed to be rigid. No elastic or dissappative elements were represented. These simplifications from reality, in the biomechanical analyses of running, were accepted for the sake of experimental tractability. No assessment of their impact on the understanding of running style was made. As such the simplifications in the model represent a source of uncontrolled and unbounded conceptual error. The relevance of the mechanical model to the assessment of effort will be discussed further in the section on biological generality.

The inertia of the segments defined for the model were obtained from a set of regression equations. These were in turn obtained empirically from cadaver studies. As such they are dependent on the methods used and the population of subjects from which they were obtained. Two types of error result. Rotations may have occurred around points other than the endpoints defined for each segment. This is compounded by investigator error in placing the filming markers. The second is that the subject may have been ill-matched to the cadaver study population. Although both of these errors are systematic the extent to which they might influence the results is unknown.

The criterion quantity was defined as a function of mechanical impulse, the integral of torque with respect to time. The impulse applied to, and the momentum of, a body is defined with respect to particular axes. The sum of measures of momentum or impulse defined for different coordinates q_i is not, strictly speaking, a valid mechanical quantity. The criterion quantity was defined not as a mechanical quantity, per se, but as a model of physiological effort stated in mechanical units. It is not a measure of mechanical impulse.

There is a disadvantage in relying on a quantity that is not valid mechanically. The variation of mechanical impulse with changes in time and place is completely determined within the laws of classical mechanics. The variation of a non-classic quantity like **e** is not. If this experiment were to be repeated with the subject running uphill the results might be different. It would be difficult to separate the variation in **e** caused by the change in conditions from the variation caused by a change in criterion. The variation in a classically defined quantity would be easier to predict and account for.

Biological Generality

The analysis performed for this investigation is based on models of two major subsystems of the human body, the physiological system and the mechanical system. Two quantities, physiological effort and muscular force, their causal roles in human motion, and their relationship to each other were also modeled. The form of the relationship between physiological effort and muscular force was applied to the torque-driven, ideal mechanical model of the body. This yielded the derived quantity **e** . Throughout the discussion in chapter 3 it has been assumed that the derived quantity **e** was the correct mechanical model of the physiological resources required for motion. This is not true of course. Each step in the derivation required simplifications and assumptions that restrict the validity of the investigation. In this section some of these steps are discussed and, where possible, their implications for the results described.

The Physiological Model

The action of the physiological system in causing human motion was represented as a physiological law of motion in analogy to the mechanical law of motion. In the following section the simplifications within and the form of the physiological law of motion are reconsidered. The impact of revising the form on the assessment of effort is also discussed.

The acceleration of the body (X) was assumed to be a reflection of the action of the physiological system. The physiological laws of motion were assumed to be second order differential system;

$$X''(t) = X''(P(t))$$

This assumption was convenient because the mechanical laws of motion are of the same order. In principle the solutions of the two laws of motion could then be used to derive a relationship between the forces F_q and the action of the physiological system;

$$F_q = F_q(P(t)) \ .$$

If the true physiological laws of motion are of a different order than the mechanical laws then a problem would arise. For example assume that the rate of change of acceleration (X'''), sometimes called jerk, is the variable driven by the action of the physiological system;

$$X'''(t) = X'''(P(t)) \ .$$

The solution of this differential system would have a dependence on the acceleration at time t_o, a term which could not be eliminated during substitution. The relationship between the mechanical forces and the physiological system would then be of the form;

$$F_q(t) = F_q(P(t) , X''(t_0)).$$

The definition for effort would therefore also have a dependence on the term $X''(t_o)$;

$$E(t) = E(\ F_q(t), X''(t_o)\ )\ .$$

Consider now the possibility that the action the physiological system controls the velocity of the body;

$$X'(t) = X'(\ P(t)\ )\ .$$

After combining with the mechanical laws of motion and eliminating the redundant terms the representation of physiological effort would have an explicit dependence on $X'(t_o)$;

$$E(t) = E(\ F_q(t), X'(t_o)\ )\ .$$

Retaining an explicit dependence on an initial condition would complicate the derivation of **e** , the mechanical model, in a mathematical sense. It is also a problem in a conceptual sense. Some relevant physiological rationale for the dependence of effort, over time, on the mechanical state of the system at a particular point in time would be needed.

There are some mechanical properties of muscle which may have some relevance to the physiological causes of motion. The well-known length and velocity dependencies of muscular force were not incorporated into the model used in this experiment and to the extent that they may influence running style the model was inadequate.

There are several ways to incorporate the length and velocity dependencies into the model. The first might be to consider the physiological system and the length and velocity dependencies of muscle as independent factors. This would, in turn, make the forces F_q

dependent on the action of the physiological system and on the position and velocity of the body;

$$F_q = F_q(\, P(t), X'(t), X(t)\,) \;.$$

Transforming this equation with E(P) would yield a form of E with a dependence on the forces F_q and on the position and velocity of the body;

$$E(t) = E(\, F_q(t), X'(t), X(t)\,) \;.$$

This equation implies that the effort required to produce a particular impulse is qualified by the position and velocity of the body.

In Chapter 3 physiological effort was represented as the definite integral E;

$$E = \int_{t_0}^{t_f} E'(\, F_q(t)\,)\, dt$$

The rate at which the j^{th} muscle consumed effort, a term inside the integral, was represented by the following;

$$E_j'(t) = g_j(t)\;\; F_j(t) \;.$$

The mechanical model $\mathbf{e}_j$ was derived by considering the factor g_j to be constant over the period of the cycle. A length and velocity dependence into the mechanical model of physiological effort for the j^{th} muscle could be introduced by altering the form of g;

$$E_j'(t) = g_j(\, q_j'(t), q_j(t)\,)\;\; F_j(t) \;,$$

where q_j is the coordinate that parallels the line of force of the muscle.

Different running styles require, by definition, different patterns of position and velocity. Failure to rescale the impulse provided by each torque generator may have contributed to the failure to support the hypothesis.

A second method for introducing a position and velocity dependence into the analysis exists. It would be to assume that the P system itself had a dependence on the position and velocity and on some input control function A(t);

$$P(t) = P(\ A(t), X'(t), X(t)\ )\ .$$

This model is attractive because it provides a framework for introducing neurophysiological control into the model. It is interesting because it does not necessarily imply that the transformation of physiological effort to force is qualified by the position and velocity of the body. At the muscular level it implies that the factor g need not have an explicit dependence on the muscle length (q_j) and velocity (q_j'). As such the dependence described in the model would not influence the results of this experiment.

While the model described above does not qualify the transformation of physiological effort to muscular force it does influence motion. It states that the availability of physiological resources is dependent on position and velocity. This describes a constraint to motion.

There is a third possibility. It is possible that both the efficacy of the transformation, expressed by the factor g, and the availability of the physiological resources required for motion, E(t), are explicit functions of the position and velocity of the system. It is clear that some decision has to be made as to where the length and velocity dependencies of muscular force fit into the transformation of physiological resources into muscular force. A better understanding of this process will have consequences for the assessment of effort and the understanding of running style.

Ignore for the moment the dependencies on position and velocity and reconsider the relationship between the P system and the input A(t). For the purposes of this discussion A(t) can be considered as the action of the neurological system in controlling motion. It may be of value to consider the relationship as a dynamic one, one in which the the neurological system acts to change the configuration of the physiological system just as forces act to change the physical configuration of the body. Consider the first order differential equation

$$P'(t) = P'(A(t)) \ .$$

The solution (P(t)) of this differential equation would have a dependency on the initial condition $P(t_o)$;

$$P(t) = P(A(t), P(t_o)) \ .$$

This form is intriguing because it provides a framework for introducing the concepts of training and fatigue. The initial condition $P(t_o)$ could be made to depend on quantities such as the amount of effort that has already been consumed. The introduction of fatigue or training effects with the term $P(t_o)$ would not necessarily alter the transformation of physiological effort to force. The factor g would remain independent of $P(t_o)$. The dependence of P(t) on $P(t_o)$ acts to alter the availability of physiological resources and therefore acts to constrain movement. If this model correctly represents the influence of fatigue or training on motion then the results of this investigation would not have been affected.

Fatigue or training would have had an impact on the results if their effects are considered independent of the action of the P system in producing force;

$$F_q = F_q(P(t), \kappa(t)) \ .$$

where $k(t)$ is some factor that represents the influence of training or fatigue. Transformation of this relationship with E(P) yields the following;

$$E(t) = E(F_q(t), \kappa(t)) \ .$$

The resulting relationship, at the single muscle level, between E_m and F_m would be of the form

$$E_j(t) = g_j(\kappa(t)) F_j(t) \ .$$

A mechanical model of $\mathbf{e}_j$ derived from this equation would require a scaling factor to account for the influence of fatigue and training on the transformation of effort to force.

As was the case with the position and velocity dependencies, fatigue and training may influence motion in two ways. It may act to modify P(t) and therefore to modify the constraints on force or it may influence the transformation of physiological effort directly. Both scenarios may be true to a certain extent. Resolution of this uncertainty will be important to the understanding of style.

Mechanical System

A mathematical representation of an ideal mechanical system was used as a basis for the analysis performed in this investigation. The most fundamental concept for any model is its state. Its state is defined in terms of a position in a coordinate system, a frame of reference (F-O-R) within which the system exists. It is legitimate to state that a system can exist in different frames-of-reference at the same time. The choice of F-O-R within which to define a model is therefore dependent on the application.

The human body consists of a very large number (N) of particles. To consider a number of degrees of freedom less than 3N is to imply that constraints act between them. While it is clear that constraints do act between the particles it is not clear how many degrees of freedom exist. Further, it is not clear how many are relevant to the control or understanding of running style. The data collected for this experiment represent a projection of the subject's true motion onto a two-dimensional plane. Like the shadowy image cast by a clever hand on a blank wall, the data may not reveal the true configuration of the body that yields it.

A conceptual link between a physiological rationale for style and the forces required to produce that style is established earlier in this and in previous chapters. Muscles are clearly implicated as having a direct role in transforming physiological action into force. Among all the forces that act in human locomotion it is the set of muscular forces alone that can be varied directly in subservience to a preference for style. Each muscle produces force with respect to its own orientation. The orientation of each muscle can be thought of as an axis in a coordinate system. The set of these coordinates can be thought of as a muscular frame of reference. The position of each muscle and, therefore, the position of the entire body can, in principle, be designated in this F-O-R. It is equally true that the inertia of the entire body should be defineable in this F-O-R. The conceptual link between muscular forces and the physiological rationale for style would imply that a mechanical model defined in a muscular F-O-R would be relevant to the study of running style.

The model defined for this investigation deviated from the true muscular F-O-R in several ways. Angular rather than curvilinear muscular coordinates were used. The number of coordinates (degrees of freedom) used was much smaller than the number of muscles acting in locomotion. Elasticity within the system was ignored. The consequences of these simplifications for the assessment of force, and therefore of effort, is detailed below.

The segments of the upper body were ignored in this investigation. This was done to simplify the analysis and to eliminate the conceptual error in assuming that the upper body consisted of rigid segments. Two questions arise. Are the differences in effort required to move the upper body across styles small in comparison to the differences in the effort required to move the lower limb? If it were so then ignoring the upper body could be justified. A second question to ask is if the error in ignoring the upper body is of a greater or lesser order than the error that would result from modelling the upper body as a set of rigid segments. Neither question can be answered yet.

Gracovetsky (1985) has hypothesized that the spine plays an dominant role in locomotion. He went so far as to suggest that locomotion would not be possible without the spine. To the extent that this may be true the model used in this experiment was inadequate.

The degrees of freedom of the leg were defined as joint angles. The generalized forces acting on the leg were, as a consequence, defined as joint torques, torques which would cause acceleration in the angular coordinates. The consequences of this for the assessment of effort is described below.

An idealized muscle acting between two rigid segments is depicted in figure 4-1. Muscular tension produces equal and opposite forces (F_j) acting at the points of origin and insertion. Each of these forces can, in turn, be broken down into two components. The components F_n act normal to the segment and tangential to an arc tracing out the joint angle. The components F_c act parallel to the segment. The components F_n produce torque about the joint (J). The components F_c on the other hand tend to compress the segments together at the joint J.

The laws of motion for the model were represented symbolically as

$$F_q = MX''$$

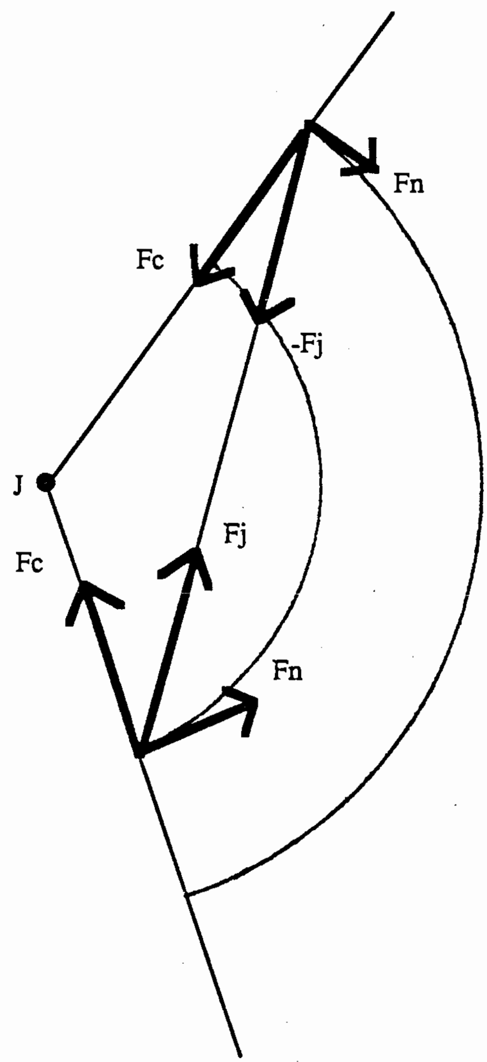
where F_q were a set of forces that act in parallel to the coordinates X. The forces of constraint were not represented implicitly. The configuration of the body was defined in terms of the joint angles. As such only F_n, the components of muscular force that are tangential to the arc of the curve and which produce the joint torques, are explicitly represented in the term F_q. Effort was hypothesized to be a function of the forces F_q. As such the effort required to produce the components F_c, which contribute to the forces of constraint, was not accounted for.

The inverse dynamic analysis used in the investigation yielded values of joint reaction force to which the components F_c contribute. It is not possible, however, to separate the magnitude of the forces F_c from the forces due to the motion of the segments relative to each other. It is not clear what impact the omission of the effort required to produce the components F_c has on the understanding of running style. If it varies across style in the same proportion as the effort required for the components F_n then the omission can be justified. If the variation is small in comparison then the omission can also be justified.

There are two methods by which this problem might be resolved. The first would be to re-write the laws of motion in terms of the linear muscular coordinates rather than the joint angles. The muscular forces would then appear explicitly. The second is to estimate or measure the distance from the point of attachments of each muscle to the joint center J.

Figure 4-1

Linear and Angular Muscular Coordinates



The components F_n could then be calculated directly from the torque values. The direction of the muscular force vector F_j would be defined by the line between the points of origin and insertion. Enough information to calculate the components F_c would then be available. Co-contraction and the action of two-joint muscles would complicate this procedure.

The failure to account for the action of two-joint muscles and for co-contraction is discussed in the previous chapter. These inadequacies are special cases of the general problem that occurs with the use of a coordinate system other than the muscular F-O-R.

Indeterminacy is the root of the inability to account for two-joint muscles and co-contraction. The number of degrees of freedom in the model is less than the number of muscles that can act to cause locomotion. The complexity of the model used in this investigation was limited by the complexity implicit in the inertial model used.

Co-contraction may occur at a joint which has more than one muscle crossing it. Such a joint can be said to be overdetermined. The existence of overdetermined degrees of freedom represents a problem for the motor control system. It must decide how to distribute the requirement for force and impulse amongst the various muscles acting in that degree of freedom. This is called the general distribution problem (GDP). Attempts to solve the GDP analytically are reviewed by Crowninshield and Brand (1982).

If a subject applies a particular criterion for movement then presumably this criterion is reflected in his solution to the GDP encountered in particular movements. Solving the GDP theoretically or empirically would be equivalent to identifying the criterion for movement.

Another shortcoming of the model is the failure to account for the elastic elements in the body, especially the series elasticity in muscle. Morgan, Proske, and Warren (1975), in a study of kangaroo hopping, found that varying the amount of elasticity in series with muscle can significantly alter the relationship between oxygen consumption and speed. As well, the stiffness of an elastic element in series with muscle will directly influence the displacement of a limb attached to that muscle. This information is relevant because the relationship between oxygen consumption and stride length, a function of the displacement of the limbs, was an important motivation for this investigation.

Models incorporating two components, a series elastic component (SEC) and contractile component (CC) have been found to be able to account for much of the behaviour of muscle (Chapman, 1985). Replacing each torque generator with two-component muscle models could accomplish two important objectives. The first would be to allow for the influence of series elasticity on oxygen consumption to be incorporated in the evaluation of effort. The second would be to increase the number of degrees of freedom of the model. A single joint angle could be replaced by two lengths, that of the CC and that of the SEC. Placing a two-component model on either side of each joint would further raise the number of degrees of freedom to four. This, along with a remodelling of the inertia of the body with a dependence on the configuration of the muscle itself could help resolve the indeterminacy. A number of problems would have to be overcome before this system could be used. The laws of motion for the model would become more complex. Redistributing the inertia of the body as a function of the length of the CC and the SEC would be problematic. Measuring the behaviour of the CC and the SEC, in situ, during locomotion is not yet feasible.

Muscles are arranged in curvilinear patterns within the body. Designating the configuration of a model of the body in terms of positions in curvilinear coordinates parallel to the muscles would make it very lifelike. It would also be very complex. Defining two-component muscle models in these curvilinear, muscular, coordinates would further enhance the validity, and the complexity, of the model.

It is possible to go on, ad infinitum, adding more and more complexity to the model. Individual motor units could be modeled for example. Pennate and longitudinal muscle architecture may be relevant to style. Hatze (1987) warned against oversimplification in biomechanical modeling. He argued that the neuromusculoskeletal system is inherently complex and for true understanding, that complexity must be dealt with. This trend must be balanced with experimental tractability of course. It remains a matter of conjecture how much more complex, if at all, the model must be before running style can be understood.

Paradigmatic Generality

This experiment was based on the premise that a preference for running style could be understood in terms of a teleological tendency in the control of human motion. The results of the experiment were equivocal. A criterion with which the preferred style could be determined was not identified. The question remains; did the subject choose his style so as to achieve the objective within the contraints and according to some criterion?

To argue successfully that the teleological theory is irrelevant to the understanding of running style would be a very difficult job. It must be argued that the empirical finding that oxygen consumption is low with preferred stride length is not relevant to that choice. It must be argued that the fact that **e** , the hypothetical measure of effort, is low in the preferred style is not relevant to the preference for style. It must be argued that the subject does not or cannot choose his style for a specific purpose.

There are some problems with the teleological view. The discussions presented previously in this chapter depict a chain of command. Forces can be said to cause motion. The action of the physiological system can be said to cause forces. A neurological input can be said to cause the action of the physiological system. Each step in the chain of command was introduced to allow for factors, the length and velocity dependence of muscular force for example, that may or do influence running behaviour. A question immediately arises which illustrates the weakness with the teleological theory; What causes the action of the neurological system? It is a problem of infinite regress. To understand locomotion fully the evergrowing conceptual chain of command must be broken. In fact it must be demonstrated that, at some level of control, the subject cannot or does not choose his style.

Kelso describes infinite regress as one of the problems of a conventional approach to the study of human movement. The goal of such an approach is to describe the nature of an implied internal representation of the movement. In Kelso's terms the present investigation is conventional. The implied internal representation of the movement is the criterion quantity. The goal of the investigation was to find the form of the criterion.

Kelso advocates another approach in the form of a question;

> " How can I understand the order and regularity that I see when people and animals generate actions as a necessary consequence - an a posteriori fact - of way the system is designed to function.? Or, said in another way, what are the constraints on the system that allow actions to arise? "

Restated in the context of this investigation Kelso's question might read: How does the design of the human biological system,with its mechanical, physiological, neurological, and other subsystems, determine style? What constraints are placed on the degrees-of-freedom of these subsystems in order to yield the preferred style?

Turvey (1982) refers to the work of N.Bernstein (1896-1966) in explaining the need to place constraints on a complex system or order to facilitate the control of its behaviour. In doing so two desirable affects are obtained, the requirements for attention and for action of a motor control executive are reduced. If enough constraints are placed on the system the requirements for an operator can be eliminated. This, in effect, solves the problem of infinite regress.

In this investigation the optimization of a criterion quantity was assummed to account for the reduction of the order of the system in order to yield the preferred style. In doing so however the hypothesis did not address the organization of the system directly but to the nature of movement made in response to an ambiguous command. In light of the comments of Kelso and Turvey it is possible to restate the objective of the present, conventional, investigation of human movement as an attempt to answer the following question. What is the rationale not for the preferred style, per se, but for the design of the biological system that yields the preferred style in the context of the experiment? Why resolve the redundancy so as to yield one particular style, the preferred, rather than any of the infinity of other possible and, in terms of the request to run, appropriate styles?

The departures from reality of the model used in this investigation have been discussed at length. The number and nature of these departures forces one to question whether the investigation can be justified. In fact the process produced some useful results. The relationship of **e** to preferred style was found to be similar in nature to an empirically derived relationship between freely chosen stride length and oxygen consumption. The potential roles of some properties of the physiological system in determining running style were illustrated.

How is it that useful information was obtained with the use of a model that is acknowledged to be incorrect? Perhaps the most important result of this investigation was illustrating the fact that as much understanding was gained by considering the form of the model as by considering the details of the results. This is justification for using a model that is a likeness, but not an exact likeness, of a biological system in order to understand its behaviour.

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APPENDIX A

EFFORT DATA

Table A-1

(e_{jk}, e) Effort (Nms)

Condition: Slow

trial	ankle		knee		hip		total
	e_{11}	e_{12}	e_{21}	e_{22}	e_{31}	e_{32}	e
PS_1	21.22	1.03	5.80	24.24	31.12	16.73	100.13
PS_2	23.19	1.00	7.81	14.17	39.33	14.08	99.58
PS_3	21.00	0.97	5.73	17.21	34.76	15.00	94.67
PS_4	25.34	1.01	9.16	14.11	43.22	14.31	107.15
PS_5	23.63	0.91	8.64	14.41	42.56	15.32	105.47
$\underline{e}^{PS}$	22.88	0.98	7.43	16.83	38.23	15.09	101.40
S^{PS}	1.80	0.05	1.60	4.34	5.19	1.05	5.00
HF_1	22.88	1.15	10.04	7.75	48.05	16.46	106.34
HF_2	21.99	1.21	18.38	3.91	67.72	17.20	130.40
KF_1							
KF_2	28.92	1.47	14.72	12.80	50.74	14.67	123.32
SL_1	27.64	1.39	13.16	11.58	52.44	19.98	126.21
SL_2	25.50	1.38	18.08	8.37	58.90	21.19	133.42
OJ_1	20.51	1.01	8.13	15.25	51.30	18.54	114.74
OJ_2	17.08	1.16	6.88	19.40	44.73	18.29	107.54

(KF_1 Discarded Trial)

Table A-2

<u>(e_{jk} , e) Effort (Nms)</u>

Condition: Medium

trial	ankle		knee		hip		total
	e_{11}	e_{12}	e_{21}	e_{22}	e_{31}	e_{32}	e
PS_1	21.52	1.62	5.37	30.16	13.46	19.50	91.62
PS_2	22.72	0.97	4.42	32.48	10.88	20.21	91.68
PS_3	16.89	1.92	5.52	36.12	9.09	19.67	89.21
PS_4	17.20	1.82	4.79	40.08	8.72	28.39	101.00
PS_5							
$\underline{e}^{PS}$	19.60	1.58	5.02	34.71	10.50	21.94	93.38
S^{PS}	2.97	0.42	0.51	4.34	2.16	4.31	5.21
HF_1	25.28	1.52	7.52	39.18	11.85	29.08	114.42
HF_2	25.27	1.64	7.95	44.75	11.72	35.06	126.38
KF_1	25.95	1.79	8.30	41.48	10.65	25.09	113.24
KF_2	23.93	1.25	7.27	32.51	11.23	23.13	99.31
SL_1	23.11	1.64	8.58	26.94	18.74	27.15	106.15
SL_2	45.00	1.35	8.94	11.33	33.01	19.84	119.47
OJ_1							
OJ_2	3.67	8.13	5.48	62.40	9.04	33.95	122.67

(PS_5 Discarded Trial)
(OJ_1 Discarded Trial)

Table A-3

(e_{jk}, e) Effort (Nms)

Condition: Fast

trial	ankle		knee		hip		total
	e_{11}	e_{12}	e_{21}	e_{22}	e_{31}	e_{32}	e
PS_1	14.73	2.21	6.23	35.21	10.00	20.35	88.73
PS_2	17.34	2.40	6.39	39.44	10.36	27.00	102.93
PS_3	18.36	2.08	7.27	38.48	10.92	23.61	100.74
PS_4	16.80	2.19	7.22	34.95	11.30	22.18	94.63
PS_5	18.31	2.09	6.55	36.17	11.42	20.36	94.89
$\underline{e}^{PS}$	17.13	2.20	6.73	36.85	10.84	22.70	96.38
S^{PS}	1.49	0.13	0.49	2.01	0.61	2.77	5.61
HF_1	25.25	1.79	9.76	40.65	15.93	28.59	121.97
HF_2	22.43	1.59	9.45	32.71	14.92	30.44	111.55
KF_1	29.47	1.74	12.14	28.95	42.59	22.97	137.85
KF_2	30.13	2.32	12.01	22.94	39.88	21.03	127.67
SL_1	33.21	2.03	20.20	16.18	80.55	33.47	185.57
SL_2	31.88	2.15	16.17	24.07	62.72	36.75	173.73
OJ_1	19.89	1.65	8.45	30.25	41.64	27.96	129.84
OJ_2	23.81	1.20	9.46	27.60	44.93	23.14	130.13

Figure A-1.1

<u>(e_{3k}) Hip Effort (Nms)</u>

Condition: Slow

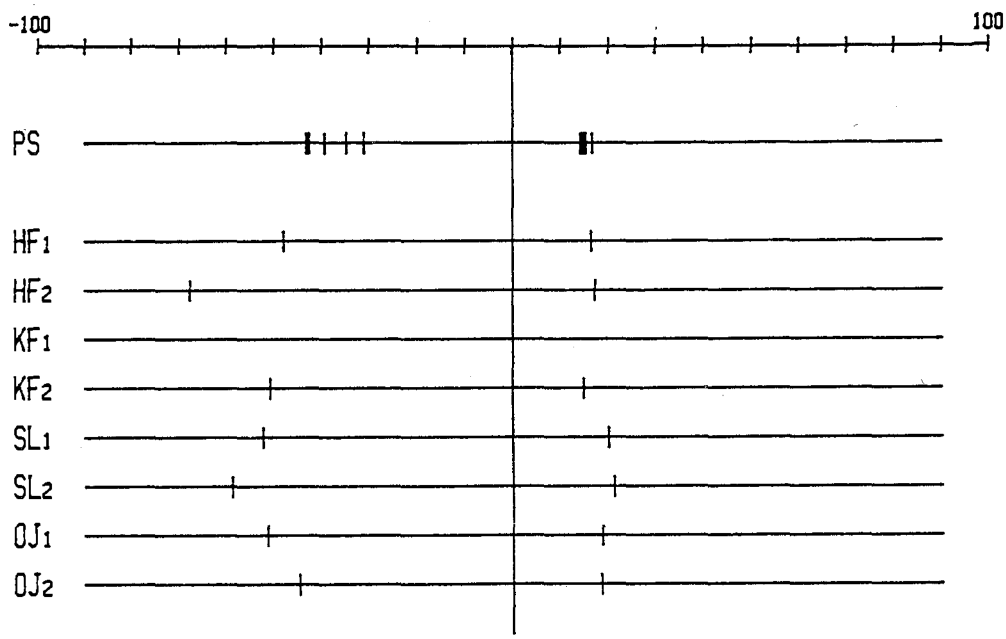


Figure A-1.2

(e_{2k}) Knee Effort (Nms)

Condition: Slow

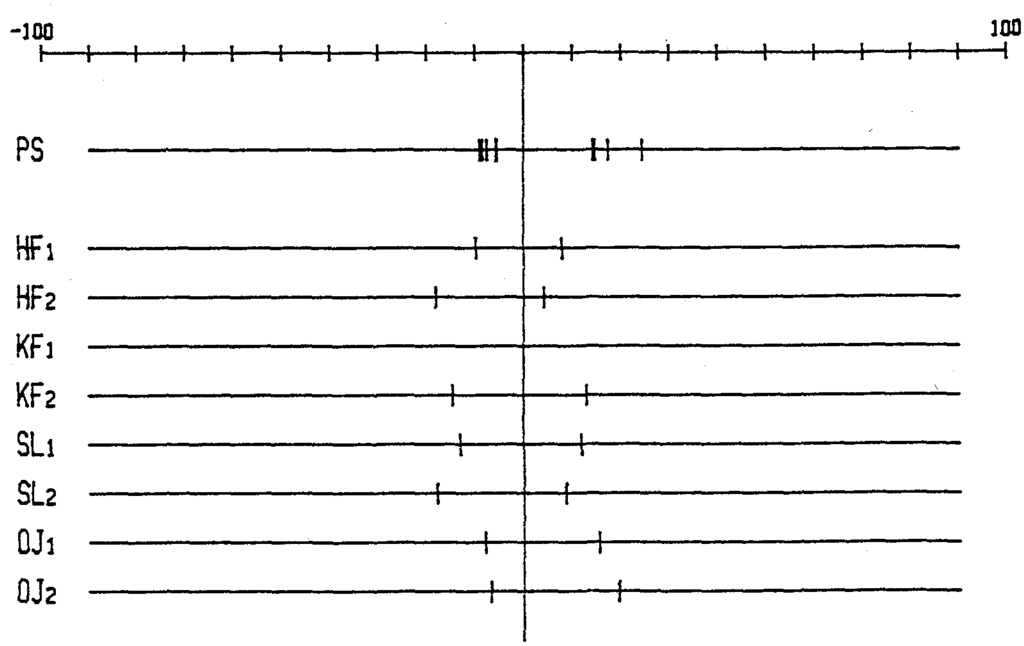


Figure A-1.3

(e_{1k}) Ankle Effort (Nms)

Condition: Slow

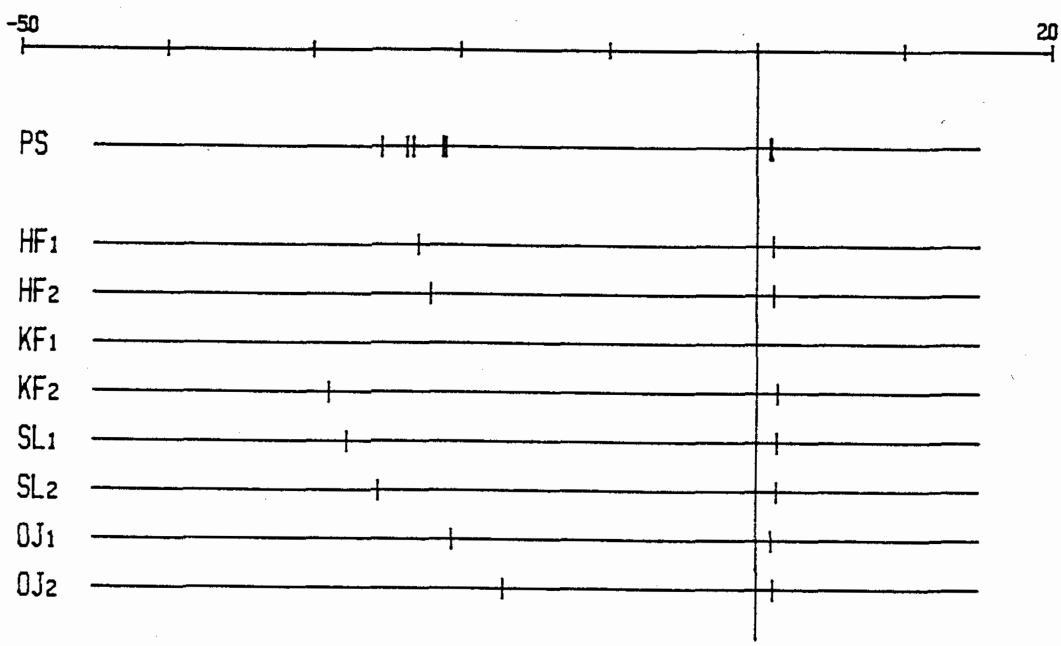


Figure A-2.1

(e_{3k}) Hip Effort (Nms)

Condition: Medium

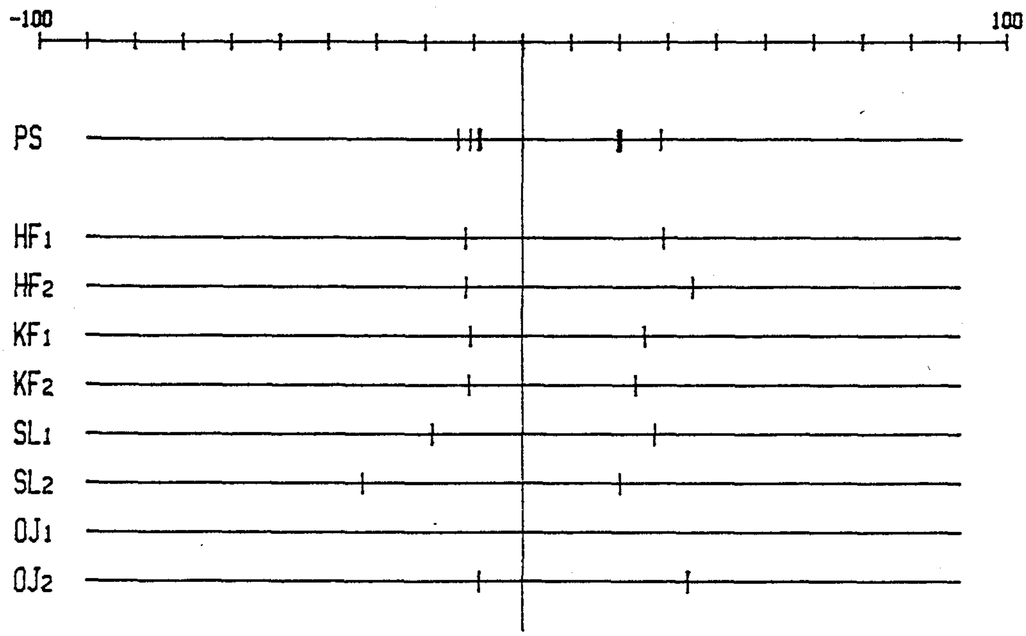


Figure A-2.2

(e_{2k}) Knee Effort (Nms)

Condition: Medium

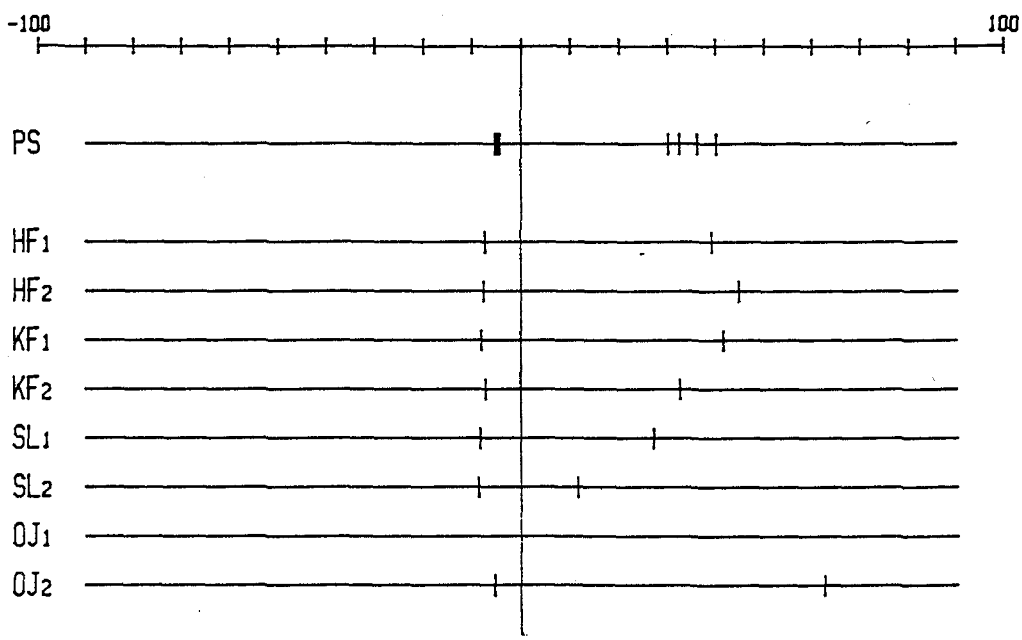


Figure A-2.3

(e_{1k}) Ankle Effort (Nms)

Condition: Medium

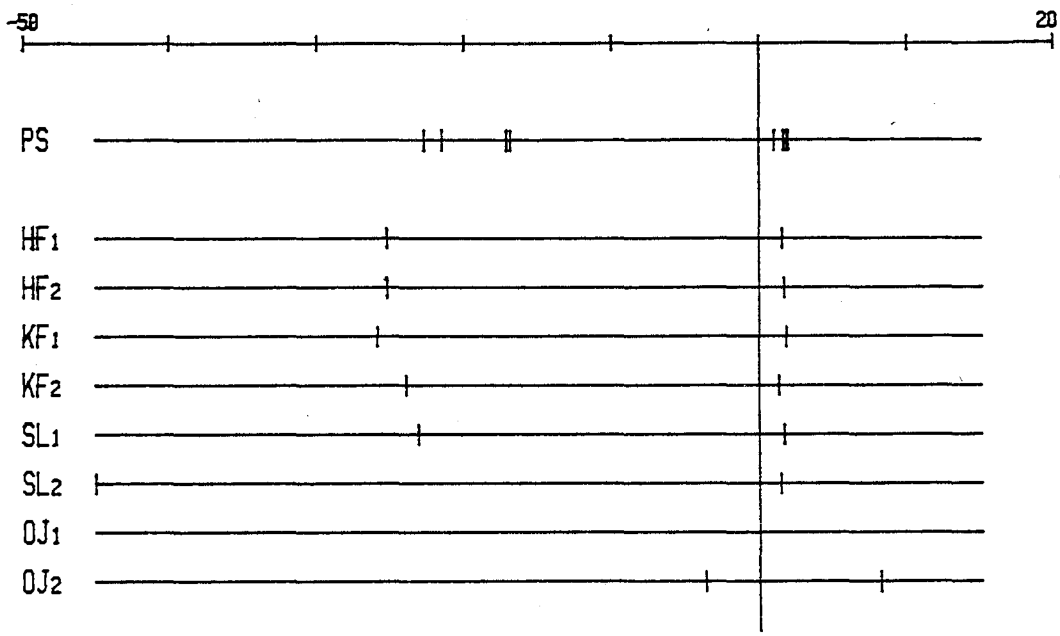


Figure A-3.1

(e_{3k}) Hip Effort (Nms)

Condition: Fast

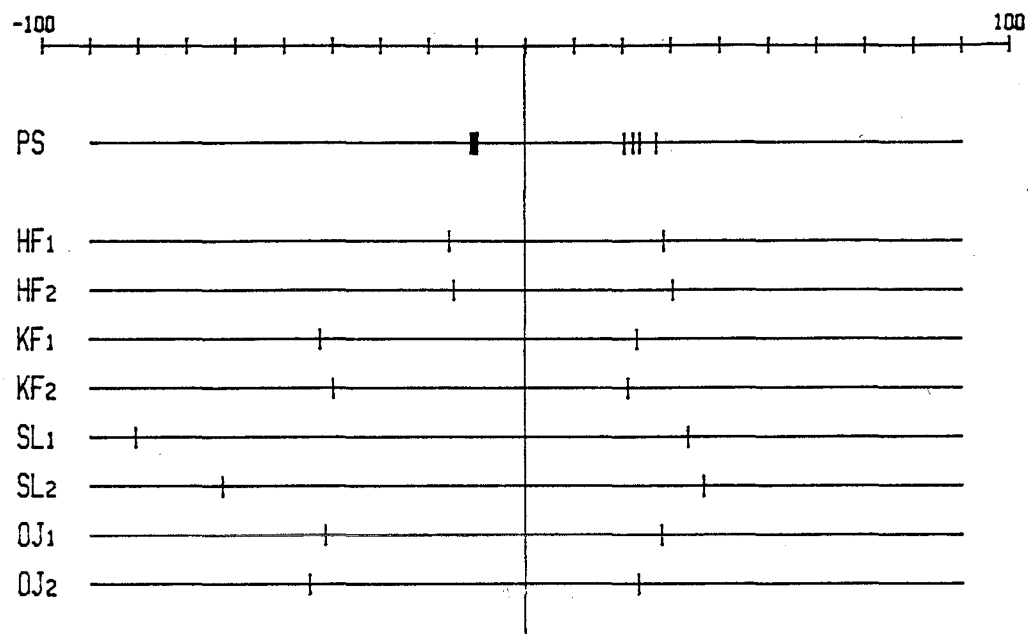


Figure A-3.2

(e_{2k}) Knee Effort (Nms)

Condition: Fast

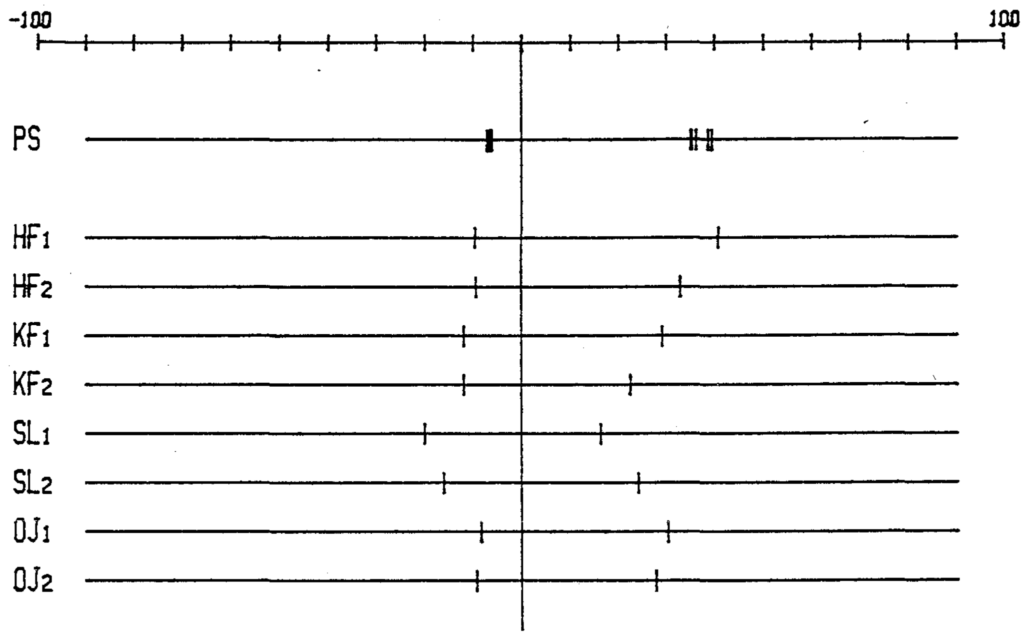
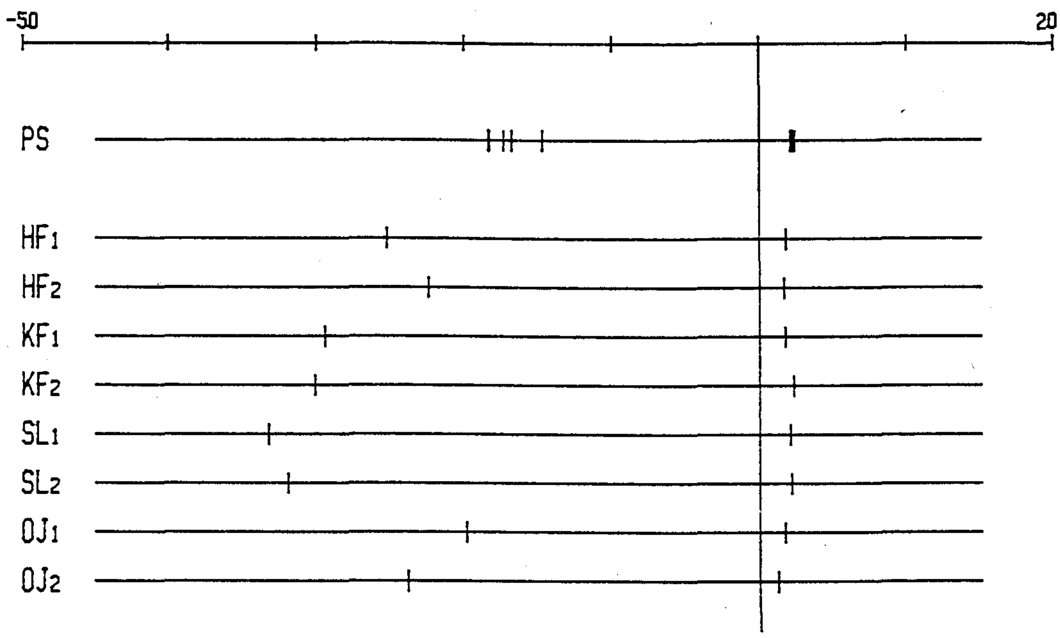


Figure A-3.3

(e_{1k}) Ankle Effort (Nms)

Condition: Fast



APPENDIX B

SPEED CORRECTED DATA

Table B-1

(e^c_{jk} , e^c) Speed Corrected Effort (Nms)

Condition: Slow

trial	ankle		knee		hip		total
	e_{11}	e_{12}	e_{21}	e_{22}	e_{31}	e_{32}	e
PS_1	21.22	1.03	5.80	24.24	31.12	16.73	100.13
PS_2	23.15	1.00	7.80	14.14	39.26	14.05	99.38
PS_3	24.42	0.99	5.84	17.55	35.45	15.30	96.56
PS_4	24.61	0.98	8.89	13.70	41.97	13.89	104.04
PS_5	28.86	0.92	8.73	14.56	42.99	15.48	106.53
$\underline{e}^{PS}$	24.45	0.98	7.41	16.84	38.16	15.09	101.33
S^{PS}	2.81	0.04	1.51	4.41	4.90	1.16	3.95
HF_1	22.79	1.15	10.00	7.72	47.86	16.40	105.91
HF_2	21.68	1.19	18.12	3.85	66.67	16.96	128.58
KF_1							
KF_2	30.36	1.55	15.46	13.44	53.28	15.41	129.49
SL_1	28.47	1.43	13.56	11.93	54.02	20.58	129.99
SL_2	25.30	1.37	17.94	8.30	58.42	21.02	132.35
OJ_1	21.33	1.05	8.46	15.86	55.35	19.28	119.33
OJ_2	17.25	1.17	6.95	19.59	45.17	18.47	108.61

(KF_1 Discarded Trial)

Table B-2

(e^c_{jk}, e^c) Speed Corrected Effort (Nms)

Condition: Medium

PS_1	20.72	1.53	5.06	28.41	12.68	18.37	86.31
PS_2	23.63	1.01	4.60	33.78	11.31	21.01	95.34
PS_2	16.62	1.89	5.43	35.55	8.95	19.36	87.79
PS_4	17.89	1.89	5.00	41.68	9.07	29.52	105.04
PS_5							
$\underline{e}^{PS}$	19.72	1.58	5.02	34.85	10.50	22.07	93.62
S^{PS}	3.12	0.42	0.34	5.47	1.81	5.09	8.58
HF_1	25.00	1.51	7.43	38.75	11.72	28.76	113.17
HF_2	26.78	1.74	8.43	47.43	12.42	37.16	133.96
KF_1	27.24	1.88	8.71	43.55	11.18	26.34	118.91
KF_2	23.88	1.25	7.25	32.44	11.21	23.09	99.12
SL_1	22.25	1.58	8.26	25.95	18.04	26.14	102.22
SL_2	45.45	1.36	9.03	11.44	33.34	20.04	120.67
OJ_1							
OJ_2	3.85	8.54	5.75	65.52	9.49	35.65	128.80

(PS_5 Discarded Trial)
(OJ_1 Discarded Trial)

Table B-3

(e^c_{jk} , e^c) Speed Corrected Effort (Nms)

Condition: Fast

trial	ankle		knee		hip		total
	e_{11}	e_{12}	e_{21}	e_{22}	e_{31}	e_{32}	e
PS_1	14.88	2.23	6.29	35.56	10.10	20.56	89.62
PS_2	18.04	2.49	6.64	41.02	10.78	28.08	107.05
PS_3	18.55	2.10	7.35	38.87	11.03	23.85	101.75
PS_4	15.92	2.08	6.85	33.13	10.71	21.02	89.71
PS_5	18.49	2.12	6.61	36.53	11.53	20.56	95.84
$\underline{e}^{PS}$	17.20	2.20	6.75	37.02	10.83	22.82	96.79
S^{PS}	1.68	0.17	0.39	3.04	0.52	3.25	7.62
HF_1	26.26	1.86	10.15	42.28	16.56	29.73	126.85
HF_2	20.95	1.49	8.83	30.55	13.94	28.44	104.19
KF_1	28.55	1.68	11.76	28.05	41.27	22.26	133.57
KF_2	31.64	2.44	12.61	23.41	41.88	22.08	134.06
SL_1	34.54	2.11	21.01	16.76	83.77	34.81	193.00
SL_2	31.56	2.13	16.00	23.83	62.09	36.39	171.99
OJ_1	18.12	1.51	7.69	27.56	37.93	25.48	118.28
OJ_2	24.04	1.21	9.55	27.87	45.38	23.37	131.43

APPENDIX C

REFILTERED DATA

Table C-1.1

(e_{jk} , e) Refiltered Effort (Nms)

Condition: Slow Filter: Low

trial	ankle		knee		hip		total
	e_{11}	e_{12}	e_{21}	e_{22}	e_{31}	e_{32}	e
PS_1	21.14	0.94	5.31	23.59	30.95	16.34	98.27
PS_2	23.20	1.00	7.37	13.53	39.29	13.79	98.18
PS_3	20.95	0.90	5.76	17.00	34.24	14.29	93.13
PS_4	25.27	0.94	8.94	13.65	42.80	13.71	105.32
PS_5	23.57	0.86	8.55	14.21	42.21	14.94	104.34
$\underline{e}^{PS}$	22.83	0.93	7.19	16.40	37.89	14.61	99.85
S^{PS}	1.80	0.05	1.62	4.26	5.15	1.08	5.01
HF_1	22.81	1.10	9.58	7.42	48.58	17.08	106.57
HF_2	21.87	1.10	17.45	3.40	68.53	17.90	130.25
KF_1							
KF_2	28.90	1.45	13.57	11.49	49.77	13.58	118.77
SL_1	27.56	1.30	11.96	10.25	51.90	19.34	122.30
SL_2	25.38	1.26	16.85	7.23	58.17	20.61	129.50
OJ_1	20.51	1.00	7.16	14.28	49.83	16.99	109.77
OJ_2	17.04	1.12	6.09	18.74	44.04	17.70	104.73

(KF_1 Discarded Trial)

Table C-1.2

(e_{jk}, e) Refilted Effort (Nms)

Condition: Medium Filter: Low

trial	ankle		knee		hip		total
	e_{11}	e_{12}	e_{21}	e_{22}	e_{31}	e_{32}	e
PS_1	21.36	1.42	5.34	30.04	12.03	18.25	88.43
PS_2	22.66	0.91	4.23	32.03	11.08	20.41	91.32
PS_3	16.80	1.81	5.35	35.56	8.57	19.18	87.28
PS_4	17.16	1.76	4.49	39.42	8.82	28.88	100.52
PS_5							
$\underline{e}^{PS}$	19.49	1.47	4.85	34.26	10.13	21.68	91.88
S^{PS}	2.96	0.42	0.58	4.13	1.70	4.88	6.00
HF_1	25.09	1.35	7.75	38.41	12.68	30.90	116.18
HF_2	25.17	1.56	7.54	43.18	11.94	36.49	125.86
KF_1	25.88	1.74	8.35	40.76	12.42	27.89	117.03
KF_2	23.83	1.20	6.90	31.12	11.67	24.12	98.83
SL_1	23.08	1.61	7.94	25.25	19.34	28.62	105.83
SL_2	44.93	1.30	8.73	10.20	32.75	20.76	118.18
OJ_1							
OJ_2	3.58	8.05	5.37	62.35	9.24	34.30	122.88

(PS_5 Discarded Trial)
(OJ_1 Discarded Trial)

Table C-1.3

(e_{jk}, e) Refiltered Effort (Nms)

Condition: Fast   Filter: Low

trial	ankle		knee		hip		total
	e_{11}	e_{12}	e_{21}	e_{22}	e_{31}	e_{32}	e
PS_1	14.62	2.12	6.01	33.89	10.85	21.63	89.12
PS_2	17.21	2.24	6.27	38.05	11.19	28.06	103.02
PS_3	18.22	1.94	7.43	37.30	10.41	23.38	98.67
PS_4	16.72	2.09	7.41	34.19	10.90	21.83	93.14
PS_5	18.25	2.02	6.37	35.33	10.61	20.09	92.66
$\underline{e}^{PS}$	17.00	2.08	6.70	35.75	10.79	23.00	95.32
S^{PS}	1.49	0.11	0.67	1.85	0.30	3.06	5.50
HF_1	25.13	1.66	8.59	38.24	16.13	30.19	119.93
HF_2	22.32	1.45	8.27	29.70	13.81	30.54	106.10
KF_1	29.34	1.60	11.67	26.54	40.99	21.92	132.05
KF_2	30.08	2.16	11.61	21.38	40.30	19.72	125.25
SL_1	33.06	1.91	18.83	14.31	80.97	33.45	182.53
SL_2	31.78	2.07	15.31	21.48	60.54	34.67	165.85
OJ_1	19.71	1.49	7.99	29.70	40.02	26.64	125.55
OJ_2	23.72	1.14	9.23	27.18	44.79	22.93	129.00

Table C-2.1

$(\underline{e_{jk}}, \underline{e})$ Refiltered Effort (Nms)

Condition: Slow Filter: High

trial	ankle		knee		hip		total
	e_{11}	e_{12}	e_{21}	e_{22}	e_{31}	e_{32}	e
PS_1	21.14	0.94	5.31	23.59	30.95	16.34	98.27
PS_2	23.20	1.00	7.37	13.53	39.29	13.79	98.18
PS_3	20.95	0.90	5.76	17.00	34.24	14.29	93.13
PS_4	25.27	0.94	8.94	13.65	42.80	13.71	105.32
PS_5	23.57	0.86	8.55	14.21	42.21	14.94	104.34
$\underline{e}^{PS}$	22.83	0.93	7.19	16.40	37.90	14.61	99.85
S^{PS}	1.80	0.05	1.62	4.26	5.15	1.08	5.01
HF_1	22.81	1.10	9.58	7.42	48.58	17.08	106.57
HF_2	21.87	1.10	17.45	3.40	68.53	17.90	130.25
KF_1							
KF_2	28.90	1.45	13.57	11.49	49.77	13.58	118.77
SL_1	27.56	1.30	11.96	10.25	51.90	19.34	122.30
SL_2	25.38	1.26	16.85	7.23	58.17	20.61	129.50
OJ_1	20.51	1.00	7.16	14.28	49.83	16.99	109.77
OJ_2	17.04	1.12	6.09	18.74	44.04	17.70	104.73

(KF_1 Discarded Trial)

Table C-2.2

<u>(e_{jk} , e) Refiltered Effort (Nms)</u>

Condition: Medium Filter: High

trial	ankle		knee		hip		total
	e_{11}	e_{12}	e_{21}	e_{22}	e_{31}	e_{32}	e
PS_1	21.58	1.70	5.29	30.11	14.58	20.66	93.93
PS_2	22.85	1.10	4.88	32.93	11.58	20.85	94.16
PS_3	16.94	1.97	6.38	37.02	10.13	20.74	93.18
PS_4	17.31	1.95	5.03	40.35	8.92	28.63	102.18
PS_5							
$\underline{e}^{PS}$	19.67	1.68	5.39	35.10	11.30	22.72	95.86
S^{PS}	2.99	0.41	0.68	4.50	2.44	3.94	4.23
HF_1	25.30	1.52	7.87	39.38	13.19	30.31	117.57
HF_2	25.34	1.71	8.03	44.72	11.82	35.14	126.76
KF_1	25.98	1.82	8.15	41.36	10.48	24.91	112.69
KF_2	24.03	1.35	7.49	32.62	12.17	24.00	101.66
SL_1	23.20	1.71	8.74	27.04	20.38	28.91	109.98
SL_2	44.99	1.35	9.57	11.73	34.69	21.40	123.73
OJ_1							
OJ_2	3.78	8.27	6.12	63.03	10.89	35.85	127.95

(PS_5 Discarded Trial)
(OJ_1 Discarded Trail)

Table C-2.3

(e_{jk}, e) Refiltered Effort (Nms)

Condition: Fast Filter: High

trial	ankle		knee		hip		total
	e_{11}	e_{12}	e_{21}	e_{22}	e_{31}	e_{32}	e
PS_1	14.99	2.50	7.37	36.41	11.30	21.76	94.33
PS_2	17.38	2.44	6.71	39.72	11.44	28.07	105.76
PS_3	18.56	2.30	7.84	39.13	13.24	26.04	107.11
PS_4	16.89	2.28	8.27	36.00	12.59	23.58	99.60
PS_5	18.47	2.24	7.99	37.57	13.13	21.94	101.34
$\underline{e}^{PS}$	17.26	2.35	7.63	37.77	12.34	24.38	101.63
S^{PS}	1.45	0.11	0.61	1.63	0.92	2.73	5.08
HF_1	25.41	1.93	10.12	40.76	14.83	27.36	120.40
HF_2	22.59	1.72	10.12	33.21	17.07	32.49	117.19
KF_1	29.72	2.00	12.60	29.34	43.41	23.71	140.77
KF_2	30.14	2.39	13.14	21.95	40.07	21.72	129.41
SL_1	33.36	2.16	21.38	16.99	83.80	36.61	194.30
SL_2	31.91	2.16	16.98	24.53	64.70	38.51	178.79
OJ_1	19.97	1.73	8.80	30.74	42.17	28.48	131.78
OJ_2	23.88	1.28	10.65	28.80	45.70	23.86	134.17