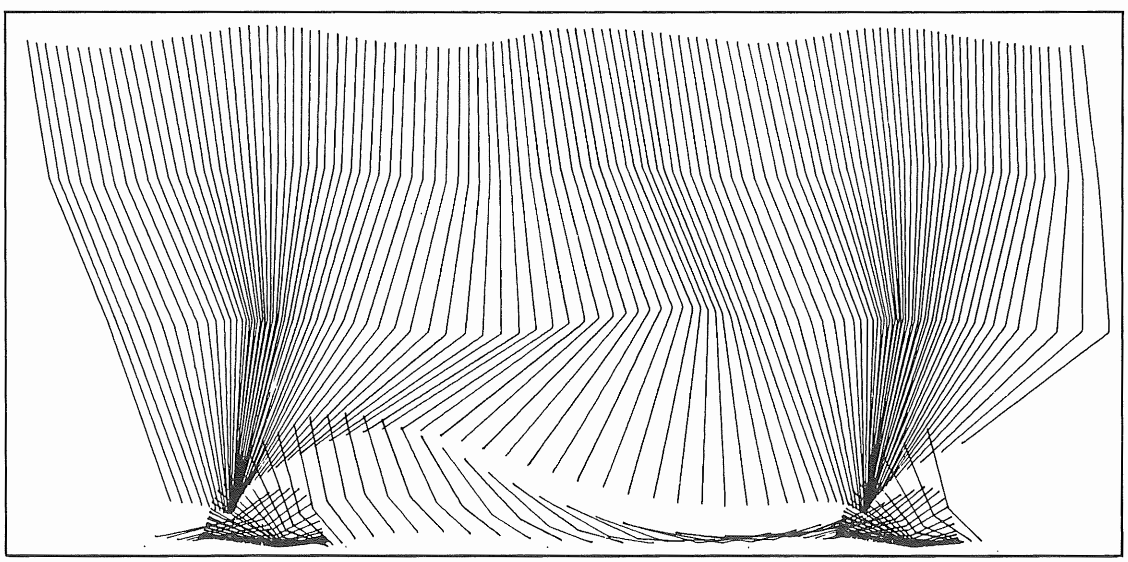
The BIOMECHANICS and MOTOR CONTROL of HUMAN GAIT



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Dedication

This book is dedicated to my wife, Judy, and my three children, Merriam, Andrew and Bruce.

Table of Contents

Acknowledgements

This book was the result of the efforts of many people over the past decade: students who gave their time as subjects in our studies of normal gait; graduate students and research assistants who collected and analysed the data; Mr. Paul Guy, who developed most of the software and hardware, and Mr. Steve Cornall, who prepared most of the graphs and tables. Also, I would like to thank Barb Carroll who typed the manuscript and painstakingly proofread the text and extended her knowledge in specialized text processing.

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FOREWORD

"Walking"

Walking is for moving from one place to another-
to go to table for breakfast, to climb stairs to
bed, to meet a friend, to walk the aisles at the
food mart.
Walking is for enjoying from one day to another-
to pl ay hopscotch, to play jump the rope, to
play cops and robbers, to go dancing through
the trees in the park, at parties for weddings
and birthdays, with the beloved at nightfall, to
meet the returning child.
Walking is for moving from place to place-is for
dancing.
Walking is very important for meeting the world,
for growing up, for retreating to solitude, for
returning to join again, for carrying the day's
tasks, for belonging.
Walking makes a big difference in how one's life
turns out, in whether one has a life at all, so
it's very important for almost everything-like
breathing.
Walking can make the difference whether we ever
stand up straight or not.
But walking does not come automatically like
breathing. It must be learned. It begins while
we are still very small, lying alone in our bed
quite dependent on the others we see walking.
We see the others walking-the mother who nurses
and feeds us, the father who holds and plays
with us, the sister or brother or neighbour who
walks toward us, smiles to us, then walks
away, leaving us there alone on our back in
our bed.
We see others walking, see what they can do with
their walking, what we cannot do, and we
want to do it too.
So lying on our back or face soon is not enough.
We struggle to crawl-then we crawl every-
where we can. Then we try pulling ourselves
to stand at table leg, at father's leg, at the
stairsteps, by holding to the drapes, by hang-
ing to the tablecloth.
We grunt and push and pull and fall and roll and
bump, then try again and keep it up over and
over again, and never quit in spite of face-
falls and nose bruises - all because we want to
be what we feel persons come to be by walk-
ing.
Walking is expensive. It is not learned without
risking, without falling so hard it hurts, with-
out slipping on the stairs so suddenly it fright-
ens us and our parents, keeps us from trying
again-for a while at least.

Focus of the Textbook

Gait (walking and running) is the most common of all human movements. It is one of the most complex totally integrated movements and yet is probably the most taken-for-granted. Also, it has been described and analysed more than any other movement, and scores of laboratories are dedicated to the analysis and assessment of walking and running, both normal and pathological.

Because this textbook is intended to be a resource of normative data and analyses, and its interpretation, it is not intended to be a text on how to measure and analyse. The techniques for calculating most gait-related variables are well covered in the literature and in textbooks. However, relating to a few methodological issues, the author reviews different techniques and comments on their advantages and disadvantages and, in some cases, documents erroneous techniques that have hindered valid interpretations. References to the more important appropriate research reports or texts are made throughout the book. Thus, for the newer researcher, it will provide a starting point for the more important references as of the date of publication.

Terminology, conventions and standards widely vary from lab to lab and this text attempts to resolve that problem area by providing a detailed and comprehensive list of terms whose definitions have been agreed upon by a working group of the International Society of Biomechanics.

It is recognized that the 3-D data analysis of level walking is not yet fully documented nor are normal profiles available for special gait patterns: walking up and down stairs, running, walking backwards, and so on. It is planned that future additions will document those additional aspects of human gait, commencing in 1989.

Detailed Topics

1) A glossary of terms, definitions and conventions.

   i) Gait specific terminology, i.e., stance, heel contact.

   ii) Biomechanical, anatomical and neurological terms that apply to gait.

   iii) Recommended conventions and abbreviations for use in the description and analysis of gait.

2) A handbook of normative data on a wide range of gait variables.

   i) Temporal, cadence and stride length measures.

   ii) Kinematic variables over the stride period: joint angles, limb velocities, accelerations and trajectories.

   iii) Kinetic variables over the stride period: reaction forces, moments-of-force, power, energy.

   iv) Electromyographic variables over the stride period: from 16 major muscles.

Full intra and inter-subject curves are presented at three walking cadences: slow, natural and fast, and measures of variability are reported. A full description of each family of curves is presented and an interpretation is offered. On some of the more important baseline curves (moments-of-force, joint angles, etc.) a table of values is reproduced so that the user may input those curves into his or her computer without the error and labour of trying to read values off the graphs.

Who Will Be Interested

The textbook will address the needs of the following professionals:

i) Rehabilitation personnel: rehab engineering, clinical kinesiology, physiotherapy, orthopaedic surgery, sports medicine.

ii) Researchers in basic human movement: kinesiology, biomechanics, bioengineering, neural control.

iii) Physical education researchers and teachers.

1.0 Gait Analysis: Considerations and Terminology

Human gait is the most common of all human movements. It is one of the more difficult movement tasks that we learn, but once learnt it becomes almost subconscious. Only when this complex neuromuscular skeletal system is disturbed by traumatic injury, neurological damage, gradual degeneration, or fatigue do we realize our limited understanding of the complex biomechanics and motor control mechanisms. The purpose of this text is to assist in the understanding of those mechanisms, and, through my sixteen years of experience in the assessment of human gait, attempt a summary of the state of knowledge as it appears in the literature, and as represented by normal data collected in my laboratory. Thus, it is hoped that the material content of this book will serve as a reference for those who need baseline measures on normal gait, for those who wish to teach the specific movement of gait, and for those who wish normative curves for mathematical modelling of gait. It will be evident from a close assessment of these normal patterns, their magnitudes, consistency and variability that there are several major sub-functions involved in walking. A look at the motor patterns reveals how the neuromuscular system achieves those tasks and gives some insight into the essential features of that control system.

1.01 What Measures Are Important?

In the several thousand papers published on human gait, the measures reported and the measurement techniques vary considerably. Each paper appears to be characteristic of the understanding and interests of the investigator and the logistic support available. Clinical investigators tend to look at output measures such as stride length, cadence, and joint angles because that is what they visually observe. Neurological researchers focus on EMG measures and biomechanical investigators analyse kinematics, reaction forces, moments-of-force, powers, energies, etc. There is no doubt that certain information is more available from certain measures than from others. For example, efficiency information will be seen in the energy and power analyses while neuropathies manifest themselves in the EMG and moment-of-force profiles. However, we must be acutely aware as to those variables that reflect the cause of the gait pattern as opposed to those that describe the effect. EMG signals, moments of force and power get close to the cause of any movement while kinematics, momentum, stride length, cadence and ground reaction forces merely reflect many output integrated effects. The time-history of knee angle, for example, is the net result of a score of muscle forces acting on both lower limbs. Also, the ground reaction forces, which reflect the net mass-acceleration products of all body segments, are the net result of all active muscular forces.

1.02 Normal Patterns and the Problem of Intra- and Inter-subject Variability

We do not walk as robots, nor do we walk the same as our neighbour. A given person will perform his or her walking pattern in a fairly repeatable and characteristic way, sufficiently unique that it is possible to recognize a person at a distance by their gait. The variability of that pattern on a stride-to-stride and day-to-day basis is moderately low. Across any group of normal subjects we have greater but not excessive variability. Because of differences of height, body mass, age, cadence and sex, we are challenged to identify methods that will "normalize" our measures so as to give us less variable and more universal patterns. For example, we have normalized our moment-of-force and power patterns by dividing by body mass and clustered our results into three cadence groups: slow, natural and fast. Others have normalized walking velocity by dividing by body height and reported velocity as statures/second. Such normalization requires access to large data bases, and, in our case, has certainly narrowed the variability. More optimum methods may ultimately yield more universal profiles.

1.03 Indeterminacy (Flexibility) at the Motor Level

Walking involves the integrated activity of muscles acting across many joints. Because of the synergistic and antagonistic nature of many of these muscles, it is quite possible to achieve the same movement, as measured kinematically, from a score of different combinations of muscle patterns. As the famous Russian scientist, Bernstein (1967), noted:

> "The coordination of a movement is the process of mastering redundant degrees of freedom of the moving organ, in other words, its conversion to a controllable system."

This redundancy drastically confounds the assessment problem such that there is no unique solution to a given kinematic pattern. For example, the ankle and knee angle histories (which are regularly reported by many laboratories) could remain the same even though there were drastic changes in the moment of force or EMG profiles. Thus, gait laboratories who specialize exclusively in kinematic measures are left, at best, to speculate about the motor patterns that are the cause of the gait patterns they are reporting. For example, a patient is observed to be walking with a stiff leg during stance. Three possible abnormally high muscle forces could prevent normal knee flexion during weight bearing: higher than normal quadriceps activity, abnormally high hip extensor forces or hyperactivity of the plantarflexors. There is nothing in the observed knee angle curve that could pinpoint which one of these abnormalities was the cause, or even if elevated activity in all three extensor groups were responsible. Only by means of a proper link segment biomechanical analysis can we put our finger on the "guilty" muscle pattern. At the outset, the investigator may be somewhat frustrated by the indeterminacy of what is observed visually. However, an examination of the total motor pattern will soon convince him that he is looking at a human control system that can accomplish the same goal many ways and therefore is extremely flexible and adaptable.

1.10 Motor Functions in Human Gait

1.11 Purpose of Gait

The sole purpose of walking or running is to transport the body safely and efficiently across the ground, on the level, uphill and downhill. In level gait, the measure of the task is "how much mass is transported how far". In units of measurement, the task is a product of mass times distance: kg.m. For uphill or downhill gait, an additional factor, that of change in altitude, must be considered. The best measure of this is change in potential energy = mgh joules. In order to transport the body safely, the neuromuscular control system must also provide appropriate shock absorption, prevent collapse and maintain balance of the upper extremity, and achieve a safe foot trajectory.

1.12 Major Motor Functions During the Gait Cycle

In order to achieve safe and efficient propulsion of the body, there are five main functions that must be performed during each stride period, independent of whether we walk or run.

i) Generation of mechanical energy to maintain the present velocity or to increase the forward velocity of the body (Winter, 1983a; 1983b).

ii) Absorption of mechanical energy for shock absorption or stability or to decrease the forward velocity of the body (Winter, 1983a; 1983b).

iii) Maintenance of support of the upper body (i.e., prevent collapse of the lower limb) during stance (Winter, 1980; 1984).

iv) Maintenance of upright posture and balance of the total body (Nashner, 1980; 1982).

v) Control of foot trajectory to achieve safe ground clearance and a gentle heel or toe landing.

All of the above functions must be performed within the anatomical constraints of the human body. These constraints have been summarized by Saunders, Inman and Eberhart (1953) who referred to them as the six determinants of gait. They are not really determinants in the true sense of the word except as they describe the constraints of the interconnected segments and therefore "determine" the anatomical framework and limits within which the neuromuscular system must operate. The CNS must also integrate desired efferent commands with peripheral feedback and vestibular and visual inputs to generate the correct patterns of moment of force at each joint.

1.20 Definitions, Terms and Conventions Related to Human Gait

1.21 Mean Plane of Progression - is the average vertical plane along which the centre of mass of the body moves during the stride period. If a person is changing direction, the term instantaneous plane of progression is suggested.

1.22 Spatial Reference System - More than one spatial reference system appears to be in use, but all adhere to the Right Hand Rule which is a convention to show the relationship in the directions of three orthogonal vectors, and the sequence of labelling them. For example, in Figure 1.22, the thumb would point forward, the index finger upwards, and the middle finger laterally. Thus, the direction of progression (anterior-posterior) is X, vertical direction is Y, and sideways direction (medial-lateral) is Z. Another common combination is to label vertical as Z, forward as Y and sideways as X. Regardless of the convention used within each laboratory, any plots should not be labelled as X, Y, or Z, but rather as vertical, anterior-posterior and medial-lateral.

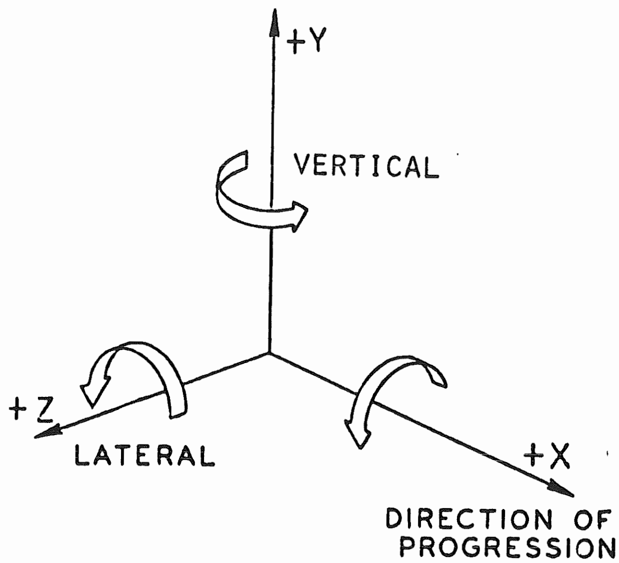

Figure 1.22

1.23a Anatomical Position - is the position of the body standing erect with feet together, arms at the sides and palms facing forward.

1.23b Sagittal Plane - is a vertical plane that divides the body into right and left parts.

1.23c Coronal (Frontal) Plane - is a vertical plane that divides the body into anterior and posterior parts.

1.23d Transverse (Horizontal) Plane - is a horizontal plane cutting through the body at right angles to the sagittal and coronal planes.

1.24a Initial Contact (IC) - is a general term to define the instant when the foot or shoe makes first contact with the ground, independent of how it makes contact with the ground.

1.24b Heel Contact (HC) - is the instant when the heel of the foot or shoe makes initial contact with the ground.

1.24c Ball Contact (BC) - is the instant when the ball of the foot (metatarsal head) or shoe makes initial contact with the ground, and is used in running or in certain pathological walking when initial contact is not made by the heel.

1.25 Foot Flat (FF) - is the first instant during stance when the foot or shoe is flat on the ground and is independent of how IC was made. If HC and BC occur simultaneously, then FF is the same as IC.

1.26 Heel Off (HO) - is the instant during stance when the heel leaves the ground. It usually is closely related to the start of push-off.

1.27 Toe Off (TO) - is the instant when the toe of the foot or shoe leaves the ground. It usually defines the end of stance and the start of swing. In certain rare gait pathologies, the end of weight bearing is not defined by TO when the term end of contact (EC) might be used.

1.28 Mid Swing (MSw) - is the mid point in time between TO and IC.

1.29a Stride Period - is the period of time for two steps, in seconds, and is measured from an event of one foot to the subsequent occurrence of the same foot. It is suggested that stride period begin with IC of one foot to the next IC of the same foot. It is also commonly expressed as 0 to 100% for purposes of comparing subjects with different stride periods, or successive strides of the same subject.

1.29b Step Period - is the period of time for one step, in seconds, and is measured from an event of one foot to the subsequent occurrence of the other foot. It is suggested that step period begin with IC of one foot to the next IC of the other foot.

1.30a Double Support - is the period of time in walking when both feet are in contact with the ground expressed in seconds or as a % of the stride period. Right double support is the time between IC of the left foot and TO of the right; left double support is the time between IC of the right foot and TO of the left.

1.30b Single Support - is the period of time when only one limb is in contact with the ground expressed in seconds or as a % of the stride period. By definition, for walking, it is exactly equal to the swing period of the contralateral limb.

1.30c Flight Period - is the period of time in running when neither foot is in contact with the ground, expressed in seconds or as a % of the stride period.

1.31 Stance Period - is the period of time when the foot is in contact with the ground, expressed in seconds or as a % of the stride period. Stance has been subdivided into several sub-events: weight acceptance, mid-stance and push-off.

1.31a Weight Acceptance (WA) - is the period of time between IC and the time of maximum knee flexion of the support limb during stance, expressed in seconds or as a % of stride. For certain pathological gaits when knee flexion does not occur, weight acceptance is defined as the time between IC of the ipsilateral limb and TO of the contralateral limb.

Rationale: During this time, energy absorption will have taken place by the ankle, knee and hip muscles. In both walking and running, the hip and knee joints flex at IC, the hip extensors absorb energy very briefly but the knee extensor continue to absorb energy until maximum knee flexion (which is 15% of the stride period for slow to fast walking and 20% of the stride period for runners). The ankle muscles absorb energy immediately after IC (regardless of HC or BC). Thus, maximum knee flexion represents the point in time when all major energy absorption ends and weight is effectively accepted by the support limb.

1.31b Push Off (PO) - is the period of time late in stance when the lower limb is pushing away from the gound and ankle plantarflexion occurs, expressed in seconds or as a % of stride.

Rationale: In slow to fast walking the plantarflexors contract concentrically to produce a major generation of energy (between about 40% of stride and TO). In running, generation takes place between about 20% of stride and TO. Thus, push-off is defined as the period in time in late stance when plantarflexion is taking place, and begins shortly after HO and ends with TO.

1.31c Mid Stance (MS) - is the period of time between WA and PO. In slow to fast walking of normals, this represents 15% to 40% of stride period. In running, WA occupies 0 to 20% and PO takes the balance of stance so there is, in effect, no mid stance period.

1.32 Swing Period - is the period of time when the foot is not in contact with the ground, usually expressed in seconds or as a % of stride period. Swing has also been broken into sub-events: lift-off and reach.

1.32a Lift-Off (Early Swing) - is the period of time during early swing, between TO and MSw expressed in seconds or as a % of stride.

1.32b Reach (Late Swing) - is the period of time during swing between MSw and IC when the knee is extending, expressed in seconds or as a % of stride.

1.33 Cadence - is the number of steps per unit time, expressed as steps/min. Cadence (steps/min) = 120/Stride Period (seconds). *Natural* or *Free* cadence is the cadence that the subject or patient achieves when given instructions to walk as naturally or freely as possible. Fast or slow cadences are forced cadences above or below natural cadence and must be specified by the researcher, e.g., fast = natural + 20 steps/min. If a metronome is used in walking or running, the controlled cadence must be reported.

1.34 Stride Length - is the horizontal distance covered along the plane of progression during one stride; it is the distance covered from IC to IC of the same foot expressed in meters. The stride length is equal to the sum of the two step lengths and will be equal for left and right limbs if the person is walking in a straight line, even in the presence of marked gait assymetry.

1.35 Step Length - is the horizontal distance covered along the plane of progression during one step; it is the distance measured from a point on one foot to the same point on the other foot, expressed in meters. Left step length, for example, would be measured from the contact position of the right heel to the contact position of the subsequent left heel when they are in contact with the ground. Specific step lengths for right and left side must be measured within the same stride. The term can also be used to specify an average step length over many strides.

1.36 Stance/Swing Ratio - is the ratio of stance period to swing period.

1.37 Gait Velocity - is the average horizontal speed of the body along the plane of progression measured over one or more stride periods. It is reported in m/s or m/min and the period of time or distance over which the average velocity was calculated should be reported.

1.38 Segment and Joint Angles (see Figure 1.38)

A. In the sagittal plane, it is assumed the the right side of the subject is being viewed as he or she progresses from left to right. Angles reported for the left limb should have the same convention as the right limb such that the segment and joint angles can be plotted on the same graph and left and right limbs compared directly. Angles can be reported in degrees or radians, but preference is in degrees.

Rationale: The convention followed below for the definition of segment angles and the joint angles fol-

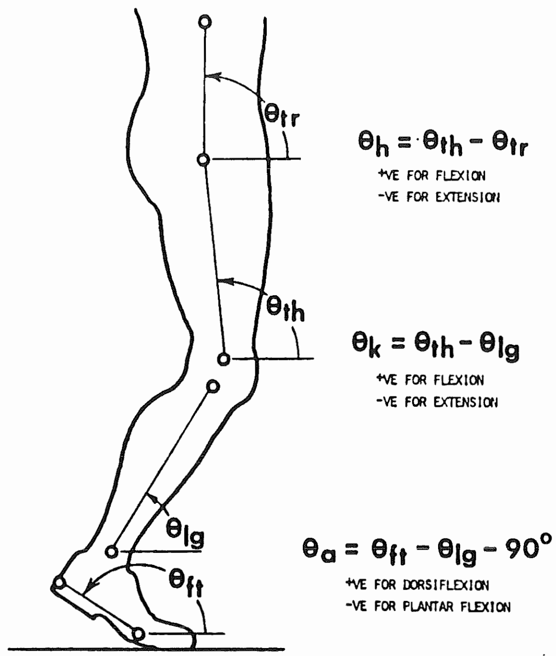

Figure 1.38

lows a consistent convention that permits subsequent biomechanical analyses (kinematic and kinetic). For example, all segments must be defined as positive in a *counter-clockwise* direction from the horizontal in order that the first and second time derivatives have the correct polarity for subsequent kinetic analyses (energy, moment of force and power). The only choice is whether the origin of segment rotation is defined at the proximal or distal ends. The distal end was chosen because it meant that the leg, thigh and trunk segments would oscilate either side of 90° rather than 270°. Similarly, consistency is necessary with the joint angle definition in order that joint mechanical power (which requires joint angular velocity) can be calculated correctly. Thus, all joint angles are designated positive if they are in flexion (dorsiflexion) and negative if they are in extension (plantarflexion).

(a) *Foot Angle* - the angle between horizontal and a line along the bottom of the foot measured from the distal end (5th metatarsal phalangeal joint), measured positive in a counter-clockwise direction.

(b) *Leg Angle* - the angle between horizontal and the long axis of the leg measured from the distal end of the leg (ankle joint) with positive in a counter-clockwise direction.

(c) *Thigh Angle* - the angle between horizontal and the long axis of the thigh measured from the distal end of the thigh (knee joint) with positive in a counter-clockwise direction.

(d) *Pelvis Angle* - the angle between horizontal and the angle of the pelvis (line joining the posterior superior iliac spine and the anterior superior iliac spine), with +ve in a counter-clockwise direction.

(e) *Trunk Angle* - the angle between the horizontal and the mean axis of the spine measured from the distal end of the spine, with positive in a counter-clockwise direction. The axis of the spine is defined as the line joining C7 to L5.

(f) *Ankle Angle* - the angle between the foot and the leg minus 90°; +ve for dorsiflexion, −ve for plantarflexion.

(g) *Knee Angle* - the angle between the thigh and leg; +ve for flexion, −ve for extension.

(h) *Hip Angle* - the angle between the thigh and trunk or pelvis; +ve for flexion, −ve for extension.

(i) *Segment and Joint Angular Velocities* - are the first time-derivatives of the segment or joint angles. Thus, a counter-clockwise rotation of any segment is a positive velocity, and a joint velocity will be positive if it is flexing, and negative if it is extending. Units are degrees/s or rad/s.

(j) *Segment and Joint Angular Accelerations* - are the second time-derivatives of the segment or joint angles. Thus, a counter-clockwise acceleration of any segment is a positive angular acceleration, and a joint angular acceleration will be positive if it is flexor, and negative if it is extensor. Units are degrees/s^2 or rad/s^2.

B. In the frontal plane (when the subject is viewed from the front), the following conventions and terms apply to joint angles. In the anatomical position hip abduction/adduction is zero, knee valgus/varus is zero and ankle eversion/inversion is zero. Adduction is positive, abduction is negative; varus is positive, valgus is negative; inversion is positive, eversion is negative. Foot pronation and supination are defined as inversion and eversion of the calcaneous with respect to the midline of the leg. Foot pronation is positive and supination is negative.

Pelvic list is the angle of the line joining identical landmarks on the left and right of the pelvis with respect to the horizontal, with positive in the counter-clockwise direction. Suggested landmarks are anterior superior spine or iliac crest.

Trunk list is the mean angle of the spine as defined by the line joining C7 to L5 measured with respect to the horizontal, with positive in a counter-clockwise direction.

C. In the transverse plane (when the subject is viewed from above), an external rotation of a distal segment relative to a proximal segment is positive, and internal rotation is negative.

1.39 Strike Index - is a term used in running to describe the ratio of distance from the heel to the point on the foot or shoe where the center of pressure first occurs to the total foot length. In Figure 1.39 it is the ratio of PS to PA.

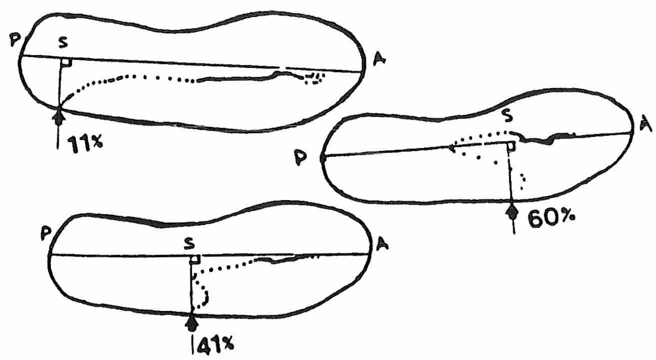

Figure 1.39

i) Rearfoot Strike - a pattern of contact between the foot and ground with a strike index between 0 and .33.

ii) Midfoot Strike - a pattern of contact between the foot and ground with a strike index between .33 and .67.

iii) Forefoot Strike - a pattern of contact between the foot and ground with a strike index between .67 and 1.0.

1.40 Biomechanical Terms - Definitions and Conventions

1.41 Kinematics - is the term used to describe the spatial movement of the body, not considering the forces that cause the movement. Included are linear and angular displacements, velocities and accelerations. The spatial reference system can be absolute or relative, and, in the case of the latter, the anatomical reference is reported. All joint angles, angular velocities and accelerations are, by definition, relative. The external (absolute) spatial reference system has been described in Section 1.22.

1.42 Kinetics - is the term used to describe the measurement and analyses of the forces, powers and energies of the movement. Kinetic variables include ground reaction forces, joint reaction forces, moments of forces, tendon forces, joint contact forces, power, work and energies.

1.43 Reaction Force - is the resultant force acting on or at any point in the skeletal system. The internal reaction forces at any point are in static or dynamic equilibrium with the externally applied forces and the inertial forces distal to that point. The unit is Newton (N). Within the skeletal system, reaction forces are usually calculated only at joint centres. However, in some research it may be desirable to calculate a reaction force at a point along a long bone or within a prosthesis, and, in such cases, the location must be clearly stated.

1.44 Bone-on-Bone Force or Joint Contact Force - is the vector summation of all forces acting at a joint. It is the summation of the reaction forces at that joint plus any compressive or shear forces due to the muscles, ligaments or structural constraints acting at that joint. The unit is Newton (N).

1.45 Moment of Force - is the name given to the product of a force acting at a distance about an axis of rotation, and which causes an angular acceleration about that axis. The unit is Newton-meters (N.m). Joint moments of force are the net result of all internal forces acting at that joint and include the moments due to muscles, ligaments, joint friction and structural constraints. Moments of force can be calculated by inverse dynamics* (Bresler & Frankel, 1950), and, are in static or dynamic equilibrium with the external moments due to externally applied and intertial forces distal to that joint. A knee flexor moment, for example, means that the knee flexors are *dominant* at the knee joint which means that the knee flexors (hamstrings and gastrocnemius) are creating a greater moment of force than are the knee extensors (quadriceps).

The term torque as a synonym for moment of force is discouraged.

1.46a Linear Momentum - is the product of a mass of a body and the velocity of its centre of mass quantified in kg.m/s or N.s.

1.46b Angular Momentum - is the product of the moment of inertia of a body about a given axis and its angular velocity about that axis quantified in $kg.m^2/s$ or N.m.s.

1.47a Linear Impulse - is the time integral of a force curve and is quantified in N.s. The time of start and end of the integretion must be reported.

1.47b Angular Impulse - is the time integral of a moment of force curve and is quantified in N.m.s. The time of start and end of the integration must be reported.

1.48 Mechanical Energy - is the energy state of any limb segment or total body system at an instant in time and represents the potential of that system to do work. It is measured in Joules (J). It comprises potential energy, translational kinetic energy and rotational kinetic energy. The zero reference datum for potential energy is ground level or the lowest point in the spatial reference system. Potential energy stored in spring mechanisms should be clearly identified in contrast to the more common gravitational potential energy.

1.49 Mechanical Power - is the work performed per unit time, and is used to quantify the rate of generating or absorbing energy by muscles or the rate of change of energy of a segment or body system. It is measured in watts (W). Muscle mechanical power is the product of the muscle force and its velocity (linear) of shortening or lengthening. The joint mechanical power (generated or absorbed) by all muscles and ligaments crossing a joint is the product of the moment of force and the angular velocity at the joint. At any given time, some muscles crossing a joint can be generating energy while others are absorbing energy; the net rate of energy generation or absorption is evident in the mechanical power.

1.50a Concentric - is the term used to describe an action when a muscle is shortening under tension.

1.50b Eccentric - is the term used to describe an action when a muscle is lengthening under tension.

1.51a Positive Work - is the work done by concentrically acting muscle and equals the time integral of the mechanical power during the time that the muscle is shortening. The net positive work done by all the muscles crossing a joint is the time integral of the joint mechanical power when it is +ve. The work is measured in joules (J).

1.51b Negative Work - is the work done on an eccentrically acting muscle and equals the time integral of the mechanical power during the time that the muscle is lengthening. The net negative work done by all the muscles crossing a joint is the time integral of the joint mechanical power when it is –ve. The work is measured in joules (J).

1.52 Types of Muscle Contraction - A wide variety of terms are used to describe the time-course of the force and length of a muscle while it is generating tension.

1.52a Isometric - is an action in which a muscle develops tension without a change in length or a moment of force without a change in joint angle. In the case of biarticulate muscles, it is difficult to define that both joint angles cannot change because a pure isometric action could take place as both joint angles changed.

1.52b Isotonic - is an action which produces constant force or constant moment of force. The force or moment should be specified. Note that lifting or lowering a mass is not isotonic unless the mass is moving at a constant velocity.

1.52c Isokinetic - is a muscle action during which the muscle lengthens or shortens at a constant velocity. During such situations, a constant joint angular velocity is not implied but may be approximated.

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* One mathematically erroneous technique is sometimes used to calculate moments, but should be used with caution as a first approximation. This method involves the calculation of the joint moment by calculating the product of the ground reaction force vector and the perpendicular distance from the joint centre to that vector. The shortcomings of this approach have been documented by Wells (1981) and are threefold. Numerically, errors are introduced because the mass-acceleration products of the foot, leg and thigh are not accounted for and, numerically, these errors compound as we calculate up from the ankle to the hip joint and higher. At the hip, the errors can be quite high. Secondly, the polarity of the calculated moment is that of a reaction moment. Thus, a knee flexor moment is calculated when this vector passes behind the knee, but the polarity must be reversed and be interpreted that the knee extensor muscles were dominant. Thirdly, the moments of force during swing cannot be calculated. Thus, when this approximation is used, it should be confined to slow non-jerky gaits and should not go beyond the ankle and knee joints.

1.52d Isometric-isotonic - is an action during which both the muscle length and the force of action remain constant over time.

1.52e Isometric-anisotonic - is an action during which muscle length remains constant but the force of action varies with time.

1.52f Anisometric-isotonic - is an action during which muscle length varies with time while the force of action is held constant.

1.52g Anisometric-anisotonic - is the most general condition, corresponding to the frequently occurring state in which both the length and force of action vary with time.

1.53 Agonist - is the term used to describe those making contributions which are dominant at a joint and therefore have the same sign as the moment of force at a joint, regardless of whether they are concentrically or eccentrically acting. For example, in the downward phase of a deep-knee bend, the quadriceps are acting eccentrically and are the agonists that control the amount of knee flexion as knee flexion increases.

1.54 Antagonist - is the term used to describe those muscles opposing agonist muscles. In the example given above, the knee flexors (hamstrings + gastrocnemius) would be antagonist muscles. Another way of defining an antagonist is that they are the co-contracting muscles that produce the lesser moment of force at a joint.

1.55 Maximum Voluntary Contraction (MVC) - is a voluntary contraction in which the subject attempts to achieve a maximum force in a muscle or muscle group.

1.60 Convention for Biomechanical Terms

1.61 Segment Angles, Angular Velocities, Angular Accelerations - are all positive when counter-clockwise in the plane of progression (when viewed from the subject's right side), in the frontal plane (when viewed from the subject's front), and in the transverse plane (when viewed from above).

1.62 Moments of Force - counter-clockwise moments as calculated at the proximal end of a segment are positive. Moments are described as flexor or extensor, abductor or adductor, etc.

1.63 Reaction Forces - in the plane of progression are positive to the right and upwards (when viewed from the subject's right), in the frontal plane they are positive upwards and to the left (when viewed from the subject's front); in the transverse plane they are positive to the right and forward (when viewed from above the subject looking forward).

1.64 Joint Angles - are described as being in extension or extended, in flexion or flexed, in plantar flexion or plantar flexed, etc. For polarities specific to gait, see Section 1.38.

1.65 Joint Angular Velocities - are described by using terms extending, flexing, dorsiflexing, etc., independent of the joint's present position. *Example*: the knee joint was flexed 20°, it was extending at 80°/sec. while a net flexor moment of 30 N.m acted to decelerate the leg and foot.

1.70 Neurological Terms - Definitions and Explanations

1.71 Motor unit (m.u.) - is the term used to describe the smallest controllable (recruitable) muscular unit. The motor unit consists of a single alpha-motoneuron, and the muscle fibres it innervates via its neuromuscular junctions.

1.72 Motor unit action potential (m.u.a.p.) - is the name given to the detected waveform resulting from a spatio-temporal summation of individual fibre action potentials from a given motor unit. Its shape and amplitude are a function of electrode type, contact area, inter-electrode spacing, material, the location with respect to the active motor unit, the electrochemical properties of the intervening muscle, fascia and fat, and the electrical characteristics of the recording equipment. Each motor unit will produce a motor unit action potential of characteristic shape and amplitude as long as the geometric relationship between the m.u. and electrode remains constant.

1.73 Motor unit firing rate - is the average firing rate of a motor unit over a given period of time.

1.74 Myoelectric signal - is the name given to the total signal seen at an electrode or differentially between electrodes. It is the algebraic summation of all m.u.a.p.'s within the pick-up area of the electrode(s). When amplified, this signal is called an electromyogram (EMG). When the amplitude is reported, it should be that value seen at the electrodes (mV or µV.).

1.75 Motor Point - is the location over a muscle where a contraction is elicited by a minimal intensity, short duration electrical stimulus.

1.76 Detector - is the name given to the electronic circuit or computer algorithm which converts the EMG into a signal of positive polarity. A linear detector is nothing more than a full-wave rectifier which reverses the sign of all negative voltages and yields the absolute value of the EMG. Non-linear detectors, such as a square law detector, result from a non-linear mathematical transformation subsequent to full wave rectification.

Linear envelope detector is the name given to a full-wave rectifier followed by a low-pass filter. The filter cut-off frequency should be specified along with its type and order. The amplitude of the detected and/or filtered signal should be reported as that seen by the electrodes followed by a detector with a gain of 1 and should be reported in mV or µV.

1.77 Integrated EMG - is the name given to the area under the full-wave rectified EMG. It has units of µV.s or mV.s and is used to calculate the mean EMG over any desired integration time. The integration period should be specified in seconds or ms.

1.80 Ensemble Average - is the name given to the average stride pattern of any variable across many repeat trials. Reporting of an average pattern on the same time base is required when the individual stride periods vary slightly. The process of determining that average pattern is called ensemble averaging and the steps required for any variable are as follows:

i) Determine the time IC to the next IC.

ii) Set each stride period = 100%, and divide the period into equal intervals (i.e., 2%, 5%).

iii) Average the measures of the variable at each interval and determine its mean and standard deviation.

iv) Plot this mean and standard deviation over the stride period.

v) Calculate the mean variability over the stride period and express it as a percentage of the mean value of the signal. In effect, calculate a variability-to-signal ratio. One suggested score that has been used for a few years is a coefficient of variation (CV) which is calculated as follows:

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

where: N is the number in intervals over the stride

X_i is the mean value of the variable at the i th interval

σ_i is the standard deviation of variable X about X_i.

1.90 References

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2.0 Temporal and Stride Measures

2.01 Introduction

Probably the most commonly measured output variables are simple temporal and length measures. Cadence data require nothing more than a stop watch and stride length or velocity requires a linear measure or calibrated walkway.

$$\text{Velocity} = \frac{\text{stride length x cadence}}{120} \quad \text{m/s} \quad \text{Eqtn.1}$$

where: stride length is distance travelled between successive stance periods (in meters).

cadence is in steps/min.

If a person is walking a straight line, the left and right stride lengths are equal; if they are unequal, they will walk in a circle. It should be noted that cadence and stride length may look as if they are independent measures. When we increase our cadence with a more forceful push-off, the swing period is reduced and cadence also increases. Also, stride length is very much a function of the subject's height, and possibly weight, age and sex.

2.10 Natural Cadence

Natural or free cadence is defined as the steps/min when a subject walks as naturally as they can. When we walk slowly or with a fast cadence, we must override our natural frequency and consciously force our cadence to a faster or slower rate. Natural cadences reported in the literature had averages varying from 101 to 122; our Laboratory has recorded 60 normal subjects (mainly university students, aged 19-32) to have a natural cadence of 107 (s.d. = 8.8). Drillis, (1958) for 936 pedestrians reported a mean cadence of 112, varying from 78 to 144. Du Chatinier, et al., (1970) revealed that females walked slightly more rapidly than males (116 vs 122) for a population of 72 males and 57 females. Finley and Cody (1970) in a massive study also reported a higher natural cadence for females (116.5, s.d. = 11.7, N=472) than for males (110.5, s.d. = 10.0, N=434). Molen and Rozendal (1972) reported for about 500 young adults that the male cadence averaged 113 compared to 124 for females. Thus, we can conclude that female natural cadence is 6 to 9 steps/min higher than that of males. This same group was classified as tall, medium and short and there was not a height-related trend. However, as will be seen in a later section, height is a dominant factor in stride length and may be a dependent variable that accounts for some gender-related velocity differences. Murray, et al., (1969) in a study on older men included subjects up to 87 years old, and, for those older than 65, both cadence and stride length decreased.

2.20 Stance and Swing Times

Stance time can be expressed in seconds or as a percent of stride period. For natural cadence, all researchers report a fairly consistent stance time: 58 to 61%. Swing time varied from 42% to 39%. Double support = stance - swing, and was calculated to vary from 16% to 22%, and if perfect symmetry is assumed, this would consist of two double support times of 8% to 11% per stride. As cadence and velocity of walking increase, both stance and swing times decrease (Andriacchi, Ogle and Galante, 1977; Grieve and Gear, 1966; Murray, 1967). Murray (1967) reported for adult males that stance decreased 3.5 times as rapidly as swing. Grieve and Gear (1966) reported a regression curve relating swing time to normalized velocity (statures/sec) for newly walking children, under 5-year olds, pre-adolescents and adults. It became quite evident that, as our gait pattern matured, the swing time decreased drastically as velocity increased. However, when swing was normalized and expressed as a percent of stance, there was a positive correlation with relative stride length (in statures). Swing increased from about 33% to 75% as the subjects increased their stride length from 0.5 statures (slow walk) to 2.0 statures (slow run).

2.30 Velocity, Cadence and Stride Length

It is evident from Equation 1 that velocity is not independent of stride length and cadence. However, stride length and cadence are dependent on each other. Lamoreux (1971) summarized the result of his work along with those of four others and noted that between cadences of 80 and 120, stride length and cadence each varied as the square root of the velocity and, therefore, had a linear relationship between them. Thus, up to a cadence of 120, speed increases are achieved equally by increasing stride length and cadence. Above 120, step length levels off and cadence only increases. Figure 2.30(a) is reproduced to show the step length/height changes as a function of cadences. In a summary of 54 trials carried out in our laboratory, we found, in the range of cadences from 80 to 125, a high linear correlation (r = .94) between velocity and cadence.

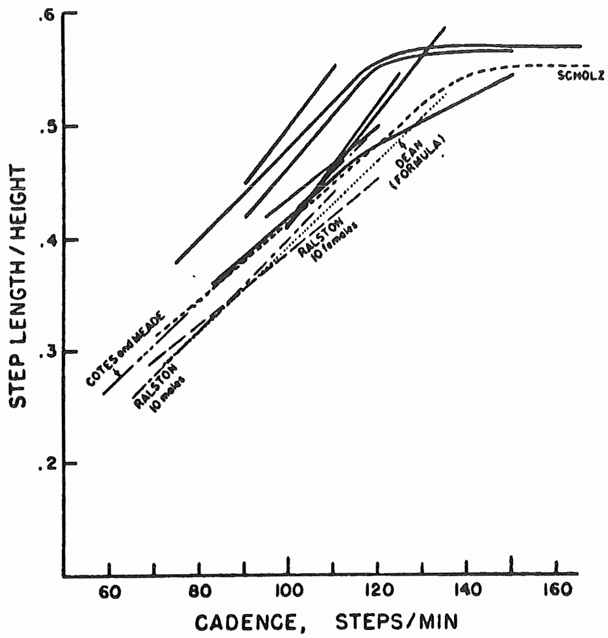

Figure 2.30(a)

Finley, et al., (1969), in a study of 23 elderly women ranging in age from 64 years to 86 years, noted significant step length and velocity differences from a group of younger women. Velocity of the older group was 1.57 mph versus 1.83 mph for the young women and this was attributed almost entirely to a reduced step length: 15 in. versus 18.5 in.

Probably the best normalized plot that summarizes all three variables may be seen in a nomogram presented by Grieve (1968) in which he plots stride length (statures) against walking speed (statures/sec) and overlays the plot with lines representing constant cycle time. A redraw of this curve is given in Figure 2.30(b) where the cycle time is replaced by its more common measure, cadence. The advantage of this graph is that body height, a major anthropometric variable, has been removed. Since stride length is well correlated with stature, it is therefore not surprising that walking speed increases with stature (Grieve & Gear, 1966; Grieve, 1968).

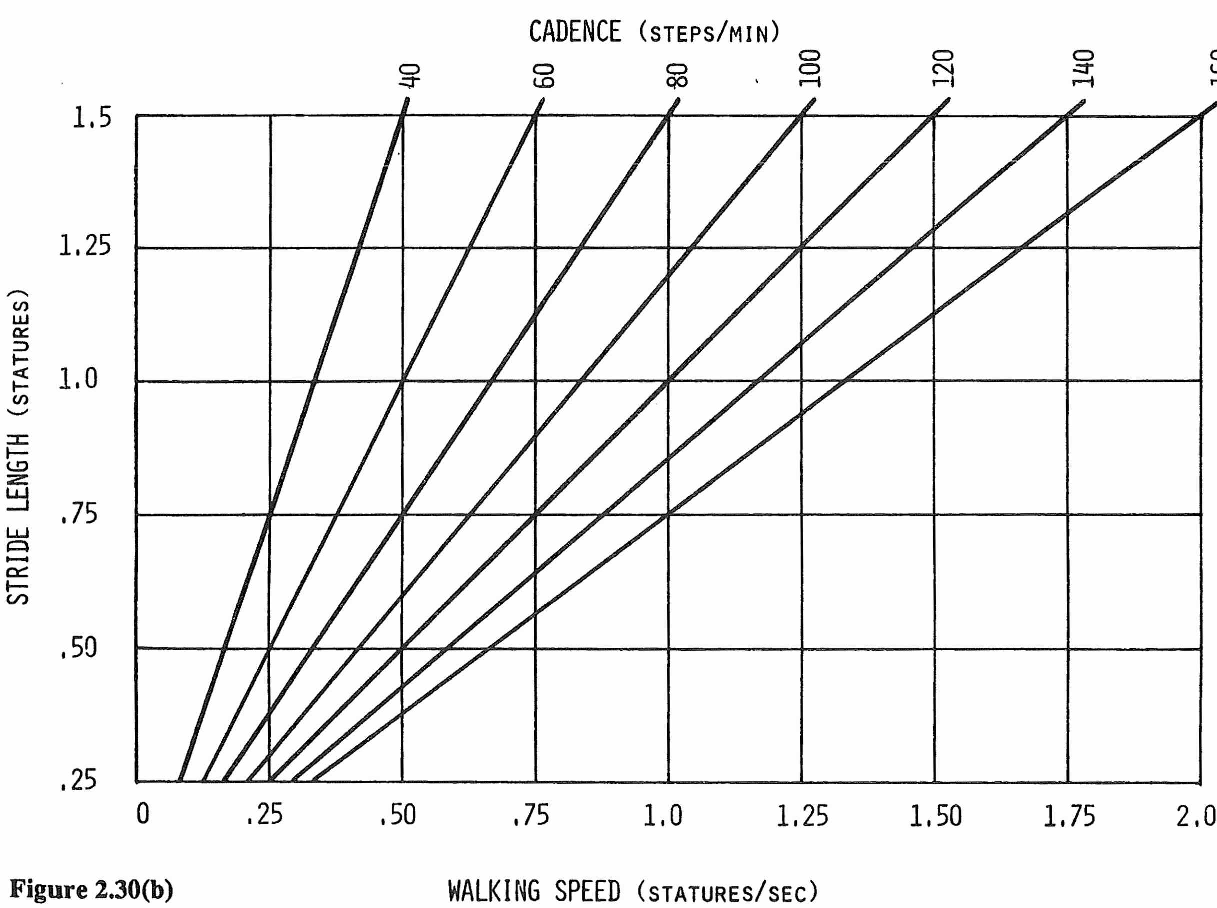

Figure 2.30(b)

2.40 References

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3.0 Kinematics

Kinematics is the term used to describe the movement itself independent of the forces, both internal and external that caused the movement. Because gait is a repetitive event, one cycle (stride) is used as the time period to describe the movement. Gait kinematics have been reported more than any other group of variables and they have been quantified using a wide variety of measurement techniques. Specially designed goniometers have been developed to give a direct record of the joint angular histories. Cinematography and special film digitizing systems have evolved to allow body coordinates to be extracted for computer analysis. Similarly, television has been used to advantage, not only as an instant play-back, but also as input to special interface electronics for conversion to computers. Also, very specialized opto-electronics systems which require active optical sources as markers are now commercially available.

Regardless of how the kinematic data are collected, it is important to realize that all such data are outcome measures and are the net result of many muscle forces. The number of kinematic variables (positions, velocities and accelerations) required to describe one gait stride is very high. For example, to describe completely the movement of the thigh in the plane of progression requires nine independent variables. Six linear measures are required to quantify the movement of the centre of mass of the thigh: the positions, velocities and accelerations in the horizontal and vertical directions. Three additional measures (angle, angular velocity and angular acceleration) are necessary to complete the description. Thus, if we consider the human body to be a 15-segment system (2 feet, legs, thighs, upper arms, forearms, hands, plus head, trunk and pelvis), we need 135 curves to describe the full body kinematics over the stride. Obviously, some serious compromise is necessary to make any gait description manageable.

3.01 Absolute vs Relative Measurement Systems

The 135 variables described above are absolute variables in the plane of progression. From these absolute variables, we can calculate all the relative variables such as joint angles and angular velocities. However, if we are given the relative variables, we cannot reverse the process and calculate the absolute variables. Thus, certain direct measurement systems such as goniometers which yield joint angles directly cannot be used to describe the movement of the total body. Similarly, accelerometers are only useful for relative acceleration measures which depend on its anatomical location and orientation and the latter is changing continuously over the stride. It is only through imaging systems (with cameras that define an absolute spatial reference system) that we can achieve a full kinematic description during gait. Cinematography and television are the two most common and important measurement systems used in gait. They have several major advantages over some recent commercial opto-electric systems: cost, ease of use, unlimited number of anatomical markers and ease of application of markers, ease of replay at a later date (for re-analysis or teaching), less restrictive on lighting. However, they have some disadvantages that may be important: cine film has delays in development and time required to digitize co-ordinates, expense in conversion of TV data, and limited sampling rate of TV (for certain athletic events only). Some researchers have commented that the human error in digitizing film data is high, but documentation for this criticism has never been presented. For twelve years, we have been using a relatively inexpensive film digitization system and we have regularly quantified the noise due to human error in digitizing to be about 1 mm r.m.s. when the cine camera was 4 meters from the subject. To our knowledge this error is somewhat less than that ever reported by researchers using commercial opto-electric systems.

3.02 Processing of Imaging Data

One of the initial drawbacks of coordinate data obtained from all imaging systems was the difficulty related to the calculation of velocities and accelerations from the displacement/time curves. This problem arose because of the noise content in the raw coordinate data which became drastically amplified as a result of the differentiation process required to calculate velocities and accelerations (Winter, et al., 1974a). As this noise was essentially random and spread equally across all harmonics, it became necessary to smooth the data prior to carrying out the first and second order differentiation. Four basic techniques have evolved.

i) The raw coordinate data can be curve fitted to some higher order polynomial; however this has not proven satisfactory because the trajectories in gait are not fitted well by specific mathematical expressions.

ii) Harmonic analyses are performed and the lower harmonics are used to reconstitute the trajectory curve. This approach assumes stationarity (that the amplitude and phase of each harmonic remains constant over the stride) and this has not been shown to be true. Thus, the reconstituted signal differs significantly from the original curve.

iii) Spline curve fitting does a piece-wise fit of all the raw coordinates assuming certain order of polynomial. A second order polynomial can achieve moderately good fits but the inflection points must be determined from the raw data and they may be in error. More recent cubic and quintic splines have been introduced with better results, but are quite expensive to run on the computer.

iv) Digital filtering can be used to reject the higher frequency noise. Low-pass zero-phase digital filters are easy to implement (Winter, et al., 1974b), the only decision required by the researcher is what cut-off frequency to use. We have found satisfactory results if the cut-off frequency is set to the 6th harmonic of the stride frequency. If the cadence were 120 (stride frequency = 1 Hz), then cut-off would be set to 6 Hz. Or, if the cadence were 90 (stride frequency = .75 Hz), it would be suitable to set the cut-off at 4.5 Hz.

The calculation of velocities and accelerations from the smoothed data are quite simply achieved through a finite difference; the only precaution that is necessary is that this slope must be calculated over two sample intervals in order that the velocity can be defined at each sample time. For example, the velocity at the ith interval of time, $V_i = (X_{i+1} - X_{i-1})/2T$, where X_{i+1} and X_{i-1} are the coordinates at the (i+1)th and (i-1)th intervals and T is the sampling interval. Similarly, accelerations can be calculated as $A_i = (V_{i+1} - V_{i-1})/2T$.

3.10 Linear Measures

Displacement data over the stride can be presented in two ways. Trajectory plots portray the movement in space: on a plot of vertical vs horizontal. Time is indicated by showing the position of each segment by a stick plot or each anatomical landmark by a dot. Figure 3.10(a) is a stick diagram plot of one stride of gait. A second way of presenting kinematic descriptors is on a time plot. Figure 3.10(b) is such a graph and depicts the vertical displacement, velocity, and acceleration of the heel over the stance period. Because of the mixture of units on the ordinate, each of these variables was plotted as a percent of its peak during stance. It should be noted that the displacement curve is nothing more than the time integral of the velocity which in turn was the integral of the acceleration. Thus, there is a phase delay for each of these integrals, such that the start of HO can vary considerably depending on how it was defined. If it were defined as that point in time when the variable exceeded 5% of its maximum value, the acceleration curve would define HO as occuring at

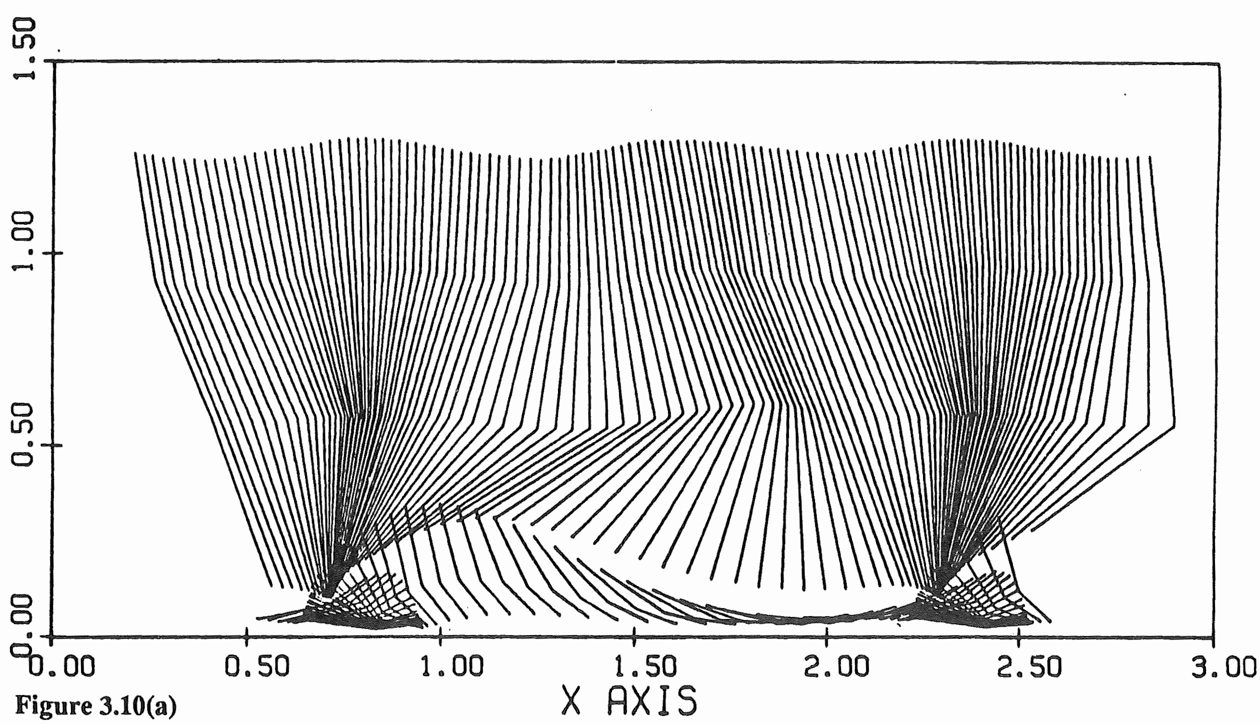

Figure 3.10(a)

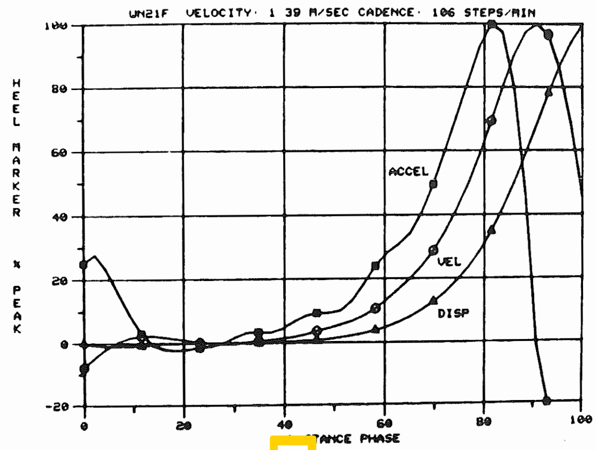

Figure 3.10(b)

40%. The velocity curve would label HO as being at 50% stance, and the displacement curve labels it at 60% stance.

3.11 Intra-Subject Foot Trajectories

Figure 3.11(a) presents the vertical displacement plus the vertical and horizontal velocity of the heel on the stride period for 14 subjects walking their natural cadences. No attempt has been made to normalize this displacement and velocity information because of the fairly low variability and because the height and limb length normalizations did not reduce the variability.

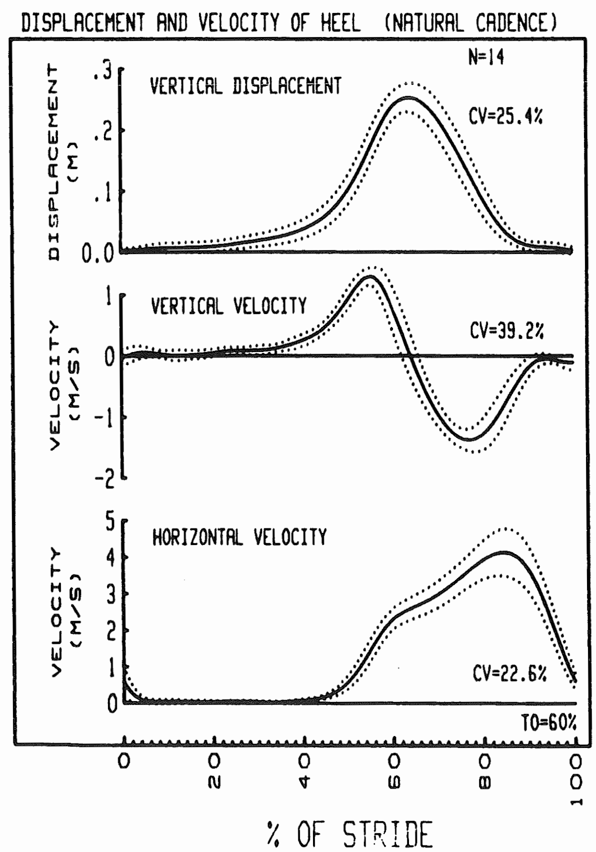

Figure 3.11(a)

Vertical displacement starts well before toe off (60%) and reaches a maximum positive (upward) velocity just prior to TO. Maximum vertical displacement is reached shortly after TO. Horizontal velocity builds up gradually after heel-off and reaches a maximum late in swing and drops rapidly to near zero just prior to HC. The vertical trajectory and velocity changes just prior to HC are particularly interesting. The vertical trajectory during mid and late swing drop rapidly but just 10% before HC the vertical drop is arrested about 1 cm above ground level. During the last 10% of swing, the heel is lowered very gently to the ground as the horizontal velocity decreases very rapidly to near zero. Thus, at HC both vertical and horizontal velocity are near zero. The reader should be aware that we cannot place markers right at the point of heel contact. Rather, we must place the marker a few cms above ground level on the rear of the heel but slightly on the lateral side. Thus, this marker appears to have a non-zero velocity just after HC and this is due to the trajectory of that marker as the foot is lowered to the ground. Thus, the trajectory just prior to HC can be described as that of an airplane just about to touch down, but with the difference that the horizontal velocity is reduced to near-zero before contact. Thus, to refer to this event as heel strike is totally erroneous.

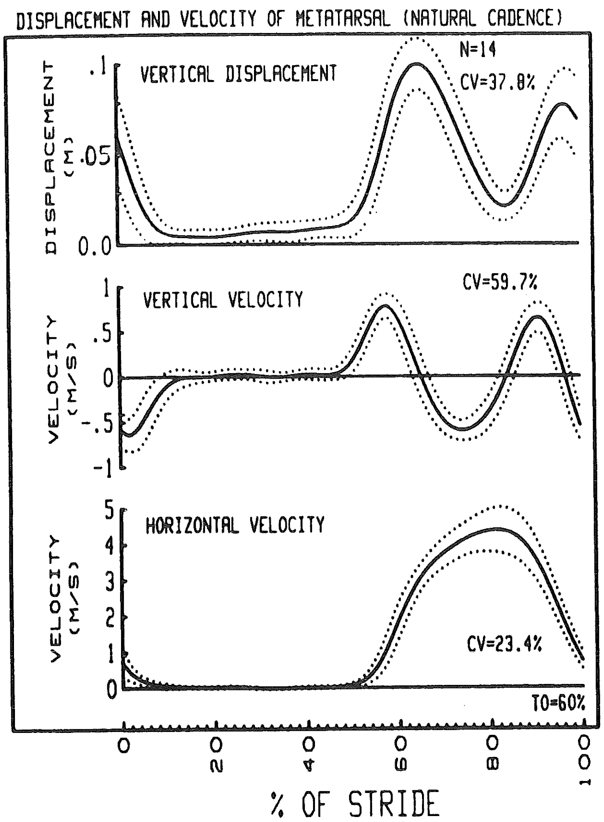

Figure 3.11(b)

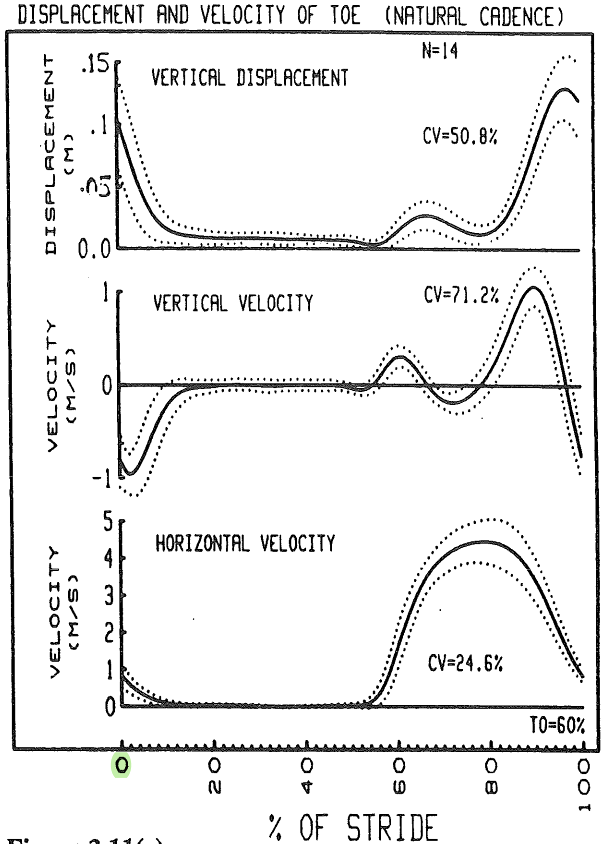

Figure 3.11(c)

Figure 3.11(b) presents similar plots for the metatareal-phalangeal marker for these same 14 natural cadence subjects. The mean clearance is 1.61 cm at about 84% of stride. There are some differences in the vertical trajectory compared with that for the heel. There is an initial rise during late push-off and early swing. Then as the leg and foot are swung forward, the fore part of the foot swings low just clear of the ground and then rises again to a second peak just prior to HC. Figure 3.11(c) shows the same displacement and velocities for the toe for the same 14 subjects. Because the toe is last to leave the ground, and because of the angle of the leg and foot during early swing, the toe rises to no more than 2.5 cms above the ground and then drops to only .87 cm clearance at mid swing. Then, as the knee extends and the foot dorsiflexes, the toe rises to a maximum of 13 cms just prior to HC.

Similar trajectory calculations we made for slow and fast walking normal subjects. The shape of the curves was essentially the same and the maximum vertical displacement was minimally different. Table 3.11 summarizes the maximum displacements and velocities for these three cadences and three markers.

Table 3.11

Marker	Cad	Vert. Disp. (m)	Vert. Vel. (m/s)	Hor. Vel. (m/s)
Heel	F N S	.27 .25 .23	1.5 1.3 1.15	5.0 4.2 3.3
M-P	F N S	.11 .10 .10	1.0 .75 .65	5.2 4.3 3.4
Toe	F N S	.13 .13 .12	1.2 1.1 .75	5.3 4.5 3.6

3.12 Trajectories of Head, Arms and Trunk (HAT)

The upper part of the body above the lower limbs represents about 2/3 of the mass of the body. It consists of the head, arms, trunk and pelvis, and is often in shortened form referred to as HAT. As it is not one segment but a combination of several segments, it is impossible to label it with a single anatomical marker. Thus, its centre of mass must be calculated as a weighted summation of the individual segments. Such a summation was done for all the subjects. The vertical displacement appears in Figures 3.12(a), (b) and (c) for the slow, natural, and fast walking cadence groups. Several biomechanical aspects of the HAT trajectory are worth noting. First, there are two cycles of up and down movement per stride, one due to the trajectory of each lower limb. During each double support phase, the trunk is at its lowest position and, during mid stance of each limb, it reaches a peak.

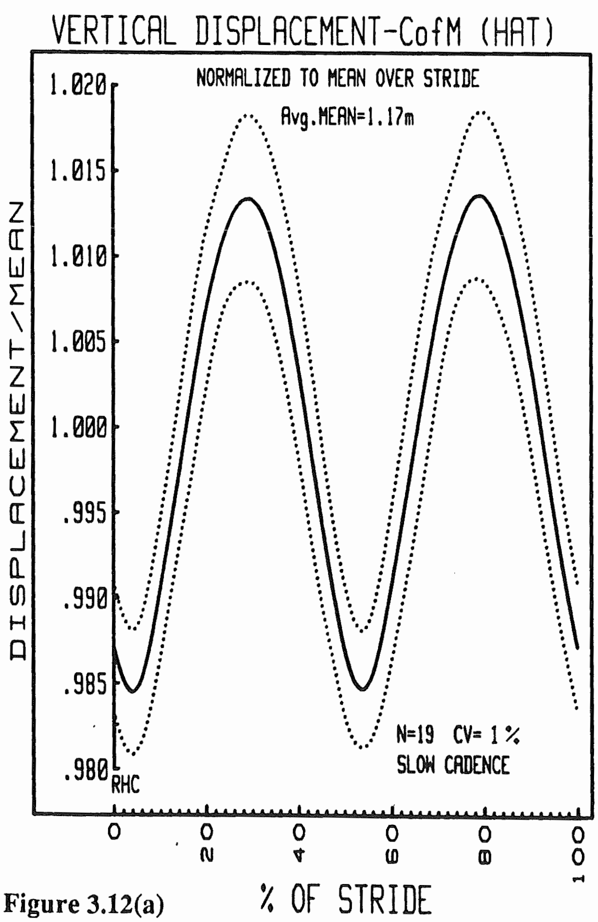

Figure 3.12(a)

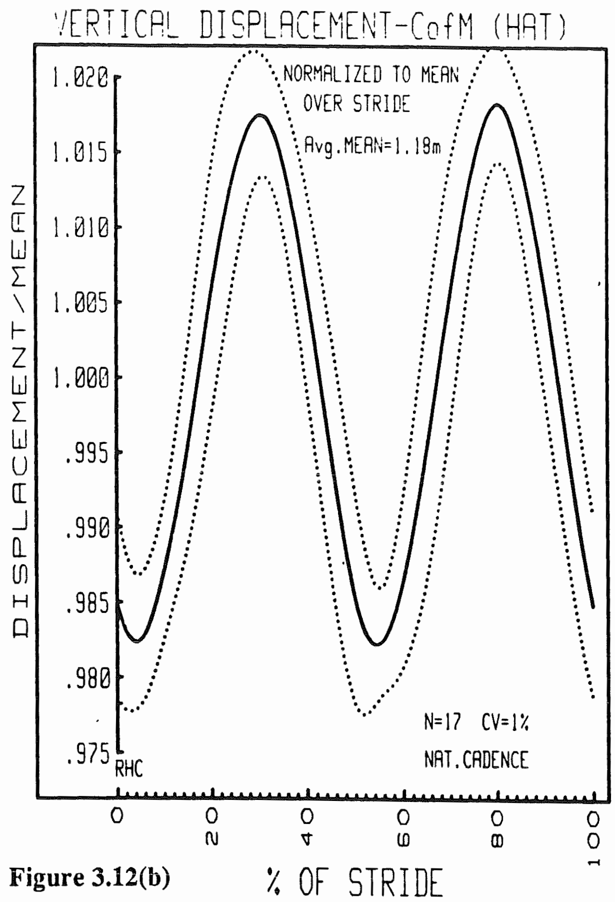

Figure 3.12(b)

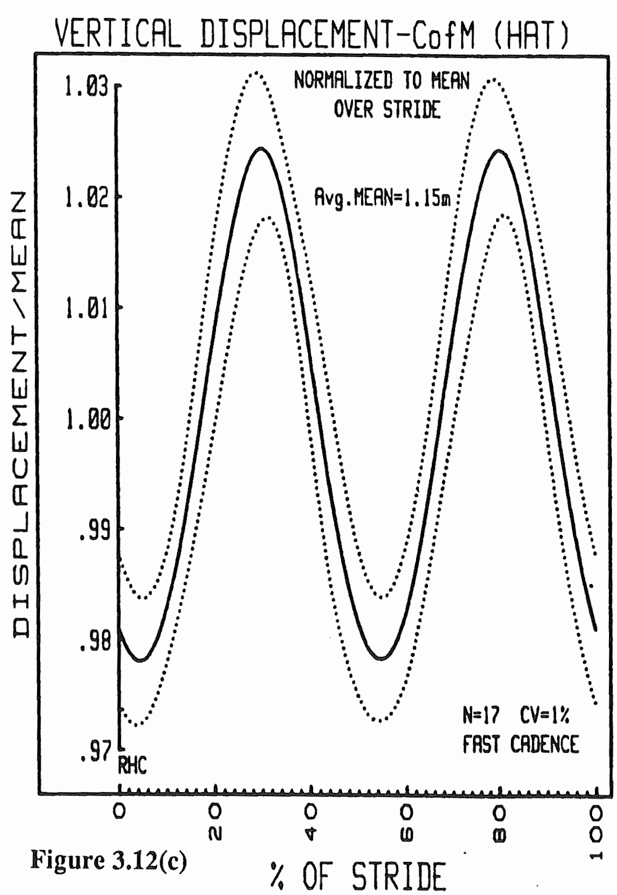

Figure 3.12(c)

This vertical displacement is almost sinusoidal and is matched by an almost equal and opposite change in horizontal velocity (see Figures 3.12(d), (e) and (f)). At the peak of vertical movement, the horizontal velocity reaches a minimum. Conversely, the reverse is true at the time of lowest trajectory. Such a pattern indicates a major conservation of energy within the trunk segment, and this has been documented in mechanical energy analyses (Winter, et al., 1976). Note that the vertical displacement plots have been normalized to the mean value over the stride period; such a normalization was necessary because of the wide range of heights of the subjects in each cadence group. The peak-to-peak vertical oscillation increases as cadence increases; about 3% at slow cadence, 3.5% for natural walking and 4.5% for the fast walkers. The CV's are quite low (≈1%) but this results from the normalization process. A similar normalization for the horizontal velocity resulted in a decreased percentage velocity variation as speed increased: 35% for slow walkers, 21% for natural cadence and 17% for fast walking. CV's decreased from 5% to 3% as cadence increased, but this was mainly due to an increase in the mean velocity.

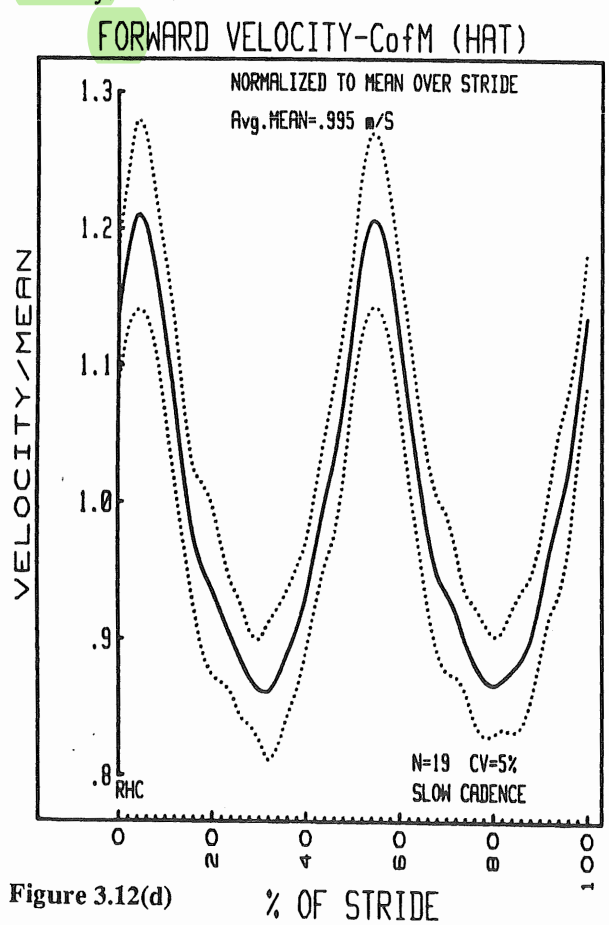

Figure 3.12(d)

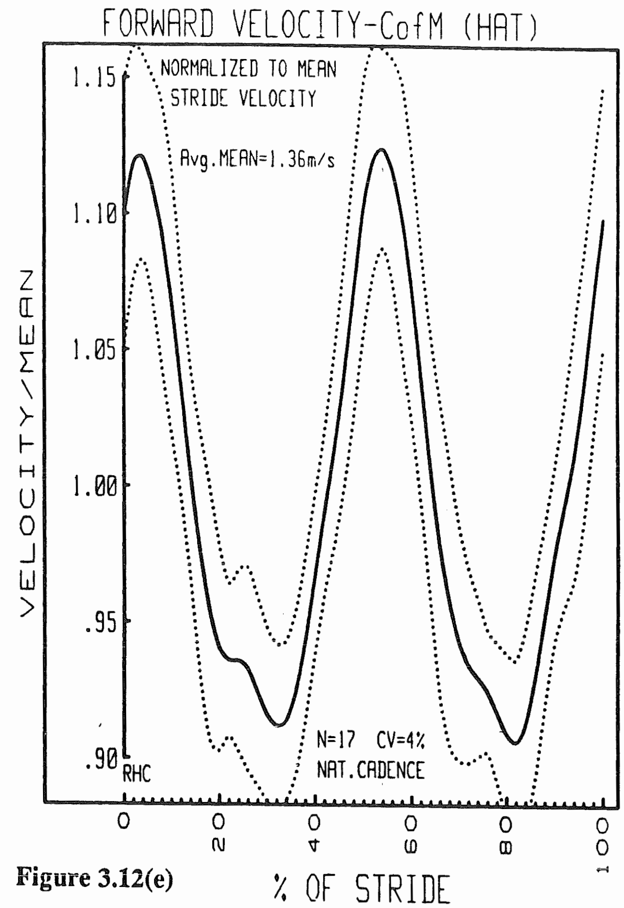

Figure 3.12(e)

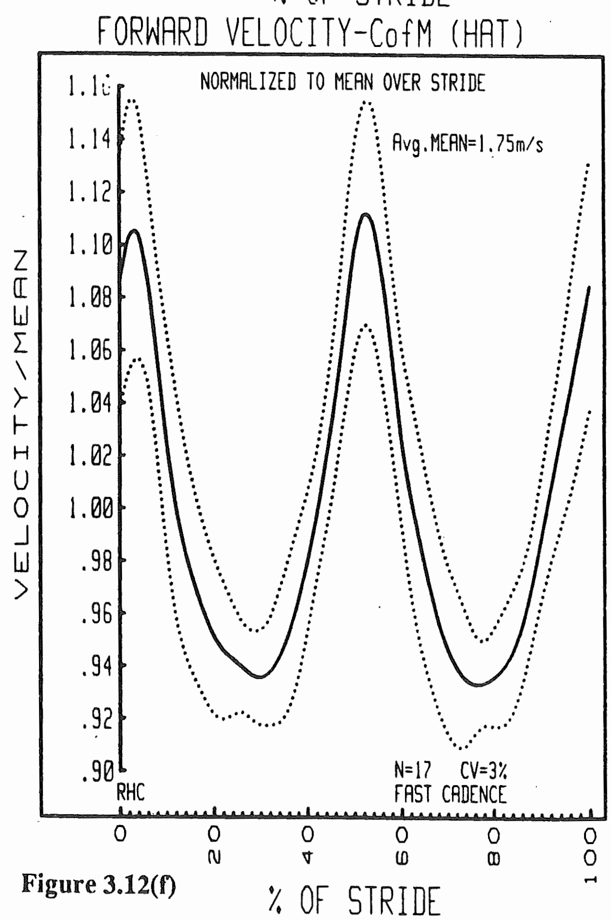

Figure 3.12(f)

3.20 Segment Angular Measures

Angular measures of individual segments in space are rarely calculated or reported. Angular displacements by themselves serve only descriptive functions as they are not used in subsequent analyses. Angular velocities (ω) are used to calculate rotational kinetic energy, $1/2I\omega^2$, and angular accelerations (α) are required in the calculation of the joint moments of force where they appear as the rotational moment, $I\alpha$.

3.21 Angular Displacements

The convention for calculation of segment displacements in space was presented in Section 1.52. All segments are measured in a counter-clockwise direction from the horizontal: Figure 3.21(a) is a repeat of Figure 1.43 and summarizes the definition of these angular measures. Thus, the trunk, thigh and leg angles oscillate either side of 90°, whereas the foot has an additional 90° in its orientation and will be seen to vary either side of 180°.

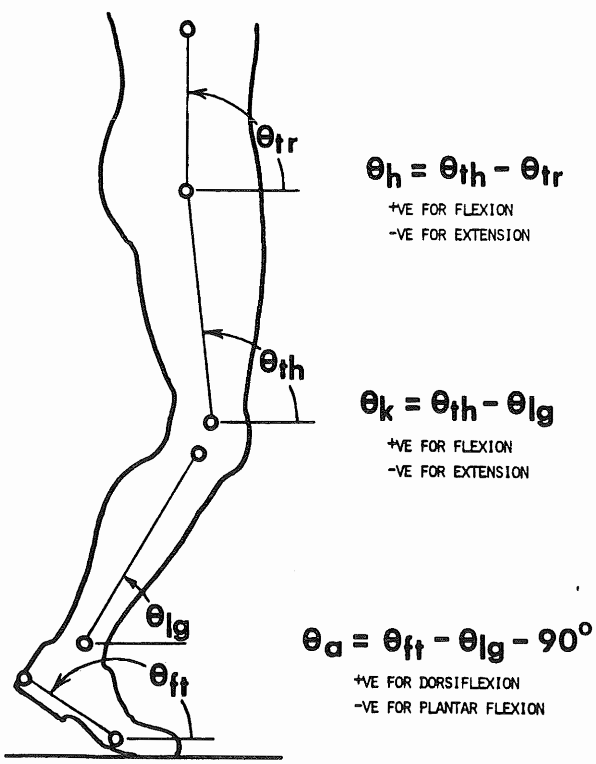

Figure 3.21(a)

3.30 Joint Angles

By definition, joint angles are relative and therefore tell us nothing about the absolute angle of each of the adjacent segments in space. In spite of this, in normal walking, the trunk can be considered to be almost vertical (actually it is biased slightly forward of vertical), thus the hip angle can be used to give a good approximation of the thigh in space, and the knee angle, in turn, would yield the leg angle and the ankle angle would give a reasonable estimate of foot angle in space.

Major early contributions were made by Murray, et al. (1964, 1967, 1969, 1970) in reports documenting adult male and female gait., Using interrupted light photography, they documented the hip, knee and ankle angles. Sutherland (1972) used cinematography to describe the ankle and knee histories. Johnston (1969), Lamoreux (1971) and Smidt (1971) and others have employed goniometers to record the joint angle histories at all three joints. Winter, et al., (1974) using a television system calculated the minimums and maximums of the hip, knee and ankle for three cadence groups (slow, natural and fast) over a range of cadences from 85 to 125. Lamoreux's 1971 study was an in-depth report on one subject who walked on a treadmill from 80 to 130 steps/min.

The results for male and female adults are the same and there are relatively small changes with cadence. The studies on the elderly women by Finley, et al., (1969) showed the range of movement to be less than for younger adults but it is not known whether they documented age-related or cadence-related differences from this slower walking group.

Age has been investigated by Murray, et al., (1966), as an independent variable and for 60 males aged from 20 to 65 there was no significant age-related trend in joint angle histories. Finley et al., (1969) also found no joint angle differences in 23 elderly females (64 to 86 years old) but they recorded shorter step lengths and velocities.

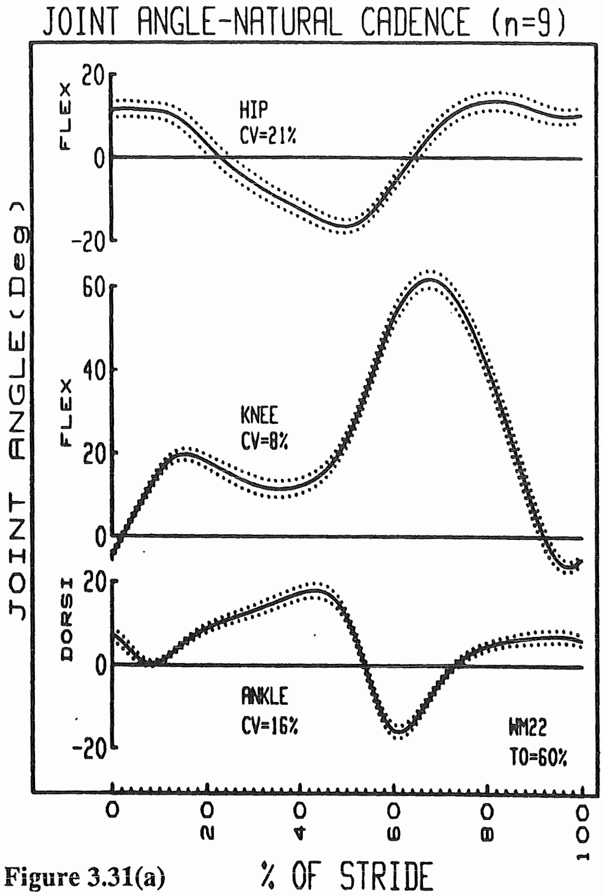

Figure 3.31(a)

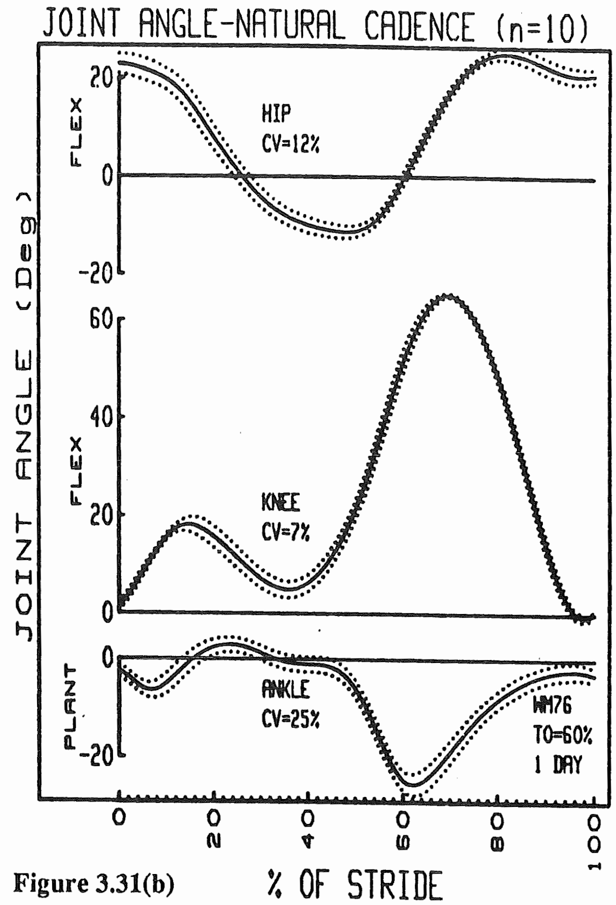

Figure 3.31(b)

3.31 Intra-Subject Joint Angles

The joint angle histories for repeat trials on the same subject done days apart are presented in Figure 3.31(a). As can be seen, the variability as measured by CV is very low. The average variability over the stride is less than 2° at all joints. In a similar manner, repeat trials done minutes apart resulted in very low variability (Figure 3.31(b)). However, it can be seen that the patterns for the two subjects represented here are somewhat different from each other and are probably the reason why we can often recognize an individual by their characteristic gait pattern.

3.32 *Inter-Subject Joint Angles*

In the curves reported here, we have documented the ankle, knee and hip angles from three cadence groups. Natural cadence was 105 ± 6, slow cadence was approximately 20 less than natural, and fast was about 20 higher than natural. Figures 3.32(a), (b) and (c) are ensemble averages of the hip, knee and ankle angles for each of these cadence groups. The solid line is the mean of the joint angles as calculated over the stride period at 2% intervals. The dotted line represents one S.D. either side of the mean. The CV's are as indicated on the curves. Tables 3.32(a), (b) and (c) in the Appendix document these mean curves and standard deviations at 2% intervals over the stride period.

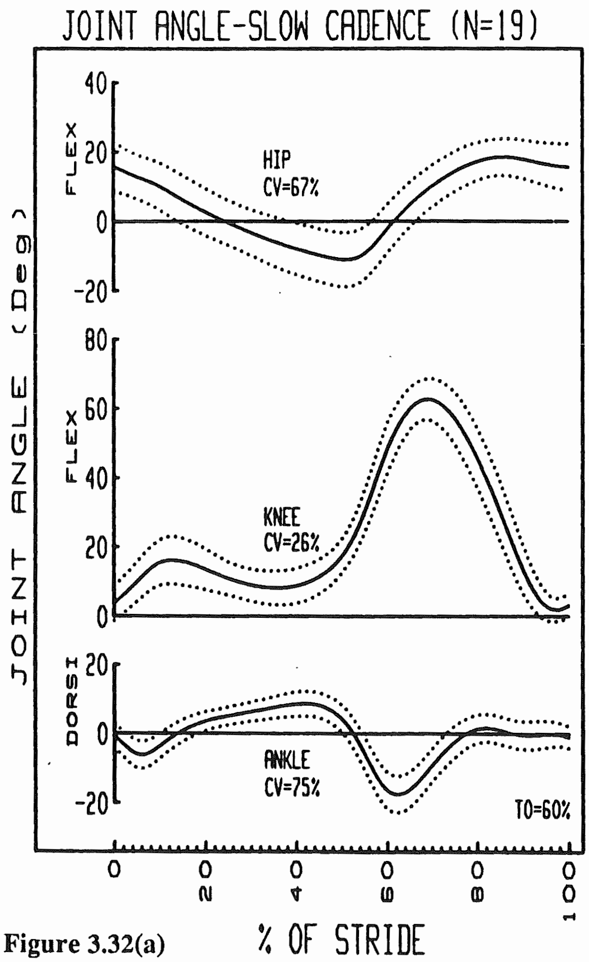

Figure 3.32(a)

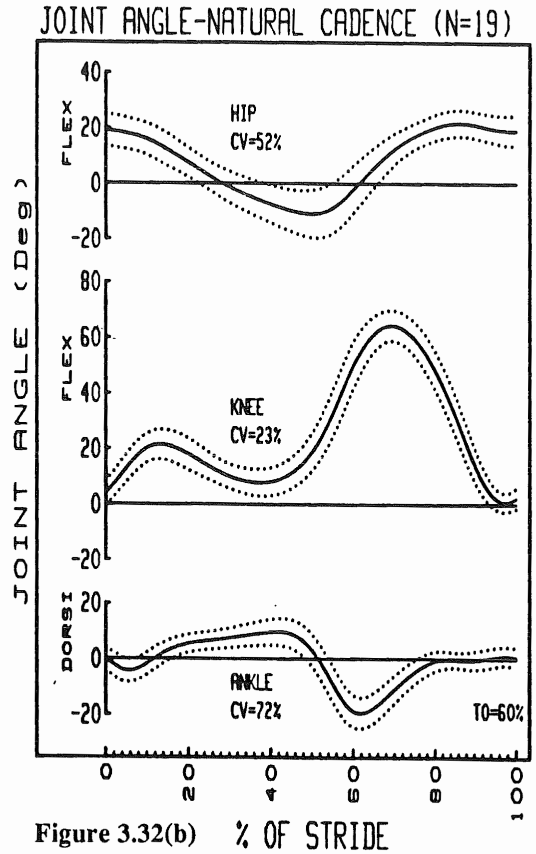

Figure 3.32(b)

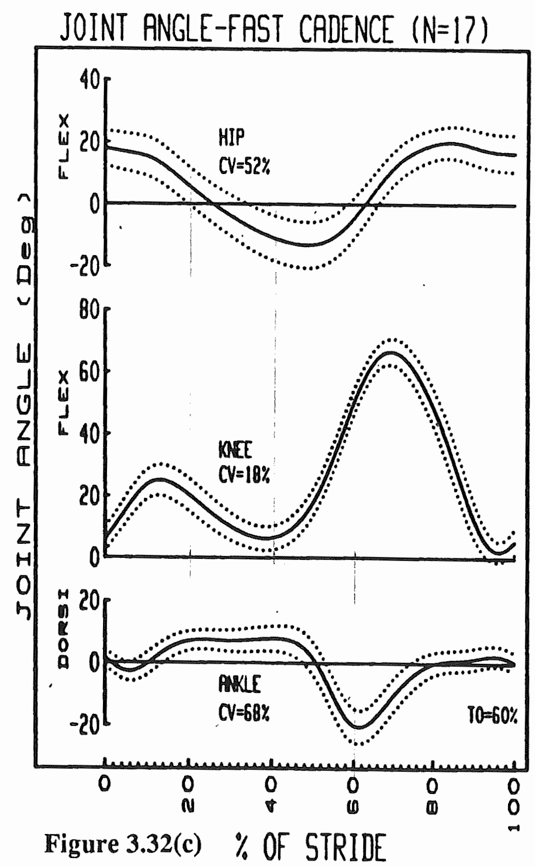

Figure 3.32(c)

To give the reader some indication of any changes that occur over the range of cadences studied, we present comparisons of the mean curves for each joint. Figure 3.32(d) compares the mean ankle, knee and hip curves across the cadences. It is evident from these joint angle plots that there is very little difference between the cadence groups. The only minor difference shows up in knee flexion during early stance. At 15% of stance, the knee reaches maximum flexion and this increases from 15° for slow walkers to 25° for the fast cadence group. The ankle plot also shows very minor differences at this time. Thus, with respect to the stride period, the joint angle histories are essentially the same. A measure of these similarities is presented in Table 3.32 where the correlations between the cadence groups are recorded along with the r.m.s. difference over the stride period.

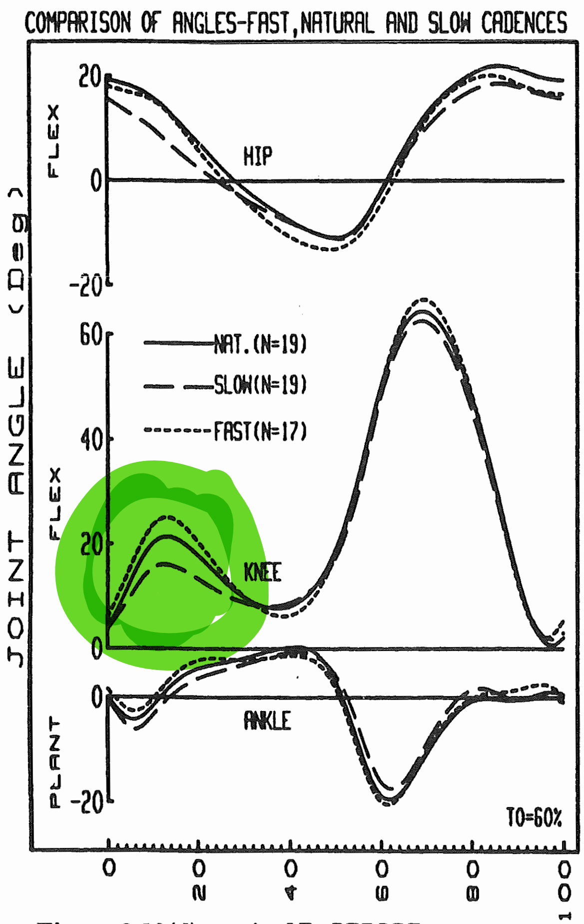

Figure 3.32(d)

Table 3.32

Comparisons of Joint Angles for Slow, Natural and Fast Walking

Cadence Comparison		Slow/ Natural	Fast/ Natural
ANKLE	Diff.	2.1°	2.8°
	Corr.	.975	.95
KNEE	Diff.	3.1°	3.3°
	Corr.	.995	.99
HIP	Diff.	4.1°	2.3°
	Corr.	.99	.99

3.33 Inter-Subject Joint Angular Velocity

For the same subject groups reported in the previous section, the joint angular velocities are presented in Figures 3.33(a), (b), (c) and (d). The first three graphs show the hip, knee and ankle angular velocities at each of the slow, natural and fast cadences.

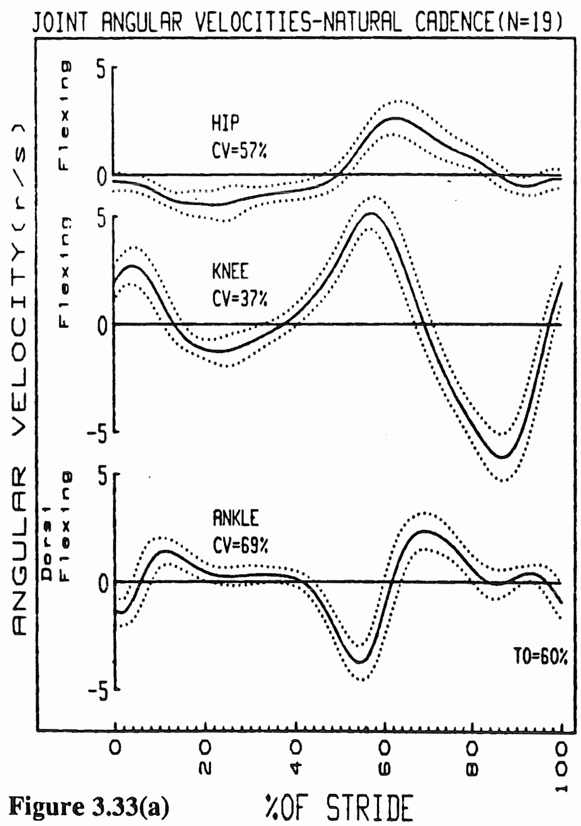

Figure 3.33(a)

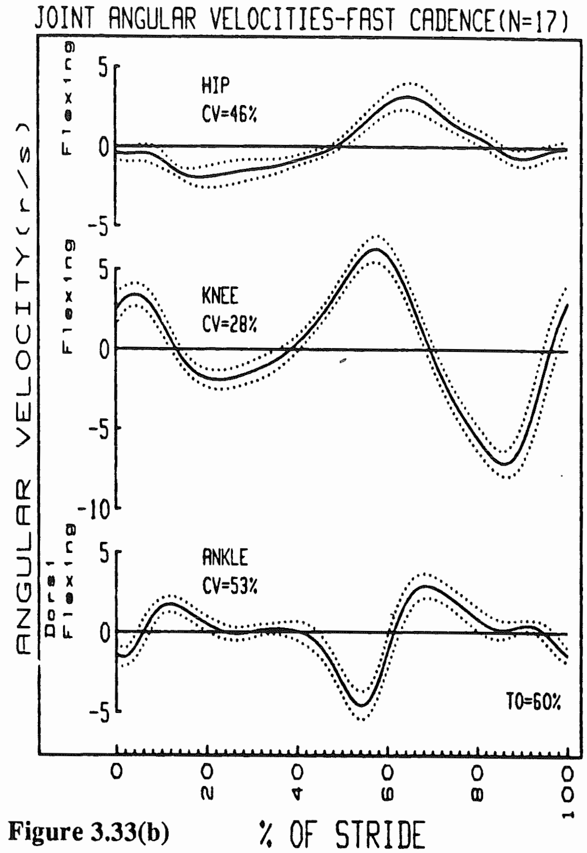

Figure 3.33(b)

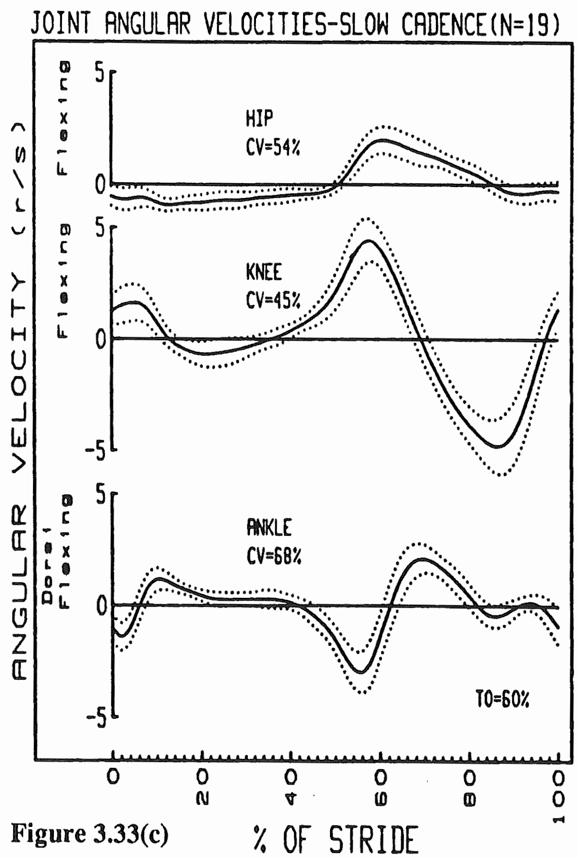

Figure 3.33(c)

Figure 3.33(d) overlays the averaged curves for all these cadence groups. A summary of the similarity in the shape of each of these curves appears in Table 3.33. As can be seen from these correlations, the shape of all the angular velocities is virtually the same at all three cadences. Also, the slope of the regression is very closely related to the ratio of cadences. For example, the fast cadence group walked 17% faster than the natural cadence group and their ankle velocity increased 18%, knee velocity 20% and hip velocity 21%. These findings have implications to the potential role of spindle receptors (which detect muscle velocity) in providing precise feedback to the CNS to control the motor patterns (moments, powers, etc.).

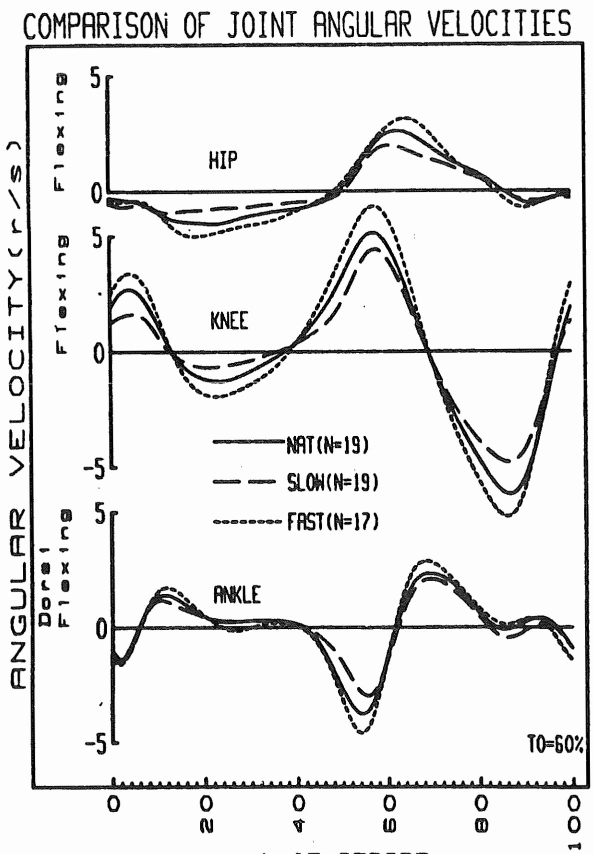

Figure 3.33(d)

Table 3.33

Comparison of Joint Angular Velocities Slow, Natural and Fast Walking

Cadence Comparison	Slow/ Natural	Fast/ Natural
Cadence	.82	1.17
Ankle Velocity	.79 (r=.98)	1.18 (r=.986)
Knee Velocity	.78 (r=.994)	1.20 (r=.995)
Hip Velocity	.73 (r=.984)	1.21 (r=.992)

3.40 References

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11. Winter, D.A., Sidwall, H.G., Hobson, D.A. Measurement and reduction of noise in kinematics of locomotion. J. Biomech. 7:157-159, 1974b.

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4.0 Kinetics

Kinetics by definition deals with those variables that are the cause of the specific walking or running pattern that we observe or measure with our cameras. As such, we are concerned with the individual muscle forces, the moments generated by those muscles across a joint, the mechanical power patterns (rate of generation or absorption by muscles, or rate of transfer between segments), or energy patterns (segment or total body). We are forced to focus on Newton's Laws and the Law of Conservation of Energy in order to interpret what is happening at each phase of the gait cycle. One other basic principle must be borne in mind as we observe these motor patterns, and that is the principle of indeterminacy. There are many combinations of muscle forces that can result in the same movement pattern. In a simple swinging movement of the leg there could be many combinations of agonist/antagonist activity that could result in the same moment of force at a given joint. Similarly, it has been shown that during stance there are an infinite number of combinations of moments of force at the hip, knee and ankle that can result in the same kinematic pattern in the stance limb (Winter, 1984). Such a principle demonstrates the tremendous flexibility and adaptability of our neuromuscular system.

4.10 Ground Reaction Forces

The ground reaction forces as measured by a force platform reflect the net vertical and shear forces acting on the surface of the platform. As such, they are an algebraic summation of the mass-acceleration products of all body segments while the foot is in contact with the platform. The vertical force reflects the accelerations due to gravity as well as the accelerations seen by the camera.

4.11 Intra-Subject Ground Reaction Forces

Figures 4.11(a) and (b) show the averaged horizontal and vertical reaction forces for the trials that were repeated days apart and minutes apart. It can be seen that these forces (normalized to N/kg) are quite consistent over stance: vertical force CV was only 10% in both trial groups and the horizontal force CV was slightly lower (21%) minutes apart as compared with days apart (26%). The shape of these reaction forces is typical of what is reported in the literature. The vertical force has the characteristic double hump: the first is related to weight-acceptance, the second is due to push-off. The horizontal force has a negative phase during the first half of stance indicating a net slowing down of the entire body, and a positive phase during the later half of stance indicating a forward acceleration of the body.

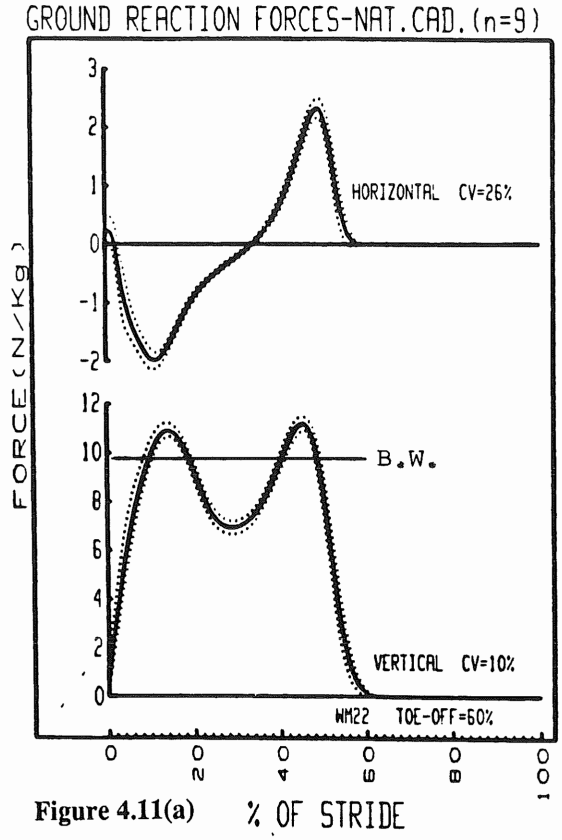


Figure 4.11(a)

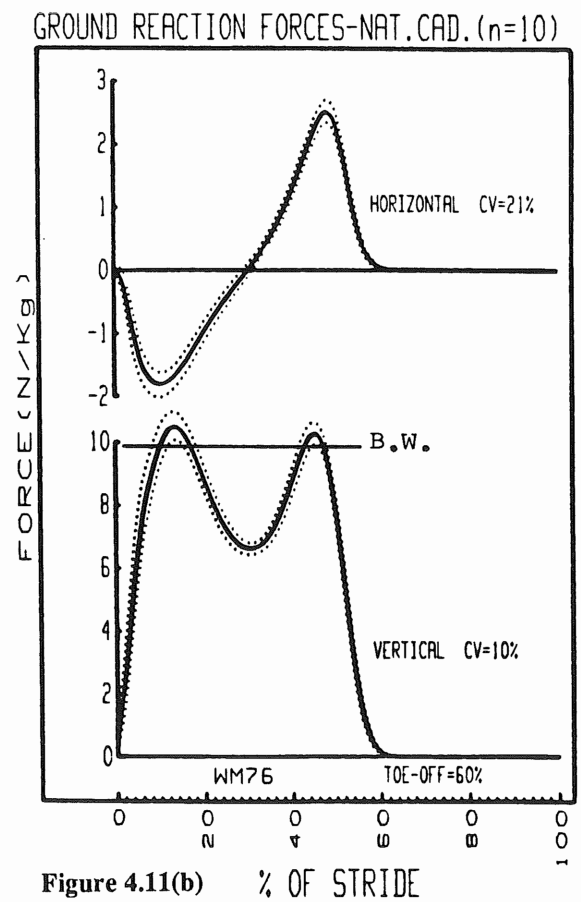


Figure 4.11(b)

4.12 Inter-Subject Ground Reaction Forces

Figures 4.12(a), (b) and (c) report the ensemble averaged ground reaction forces for slow, natural and fast cadence walking. These forces are normalized to body mass (N/kg). The CV for the vertical forces ranges from 15% to 20%, and for the horizontal, it ranges from 43% to 64%. Several differences are apparent between the cadence groups. The positive and negative peaks are about the same for any given speed and the magnitude of these peaks increases with cadence: 1.5 N/kg for slow walking, 2 N/kg for natural cadence and 2.5 N/kg for the fast walkers. The vertical force profiles also showed distinct speed-related changes. Tables 4.12(a), (b) and (c) in the Appendix document these mean reaction forces and standard deviations at 2% intervals over the stride period.

In all cases, the mean vertical force over stance was about the same: about 9 N/kg, but the amount of change as seen in the peak-to-peak differences increased drastically with speed. For slow cadences, maximum vs minimum force was 9.9 vs 8.5 N/kg, for natural walkers it was 10.7 vs 7 N/kg, and for fast walkers it was 12.5 vs 5.5 N/kg. Such major differences result from relatively small differences in knee flexion during weight-acceptance and midstance (as reported in Section 3.32 for these same subjects), and reflect the increased vertical accelerations of the body as cadence increases.

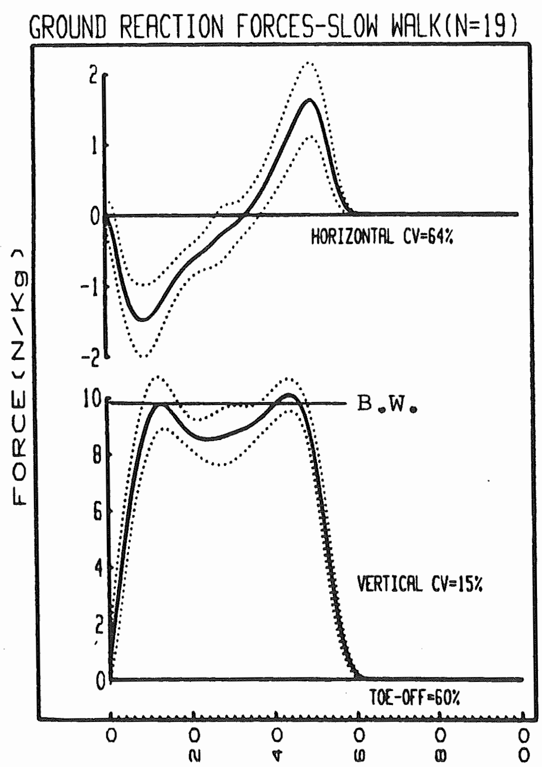


Figure 4.12(a) % OF STRIDE

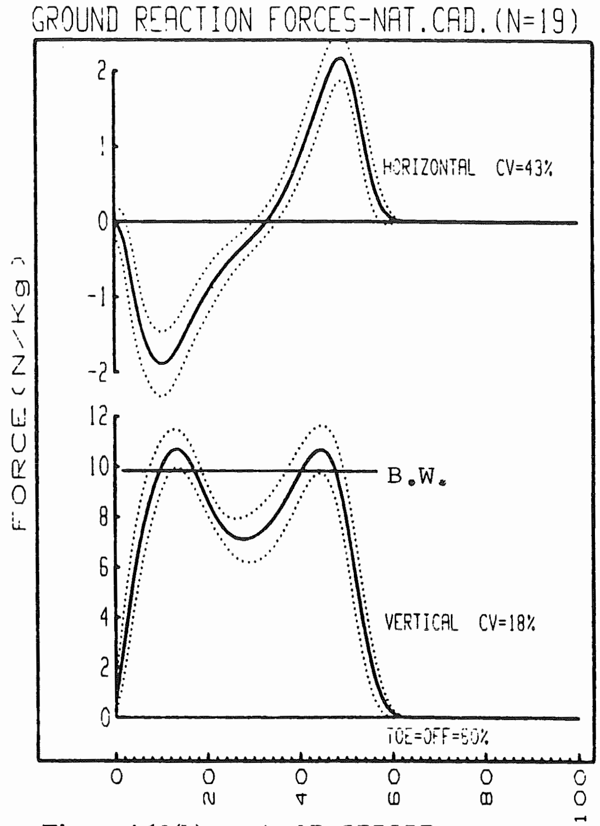


Figure 4.12(b) % OF STRIDE

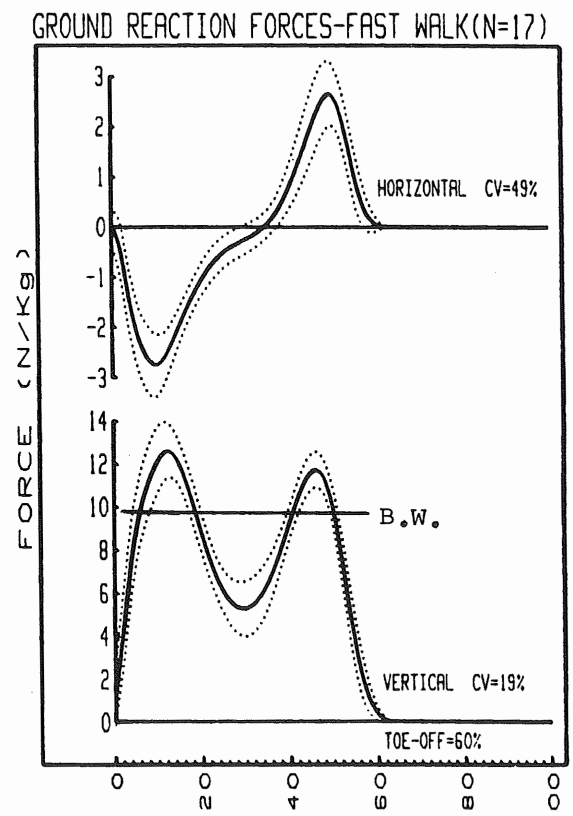


Figure 4.12(c) % OF STRIDE

4.20 Calculation of Moment of Force Patterns

Moments of force (torque) are the net result of all muscular, ligament and friction forces acting to alter the angular rotation of a joint. In normal gait, the joint angles do not reach their extreme limits and the friction forces are minimal. Thus, the net moment, as calculated from link segment analysis, can be interpreted as being due to muscle forces only. Using an inverse dynamics solution of the link segment model, the moments are readily calculated (Bresler & Frankel, 1950; Winter, 1979b). Inputs for this calculation are: horizontal and vertical ground reaction forces and the centre of pressure of those forces; a full kinematic description in absolute coordinates of the lower limbs up to the hip joint; a full anthropometric model (segments, lengths, masses, moments of inertia and locations of centre of mass). The ground reaction force patterns in the plane of progression (vertical and forward) have been described previously. Anthropometric tables are available in the literature or in textbooks (Winter, 1979b). Comments are made in Section 1.55 concerning the erroneous ground reaction vector technique for the calculation of joint moments of force.

To date, there have been very few analyses of moments of force. Elftman, in 1939, presented a pioneering study on methods to analyse and interpret these moments, and not until Paul (1966, 1967) and Morrison (1968, 1970) were computers used to remove the errors and drudgery of hand analyses. Unfortunately, all of these analyses were done on a handful of subjects so little could be inferred from their results as to what is a normal pattern. In a survey reported by Winter in 1980, analyses on only 18 subjects had been reported, but some trends were evident. The ankle moment was consistent: a small dorsiflexor moment at heel contact followed by a plantarflexor moment increasing from foot flat to a peak during push-off. Further studies by Cappozzo, et al. (1976) and Pedotti (1977) focussed on possible mechanisms involved, the relationships between these moment profiles and other variables such as energy and EMG, and then made some estimates of the range of error in such analyses. More recently, Winter (1980, 1984) reports on a large number of subjects at different cadences and identifies a total extensor pattern evident in all subjects during stance (reported in Section 4.222), and a neurologically controlled trade-off between the hip and knee muscles (see Section 4.24).

4.21 Normalization of Moment of Force Patterns

Individual trials can be reported in moments (N.m) vs time (sec) or vs stride time (%). However, when we compare intra or inter-subject results, our time base changes and we must report the results on a time base as a percent of the stride period. Because of the rapid changes taking place at heel contact and toe-off, we must ensure that the stance time is not too different. Thus, it is necessary to compare subjects walking approximately the same cadence (within ± 5%). In this way, we prepare separate ensemble averages for different walking cadences (fast, natural and slow).

The ordinate of our time pattern plots must also be normalized for inter-subject comparisons or averages because of different body masses of each subject. Thus, all kinetic plots are normalized by dividing by body mass. Thus, inter-subject patterns of moment of force are reported as N.m/kg, or power patterns as W/kg.

4.22 Biomechanical Convention for Moments of Force

Figure 4.22 shows the standard convention for moments of force in the plane of progression. Counter-clockwise moments, as calculated at the proximal end of each segment are positive, and clockwise moments are negative. Thus, a knee extension moment is positive while an ankle plantarflexor and hip extensor moment are negative. These moments of force are the net effect of all internal forces (muscle + ligament + friction) acting at that joint.

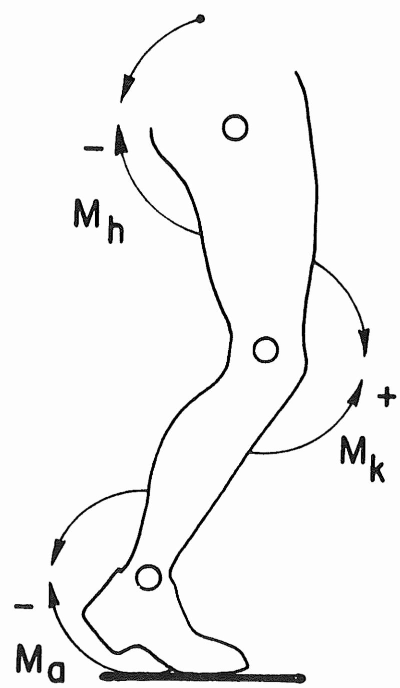


Figure 4.22

4.221 Convention for Reporting Moments of Force

Because of different roles of these joint moments, especially during stance, it is advantageous to report all extensor moments as being positive because they are attempting to push the body away from the ground (i.e., upwards). Conversely, flexor moments are attempting to collapse the lower limb during stance and, because they are attempting to bring the body downwards, it makes sense to report them as negative. Thus, all moments of force are reported on this basis. Also, such a convention makes the interpretation of the support moment much more easily understood, as will be indicated in Section 4.222.

4.222 Support Moment: Calculation and Interpretation

As a result of the analysis of scores of subjects and patients, we have observed one basic pattern evolving from the total lower limb during stance. In spite of variations evident at individual joints, it is seen that there is a net extensor (positive) pattern during stance. We have called that algebraic summation the support moment (Winter, 1980) and, subsequently, we have documented the reason why the support moment is so consistently positive (Winter, 1984). With all extensor moments reported as positive the support moment, $M_s = M_a + M_k + M_h$. In a nutshell, the reason goes as follows. On all studies, the ankle patterns are quite consistent and show a plantarflexor moment during stance. However, there is considerable but highly correlated variability at the hip and knee. For example, on a given stride, the knee extensors (quadriceps) may be dominant in supporting the lower limb (preventing knee collapse) and therefore contributing to the support moment. On a subsequent stride on the same subject, the quadriceps are inactive and the control of the knee results primarily from hip extensor activity (hamstrings + gluteus maximus) and, in this case, the support moment pattern remains about the same. In effect, there is a flexible stride-to-stride trade-off between the muscles crossing these two joints and, recently, we have shown that the trade-off is almost exactly one-for-one during stance (Winter and White, 1985). This means that the knee extensors may decrease their average contribution to support by 10 N.m and the hip extensors increase their contribution by 10 N.m. Thus, the support moment pattern during stance remains virtually the same for the same subject and also is very similar when we compare across subjects.

4.23 Intra-Subject Trials of Moments of Force

Very few studies have dealt with the issue of variability of any kinetic pattern across repeat trials on the same subject. We have completed only two such projects, the first involving 9 repeat trials on the same subject separated by days, and the latter involving 10 repeat trials on another subject done minutes apart. Based on the results of these projects, we can get an indication of the within-subject variability. Figures 4.23(a) and 4.23(b) present the results of the ensemble average of the joint moments of force and the net support moment for the between-day trials and within-day trials respectively. The CV's indicate that the trials done minutes apart are almost half as variable as those done days apart.

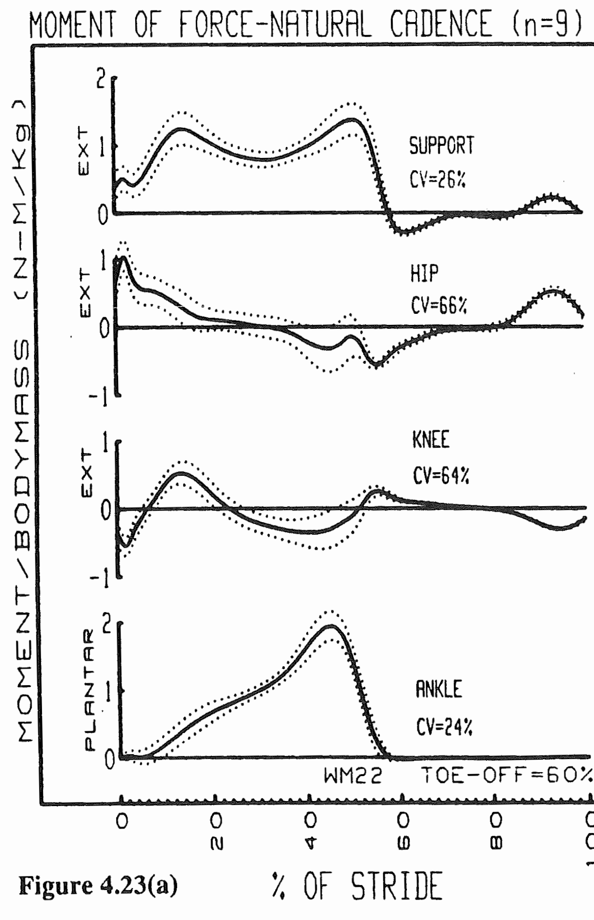


Figure 4.23(a)

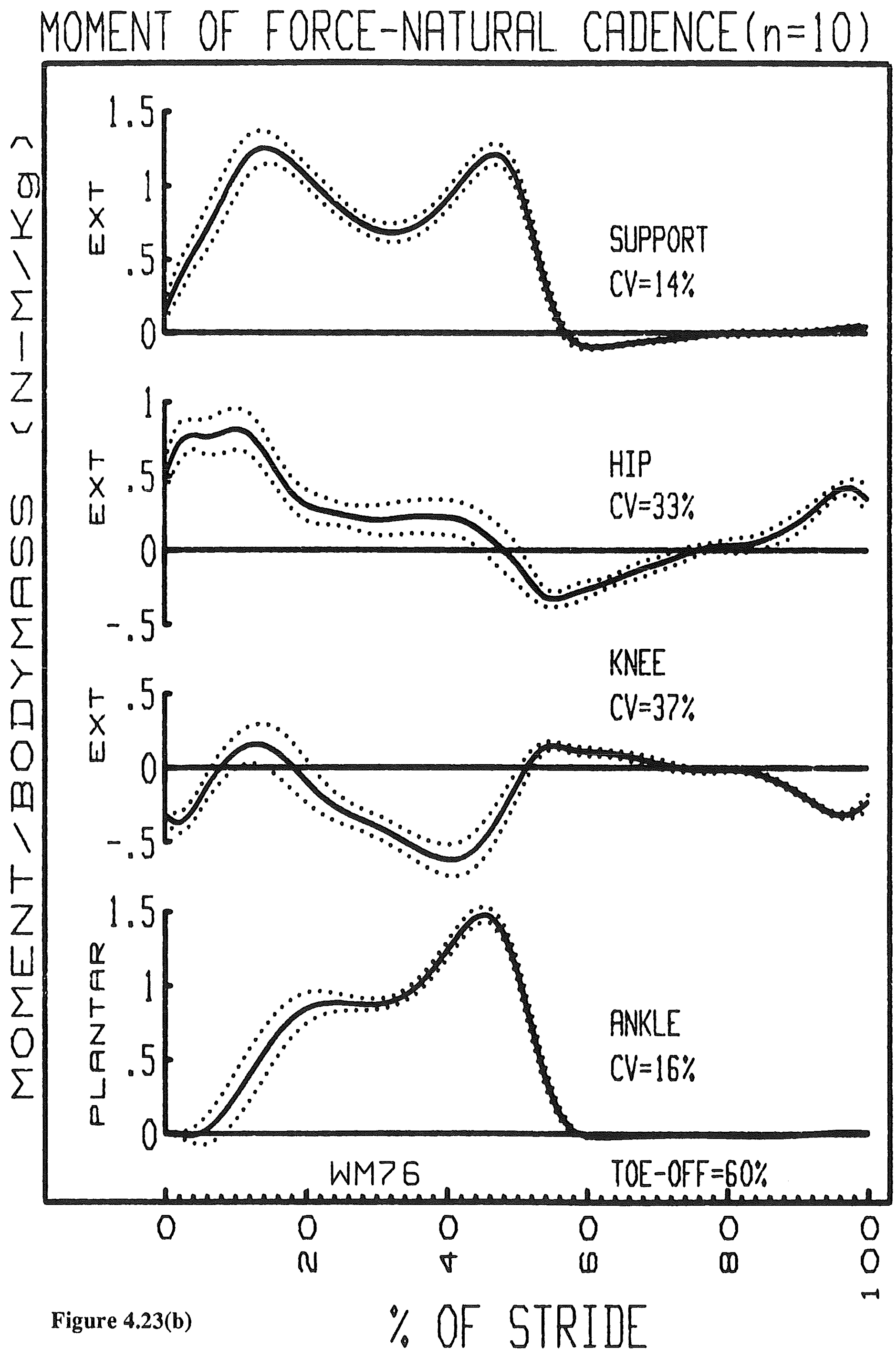

MOMENT OF FORCE-NATURAL CADENCE(n=10)
MOMENT/BODYMASS (N-M/Kg)
EXT
1.5
1
.5
0
SUPPORT
CV=14%
EXT
1
.5
0
-.5
HIP
CV=33%
KNEE
CV=37%
EXT
.5
0
-.5
PLANTAR
1.5
1
.5
0
ANKLE
CV=16%
WM76
TOE-OFF=60%
0
20
40
60
80
100
% OF STRIDE

Figure 4.23(b)

4.24. Inter-Subject Trials of Moments of Force: Variability

As explained in Section 4.21, inter-subject moment of force patterns must be normalized not only for body weight (N.m/kg), but also for cadence. Thus, there are three sets of ensemble average curves to report, one for slow walking (cadence = 86.8), natural walking (cadence = 105.3), and fast walking (cadence = 123.1). These are reported in Figures 4.24(a), (b) and (c), respectively and the values that went into these plots are given in Tables 4.24(a), (b) and (c). The purpose of these tabulated results is so that basic and clinical researchers may readily input these baseline curves into their computer systems without the problem of trying to extract accurate values from the graphs.

The interpretation of these ensemble averages reinforces many factors that have already been discussed. First, we see quite consistent ankle patterns at all three speeds with low CV's (42%-45%). However, at the knee and hip, the variability is extremely high but decreasing with increased cadence. The hip CV decreases from 207% to 76% and knee CV decreases from 171% to 95%. Most of this decrease is due to a general increase in the mean moment of force as we walk faster. For example, the peak extensor moment at the knee early in stance increases from .27 N.m/kg at slow cadences to 1.1 N.m/kg for the fast walkers. Finally, in spite of the tremendous variability seen at the hip and knee, the support moment CV is consistently low (49%-55%). Such evidence further supports the flexible trade-off between the hip and knee muscles to the total support of the lower limb. Tables 4.24(a), (b) and (c) in the Appendix document these mean moment curves and standard deviations at 2% intervals over the stride period.

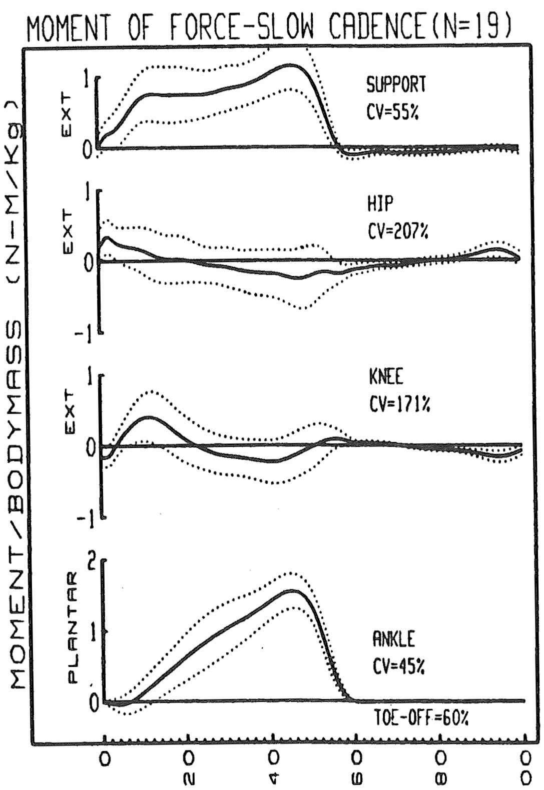


Figure 4.24(a) % OF STRIDE

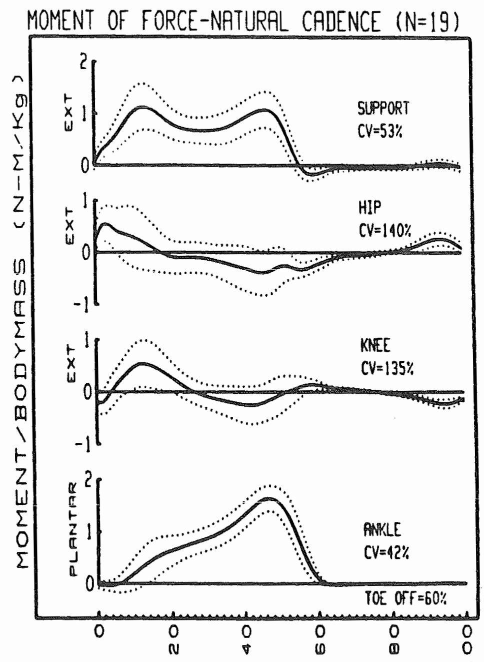


Figure 4.24(b) % OF STRIDE

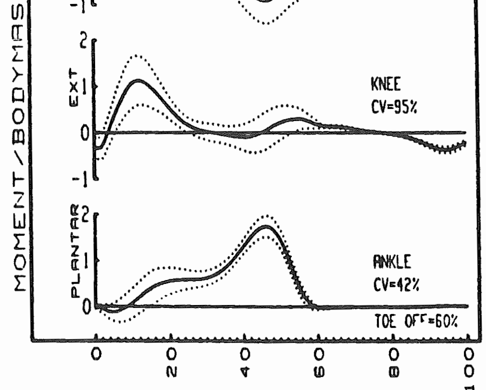


Figure 4.24(c) % OF STRIDE

4.25 Inter-Subject Moment of Force Patterns: Similarity Across Cadences

The variability reported in the previous section results from compensations that were present during the particular stride that was analysed on each subject. However, when we can average 15 or 20 subject patterns we assume that those stride-specific variations have been largely averaged out, and what we see the basic pattern that is characteristic of human gait.

The ensemble average patterns for the ankle, knee and hip for the three cadences are presented in superimposed graphs in Figures 4.25(a), (b) and (c), respectively. Also, the support moment ensemble averages is presented in Figure 4.25(d).

The comparison of the ankle moment patterns (Figure 4.25(a)) is particularly interesting in the way the patterns change from early and mid stance (5%-40%) compared to push-off late in stance (40%-60%). As will be seen later in Section 4.3, the plantarflexors absorb energy during early and mid stance as the leg rotates over the foot, and late in stance, these same muscles cause the foot to plantar flex rapidly and generate an explosive burst of energy (i.e., push-off). The fast walking subjects had a lower moment than the natural and slow walkers during the energy absorption phase while the reverse was seen during push-off. Such a reversal is strong evidence of muscle synergy: in order to walk faster we use the "brakes" less and the "accelerators" more. Conversely, in order to maintain a slow cadence, we increase the braking action and reduce the generator activity. The cross-over of all three of these curves is seen to be at 40% of stride and, as will be seen in Section 4.3, this is precisely the point of transition from energy absorption to energy generation.

The knee moment patterns (Figure 4.25(b)) show that the shape of the motor patterns remains essentially the same at all cadences, except that "gain" is higher as

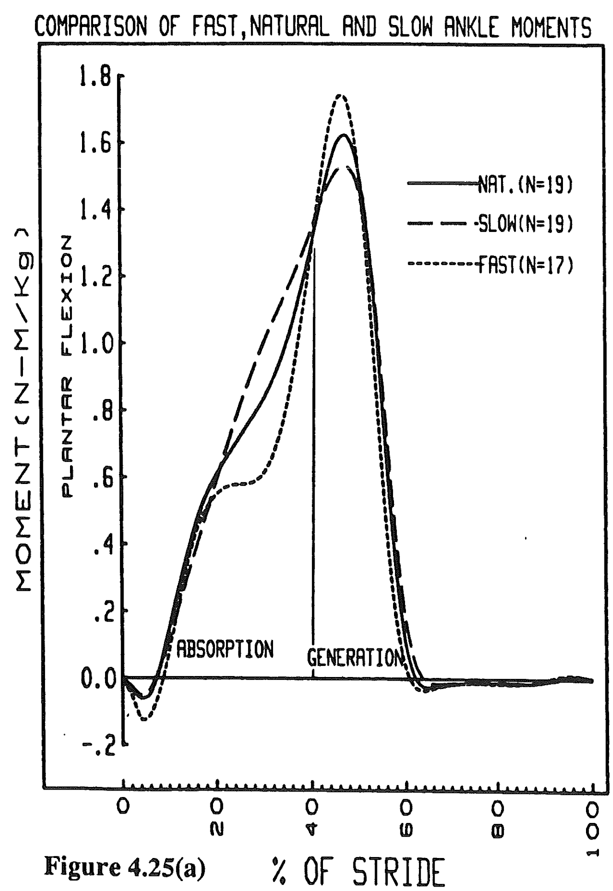


Figure 4.25(a)

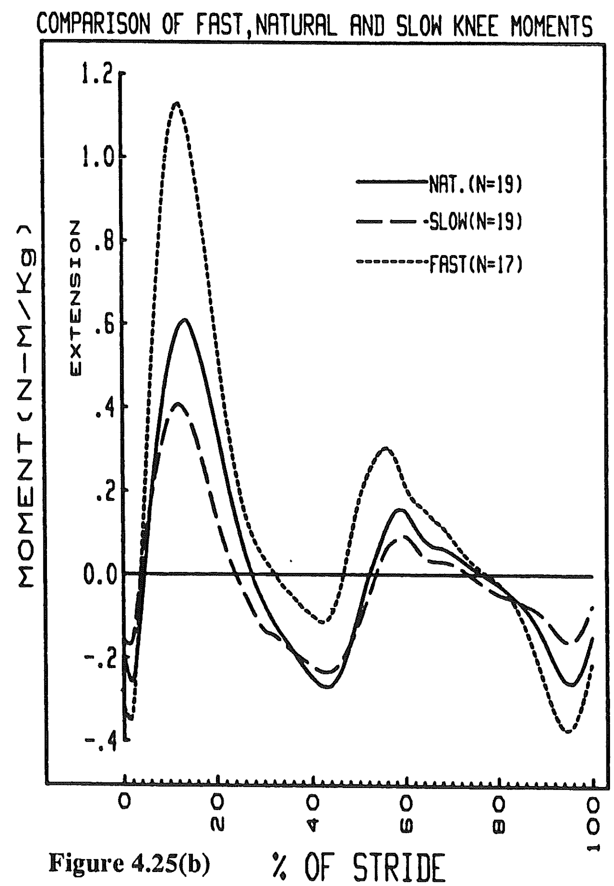


Figure 4.25(b)

the cadence increases. At HC, the knee has a flexor moment but very rapidly (within 4% of stride) the knee extensors begin a major burst of activity, initially to absorb energy and control the amount of knee flexion (until 15% of stride), and then to assist in extending the knee and adding potential energy to the body (which, later in stance, is converted to kinetic energy). Between 30% and 50% of stance, the knee moment becomes slightly flexor and this is mainly due to the action of the gastrocnemius muscles whose forces are becoming quite high at this point in time (start of push-off). However, knee muscles rapidly shift to extensor during late stance through to mid swing. The function of these extensors is two-fold. During the powerful ankle push-off, the knee is starting to collapse (from Figure 3.32 (b) the knee flexes from 5° to 40° during 40% to 60% of stride), thus, the degree of collapse is being controlled by these eccentrically contracting quadriceps. Then, in early swing, these same muscles create an extensor moment to decelerate the backward swinging leg. However, they provide negligible energy to actually accelerate the leg forward (see Section 4.3 on power). During the later half of swing, a very distinct knee flexor moment is seen and this results from the hamstring muscles decelerating the swinging leg. It will be noted that a similarly shaped extensor moment curve, is seen at the hip (Figure 4.25(c)), confirming the synergistic role of the hamstring muscles at these two joints.

The hip moment patterns (Figure 4.25(c)) for the three speed groups are somewhat similar in shape over the stride but do show some timing differences, especially for the fast cadence group. Initially, the hip extensors are active to assist in absorbing energy at IC by controlling hip flexion (and, ultimately, control knee collapse) and also the forward rotation of the pelvis and thereby stabilize the trunk. At fast cadences, this event is over by 10% of the stride while at slow cadences, it lasts until 25% of the stride period. Then, during the remainder of the stride, the hip flexors become active. During mid stance (15%-50%), these hip flexors serve to control the backward rotating thigh and thus arrest its backward rotation. Finally, at 50% of stride, it reverses and the hip flexors now contract concentrically to start a "pull-off" of the lower limb. This continues into mid swing. Pull-off cannot have much effect until the contralateral limb is on the ground; it is during the double support phase that the new base of support can be used to advantage and allow the hip flexors to pull the swinging limb upward and forwards. Many gait-related pathologies use this pull-off to advantage to compensate for a weak or non-existent push-off.

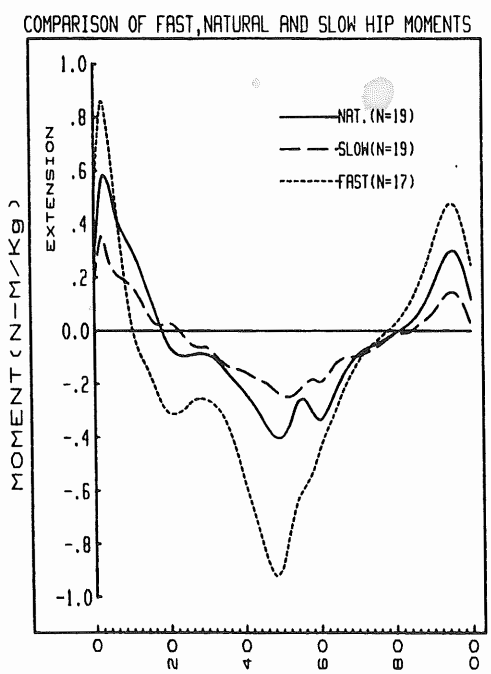


Figure 4.25(c) % OF STRIDE

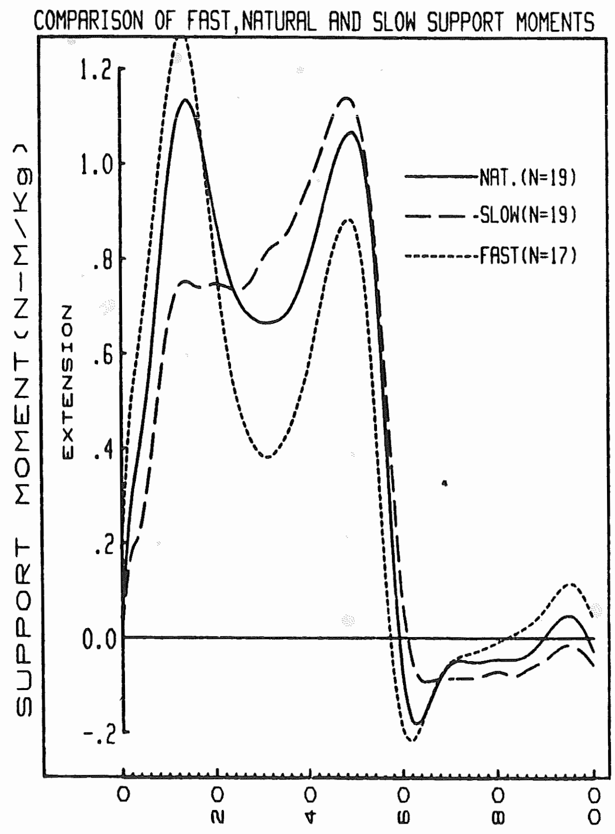


Figure 4.25(d) % OF STRIDE

4.30 Energetics of Walking

Considerable research has reported on the energetics of walking and running. Unfortunately, much of the research has not properly recognized the laws of mechanics or has ignored certain energy saving phenomena that occur during various phases of the gait cycle. For example, Fenn (1929) in a study of runners ignored the potential and kinetic energy exchanges within each segment he analysed, and this resulted in a much exaggerated assessment of internal work. Likewise, Cavagna and co-workers (1963, 1964) introduced a series of papers focussed on the potential and kinetic energies of the body's centre of mass. In doing so, they were looking at the *vector* summation of the mass-acceleration products of all segments, and they failed to realize that energies are *scalars*. Thus, movements of legs and arms in the opposite direction would tend to cancel in spite of the fact that energy was required to accelerate and decelerate those limbs. Thus, all their analyses result in erroneously low values of internal work and they also give results that cannot be traced to the source or sink of the energy. A major improvement was made by Ralston and Lukin (1973), Winter, et al. (1976) and Cappozzo,et al (1976) when they analysed the mechanical energy segment-by-segment. They all concluded, in walking, that the energy conservation of H.A.T. was very high and that the lower limbs dominated the work required. A comparison between this segment-by-segment approach with the centre-of-gravity approach confirmed predicted errors of the latter technique (Winter, 1979a).

The segment-by-segment approach still has some potential errors in it for the calculation of internal work and it still does not identify the source and sink of the energy. Only through a mechanical power analysis can we determine which muscle groups are generating energy and which are absorbing. Elftman (1939) pioneered such ideas and this approach was exploited by Quanbury, et al. (1975), Winter and Robertson (1978), Robertson and Winter (1980), Winter (1983a, 1983b) and Chapman, et al. (1983) in a range of studies on walking and running. Two major energy-saving mechanisms became evident in these segment-by-segment and joint-by-joint analyses. Firstly, an accounting of all the energy changes in each segment showed that passive flow of energy across joints was a very major component which almost entirely accounted for the energy changes of the more distal segments at the beginning and end of swing. Secondly, the magnitudes of active energy transfers across muscles was quite significant and this occurred at several points during the gait cycle when adjacent segments were rotating in the same direction.

4.31 Mechanical Power

If one were pressed to give the most important role of muscles during any movement, one would have to consider the function of muscles as they shorten and lengthen under tension. It is during these concentric and eccentric phases that muscles generate and absorb the mechanical energy necessary to accomplish the movement that we observe. The single variable that summarizes that role is mechanical power, and it is the product of the joint moment of force and joint angular velocity.

$$P_j = M_j \cdot \omega_j \quad \text{watts}$$

where: M_j is the joint moment of force (N.m)

ω_j is the joint angular velocity (r/s).

The convention of M_j and ω_j is such that P_j is positive if M_j and ω_j have the same polarity (i.e., a concentric contraction). Conversely, when M_j and ω_j have opposite polarities we have an eccentric contraction and P_j is negative. The assumption inherent in the above calculation is that we have a torque generator at each joint which is independent of what is happening at adjacent joints. However, it is possible with close scrutiny to see the influence of biarticulate muscles acting across adjacent joints. For example, during a given phase of the gait cycle, we may note generation of energy at one joint and absorption at an adjacent joint. If the moment of force patterns at that time appear to be related to activity of biarticulate muscles, then there is a good chance that those muscles appear to have generated energy at one end and absorbed energy at the other end; in fact, they may be contracting in a near isometric fashion and are merely transferring energy between the segments attached to the origin and insertion.

4.32 *Example Patterns of Ankle and Knee Powers*

Figure 4.32(a) shows how the ankle powers are calculated. Plotted are curves of ankle angle broken

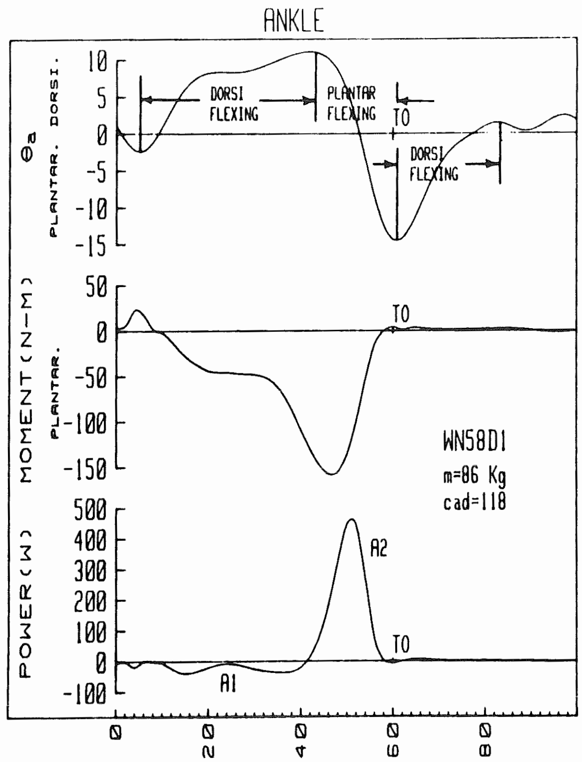


Figure 4.32(a) % of STRIDE

into phases showing when plantarflexion and dorsiflexion are taking place, the ankle moment of force showing plantarflexor and dorsiflexor moments and, finally, the power generation (+ve) or absorption (–ve). The more important power phases are labelled as A1, A2. Between HC and 5% of stride, the foot plantarflexes under the control of a small dorsiflexor moment. The resultant power absorption is small and is not labelled. From 5% to 40% of stride, the leg rotates over the foot (ankle dorsiflexes) under the control of an increasing plantarflexor moment. This results in A1 phase of energy absorption. Then at 35% of stride, the plantarflexor moment has increased sufficiently to cause heel off and then by 40% of stride to cause active and rapid plantarflexion of the ankle. The product of Ma and ωa is quite high and A2 power burst results. This is the single most important energy generation phase and results in 80%-85% of that generated during the entire gait cycle (Winter, 1983). At 60% of stride, TO occours and the strong plantarflexor moment ends. A very small dorsiflexor moment then causes rapid dorsiflexion of the foot until 75% of stride when it is sufficiently dorsiflexed to clear the ground in mid swing. However, the power generation associated with this low mass ballistic movement is extremely small.

Figure 4.32(b) presents a typical knee power analysis. The knee angle shows when the knee is flexing and extending; the knee moment of force is extensor during stance into swing, and flexor during late stance;

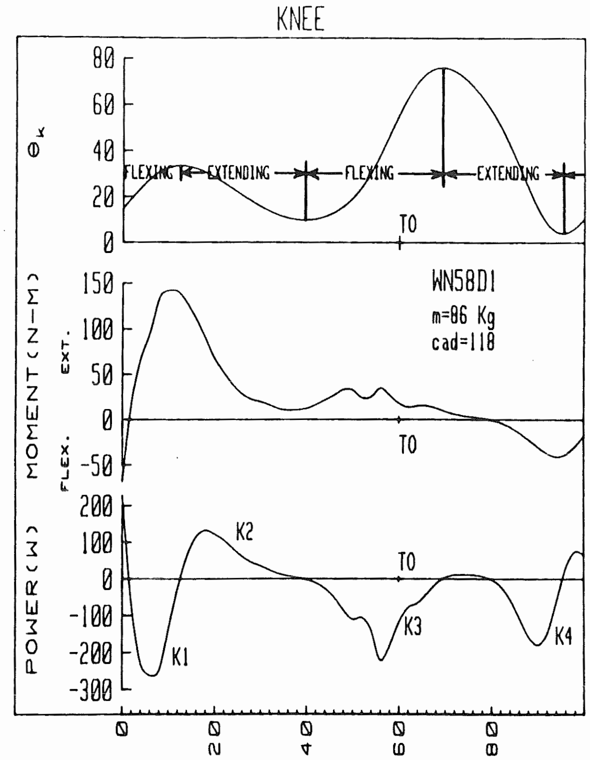


Figure 4.32(b) % OF STRIDE

the power curve shows four distinct phases of absorption and generation. From HC to 15% stride, the knee flexes under the control of knee extensors and results in K1, the first major absorption phase. From 15% to 40% stride, the knee extends partially under the control of quadriceps (the knee may also be controlled by hip extensors at this time). This concentric contraction results in K2, the only positive burst of power from the knee extensors and represents only about 10%-15% of the energy generation in level walking. At 40% of stride, the knee starts to flex and continues to do so into early swing (70% stride). The small knee extensor moment at this time has two functions associated with the K3 absorption burst. Prior to TO, the quadriceps absorb energy and provide some control of the collapsing knee joint. After TO, these same muscles continue to absorb energy by decelerating the backward swinging leg and preparing it for extension during most of swing phase. It should be noted that the quadriceps do not provide any more than token power to swing the leg forward. The actual energy to swing the leg comes from pendulum action (converting potential energy to kinetic) and due to hip moments providing an appropriate couple at the knee joint (Winter, 1978). Finally, in the latter half of swing, the knee flexors (hamstrings) turn on during K4 burst they absorb most of the energy from the swinging leg and foot.

Hip powers are seen to be quite variable across subjects and, therefore, to describe the power patterns from the data from one stride of any given subject may not be representative of the average population. Thus, a description of such patterns will be given in the section on inter-subject averages.

4.33 Intra-Subject Averages of Mechanical Power

As indicated in Section 4.23, there have been very few studies with repeat trials on the same subject. We have completed only two such studies. Figure 4.33(a) reports the ensemble average of 9 repeat trials on the same subject done days apart and Figure 4.33(b) summarizes the same curves for a second subject whose trials were done minutes apart. Both plots normalize the power and report them as W/kg. The range of variability is as expected; repeat trials done minutes apart are somewhat less variable than those done days apart. Also, there are some minor differences in the magnitude and phase of some of the peaks. Subject WM22 (repeat trials across days) had fairly characteristic peaks except K2 was quite reduced. WM76 (repeat trials minutes apart) had no K2, A1 momentarily reversed at mid stance, and H1 was very large compared to the normal population. These typical differences between individuals are typical and are the reason why the inter-subject profiles of power are so variable.

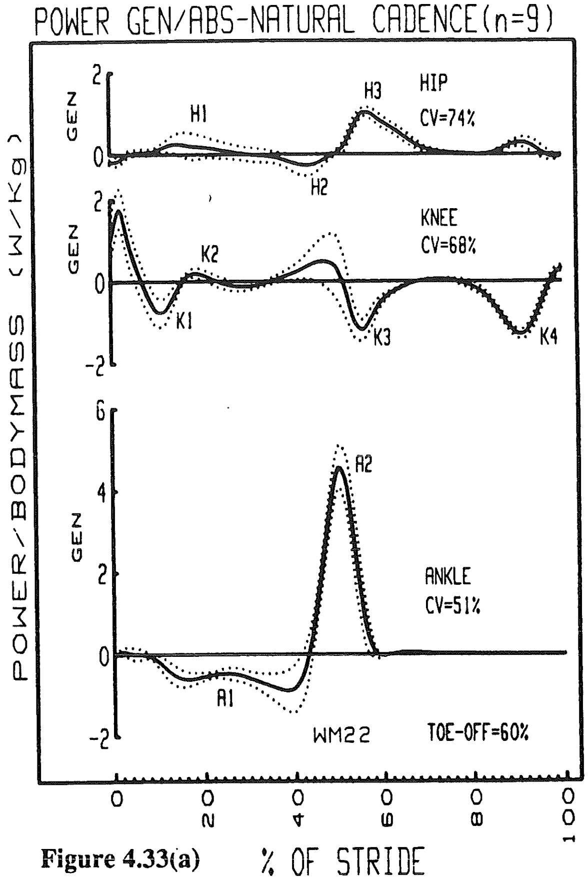


Figure 4.33(a)

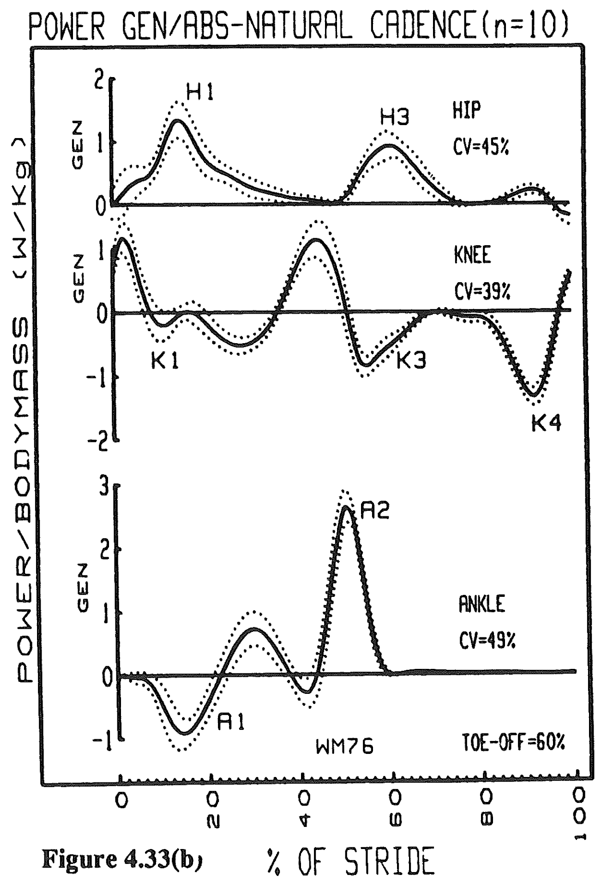


Figure 4.33(b)

4.34 Inter-Subject Averages of Mechanical Power

As explained in Section 4.21, any inter-subject ensemble averages must be normalized for cadence and for body weight. Thus, subject trials are grouped as natural, fast or slow and the powers are divided by body mass and reported as W/kg.

Figures 4.34(a), (b) and (c) show the ankle, knee and hip powers for slow, natural and fast cadence groups. The variability of these power curves at different cadences gives us some insight into the varying degrees of flexibility at each joint. The ankle power curves show a slight reduction in CV's as cadence increases and this trend is entirely due to the increase in the A2 burst at push-off as cadence increases (from a peak of 2.1 W/kg at slow walking to 4.5 W/kg at fast cadences). The knee and hip variability at slow and natural walking is extremely high and is partially predicted by a similarly high variability in the moments of force that were responsible for these powers (see Fig. 4.24(a) and (b)). At the hip, for example, the power levels for most of stance (H1 and H2 phases) are very low and not well defined; thus, the CV measures for the stride are very high. At these slower cadences, the hip powers are mainly to maintain trunk balance, and this task is quite variable across individual strides of each subject. It is not until we achieve a fast cadence that the hip power patterns become well defined. H2 is the absorption of energy by the hip flexors as the thigh is decelerated as it rotates backwards. Also, H3, the "pull-off" burst, increases rapidly as the inertial load of the swinging limb becomes important, and the CV at the hip is as low as that seen at the ankle. Tables 4.34(a), (b) and (c) document these mean power profiles and standard deviations at 2% intervals over the stride period.

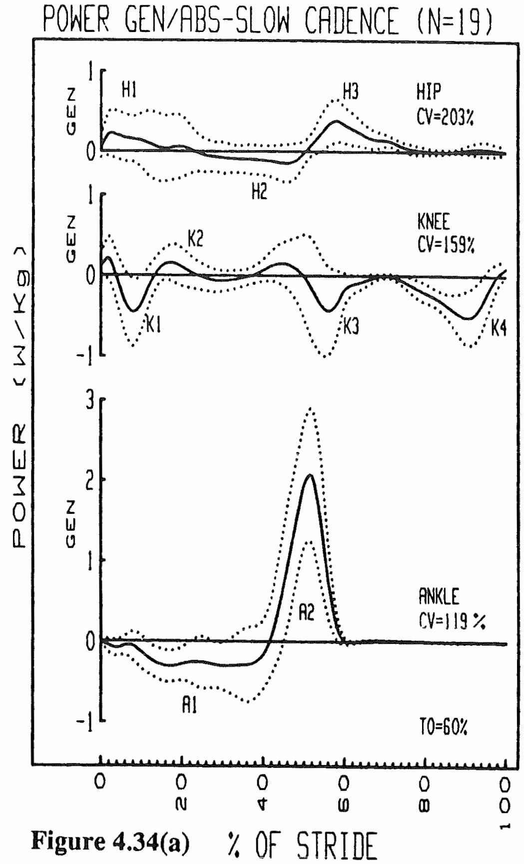


Figure 4.34(a)

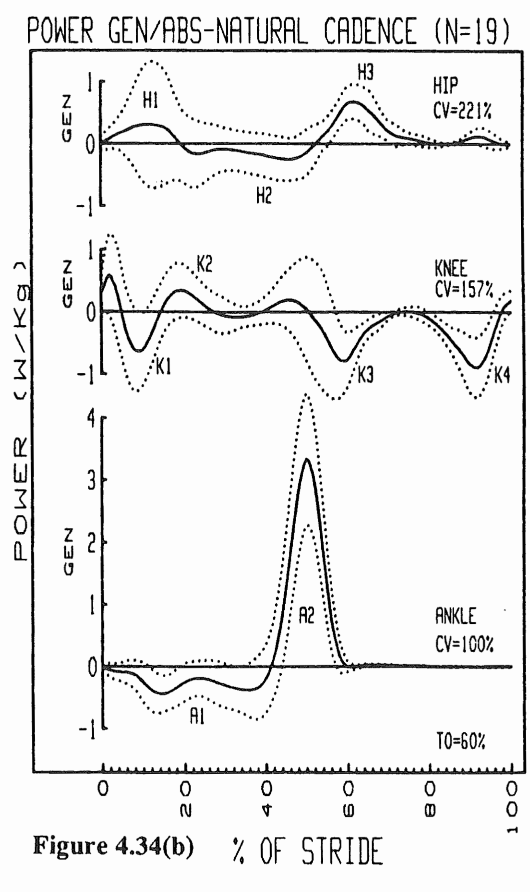
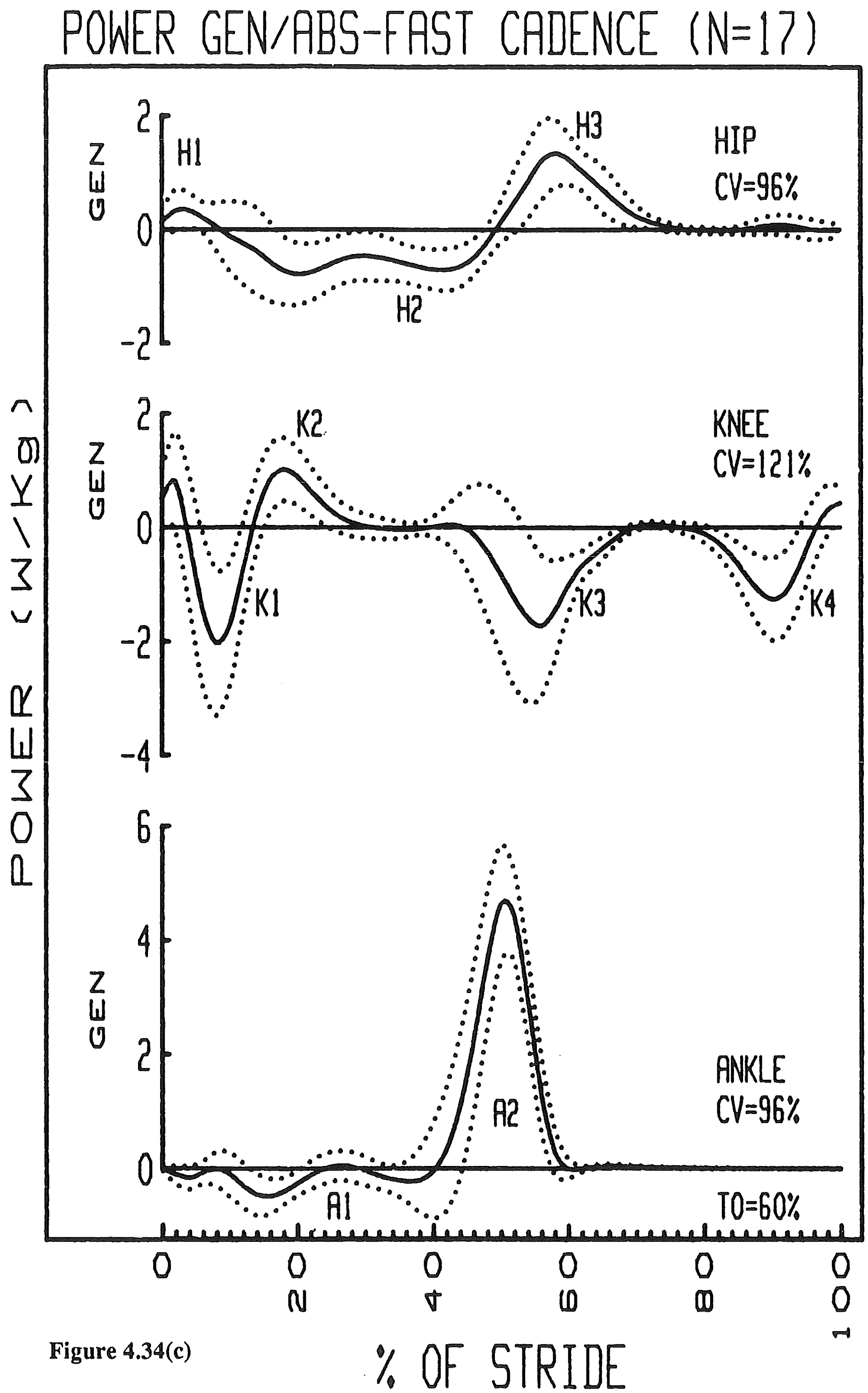


Figure 4.34(b)



POWER GEN/ABS-FAST CADENCE (N=17)
POWER (W/Kg)
GEN
H1
H2
H3
HIP
CV=96%
2
0
-2
K1
K2
K3
K4
KNEE
CV=121%
-4
A1
A2
ANKLE
CV=96%
TO=60%
6
4
0
20
40
60
80
100
% OF STRIDE

Figure 4.34(c)

It is useful to compare the power patterns at the same joint across all three speeds. This can be accomplished in spite of cadence differences because the time bases are already normalized to 100%. Figure 4.34(d) shows the ankle joint power patterns. It can be seen that the timing of the patterns is almost identical at all speeds and that the "gain" increases with cadence. The correlation between these curves is a measure of their similarity in shape. The slow/natural curves have a correlation of .98 while the fast/natural curves have a correlation of .99. Figure 4.34(e) presents the knee joint powers, and again we see that the timing of the power bursts within the stride period is almost the same for all three cadence groups, and that increased cadence is associated with increased "gain". Correlations for the slow/natural curves was .94 and for the the fast/natural powers was .87. Finally, in 4.34(f), the hip power comparisons is a further demonstration of the similarity in shape and increase in gain as cadence increases. The slow/natural correlation was .91 and fast/natural was .87.

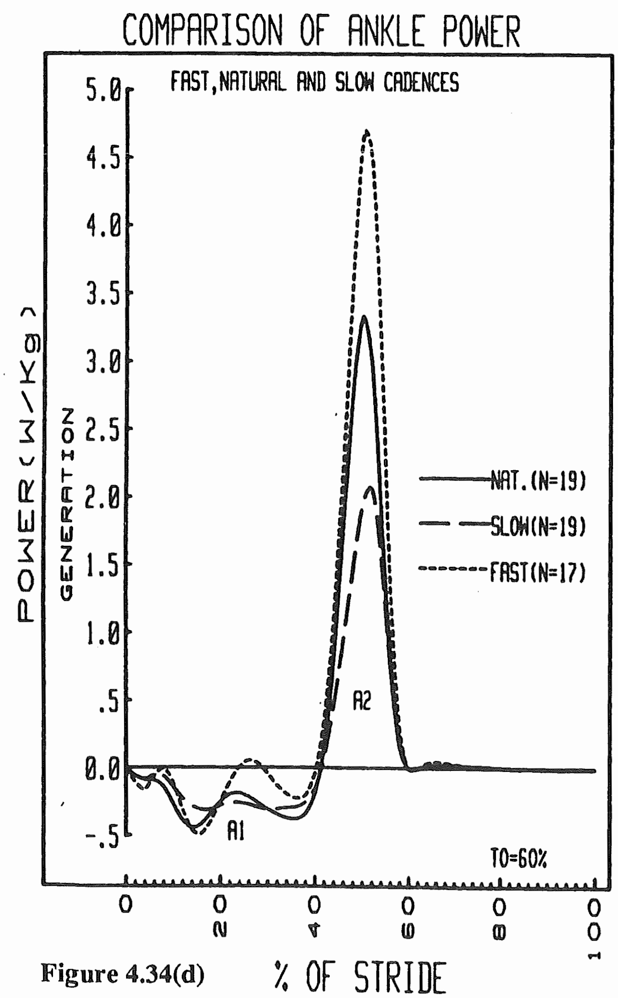


Figure 4.34(d)

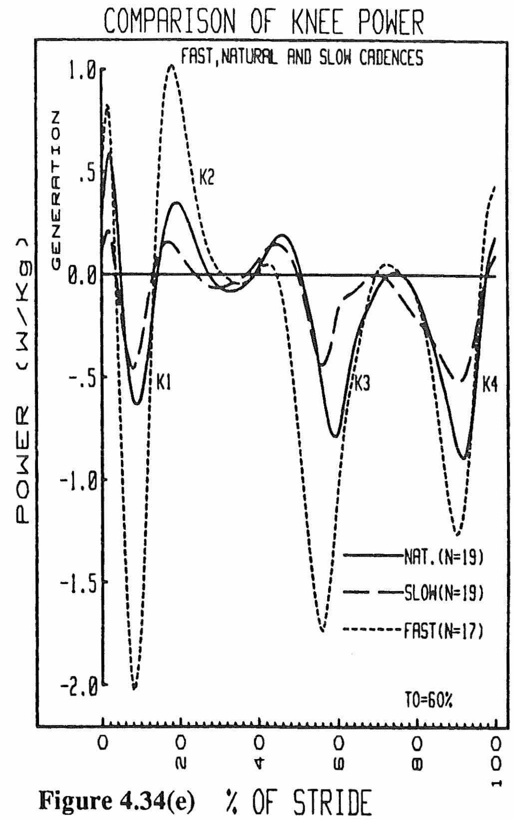


Figure 4.34(e)

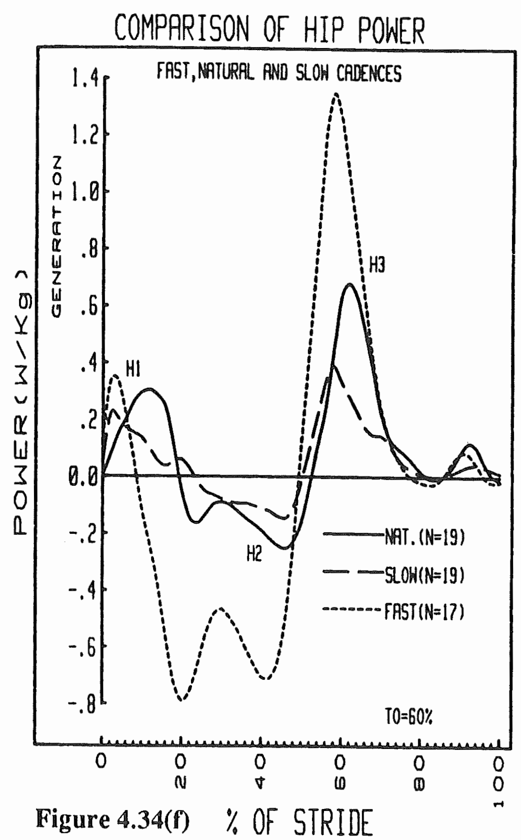


Figure 4.34(f)

4.40 References

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5.0 Electromyography in Human Gait

The electromyogram (EMG) is the single best representation of the neurological control (activation) of skeletal muscle. Each motor unit, when activated, produces a motor unit action potential (m.u.a.p) that is characteristic of that motor unit and the position of the pick-up electrodes. During any voluntary movement such as gait, the net EMG signal is merely an algebraic summation of all m.u.a.p.'s that are active at that time. It has been shown by many researchers that many parameters affect the magnitude and harmonic content of the EMG: electrode location, electrode size and shape, spacing between electrodes, fat tissue overlying the muscle, muscle temperature, cross-sectional area and length, etc. It has been shown that surface electrodes are more reliable than indwelling electrodes and that they detect m.u.a.p.'s from a wider area of the muscle. If care is taken to minimize cross-talk from adjacent muscles, surface electrodes are the best choice for superficial muscles. For deep muscles, the only route to go is with indwelling electrodes. In gait, we have been able to isolate 16 muscles and record their patterns, and these muscles represent the more important muscle groups.

The various problems of recording and reporting the EMG have been addressed by many research papers. The reader is referred to a document published by the International Society of Electrophysiological Kinesiology (ISEK) which makes recommendations regarding units, terms and standards in EMG research (Winter, et al., 1980).

5.01 Processing of the EMG

Many researchers have reported their EMG as a raw (unprocessed) signal. Unfortunately, it is difficult to interpret the amplitude and shape of these raw signals, and it is impossible to quantify their stride-to-stride variability. Thus, there is a need to process these raw EMG signals to provide the assessor with a "pattern" that can be justified on some biophysical basis.

Many studies (Bouisset, et al., 1973; Close, et al., 1960; de Jong and Freund, 1967; Hof and van den Berg, 1977; Komi, 1973; Lippold, 1952) found linear relationships between muscle tension and various measures of EMG amplitude (linear envelope, integrator resets/unit time, average full-wave rectified) during isometric contractions. Other studies (Hof and van den Berg, 1977; Komi, et al., 1978; Maton, 1976; Vredenbregt and Rau, 1973; Zuniga, et al., 1970) showed varying degrees of non-linearity in some or all of the muscles tested. Fortunately, the non-linearity occurs only at high tension levels, well above that seen in most muscles during normal walking.

A number of smoothing techniques are available that follow the rises and falls of EMG as the muscle increases and decreases its tension. It is desirable that the technique chosen bears some relationship to the biomechanics of the movement and, specifically, to the muscle force or joint moment of force. In this way, the observer can make intelligent correlations with the observed movement. For example, in a ballistic movement, a suitably processed EMG should be closely related to the rotational acceleration or deceleration of the joint controlled by those active muscles. Because of the known electromechanical delay between the m.u.a.p. and the resultant twitch waveform, the muscle tension waveform will have a significant phase lag behind the associated EMG profile. Thus, any processing scheme must have an appropriate delay. Inman and associates (1952) noted more than three decades ago that the linear envelope (full wave rectifier followed by a low pass filter) of the EMG followed closely the rises and falls of muscle tension. A number of other researchers have determined that the transfer function relating a m.u.a.p. (considered to be an impulse) to the twitch of tension is a second order system, critically damped or slightly overdamped. The cut-off frequency was in the range of 2.3 Hz to 8.5 Hz (Aaron and Stein, 1976; Crosby, 1978; Milner-Brown, et al., 1973). Such responses are compatible with the 30 ms to 90 ms range of contraction times reported in the twitch waveforms (Buchthal & Schmalbruch, 1970; Milner-Brown, et al., 1973). Good correlations have been reported between muscle force and the EMG linear envelope during isometric anisotonic contractions (Calvert and Chapman, 1977; Crosby, 1978; Winter, 1976). Recently, we have reported an optimized EMG-driven muscle model (Olney & Winter, 1985) to predict the moments of force during walking, and the optimal cut-off frequency of the critically damped filters ranged from 1.0 Hz for the soleus to 2.8 Hz for the rectus femoris. Thus, for the EMG profiles reported here, linear envelope processing with the filter cut-off frequency of 3 Hz is employed.

5.02 Normalization of the EMG Linear Envelope

A wide range of attempts has been made to identify normal EMG patterns. Reports vary from data from a single subject (Chong, et al., 1978; Thorstensson, et al., 1982), a small number of strides on a large group of normals (Dubo, et al., 1976), to a range of inter-subject normalization protocols using a large number of strides and subjects (Arsenault, et al., 1985; Knutsson and Richards, 1979; Yang and Winter, 1984). For inter-subject averages, it is assumed that reliable estimates of the patterns for each subject is available as a result of averaging out stride-to-stride perturbations. This is because stride-to-stride variability of the linear envelope EMG's are in the 20% to 70% range in most gait-related muscles. Most stride-to-stride perturbations can be removed through ensemble averaging of the linear envelope patterns for at least 10 strides.

Inter-subject profiles can only be obtained after we attempt some form of normalization. A recent study (Yang and Winter, 1984) tested four different amplitude normalization techniques and compared them against the unnormalized signal (reported in microvolts). Biomechanical techniques (calibration of each muscle against a maximum voluntary contraction or against a known torque) actually resulted in somewhat greater variability in the inter-subject ensemble averages. Because of the sensitivity of the EMG to the velocity of walking (Yang & Winter, 1985), only natural cadence trials are reported.

5.10 Detailed Profiles on Sixteen Gait Related Muscles

The following pages contain the inter-subject ensemble averages of the EMG profiles of sixteen muscles. Two plots are contained in each figure, the upper being the unnormalized patterns in microvolts and the lower in which each subject's mean EMG was normalized to 100% prior to averaging. Associated with each muscle is a description of the placement of the electrodes and a functional explanation of its EMG pattern over the gait cycle. Bipolar recordings were made with the electrodes placed 2 cm apart. Tables 5.11 through to 5.26 in the Appendix document the mean profiles for these sixteen muscles at 2% intervals over the stride period. TO in all plots is at 60% of the stride period.

5.11 Gluteus Medius

Placement - about 3 cms below the midpoint of the iliac crest.

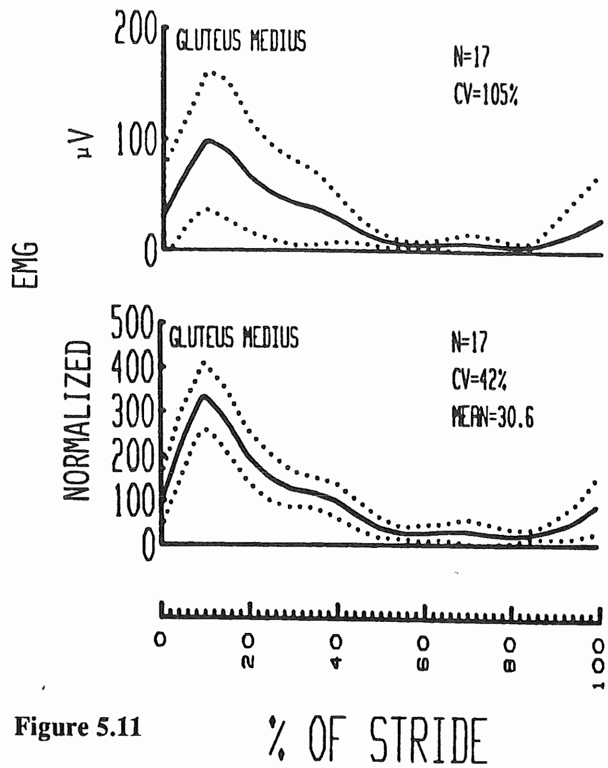

Figure 5.11

The major activity begins late in swing and peaks during weight acceptance (15% stride) and declines during mid stance to near zero by the end of single support (50% stride). Gluteus medius serves three functions. The anterior fibres act mainly as a hip abductor to control the drop of the pelvis during weight acceptance. Secondly, the medial fibres serve as a hip extensor to assist in controlling hip flexion (and thereby controlling knee collapse). Finally, the anterior fibres act as a medial rotator of the thigh but with the thigh in weight bearing, it serves to anteriorly rotate the pelvis and assist in swinging the contralateral limb.

5.12 *Gluteus Maximus*

Placement - over the area of greatest muscle bulk proximal to a line between the greater trochanter and ischial tuberosity.

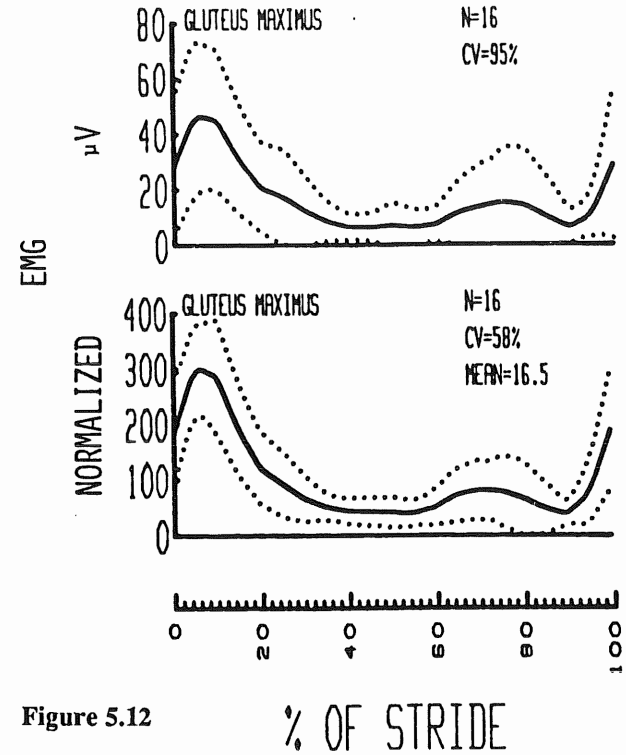

Figure 5.12

The major activity begins late in swing and peaks during weight acceptance (10% stride), decreasing to a low level by the end of mid-stance. A second minor burst of activity occurs during the first half of swing. The gluteus maximus is a hip extensor and acts during weight acceptance to control hip flexion. As such, it assists in controlling forward thigh rotation and, thus, the amount of knee flexion. Also, during weight acceptance, the trunk forward rotation must be controlled and the pelvis must be stabilized so that the erector spinae can act to control forward trunk rotation. The second burst early in swing appears to serve to decelerate the forward swinging thigh and actually reverse the thigh rotation by 85% of stride (see Figure 3.31(a)).

5.13 *Medial Hamstrings (Semitendinosus)*

Placement - midway on a line between the ischial tuberosity and the medial epicondyle of the tibia.

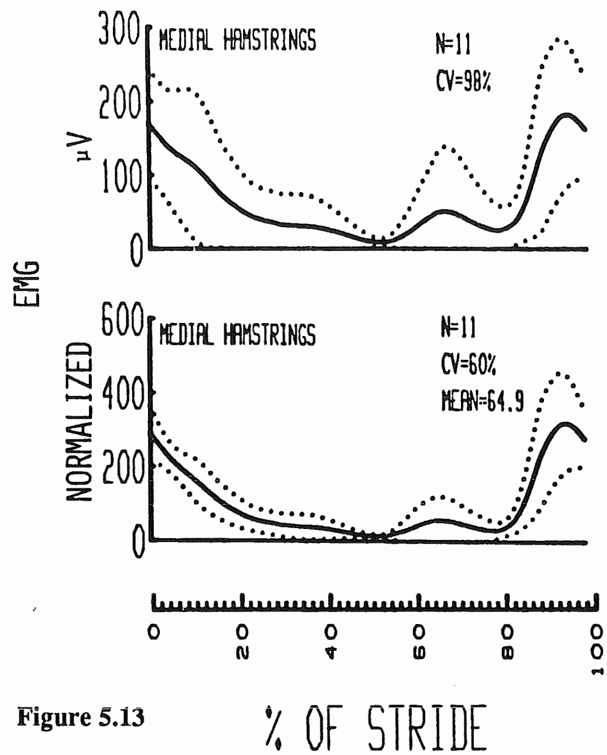

Figure 5.13

The major activity begins in the latter half of swing peaking at 95% stride and continues into weight acceptance. A second minor phase of activity takes place early in swing. The major burst serves to decelerate the swinging lower limb, mainly as a knee flexor to slow down the leg and foot, because the thigh has actually reversed its direction at this time. When heel contact occurs, the medial hamstrings serve as hip extensors to assist the gluteus maximus in controlling the forward rotation of the thigh and stabilizing the pelvis.

5.14 Lateral Hamstrings (Biceps Femoris)

Placement - midway on a line between the ischial tuberosity and the head of the fibula.

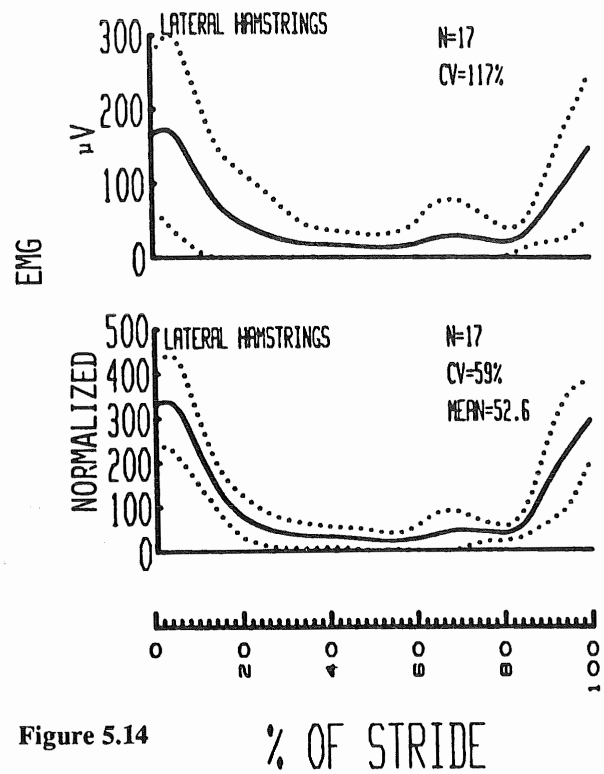

Figure 5.14

The activity pattern of the lateral hamstrings is almost exactly the same as that seen on the medial hamstrings. The only minor difference is the time of peakings of the major peak, which for this muscle is about 4% of stride. The functions of both lateral and medial hamstrings is the same. However, the slight difference in their peak activity suggests that the role of the lateral hamstrings in decelerating the leg is slightly less important and its role as a hip extensor during weight acceptance is slightly more important.

5.15 Erector Spinae

Placement - two cms lateral to the spinous process at the level of the iliac crest.

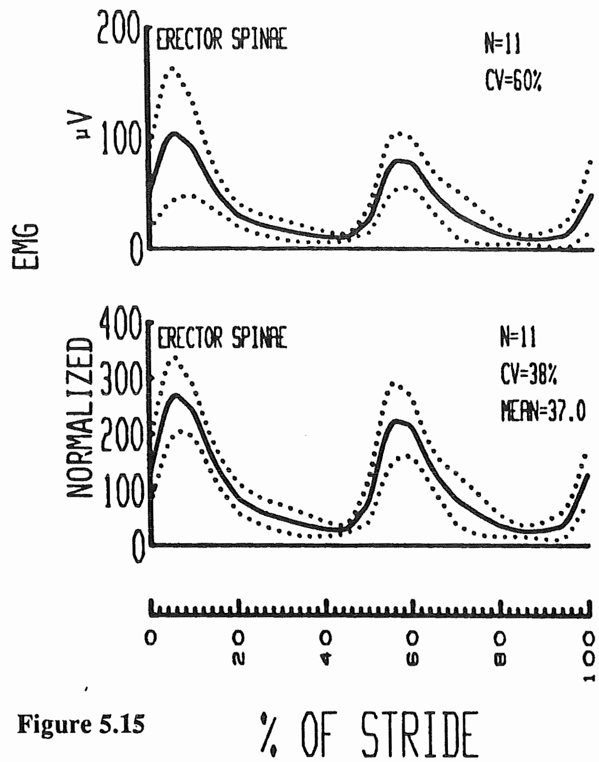

Figure 5.15

Two distinct peaks of activity are evident, the higher at 10% and one slightly smaller at 60%. Both bursts serve the same role, that of controlling the forward rotation of the trunk as each limb accepts weight. The slightly higher peak is associated with weight acceptance of the ipsilateral limb, the second peak with the contralateral limb.

5.16 Sartorius

Placement - eight cms distal to the ASIS along a line to the medial epicondyle of the tibia.

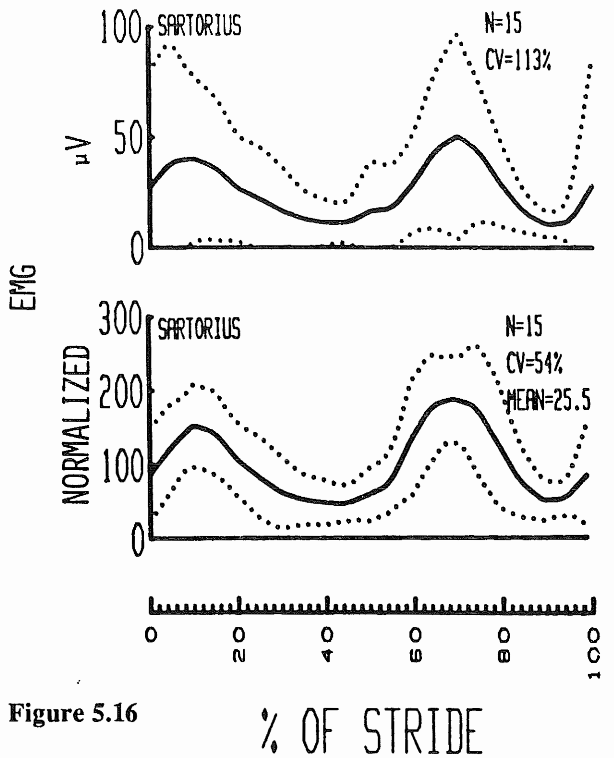

Figure 5.16

Two equal peaks of activity are evident in this long biarticulate muscle. The first peak occurs during weight acceptance (15% stride). Its role at this point in time is not completely clear. It may be acting as a hip flexor as a stabilizing co-contraction to all the hip extensors that are on at this time (hamstrings, gluteus maximus and medius). With the knee flexed about 20° during weight bearing, it may be acting at the distal end as a lateral rotator of the thigh, but as Basmajian has suggested, again as a stabilizer. The second burst occurs early in swing and assists in a minor way as a hip flexor to help swing the lower limb.

5.17 Rectus Femoris

Placement - midway between the ASIS and the superior border of the patella.

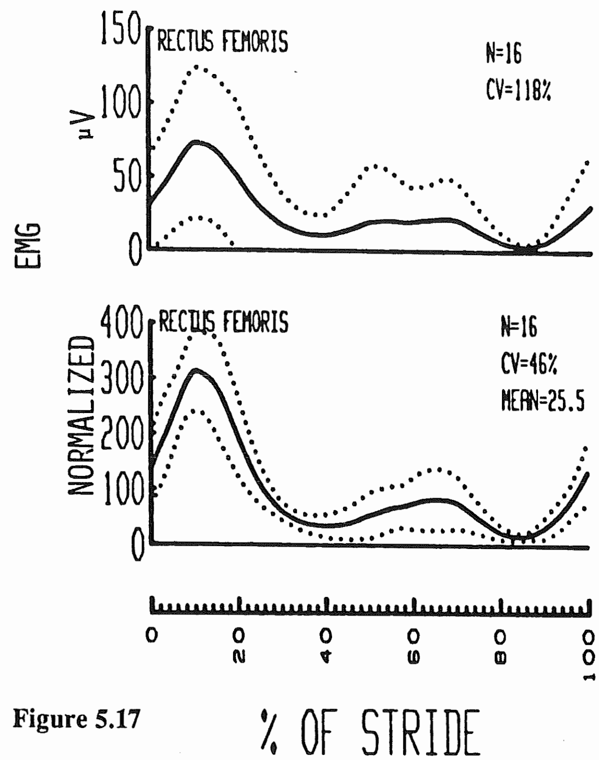

Figure 5.17

One major and one minor burst of activity are present. The major activity begins before HC to extend the leg and foot just prior to HC and continues to a maximum during weight acceptance (10% stride) when it acts as a knee extensor to control knee flexion and causes the knee to extend in mid stance. The second minor activity peaks just after TO, and this has two simultaneous functions: hip flexion to pull the swinging limb forward, and knee extension to decelerate the backward swinging leg and foot.

5.18 Vastus Lateralis

Placement - over the area of greatest muscle bulk just lateral of the rectus femoris on the distal half of the thigh.

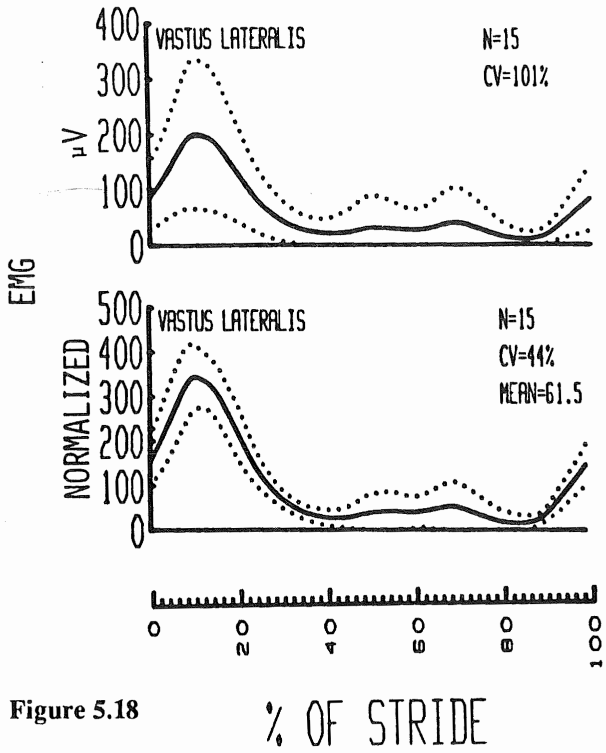

Figure 5.18

One major peak of activity at weight acceptance (10% stride) serves to control the amount of knee flexion and to assist in the extension of the knee in mid stance. On some subjects, there is a second minor phase of activity just after TO in order to assist the rectus femoris in arresting the backward swinging leg and foot.

5.19 Adductor Longus

Placement - eight cms below the pubic tubercle along a line to the tendon of insertion.

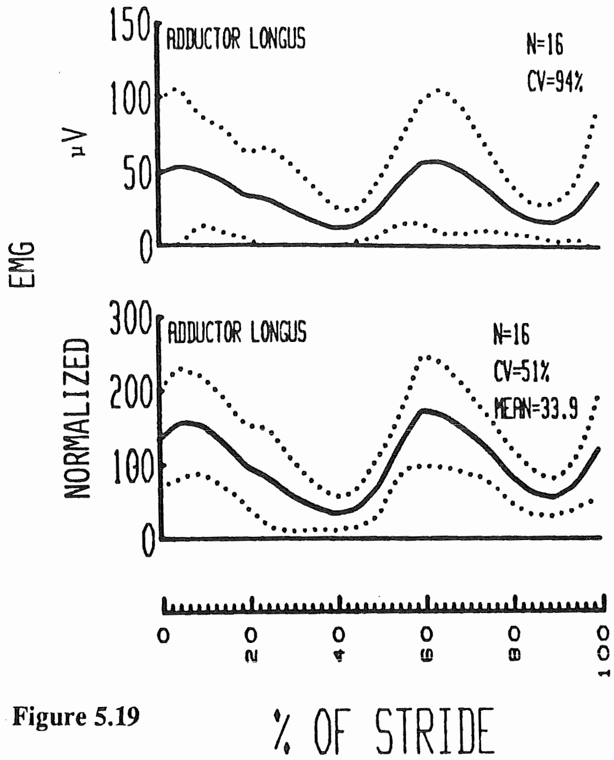

Figure 5.19

Two approximately equal peaks of activity occur, the first at 10% stride, the second just after TO. The first appears to be entirely as a co-contraction to the hip abductors and hip extensors which are dominant during weight acceptance. The activity early in swing is mainly hip flexor activity to assist the rectus femoris and iliopsoas to accelerate the thigh forward.

5.20 Adductor Magnus

Placement - centered on the proximal one third of the thigh between the gracilis and the medial hamstrings.

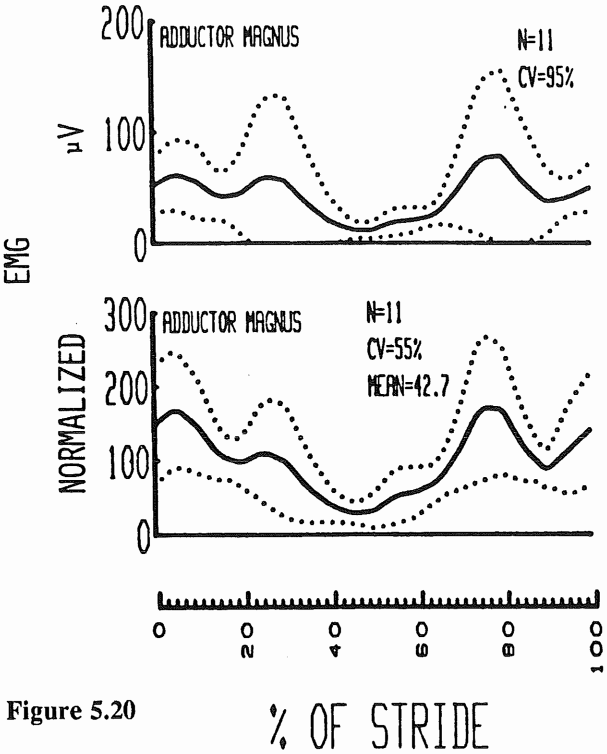

Figure 5.20

This muscle has very similar activity to that seen in adductor longus except the second peak is about 15% later (mid stance). During stance, its functions appear to be as a co-contraction to the hip abductors and hip extensors. During swing, its activity appears to be that of hip flexion plus the additional role of medial rotation and adduction of the thigh in order to keep the swinging limb towards the centre line of progression.

5.21 Tibialis Anterior

Placement - over the area of greatest muscle bulk just lateral to the crest of the tibia on the proximal half of the leg.

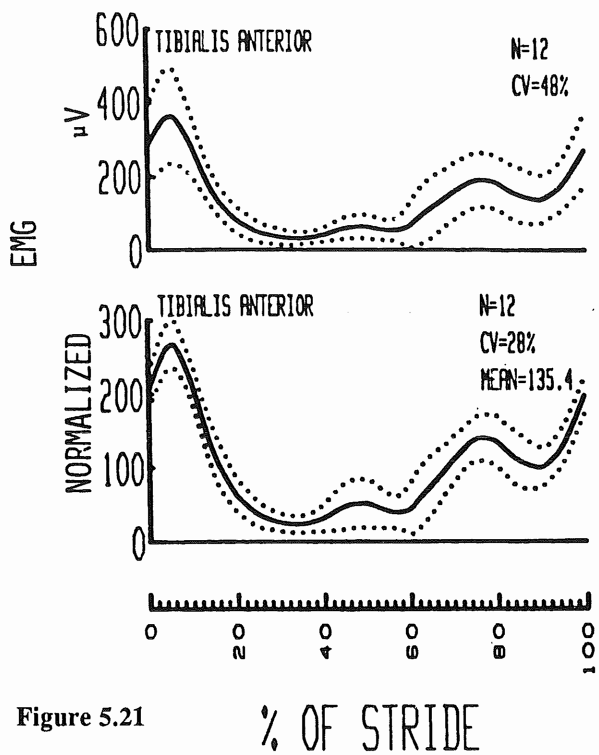

Figure 5.21

This ankle muscle commences its major activity at the end of swing to keep the foot dorsiflexed during reach phase. Immediately after HC, it peaks and this generates forces to lower the foot to the ground in opposition to the plantarflexing ground reaction forces. Shortly after FF, its activity decreases and, in some subjects, it plays a minor role in pulling the leg forward over the foot. The second burst of activity commences at TO and results in dorsiflexion of the foot for foot clearance during mid swing.

5.22 Extensor Digitorum Longus

Placement - on the distal half of the leg just lateral to the tibialis anterior.

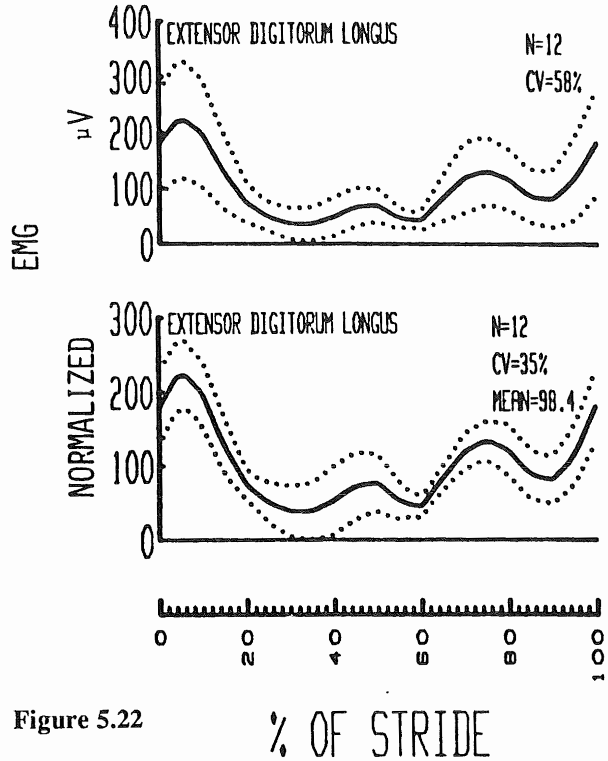

Figure 5.22

This muscle has almost identical activity as the tibialis anterior and its functions in lowering the foot after HC and in dorsiflexing the toes (and foot) for clearance during swing. A minor third phase is seen during push-off and this appears to be a co-contraction to stabilize the ankle joint during this phase of rapid energy generation.

5.23 Medial Gastrocnemius

Placement - over the area of greatest muscle bulk on the medial calf.

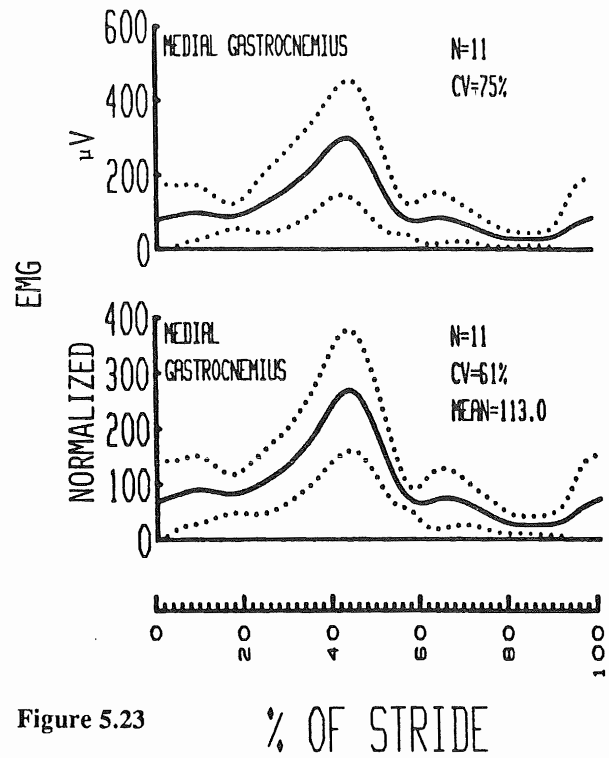

Figure 5.23

One major long duration phase of activity is evident and it begins just prior to HC and rises during stance reaching a peak at mid push-off (50% stride). At 5% stride we have FF, and until 40% stride this muscle lengthens as the leg rotates forward under its control. Then, during push-off, the muscle shortens along with the other plantar flexors to cause the foot to actively plantar flex and generate the most important impulse of energy. Activity drops rapidly until TO where low level activity remains into swing, presumably as a knee flexor to cause adequate knee flexion prior to swing through. It will be co-contracting against the rectus femoris and vastii muscles at this time.

5.24 Lateral Gastrocnemius

Placement - over the area of greatest muscle bulk on the lateral calf.

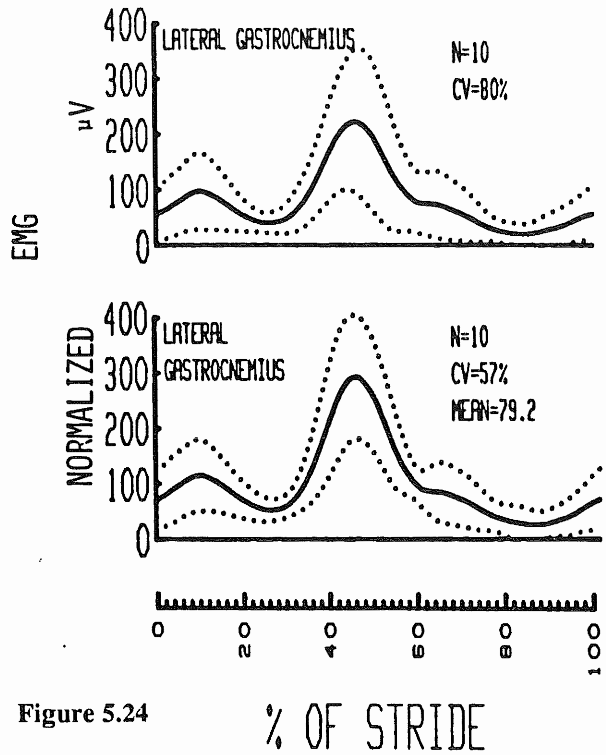

Figure 5.24

Almost identical activity is seen here as was recorded on the medial gastrocnemius and the function of this muscle is identical.

5.25 Soleus

Placement - proximal electrode is placed one cm. distal to the medial head of the gastrocnemius

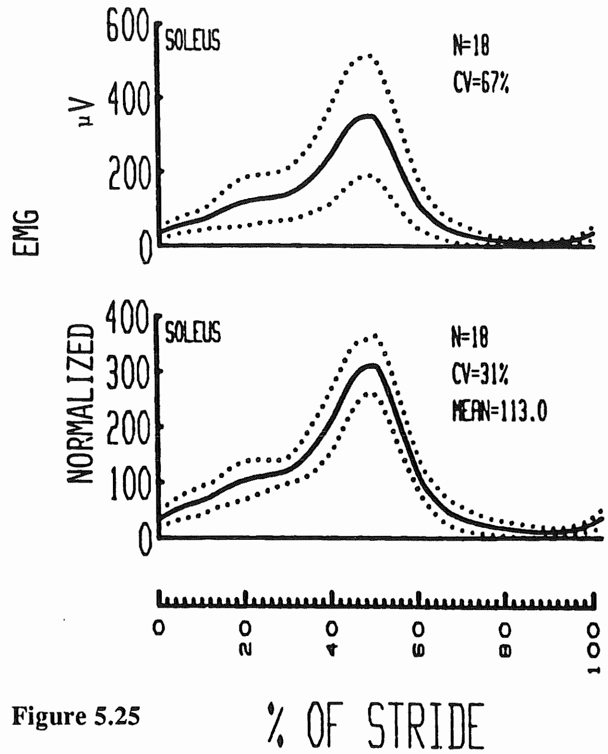

Figure 5.25

As a plantar flexor, the soleus is active during stance, initially to control the forward rotating leg and finally, between 40% and 60% of the stride period, to generate an explosive push-off.

5.26 Peroneus Longus

Placement - midway along the line between the head of the fibula and the lateral malleolus.

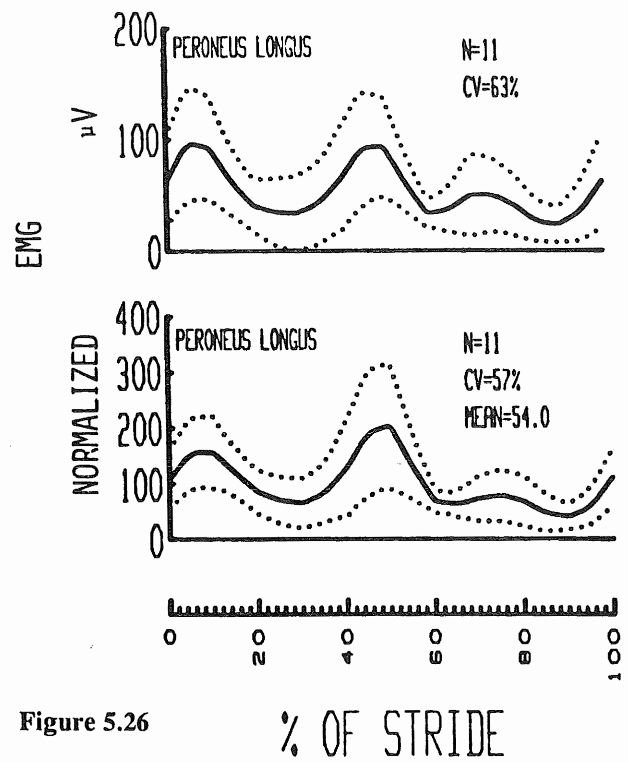

Figure 5.26

Two bursts of activity occur. The smaller occurs during weight acceptance (10% stride), the larger is during push-off (50% stride). The first burst appears to stabilize the ankle against foot inversion (possibly as a co-contraction to the tibialis anterior), and the remainder of the activity through to and including push-off is as a plantar flexor. During early swing, there is some low level activity that is likely a co-contraction to tibialis anterior to control the amount of foot dorsiflexion and inversion.

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Table 3.32(a)

JOINT ANGLES(Deg)-SLOW CADENCE (N=19)

	HIP		KNEE		ANKLE	
% STRIDE	Mean	Std.Dev.	Mean	Std.Dev.	Mean	Std.Dev.
0	15.73	7.01	3.74	5.06	-.57	3.37
2	14.69	6.69	5.96	5.20	-2.83	3.67
4	13.52	6.43	8.33	5.44	-5.27	3.95
6	12.49	6.35	10.88	5.79	-6.46	3.96
8	11.60	6.45	13.39	6.16	-5.91	3.93
10	10.55	6.64	15.31	6.48	-4.18	3.77
12	9.14	6.84	16.20	6.73	-2.19	3.60
14	7.54	6.99	16.20	6.87	-.46	3.46
16	6.06	7.01	15.75	6.81	.96	3.33
18	4.77	6.91	15.07	6.54	2.13	3.21
20	3.50	6.76	14.16	6.14	3.07	3.01
22	2.19	6.64	13.10	5.73	3.79	2.80
24	.94	6.55	12.04	5.34	4.35	2.66
26	-.16	6.49	11.10	4.99	4.84	2.62
28	-1.20	6.49	10.28	4.73	5.34	2.66
30	-2.27	6.54	9.54	4.60	5.84	2.76
32	-3.34	6.64	8.93	4.61	6.29	2.89
34	-4.35	6.80	8.47	4.70	6.71	3.01
36	-5.24	7.00	8.23	4.84	7.16	3.11
38	-6.07	7.18	8.21	4.97	7.66	3.27
40	-6.89	7.26	8.36	5.02	8.11	3.47
42	-7.69	7.32	8.71	5.01	8.43	3.57
44	-8.43	7.43	9.33	5.04	8.56	3.60
46	-9.12	7.55	10.25	5.09	8.46	3.60
48	-9.77	7.63	11.53	5.08	8.03	3.63
50	-10.39	7.69	13.21	5.03	7.14	3.76
52	-10.93	7.75	15.47	5.02	5.55	4.01
54	-11.27	7.76	18.54	5.17	3.06	4.49
56	-11.05	7.66	22.74	5.51	-.34	5.04
58	-9.95	7.41	28.32	5.95	-4.47	5.41
60	-7.96	7.11	35.05	6.36	-8.91	5.29
62	-5.27	6.84	42.28	6.49	-13.12	4.64
64	-2.23	6.66	49.12	6.33	-16.27	4.22
66	.86	6.44	54.89	5.94	-17.55	4.74
68	3.80	6.10	59.14	5.55	-16.64	5.55
70	6.45	5.70	61.71	5.60	-14.00	5.78
72	8.84	5.44	62.55	6.14	-10.48	5.61
74	11.04	5.31	61.77	6.95	-6.91	5.47
76	13.06	5.25	59.52	7.81	-3.81	5.36
78	14.80	5.29	55.96	8.62	-1.34	5.08
80	16.20	5.33	51.26	9.21	.51	4.64
82	17.27	5.24	45.57	9.54	1.59	4.28
84	18.07	5.15	39.10	9.63	1.72	4.03
86	18.53	5.21	31.99	9.46	1.11	3.91
88	18.55	5.38	24.44	8.94	.25	3.93
90	18.17	5.54	16.91	8.02	-.34	4.03
92	17.55	5.68	10.13	6.68	-.45	4.09
94	16.88	5.90	4.98	5.11	-.19	4.01
96	16.30	6.23	2.11	3.72	.14	3.77
98	15.90	6.51	1.73	3.20	.04	3.38
100	15.53	6.70	3.21	3.44	-.89	3.13

Table 3.32(b)

JOINT ANGLES(Deg)-NATURAL CADENCE (N=19)

	HIP		KNEE		ANKLE	
% STRIDE	Mean	Std.Dev.	Mean	Std.Dev.	Mean	Std.Dev.
0	19.33	5.64	3.97	4.19	.02	3.93
2	18.92	5.79	7.00	4.58	-2.06	4.36
4	18.45	5.77	10.52	4.79	-3.88	4.32
6	17.94	5.64	14.12	4.98	-4.60	4.02
8	17.30	5.56	17.38	5.11	-3.98	3.95
10	16.40	5.63	19.84	5.12	-2.40	4.05
12	15.18	5.81	21.27	5.17	-.45	4.01
14	13.67	5.94	21.67	5.34	1.45	3.81
16	11.97	5.90	21.22	5.61	3.04	3.56
18	10.21	5.73	20.20	5.85	4.27	3.36
20	8.48	5.57	18.86	5.95	5.13	3.24
22	6.74	5.47	17.35	5.85	5.71	3.19
24	4.94	5.38	15.73	5.57	6.10	3.18
26	3.13	5.36	14.08	5.18	6.43	3.18
28	1.42	5.51	12.50	4.84	6.76	3.21
30	-.13	5.75	11.09	4.69	7.12	3.30
32	-1.54	5.96	9.91	4.64	7.54	3.44
34	-2.87	6.14	8.97	4.66	7.99	3.60
36	-4.12	6.34	8.28	4.74	8.44	3.79
38	-5.30	6.58	7.86	4.86	8.86	4.00
40	-6.40	6.86	7.72	4.95	9.23	4.25
42	-7.43	7.14	7.94	4.98	9.51	4.51
44	-8.39	7.40	8.60	4.97	9.62	4.75
46	-9.27	7.68	9.76	4.96	9.43	4.98
48	-10.02	7.97	11.50	4.97	8.70	5.26
50	-10.61	8.25	13.86	5.05	7.20	5.65
52	-10.95	8.51	16.97	5.22	4.69	6.12
54	-10.91	8.71	20.96	5.49	1.15	6.56
56	-10.31	8.81	26.00	5.86	-3.26	6.87
58	-9.00	8.72	32.03	6.20	-8.17	6.93
60	-6.95	8.39	38.74	6.30	-13.05	6.64
62	-4.25	7.84	45.60	6.05	-17.13	6.19
64	-1.05	7.17	52.05	5.53	-19.52	5.91
66	2.42	6.47	57.54	4.99	-19.77	5.81
68	5.93	5.80	61.66	4.75	-18.12	5.57
70	9.22	5.22	64.12	4.93	-15.29	5.06
72	12.11	4.75	64.86	5.41	-12.04	4.45
74	14.55	4.44	63.95	5.99	-8.85	3.99
76	16.53	4.35	61.59	6.51	-5.96	3.76
78	18.13	4.43	57.97	6.93	-3.51	3.63
80	19.45	4.59	53.27	7.22	-1.64	3.49
82	20.54	4.76	47.58	7.42	-.50	3.27
84	21.38	4.84	40.94	7.55	-.07	2.94
86	21.84	4.83	33.46	7.54	-.16	2.65
88	21.87	4.75	25.38	7.29	-.42	2.71
90	21.50	4.68	17.27	6.69	-.52	3.09
92	20.84	4.72	9.94	5.73	-.26	3.45
94	20.09	4.84	4.31	4.54	.36	3.53
96	19.50	5.02	1.12	3.55	1.00	3.40
98	19.18	5.23	.54	3.28	1.20	3.33
100	19.01	5.43	2.21	3.60	.58	3.52

Table 3.32(c)

JOINT ANGLES(Deg)-FAST CADENCE (N=17)

	HIP		KNEE		ANKLE	
% STRIDE	Mean	Std.Dev.	Mean	Std.Dev.	Mean	Std.Dev.
0	18.06	5.65	5.91	4.09	1.56	2.71
2	17.58	5.75	9.63	3.83	-.48	2.76
4	17.06	5.89	13.42	4.26	-2.24	2.86
6	16.62	6.03	17.30	4.60	-2.94	3.08
8	16.21	6.12	20.82	4.76	-2.32	3.45
10	15.62	6.18	23.51	4.83	-.74	3.73
12	14.58	6.18	25.00	4.87	1.28	3.77
14	13.02	6.09	25.24	4.94	3.25	3.65
16	11.05	5.94	24.48	5.01	4.92	3.49
18	8.87	5.81	22.99	5.06	6.13	3.35
20	6.66	5.73	21.07	5.08	6.92	3.24
22	4.51	5.71	18.93	5.06	7.35	3.16
24	2.47	5.73	16.72	5.00	7.47	3.14
26	.55	5.79	14.56	4.89	7.36	3.18
28	-1.25	5.91	12.56	4.73	7.16	3.26
30	-2.94	6.10	10.77	4.51	7.02	3.38
32	-4.56	6.30	9.23	4.26	7.04	3.53
34	-6.12	6.49	7.95	4.02	7.22	3.68
36	-7.61	6.67	6.98	3.85	7.49	3.80
38	-8.99	6.82	6.36	3.78	7.74	3.86
40	-10.20	6.94	6.18	3.82	7.91	3.91
42	-11.22	7.04	6.56	3.95	7.95	4.00
44	-12.04	7.14	7.58	4.15	7.74	4.19
46	-12.66	7.23	9.33	4.37	7.11	4.45
48	-13.05	7.32	11.88	4.58	5.80	4.76
50	-13.18	7.42	15.31	4.78	3.53	5.17
52	-12.94	7.56	19.73	5.01	.07	5.69
54	-12.24	7.70	25.14	5.26	-4.43	6.15
56	-10.95	7.79	31.48	5.51	-9.48	6.25
58	-9.05	7.82	38.48	5.73	-14.28	5.83
60	-6.58	7.76	45.63	5.84	-17.99	5.21
62	-3.62	7.63	52.37	5.75	-19.94	5.11
64	-.30	7.46	58.15	5.42	-19.95	5.73
66	3.20	7.23	62.58	4.90	-18.28	6.45
68	6.67	6.92	65.39	4.36	-15.49	6.74
70	9.89	6.58	66.52	4.08	-12.26	6.59
72	12.65	6.27	66.05	4.25	-9.09	6.16
74	14.87	5.96	64.09	4.75	-6.25	5.58
76	16.60	5.66	60.84	5.40	-3.84	4.96
78	17.92	5.37	56.46	6.07	-1.92	4.36
80	18.95	5.13	51.08	6.71	-.52	3.85
82	19.69	5.01	44.79	7.30	.38	3.49
84	20.06	5.01	37.64	7.76	.82	3.31
86	19.99	5.12	29.81	7.91	.97	3.28
88	19.49	5.30	21.74	7.61	1.11	3.32
90	18.70	5.49	14.14	6.79	1.44	3.41
92	17.86	5.63	7.86	5.54	1.95	3.46
94	17.18	5.73	3.65	4.16	2.43	3.42
96	16.76	5.80	2.02	3.25	2.52	3.27
98	16.55	5.83	2.80	3.34	1.93	3.13
100	16.46	5.84	5.47	4.04	.56	3.08

Table 4.12(a)

GROUND REACTION FORCES(N/Kg)
SLOW CADENCE (N=19)

% STRIDE	VERTICAL Mean	VERTICAL Std.Dev.	HORIZONTAL Mean	HORIZONTAL Std.Dev.
0	.58	.71	.06	.18
2	2.60	1.31	-.22	.42
4	4.69	1.28	-.85	.41
6	6.62	1.16	-1.28	.42
8	8.01	1.12	-1.47	.49
10	9.03	1.23	-1.52	.53
12	9.67	1.14	-1.43	.47
14	9.81	.95	-1.26	.40
16	9.61	.78	-1.08	.34
18	9.23	.67	-.91	.28
20	8.87	.58	-.75	.26
22	8.64	.60	-.64	.25
24	8.52	.69	-.56	.25
26	8.50	.81	-.47	.31
28	8.53	.93	-.36	.39
30	8.62	1.00	-.25	.42
32	8.71	.99	-.21	.30
34	8.80	.90	-.11	.28
36	8.92	.76	.02	.29
38	9.10	.65	.16	.31
40	9.33	.62	.33	.34
42	9.59	.65	.53	.40
44	9.85	.64	.76	.46
46	10.01	.59	.99	.52
48	9.99	.58	1.22	.57
50	9.63	.77	1.44	.57
52	8.74	1.20	1.59	.54
54	7.14	1.75	1.53	.50
56	5.00	2.00	1.23	.58
58	2.97	1.64	.80	.53
60	1.42	1.04	.36	.32
62	.56	.56	.11	.15
64	.14	.25	.02	.05
66	.03	.09	.00	.01
68	.00	.02	.00	.00
70	0.00	0.00	0.00	0.00
72	0.00	0.00	0.00	0.00
74	0.00	0.00	0.00	0.00
76	0.00	0.00	0.00	0.00
78	0.00	0.00	0.00	0.00
80	0.00	0.00	0.00	0.00
82	0.00	0.00	0.00	0.00
84	0.00	0.00	0.00	0.00
86	0.00	0.00	0.00	0.00
88	0.00	0.00	0.00	0.00
90	0.00	0.00	0.00	0.00
92	0.00	0.00	0.00	0.00
94	0.00	0.00	0.00	0.00
96	0.00	0.00	0.00	0.00
98	0.00	0.00	0.00	0.00
100	0.00	0.00	0.00	0.00

Table 4.12(b)

GROUND REACTION FORCES(N/Kg)
NATURAL CADENCE (N=19)

% STRIDE	VERTICAL Mean	VERTICAL Std.Dev.	HORIZONTAL Mean	HORIZONTAL Std.Dev.
0	.64	.57	.03	.14
2	2.92	1.31	-.16	.35
4	5.35	1.49	-.82	.53
6	7.31	1.46	-1.42	.47
8	8.69	1.28	-1.75	.42
10	9.76	1.07	-1.92	.45
12	10.51	.87	-1.90	.44
14	10.82	.74	-1.76	.40
16	10.57	.74	-1.52	.37
18	9.95	.78	-1.28	.35
20	9.14	.75	-1.06	.33
22	8.31	.66	-.86	.31
24	7.70	.61	-.69	.27
26	7.29	.70	-.55	.24
28	7.10	.85	-.42	.21
30	7.07	.98	-.31	.20
32	7.19	1.08	-.20	.19
34	7.45	1.13	-.07	.19
36	7.84	1.13	.08	.20
38	8.37	1.11	.28	.24
40	8.97	1.07	.52	.30
42	9.61	1.03	.79	.35
44	10.20	.97	1.11	.39
46	10.62	.94	1.44	.43
48	10.63	1.07	1.77	.42
50	10.13	1.34	2.05	.36
52	8.97	1.64	2.19	.29
54	6.92	1.81	2.02	.42
56	4.55	1.65	1.50	.51
58	2.42	1.16	.81	.41
60	1.03	.60	.29	.21
62	.33	.24	.07	.09
64	.06	.06	.00	.03
66	.00	.01	-.00	.00
68	0.00	0.00	0.00	0.00
70	0.00	0.00	0.00	0.00
72	0.00	0.00	0.00	0.00
74	0.00	0.00	0.00	0.00
76	0.00	0.00	0.00	0.00
78	0.00	0.00	0.00	0.00
80	0.00	0.00	0.00	0.00
82	0.00	0.00	0.00	0.00
84	0.00	0.00	0.00	0.00
86	0.00	0.00	0.00	0.00
88	0.00	0.00	0.00	0.00
90	0.00	0.00	0.00	0.00
92	0.00	0.00	0.00	0.00
94	0.00	0.00	0.00	0.00
96	0.00	0.00	0.00	0.00
98	0.00	0.00	0.00	0.00
100	0.00	0.00	0.00	0.00

Table 4.12(c)

GROUND REACTION FORCES(N/Kg)
FAST CADENCE (N=17)

% STRIDE	VERTICAL Mean	VERTICAL Std.Dev.	HORIZONTAL Mean	HORIZONTAL Std.Dev.
0	1.06	1.22	.04	.33
2	4.12	1.78	-.34	.57
4	7.66	1.77	-1.46	.67
6	9.87	1.66	-2.23	.65
8	11.16	1.66	-2.63	.68
10	12.11	1.57	-2.80	.64
12	12.67	1.36	-2.67	.51
14	12.56	1.08	-2.32	.41
16	11.75	.88	-1.92	.36
18	10.49	.90	-1.53	.33
20	9.04	.95	-1.19	.33
22	7.66	.96	-.90	.33
24	6.57	1.02	-.68	.33
26	5.77	1.11	-.51	.32
28	5.34	1.22	-.40	.31
30	5.20	1.27	-.31	.30
32	5.34	1.26	-.22	.28
34	5.78	1.18	-.11	.26
36	6.50	1.03	.03	.26
38	7.55	.88	.23	.32
40	8.76	.88	.52	.43
42	10.02	.99	.93	.56
44	11.07	1.05	1.39	.69
46	11.71	.99	1.86	.80
48	11.70	.74	2.29	.85
50	10.81	.70	2.57	.79
52	8.83	1.43	2.52	.66
54	6.26	2.05	2.03	.68
56	3.71	2.12	1.26	.76
58	1.90	1.70	.62	.62
60	.77	1.11	.24	.33
62	.26	.59	.07	.17
64	.08	.25	.01	.04
66	.00	.00	-.00	.00
68	0.00	0.00	0.00	0.00
70	0.00	0.00	0.00	0.00
72	0.00	0.00	0.00	0.00
74	0.00	0.00	0.00	0.00
76	0.00	0.00	0.00	0.00
78	0.00	0.00	0.00	0.00
80	0.00	0.00	0.00	0.00
82	0.00	0.00	0.00	0.00
84	0.00	0.00	0.00	0.00
86	0.00	0.00	0.00	0.00
88	0.00	0.00	0.00	0.00
90	0.00	0.00	0.00	0.00
92	0.00	0.00	0.00	0.00
94	0.00	0.00	0.00	0.00
96	0.00	0.00	0.00	0.00
98	0.00	0.00	0.00	0.00
100	0.00	0.00	0.00	0.00

Table 4.24(a)

JOINT MOMENT OF FORCE(N.M/Kg)-SLOW CADENCE (N=19)

	SUPPORT		HIP		KNEE		ANKLE	
% STRIDE	Mean	Std.Dev.	Mean	Std.Dev.	Mean	Std.Dev.	Mean	Std.Dev.
0	.018	.132	.179	.224	-.159	.109	-.002	.020
2	.187	.188	.385	.255	-.176	.156	-.022	.064
4	.216	.247	.252	.220	.018	.158	-.054	.110
6	.342	.283	.208	.266	.170	.201	-.036	.148
8	.512	.300	.190	.282	.290	.263	.031	.174
10	.661	.332	.169	.307	.379	.312	.113	.198
12	.742	.365	.125	.345	.413	.342	.204	.222
14	.756	.389	.070	.362	.388	.361	.299	.244
16	.741	.397	.025	.348	.326	.361	.390	.264
18	.739	.396	.020	.333	.237	.343	.482	.288
20	.750	.397	.032	.341	.140	.331	.578	.321
22	.744	.383	.013	.321	.063	.321	.668	.343
24	.736	.366	-.023	.283	.002	.304	.756	.361
26	.735	.342	-.055	.259	-.048	.285	.838	.367
28	.754	.322	-.065	.248	-.096	.277	.915	.376
30	.794	.309	-.059	.247	-.137	.259	.990	.365
32	.824	.336	-.087	.284	-.143	.269	1.054	.359
34	.836	.318	-.118	.279	-.162	.260	1.116	.340
36	.857	.306	-.137	.293	-.184	.265	1.179	.322
38	.901	.315	-.144	.328	-.202	.283	1.246	.306
40	.947	.325	-.153	.341	-.219	.294	1.319	.292
42	.993	.330	-.171	.331	-.234	.299	1.398	.284
44	1.051	.340	-.186	.333	-.233	.311	1.470	.281
46	1.110	.350	-.197	.358	-.214	.320	1.521	.269
48	1.142	.362	-.224	.416	-.168	.330	1.535	.243
50	1.136	.346	-.244	.439	-.116	.323	1.497	.222
52	1.063	.320	-.251	.427	-.054	.307	1.369	.238
54	.898	.307	-.231	.374	.013	.276	1.116	.295
56	.635	.335	-.200	.291	.066	.228	.768	.334
58	.345	.314	-.178	.214	.084	.160	.438	.282
60	.078	.208	-.198	.147	.096	.117	.179	.176
62	-.052	.114	-.169	.109	.071	.074	.045	.088
64	-.091	.061	-.124	.069	.039	.037	-.006	.034
66	-.087	.082	-.104	.091	.032	.031	-.015	.009
68	-.082	.067	-.099	.078	.031	.034	-.014	.005
70	-.082	.043	-.097	.060	.027	.033	-.012	.003
72	-.081	.028	-.087	.035	.015	.026	-.010	.002
74	-.082	.029	-.072	.039	-.002	.027	-.009	.002
76	-.082	.050	-.055	.051	-.018	.032	-.008	.002
78	-.073	.048	-.031	.051	-.034	.033	-.009	.003
80	-.067	.038	-.011	.051	-.047	.032	-.009	.003
82	-.074	.044	-.010	.057	-.054	.030	-.009	.003
84	-.079	.049	-.007	.059	-.062	.025	-.010	.003
86	-.065	.039	.020	.052	-.074	.027	-.011	.004
88	-.054	.036	.040	.056	-.084	.034	-.011	.003
90	-.044	.045	.066	.081	-.102	.050	-.007	.003
92	-.026	.049	.109	.096	-.132	.065	-.003	.003
94	-.012	.041	.145	.097	-.158	.073	.001	.004
96	-.014	.052	.143	.109	-.161	.079	.004	.004
98	-.029	.046	.098	.108	-.130	.079	.003	.004
100	-.054	.051	.020	.094	-.073	.061	-.001	.005

Table 4.24(b)

JOINT MOMENT OF FORCE(N.M/Kg)-NATURAL CADENCE (N=19)

	SUPPORT		HIP		KNEE		ANKLE	
% STRIDE	Mean	Std.Dev.	Mean	Std.Dev.	Mean	Std.Dev.	Mean	Std.Dev.
0	.044	.123	.249	.222	-.196	.126	-.009	.041
2	.285	.164	.600	.317	-.281	.223	-.034	.087
4	.402	.194	.556	.334	-.090	.272	-.064	.103
6	.538	.290	.416	.402	.173	.291	-.051	.137
8	.749	.354	.359	.497	.362	.328	.028	.184
10	.956	.427	.305	.555	.508	.380	.143	.262
12	1.097	.447	.245	.541	.593	.412	.260	.313
14	1.142	.441	.159	.491	.615	.428	.368	.345
16	1.110	.424	.084	.438	.556	.430	.469	.360
18	1.013	.376	-.000	.364	.469	.384	.545	.328
20	.899	.335	-.064	.310	.362	.320	.601	.293
22	.802	.300	-.092	.291	.244	.271	.650	.256
24	.735	.272	-.098	.281	.141	.244	.692	.229
26	.697	.250	-.092	.266	.052	.232	.736	.218
28	.677	.242	-.085	.267	-.019	.229	.780	.214
30	.666	.239	-.088	.264	-.070	.230	.825	.218
32	.667	.250	-.100	.272	-.114	.239	.881	.225
34	.672	.262	-.130	.272	-.149	.254	.951	.230
36	.689	.283	-.168	.287	-.181	.275	1.037	.234
38	.728	.297	-.199	.312	-.217	.301	1.144	.238
40	.782	.309	-.231	.349	-.247	.326	1.260	.239
42	.851	.319	-.269	.376	-.269	.342	1.388	.243
44	.930	.328	-.312	.384	-.270	.343	1.513	.247
46	1.006	.336	-.364	.385	-.237	.347	1.608	.255
48	1.056	.333	-.401	.394	-.171	.359	1.628	.254
50	1.073	.347	-.404	.413	-.087	.367	1.565	.265
52	1.028	.381	-.356	.389	-.004	.340	1.388	.294
54	.865	.408	-.262	.335	.054	.288	1.073	.307
56	.555	.381	-.251	.265	.116	.214	.690	.276
58	.182	.264	-.310	.168	.157	.137	.335	.192
60	-.086	.140	-.344	.137	.156	.091	.102	.092
62	-.183	.095	-.295	.125	.114	.061	-.001	.031
64	-.176	.103	-.228	.108	.080	.034	-.028	.012
66	-.127	.116	-.169	.122	.066	.031	-.023	.007
68	-.081	.073	-.126	.082	.064	.032	-.019	.006
70	-.052	.059	-.089	.066	.053	.031	-.015	.004
72	-.044	.078	-.069	.087	.037	.031	-.012	.003
74	-.047	.052	-.057	.063	.020	.030	-.010	.003
76	-.049	.035	-.044	.051	.004	.031	-.010	.002
78	-.044	.038	-.026	.053	-.009	.031	-.010	.003
80	-.042	.046	-.009	.050	-.023	.029	-.011	.003
82	-.044	.051	.008	.047	-.040	.025	-.012	.003
84	-.044	.048	.029	.053	-.059	.026	-.013	.003
86	-.036	.047	.060	.069	-.082	.032	-.013	.003
88	-.018	.056	.106	.085	-.114	.037	-.011	.003
90	.006	.068	.170	.105	-.158	.048	-.006	.005
92	.033	.076	.242	.122	-.211	.061	.001	.006
94	.050	.082	.296	.139	-.253	.073	.007	.005
96	.049	.081	.301	.144	-.263	.077	.011	.005
98	.022	.079	.237	.139	-.224	.072	.010	.005
100	-.025	.078	.118	.123	-.147	.058	.004	.005

Table 4.24(c)

JOINT MOMENT OF FORCE(N.M/Kg)-FAST CADENCE (N=17)

	SUPPORT		HIP		KNEE		ANKLE	
% STRIDE	Mean	Std.Dev.	Mean	Std.Dev.	Mean	Std.Dev.	Mean	Std.Dev.
0	.190	.157	+.515	.312	-.318	.178	-.007	.049
2	.494	.207	+.914	.392	-.373	.272	-.047	.102
4	.621	.187	+.716	.334	.035	.339	-.131	.166
6	.784	.247	+.447	.340	.450	.424	-.113	.241
8	.964	.306	+.206	.390	.798	.510	-.039	.297
10	1.135	.356	+.006	.420	1.057	.563	+.072	.332
12	1.263	.375	-.088	.381	1.144	.530	+.207	.345
14	1.286	.385	-.137	.340	1.083	.487	+.340	.375
16	1.181	.353	-.207	.271	.947	.407	+.441	.361
18	.998	.323	-.280	.221	.774	.307	+.505	.314
20	.812	.299	-.314	.211	.578	.239	+.547	.280
22	.653	.268	-.311	.212	.393	.196	+.570	.246
24	.534	.228	-.295	.217	.251	.173	+.578	.213
26	.460	.197	-.262	.237	.143	.168	+.579	.188
28	.412	.180	-.252	.260	.080	.168	+.584	.171
30	.385	.175	-.258	.294	.039	.181	+.604	.156
32	.381	.168	-.279	.303	.006	.180	+.653	.147
34	.404	.161	-.309	.292	-.028	.181	+.741	.142
36	.438	.168	-.372	.283	-.051	.187	+.861	.134
38	.496	.192	-.458	.301	-.078	.212	+1.032	.141
40	.562	.225	-.567	.352	-.096	.256	+1.225	.168
42	.659	.247	-.662	.404	-.117	.305	+1.438	.209
44	.757	.278	-.761	.445	-.104	.351	+1.622	.244
46	.837	.299	-.864	.454	-.038	.380	+1.739	.250
48	.883	.321	-.929	.472	.071	.396	+1.741	.221
50	.885	.327	-.894	.474	.180	.375	+1.598	.202
52	.786	.356	-.745	.486	.242	.331	+1.289	.256
54	.536	.403	-.628	.452	.287	.287	+.877	.322
56	.184	.406	-.583	.404	.309	.259	+.459	.306
58	-.073	.309	-.531	.279	.282	.209	+.176	.217
60	-.197	.140	-.437	.189	.211	.179	+.029	.106
62	-.220	.075	-.366	.138	.174	.126	-.027	.038
64	-.184	.067	-.306	.113	.158	.086	-.036	.008
66	-.124	.071	-.229	.083	.133	.039	-.028	.006
68	-.077	.078	-.171	.095	.116	.036	-.022	.005
70	-.048	.064	-.121	.083	.089	.031	-.016	.003
72	-.036	.041	-.082	.058	.059	.028	-.013	.002
74	-.030	.042	-.052	.057	.033	.029	-.012	.002
76	-.025	.043	-.027	.057	.014	.030	-.012	.003
78	-.016	.039	-.002	.047	-.001	.030	-.013	.004
80	-.007	.046	+.025	.051	-.018	.030	-.014	.004
82	.003	.043	+.060	.053	-.041	.031	-.015	.004
84	.013	.031	+.105	.048	-.077	.035	-.015	.004
86	.025	.038	+.164	.058	-.126	.042	-.013	.005
88	.040	.059	+.237	.083	-.188	.053	-.008	.005
90	.063	.071	+.325	.105	-.261	.063	-.001	.006
92	.091	.081	+.416	.124	-.331	.070	+.007	.006
94	.115	.093	+.477	.145	-.375	.074	+.013	.005
96	.119	.096	+.471	.157	-.366	.078	+.014	.004
98	.092	.081	+.384	.145	-.303	.079	+.011	.004
100	.043	.059	+.247	.115	-.208	.069	+.003	.005

Table 4.34(a)

POWER GENERATED AND ABSORBED(W/Kg)-SLOW CADENCE (N=19)

	HIP		KNEE		ANKLE	
% STRIDE	Mean	Std.Dev.	Mean	Std.Dev.	Mean	Std.Dev.
0	.059	.134	.115	.152	-.004	.017
2	.250	.296	.268	.306	-.053	.096
4	.208	.298	-.003	.303	-.095	.110
6	.175	.313	-.309	.386	-.042	.108
8	.158	.276	-.485	.450	-.032	.181
10	.146	.322	-.428	.387	-.110	.214
12	.114	.405	-.177	.279	-.203	.197
14	.065	.444	.025	.229	-.267	.188
16	.028	.415	.128	.195	-.303	.196
18	.050	.392	.163	.208	-.316	.199
20	.084	.421	.130	.243	-.287	.193
22	.057	.367	.074	.233	-.254	.221
24	.004	.270	.025	.200	-.244	.280
26	-.041	.214	-.012	.173	-.262	.337
28	-.053	.200	-.047	.166	-.281	.321
30	-.051	.176	-.069	.147	-.304	.284
32	-.075	.205	-.068	.126	-.296	.291
34	-.093	.176	-.054	.114	-.291	.353
36	-.096	.169	-.031	.108	-.311	.445
38	-.082	.186	-.010	.097	-.303	.469
40	-.084	.195	.016	.104	-.251	.419
42	-.102	.198	.055	.131	-.145	.403
44	-.120	.195	.100	.172	.042	.464
46	-.137	.195	.146	.228	.323	.662
48	-.151	.223	.163	.291	.702	.858
50	-.139	.247	.142	.344	1.192	.919
52	-.076	.210	.065	.435	1.774	.751
54	.027	.119	-.081	.576	2.060	.732
56	.141	.175	-.268	.669	1.836	1.031
58	.238	.281	-.378	.610	1.295	1.010
60	.353	.296	-.444	.549	.617	.602
62	.362	.269	-.344	.408	.166	.284
64	.280	.178	-.156	.139	.012	.088
66	.255	.257	-.106	.105	-.001	.027
68	.233	.219	-.078	.099	.014	.017
70	.200	.135	-.039	.067	.022	.011
72	.163	.083	-.006	.034	.022	.008
74	.127	.070	-.008	.026	.018	.004
76	.088	.079	-.046	.065	.015	.005
78	.042	.067	-.098	.098	.012	.006
80	.017	.051	-.157	.122	.009	.005
82	.012	.049	-.209	.126	.005	.005
84	.007	.039	-.264	.114	.001	.006
86	.001	.024	-.338	.136	-.002	.007
88	.006	.020	-.405	.190	-.002	.008
90	.019	.036	-.485	.279	-.000	.006
92	.034	.055	-.532	.344	.000	.002
94	.043	.083	-.439	.367	-.001	.002
96	.038	.096	-.209	.321	-.001	.003
98	.025	.077	.008	.169	-.001	.005
100	.013	.042	.094	.104	-.003	.004

Table 4.34(b)

POWER GENERATED AND ABSORBED(W/Kg)-NATURAL CADENCE(N=19)

	HIP		KNEE		ANKLE	
% STRIDE	Mean	Std.Dev.	Mean	Std.Dev.	Mean	Std.Dev.
0	.009	.106	.332	.320	-.020	.056
2	.093	.192	.746	.656	-.061	.122
4	.147	.215	.306	.717	-.090	.133
6	.186	.369	-.393	.696	-.083	.209
8	.242	.576	-.758	.753	-.102	.225
10	.272	.822	-.759	.652	-.224	.323
12	.275	.942	-.460	.545	-.358	.365
14	.243	.926	-.061	.475	-.442	.336
16	.203	.828	.231	.449	-.436	.288
18	.075	.622	.371	.427	-.367	.305
20	-.061	.511	.386	.400	-.275	.317
22	-.143	.509	.314	.382	-.204	.291
24	-.155	.475	.204	.349	-.173	.266
26	-.125	.408	.083	.314	-.184	.291
28	-.096	.364	-.012	.299	-.230	.340
30	-.080	.317	-.072	.279	-.289	.362
32	-.079	.301	-.103	.227	-.338	.356
34	-.108	.293	-.104	.177	-.370	.346
36	-.145	.303	-.089	.165	-.385	.366
38	-.169	.307	-.049	.175	-.382	.452
40	-.185	.313	.012	.196	-.333	.519
42	-.209	.319	.096	.267	-.200	.547
44	-.236	.319	.180	.363	.088	.604
46	-.254	.308	.227	.467	.619	.824
48	-.231	.311	.211	.613	1.441	1.082
50	-.166	.346	.122	.785	2.448	1.175
52	-.042	.311	-.042	.895	3.266	1.106
54	.104	.213	-.230	.898	3.331	1.024
56	.238	.223	-.516	.766	2.552	1.040
58	.456	.226	-.757	.565	1.350	.804
60	.672	.266	-.778	.413	.408	.389
62	.713	.273	-.555	.286	.010	.103
64	.647	.262	-.354	.159	-.032	.039
66	.513	.339	-.241	.143	.008	.023
68	.380	.257	-.171	.124	.033	.013
70	.244	.191	-.078	.062	.036	.014
72	.161	.209	-.010	.035	.030	.013
74	.113	.123	.014	.056	.023	.010
76	.076	.083	.001	.079	.019	.007
78	.045	.089	-.034	.102	.017	.007
80	.016	.076	-.089	.112	.013	.007
82	-.005	.054	-.182	.117	.008	.008
84	-.010	.024	-.323	.174	.004	.009
86	.000	.022	-.502	.258	.002	.010
88	.027	.049	-.675	.350	.000	.009
90	.076	.084	-.881	.426	.001	.007
92	.117	.122	-.976	.476	-.001	.003
94	.109	.143	-.798	.467	-.004	.005
96	.057	.145	-.366	.394	-.003	.007
98	.019	.124	.061	.273	.001	.010
100	.011	.092	.218	.181	.001	.004

Table 4.34(c)

POWER GENERATED AND ABSORBED(W/Kg)-FAST CADENCE (N=17)

	HIP		KNEE		ANKLE	
% STRIDE	Mean	Std.Dev.	Mean	Std.Dev.	Mean	Std.Dev.
0	.140	.217	.516	.522	-.003	.042
2	.359	.404	.977	.881	-.106	.171
4	.354	.323	-.145	1.178	-.180	.214
6	.232	.263	-1.436	1.415	-.055	.204
8	.074	.421	-2.106	1.301	.025	.296
10	-.107	.629	-1.874	1.087	-.071	.396
12	-.233	.729	-.889	.872	-.287	.393
14	-.350	.767	.152	.746	-.464	.376
16	-.532	.696	.831	.655	-.501	.315
18	-.717	.604	1.062	.555	-.414	.247
20	-.796	.556	.959	.486	-.277	.230
22	-.769	.509	.699	.419	-.135	.249
24	-.702	.481	.447	.352	-.015	.250
26	-.581	.459	.235	.306	.055	.244
28	-.503	.439	.108	.262	.060	.253
30	-.458	.432	.031	.230	-.002	.259
32	-.460	.399	-.014	.194	-.100	.252
34	-.492	.353	-.044	.168	-.198	.255
36	-.568	.321	-.050	.128	-.255	.313
38	-.647	.321	-.045	.096	-.249	.473
40	-.722	.347	-.017	.124	-.129	.707
42	-.732	.365	.042	.249	.179	1.012
44	-.700	.372	.097	.452	.765	1.272
46	-.603	.330	.057	.718	1.743	1.494
48	-.389	.305	-.186	1.046	3.040	1.680
50	-.062	.355	-.590	1.275	4.249	1.602
52	.295	.446	-.996	1.359	4.582	1.040
54	.639	.590	-1.411	1.353	3.735	1.104
56	1.018	.747	-1.726	1.364	2.163	1.459
58	1.245	.688	-1.668	1.227	.904	1.106
60	1.238	.551	-1.260	1.125	.221	.487
62	1.184	.487	-.950	.786	.016	.112
64	1.037	.511	-.691	.525	.017	.058
66	.776	.413	-.390	.207	.049	.037
68	.529	.408	-.180	.158	.052	.026
70	.318	.318	-.027	.127	.041	.023
72	.184	.181	.038	.087	.031	.017
74	.099	.124	.041	.053	.024	.012
76	.040	.093	.015	.072	.020	.010
78	.002	.058	-.025	.106	.017	.010
80	-.016	.065	-.084	.150	.014	.011
82	-.026	.051	-.200	.217	.010	.012
84	-.027	.049	-.415	.316	.007	.011
86	-.004	.092	-.723	.447	.004	.008
88	.041	.136	-1.055	.594	.002	.005
90	.083	.164	-1.267	.706	-.001	.004
92	.086	.181	-1.179	.741	-.004	.004
94	.052	.185	-.729	.724	-.004	.007
96	-.001	.185	-.123	.648	.001	.009
98	-.022	.156	.317	.473	.006	.008
100	-.017	.105	.432	.297	.003	.007

Tables 5.11-5.14

EMG (μV Linear Envelope)

% STRIDE	GLUT.MED. Mean	GLUT.MED. Std.Dev.	GLUT.MAX. Mean	GLUT.MAX. Std.Dev.	MED.HAMS. Mean	MED.HAMS. Std.Dev.	LAT.HAMS. Mean	LAT.HAMS. Std.Dev.
0	30.0	42.6	27.7	24.6	169.3	76.6	165.3	108.5
2	45.0	43.4	35.4	25.3	155.7	74.3	171.3	115.2
4	60.0	46.0	43.1	27.4	140.9	74.4	172.5	130.7
6	73.9	48.9	46.6	27.3	130.1	84.3	161.7	132.6
8	86.7	54.1	45.8	25.5	122.4	93.7	140.7	119.6
10	99.5	62.3	45.0	25.7	114.0	101.8	118.0	109.1
12	95.4	62.3	39.2	23.1	103.5	99.8	97.9	93.6
14	91.4	62.8	33.5	21.0	89.7	88.3	78.6	80.6
16	84.5	60.3	28.6	19.1	76.0	74.5	64.3	73.7
18	74.8	54.7	24.5	17.3	64.8	64.0	54.9	71.2
20	65.2	49.4	20.5	16.2	56.2	55.9	46.7	69.4
22	59.4	46.2	19.3	16.5	47.5	48.6	40.9	65.8
24	53.6	43.2	18.1	17.2	41.7	43.7	35.5	62.5
26	49.1	40.9	16.5	16.5	38.0	42.5	30.7	57.0
28	45.9	39.2	14.4	14.5	35.3	41.9	26.8	49.1
30	42.7	37.9	12.4	12.6	32.3	41.6	23.1	41.2
32	40.8	35.5	10.9	10.3	31.2	42.3	20.8	34.6
34	38.8	33.5	9.3	8.1	30.5	43.0	18.5	28.2
36	36.0	30.1	8.1	6.5	30.1	42.3	17.5	23.7
38	32.3	25.1	7.3	5.5	29.1	39.8	17.4	21.7
40	28.6	20.4	6.5	4.7	26.7	35.3	16.8	19.8
42	24.1	16.2	6.4	4.6	23.8	29.7	16.2	19.0
44	19.6	12.2	6.4	4.8	20.4	23.7	15.6	18.3
46	15.8	9.4	6.5	5.5	16.6	18.0	14.9	17.8
48	12.7	7.4	6.8	6.7	12.8	12.7	14.1	17.1
50	9.5	5.8	7.0	8.3	10.1	8.3	13.2	16.6
52	8.2	4.7	6.7	7.5	8.5	5.6	13.0	17.6
54	6.9	3.6	6.4	6.7	8.3	5.1	13.0	19.0
56	6.2	3.1	6.4	6.1	11.7	10.6	14.2	20.3
58	6.1	3.0	6.9	5.9	17.2	21.5	15.8	23.2
60	6.0	3.2	7.3	6.3	24.4	35.0	17.4	27.8
62	6.6	3.6	9.0	8.2	33.5	49.4	21.5	38.4
64	7.1	4.6	10.8	10.5	41.7	63.0	24.8	46.3
66	7.5	5.8	12.1	12.3	49.0	81.0	27.2	49.3
68	7.7	7.3	12.9	13.8	50.8	89.2	28.3	48.9
70	7.8	8.8	13.7	15.6	47.7	85.0	28.9	48.5
72	7.1	8.0	14.4	15.6	42.0	74.2	27.3	43.1
74	6.5	7.3	15.0	18.2	35.8	62.1	25.6	38.0
76	5.8	6.3	15.1	20.1	30.5	51.3	23.9	32.1
78	5.3	5.0	14.7	20.1	26.4	41.3	21.9	25.7
80	4.7	3.7	14.2	20.6	24.1	33.2	20.0	20.0
82	5.0	3.1	12.5	17.8	28.1	28.7	22.2	17.3
84	5.3	3.2	10.9	15.5	36.2	30.7	26.1	16.3
86	6.8	6.3	9.3	12.9	57.9	46.8	35.7	20.9
88	9.3	12.2	7.7	9.5	93.1	77.5	50.1	31.8
90	11.9	18.3	6.2	6.2	133.8	110.0	64.6	44.5
92	14.7	23.9	7.9	5.7	159.3	108.5	81.2	58.2
94	17.5	29.6	9.6	7.1	178.0	108.0	96.5	72.7
96	21.2	34.2	14.1	10.8	183.6	98.0	112.8	80.2
98	25.6	38.1	21.4	18.2	176.2	81.9	130.6	86.0
100	30.0	42.5	28.7	26.5	161.4	71.2	147.1	98.1

Tables 5.15-5.18

EMG (μV Linear Envelope)

	ERECTOR SPINAE		SARTORIUS		REC.FEM.		VAS.LAT.	
% STRIDE	Mean	Std.Dev.	Mean	Std.Dev.	Mean	Std.Dev.	Mean	Std.Dev.
0	49.3	34.2	27.5	51.4	30.5	31.9	84.6	57.4
2	73.2	44.9	32.1	52.9	38.3	34.7	105.4	68.5
4	97.1	57.6	36.6	55.1	46.2	37.7	129.3	80.3
6	105.6	59.7	39.2	51.3	56.1	41.6	156.5	98.2
8	98.7	51.0	39.7	42.7	66.1	46.4	183.5	118.1
10	91.8	43.5	40.3	37.6	73.2	51.3	201.1	135.2
12	75.9	32.7	38.6	34.9	72.3	51.0	198.7	133.6
14	60.0	22.6	36.9	33.3	70.0	50.2	193.0	130.6
16	47.7	16.0	34.2	31.0	65.2	49.9	177.4	119.9
18	39.1	12.3	30.4	27.2	57.9	50.2	154.4	104.0
20	30.6	10.4	26.6	24.0	50.4	51.0	131.1	88.4
22	27.1	9.9	24.7	23.3	41.5	44.5	107.5	74.4
24	23.7	9.3	22.7	23.6	33.4	38.1	85.6	61.0
26	21.1	9.0	20.7	23.1	27.1	32.4	68.8	51.3
28	19.3	8.9	18.6	21.4	22.1	26.9	55.9	44.0
30	17.5	9.0	16.5	19.9	17.5	21.5	44.1	36.9
32	15.8	8.2	14.9	16.8	14.8	18.3	36.4	31.8
34	14.1	7.6	13.3	13.8	12.5	15.3	29.7	27.6
36	12.7	6.7	12.2	11.9	11.1	13.6	25.5	25.3
38	11.7	5.6	11.7	11.0	10.8	13.4	23.0	24.5
40	10.7	4.7	11.1	10.1	10.5	13.2	21.2	25.3
42	10.3	4.0	11.1	8.9	11.9	17.5	21.6	29.5
44	9.9	3.8	11.1	8.7	13.4	22.2	22.6	36.0
46	13.1	3.7	12.1	11.3	15.3	27.3	24.9	42.8
48	19.9	7.3	14.2	16.7	17.8	33.1	28.7	52.0
50	26.7	12.2	16.3	22.7	20.2	38.6	31.7	58.5
52	48.3	17.1	16.8	21.0	20.5	36.9	31.3	56.5
54	69.9	23.9	17.2	19.6	21.0	34.8	30.7	52.1
56	80.1	25.9	19.9	18.9	20.7	30.6	29.5	46.0
58	78.9	23.3	24.7	19.9	19.9	24.7	27.9	39.4
60	77.6	22.4	29.5	22.3	20.4	22.3	27.2	36.1
62	65.8	19.8	35.4	26.9	21.3	23.1	28.5	38.0
64	53.9	18.7	41.3	32.7	21.6	24.1	30.9	44.5
66	44.6	18.9	45.5	37.7	22.1	27.0	35.4	53.9
68	37.9	19.8	48.0	42.2	22.3	27.4	39.9	60.9
70	31.1	21.7	50.5	47.3	21.1	25.2	41.3	63.9
72	26.9	19.3	47.6	39.4	17.9	20.7	37.1	59.0
74	22.8	17.0	44.7	33.4	14.5	16.3	31.9	51.4
76	19.1	14.3	40.1	28.5	11.4	12.5	26.3	42.4
78	16.0	11.0	33.9	22.9	8.5	8.9	20.8	33.4
80	12.8	7.7	27.6	18.3	6.1	6.0	15.7	25.5
82	11.4	5.9	22.7	14.4	4.6	3.9	12.8	19.9
84	9.9	4.4	17.7	10.6	3.6	2.3	10.6	15.6
86	9.3	4.3	14.3	8.0	3.5	2.0	10.5	12.5
88	9.6	5.2	12.3	6.5	4.8	3.6	12.7	12.1
90	10.0	6.4	10.2	5.7	6.7	5.7	17.2	14.7
92	11.4	8.8	10.8	5.9	10.7	11.9	27.7	20.1
94	12.8	11.4	11.3	8.0	15.1	18.3	40.9	28.2
96	20.7	14.7	14.8	18.3	20.0	24.3	55.3	38.0
98	35.0	22.5	21.4	36.8	25.3	30.1	69.1	48.5
100	49.4	32.2	27.9	55.4	31.2	35.7	83.3	57.8

Tables 5.19-5.22

EMG (μV Linear Envelope)

% STRIDE	ADD.LONG.		ADD.MAG.		TIB.ANT.		EXT.DIG.LONG.	
	Mean	Std.Dev.	Mean	Std.Dev.	Mean	Std.Dev.	Mean	Std.Dev.
0	48.1	48.9	51.7	25.4	279.5	102.7	179.1	93.3
2	50.4	50.6	56.0	27.3	322.5	118.5	199.1	97.1
4	52.7	53.0	60.3	30.6	358.7	130.4	219.0	104.2
6	53.1	49.7	61.1	32.4	365.8	128.3	222.8	105.5
8	51.6	41.6	58.3	32.7	328.7	105.4	210.5	100.3
10	50.2	36.5	55.6	34.1	283.7	82.9	198.2	97.0
12	47.6	35.3	49.8	28.2	225.5	62.9	169.6	84.6
14	45.0	35.5	43.9	22.8	171.9	47.9	141.1	72.8
16	41.7	34.3	41.6	21.1	131.8	41.5	116.2	60.7
18	37.6	31.2	42.9	26.7	104.5	38.5	95.0	48.0
20	33.6	29.0	44.2	35.9	80.0	36.3	73.9	36.2
22	32.7	30.6	50.4	49.3	66.2	33.9	63.6	30.3
24	31.9	33.6	56.6	63.2	52.8	30.1	53.3	27.0
26	29.9	34.7	59.1	71.5	43.7	25.9	45.8	26.7
28	26.8	33.1	57.9	74.1	38.8	22.9	41.2	27.1
30	23.7	31.8	56.6	76.9	34.4	21.5	36.6	28.2
32	21.0	28.5	48.4	65.3	31.9	19.3	36.2	28.6
34	18.2	25.4	40.1	53.7	30.5	17.4	35.8	29.3
36	15.8	22.0	32.9	42.8	32.1	16.3	38.2	30.5
38	13.7	18.1	26.7	32.4	35.4	16.2	43.5	32.4
40	11.6	14.4	20.4	22.2	40.2	18.8	48.8	35.0
42	12.0	12.0	17.0	16.4	48.3	23.0	56.1	34.4
44	12.4	10.5	13.6	10.8	56.1	27.9	63.4	34.9
46	14.9	11.6	11.8	7.5	61.3	30.2	67.6	33.8
48	19.4	14.8	11.7	6.8	62.9	31.4	68.9	31.2
50	23.9	18.4	11.5	7.4	63.7	33.0	70.2	30.1
52	31.2	21.0	14.5	9.2	58.1	30.6	60.7	25.6
54	38.5	25.3	17.5	11.6	53.9	27.3	51.1	21.8
56	45.0	29.8	19.4	12.4	52.4	27.2	45.5	16.5
58	50.7	35.1	20.2	11.5	55.0	36.9	43.7	13.3
60	56.3	41.4	20.9	11.0	63.1	59.7	41.9	17.9
62	56.7	44.5	22.8	8.5	85.5	73.9	58.0	25.1
64	57.0	48.2	24.6	8.9	104.2	81.1	74.2	35.8
66	55.7	47.9	30.2	13.3	121.1	81.7	89.7	45.3
68	52.6	44.0	39.5	23.3	138.1	76.1	104.6	53.8
70	49.6	41.3	48.7	34.7	155.1	77.1	119.5	63.2
72	45.2	35.3	59.6	48.0	170.2	73.7	124.7	62.0
74	40.7	30.1	70.5	61.9	184.5	74.0	129.8	61.4
76	35.9	25.6	76.5	70.7	191.0	73.9	129.2	59.1
78	30.8	21.2	77.6	74.5	189.0	72.7	122.7	56.0
80	25.7	17.3	78.8	78.8	184.2	72.5	116.3	54.4
82	22.4	14.4	69.0	68.7	169.0	71.3	103.6	50.9
84	19.2	11.8	59.3	58.9	155.0	73.9	90.9	48.5
86	17.3	10.8	50.9	49.2	145.6	73.8	83.8	47.8
88	16.8	11.7	43.9	39.6	139.6	69.3	82.2	48.9
90	16.3	13.2	36.9	30.8	134.9	64.9	80.7	51.5
92	19.0	14.6	38.3	23.0	147.2	62.4	95.0	60.3
94	21.8	17.1	39.8	17.5	164.1	65.1	109.2	70.4
96	27.2	22.7	42.3	16.3	191.9	72.6	129.5	78.4
98	35.3	35.4	45.9	18.0	229.9	84.3	155.8	85.3
100	43.5	50.3	49.6	21.7	270.6	98.9	182.2	94.1

Tables 5.23-5.26

EMG (μV Linear Envelope)

% STRIDE	MED.GASTROCS. Mean	MED.GASTROCS. Std.Dev.	LAT.GASTROCS. Mean	LAT.GASTROCS. Std.Dev.	SOLEUS Mean	SOLEUS Std.Dev.	PER.LONG. Mean	PER.LONG. Std.Dev.
0	79.3	100.4	55.4	48.6	36.1	16.4	61.5	41.2
2	84.4	90.8	62.2	51.1	45.2	18.4	76.0	42.8
4	89.6	82.9	72.4	56.3	53.3	20.4	90.6	48.6
6	94.4	78.5	81.8	59.1	60.4	22.7	96.9	50.8
8	98.3	77.0	92.2	64.7	65.2	24.7	94.9	48.0
10	98.9	72.4	98.2	70.7	71.1	27.1	92.9	46.8
12	95.7	61.7	93.7	66.4	80.3	32.2	80.6	39.2
14	91.2	48.0	85.6	58.5	92.5	43.6	68.3	33.1
16	87.1	36.7	75.0	48.5	103.3	53.3	57.9	28.8
18	89.0	31.8	63.7	38.1	112.9	60.0	49.4	25.5
20	96.0	41.4	54.0	29.3	120.2	65.9	40.9	24.0
22	105.5	56.6	46.4	22.5	125.2	65.3	38.5	26.0
24	118.5	74.1	41.0	17.8	128.3	64.7	36.0	28.8
26	132.4	86.7	39.2	17.9	131.3	64.6	34.6	31.0
28	145.2	95.0	40.9	20.4	134.8	65.7	34.3	32.2
30	160.4	102.5	45.7	25.8	140.8	69.3	34.0	33.5
32	179.3	111.0	57.4	34.8	153.2	76.5	37.5	34.4
34	198.1	116.1	74.7	45.2	169.5	84.1	41.0	36.1
36	220.0	121.4	99.0	55.7	191.1	95.8	47.1	37.9
38	247.4	128.1	130.4	69.3	218.3	113.1	55.8	41.3
40	273.8	136.4	164.1	85.8	247.7	130.2	64.5	47.0
42	292.9	145.8	195.0	100.4	286.5	142.6	75.7	47.6
44	300.9	155.8	214.9	113.7	322.4	156.0	87.0	50.8
46	288.7	159.7	223.3	127.9	344.3	161.0	92.9	50.1
48	256.1	152.7	215.7	136.5	352.0	159.7	93.7	46.1
50	212.6	134.9	198.5	137.2	353,2	163.8	94.4	45.2
52	164.0	108.6	168.2	125.6	312.3	149.6	81.0	36.4
54	121.0	80.3	135.2	108.5	262.5	134.6	67.6	28.9
56	93.4	51.5	108.9	85.0	209.8	114.3	55.5	23.0
58	79.5	39.2	89.5	64.0	158.6	89.9	44.5	17.3
60	75.1	50.7	75.6	54.5	113.5	66.7	33.5	12.7
62	79.7	65.8	73.8	55.6	89.5	52.5	34.9	14.6
64	84.5	70.8	73.5	60.6	67.2	39.9	36.3	18.5
66	85.5	67.9	70.1	59.7	51.4	32.7	39.7	23.4
68	80.6	60.6	63.3	54.8	42.2	28.5	45.0	29.7
70	73.6	51.5	57.2	49.8	34.1	25.7	50.2	37.0
72	64.6	44.2	49.4	43.9	29.0	20.9	50.5	35.2
74	55.0	39.6	40.7	35.3	24.4	16.6	50.8	33.9
76	44.9	34.6	32.8	25.8	20.6	13.1	48.9	31.7
78	36.3	28.4	27.1	20.1	17.9	10.7	45.1	29.0
80	30.2	22.6	23.0	19.6	15.5	9.6	41.2	27.2
82	28.7	19.2	20.7	19.2	13.6	9.2	35.3	23.7
84	28.4	17.4	19.2	18.1	11.7	9.1	29.5	20.4
86	28.5	16.6	20.0	18.7	10.7	8.3	26.0	17.9
88	28.6	17.1	23.0	24.1	11.1	6.7	24.8	16.4
90	30.3	20.2	27.5	30.0	11.7	6.3	23.5	15.8
92	36.5	33.0	31.1	31.6	13.9	7.0	28.1	19.7
94	50.5	69.4	36.6	35.6	16.3	8.6	32.7	24.0
96	66.6	101.8	44.5	37.2	20.8	10.5	40.6	29.1
98	76.7	112.8	51.7	42.7	27.6	13.1	51.7	35.3
100	85.6	104.8	56.6	49.3	38.6	17.3	62.9	41.9