AN ABSTRACT OF THE THESIS OF

Jonathan B. Fewster for the degree of Master of Science in Human Performance presented on November 30, 1995. Title: The Role of Musculoskeletal Forces in the Human Walk-Run Transition.

Abstract approved: Redacted for Privacy

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This investigation examined the possible role of musculoskeletal forces in the human walk-run transition. In order to measure these forces a treadmill was constructed which allowed the measurement of vertical ground reaction forces while subjects walked and ran at prescribed speeds. Validation proved the device to be accurate and reliable in measuring the midstance vertical ground reaction forces which were analyzed in this study.

Ten untrained runners were recruited from the University population and paid for their participation in this study. To differentiate the roles of speed of locomotion and musculoskeletal force, both speed and subject weight were manipulated. Speed was controlled by the treadmill operator and weight was added to the subjects in the form of a weight vest of approximately 15% body weight. Each subject's preferred transition speed was determined for the weighted and unweighted conditions. Following this determination, each subject's midstance vertical ground reaction forces were measured while walking and running over a range of speeds for both weight conditions.

The force at transition was consistent for the two conditions for the subjects measured, indicating that musculoskeletal force may have a role in the transition. However, speed of transition was also consistent, not allowing differentiation of the two variables. Mapping the midstance forces of each gait versus speed of locomotion illustrated running to have a significant increase in force at the preferred transition speed. A trend of

increasing variability of force at and above the preferred transition speed was evident for walking. This instability may facilitate or prompt the change from walking to running. As a result of this investigation, musculoskeletal forces may be considered to have some influence on the human walk-run transition.

The Role of Musculoskeletal Forces in the Human Walk-Run Transition

by

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A THESIS

submitted to

Oregon State University

in partial fulfillment of
the requirements for the
degree of

Master of Science

Completed November 30, 1995
Commencement June 1996

Master of Science thesis of Jonathan B. Fewster presented on November 30, 1995

APPROVED:

Redacted for Privacy

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ACKNOWLEDGMENTS

I would like to thank my advisor, Dr. Gerald Smith, for his help and support during this project. He spent a great amount of time with me discussing this research and helping trouble-shoot some of the difficulties I had along the way. His frequent ideas, insights and continual support helped immensely in this project. Thank you for your help.

I would like to thank each member of my committee for their contributions to this work. Dr. Terry Wood spent many hours helping me with issues of measurement, treadmill validation, statistical analysis and final format. Dr. Debra Rose helped me understand the basics of Dynamic Systems theory, allowing me to take a research question from theory to experimental design to conclusions. Dr. Michael Schuyler, my minor advisor, was the person I went to with most of my computer issues. He helped me with the complex interfacing and programming which was necessary to make the treadmill work. Dr. Rebecca Callison, my graduate representative, also helped with some computer issues, particularly what may be possible for the treadmill device in the future. Thank all of you for your help in this project.

My parents, Lowell and Julie Fewster, have been a great support to me throughout my life. Their unconditional support and quiet encouragement have allowed me to make my dreams reality. I was extremely happy that my father was able to attend my oral defense of this thesis. Thank you. I love both of you very much.

I wish to thank my grandfather and grandmother, J.D. and Alma Fewster, for their support. Their assistance help make my education possible. Thank you.

Many friends have helped me throughout the course of this project. I would like to thank Jennifer Beck for her help with many of the pilot tests, as well as being a good friend in the lab. Sean Clark, a friend in the Motor Behavior Lab across the hall, helped me with Dynamic Systems, and provided great assistance with the programming of SPSS. Lissa Zyromski produced the line drawing of the treadmill which appears in this thesis. There are many other people who have helped me with moving the treadmill, conducting the pilot studies, by listening to me practice presentations and lending an ear when I needed one. This group includes Morris Levy, Ki Kwang Li, Darren Dutto, Suwat Sidthilaw, and Jane LaRiviere. Thank you. Two other friends helped beyond these roles: Chris Lowry, my roommate for a significant portion of the time I was working on this thesis, was a sounding board for many of my thoughts and his high standards for research encouraged me to strive for the ideal. Louise Yount was very patient with me when my extreme focus on this project threatened to turn me into a machine. Thank you for your friendship.

Outside of the laboratory, many friends have helped me temporarily escape from this otherwise ever-present project. Some of these friends have been training partners for cross-country skiing, biking or running. Without those workouts, the mental break, and the associated endorphins I wouldn't be sane now. Thank you.

This research was supported by a Reebok Graduate Student Research Grant on Human Performance and Injury Prevention from the American College of Sports Medicine Foundation.

This research was further supported by equipment donations from PCB Piezotronics, Inc. At PCB, I would like to thank Lou Zagst for his assistance in this project.

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CHAPTER 1
INTRODUCTION

Human locomotion has been studied and modeled for many years. Despite this scrutiny many questions as to cause and function still remain unanswered. One such aspect of gait not sufficiently defined or understood is the mechanism controlling transitions between walking and running. Because of its nature, any findings on the control mechanism of gait will supplement the current understanding of the human control system. Therefore, a study of gait transitions will also provide insight into the underlying theories of control.

Human methods of locomotion have evolved from quadrupedal to bipedal locomotion. Once adapted to bipedal locomotion, humans have used primarily two gaits: walking and running, with walking generally performed at lower speeds. These gaits may be compared to the wider range of options available in quadrupedal locomotion such as walking, trotting, cantering, and galloping (in the general order of increasing speed).

When speed is not essential, metabolic cost may be expected to be minimized. Therefore, it follows that walking, the gait of choice at low speeds, generally produces a lower metabolic cost per unit of time than would running at these low speeds. The opposite is true, however, when one chooses to walk at high speeds, as is the case in race walking. By forcing the body to walk at speeds which normally would be reserved for running, the metabolic cost exceeds that of running at the same speed (Menier & Pugh, 1968).

As velocity is increased, a speed is reached at which running has a lower metabolic cost than walking. The switch in relative cost has been thought to be responsible for the change in gait between walking and running; the human body opting for the gait yielding the lower toll on the metabolic system. However, some researchers have not been convinced by these dynamics, believing the differentiation in cost to be too minute to cause the acute switch in gait (Hreljac 1993b; Thorstensson & Roberthson, 1987). Hreljac's study of the gait-metabolic cost relationship determined that gait transitions are probably not for the objective of minimizing metabolic cost.

Other factors examined have included stride frequency and amplitude of leg movements (Nilsson, Thorstensson, & Halbertsma, 1985). Thorstensson and Roberthson (1987) compared the duration of the stance phase and leg length with the speed of transition. While they observed a tendency for subjects with longer legs to have higher speeds of transition, they concluded that the reasons for gait transitions at speeds not extreme for either walking or running were "unclear."

These investigations have suggested that kinetic factors should be investigated for their role in the transition between walking and running. Evidence for a kinetic factor being a control parameter comes from research on horses. By analyzing the kinetics of the trot-gallop transition in horses, Farley and Taylor (1991) identified a critical musculoskeletal force level at which the transition takes place. Once the critical force level was achieved, the horses would switch from trot to gallop despite an increase in metabolic cost. By adding weight to the horses, Farley and Taylor were able to produce the critical force and the resulting transition at lower velocities. From this evidence and the suggestions of other research, kinetic factors need to be properly investigated for their role in the human walk-run gait transition.

1.1 Statement of the Problem

The purpose of this study was to investigate a possible kinetic control parameter governing human locomotion and the transition between walking and running in physically active, college-aged males and females. More specifically, the midstance vertical ground reaction force ($VGRF_M$), assumed to be proportional to the force on the Achilles tendon, was measured with the goal of identifying a threshold level of force which may prompt the subject to switch from walking to running. If $VGRF_M$ level is indeed a control parameter for an individual, then achieving that force level would cause the transition to occur independent of other factors. Some of these factors, notably speed, were manipulated in an attempt to isolate a determining force level.

This study took the form of three tests repeated for two conditions of weight. The first test was to determine the subject's preferred transition speed: Speed was increased in discrete intervals and the subject was free to switch between walking and running. The speed at which the subject switched to running and maintained the running gait for 30 s was labeled the preferred transition speed (PTS). The second test was to measure forces while the subject walked over a range of speeds: The subject was asked to maintain a walking gait while speed was increased and vertical ground reaction forces were measured. In the third test, forces were measured over the range of speeds while the subject maintained a running gait. The three tests described here were repeated with the subject unweighted and with the subject wearing a weight vest of approximately 15% of his or her body weight. In subsequent data analysis, the midstance forces were picked from the vertical ground reaction force curves. To test the possible control parameter, the $VGRF_M$ and PTS values were compared between the two conditions.

1.2 Research Hypotheses

It was hypothesized that switching from walking to running is due to the subject reaching a critical musculoskeletal force and not a critical velocity. If this is true, an addition of weight to the subjects would elicit a change in the preferred transition speed, but would not alter the midstance vertical ground reaction force at the preferred transition speed. Such an identification would point to musculoskeletal force as a control parameter affecting the collective variable of human gait.

In addition, if musculoskeletal force is a control parameter, midstance vertical ground reaction force would significantly change at walking speeds at and above the preferred transition speed.

After the data were initially examined, further support for musculoskeletal force as a control parameter was tested by quantifying the changes in variability of force with increasing speed, particularly about the preferred transition speed. The variability in midstance vertical ground reaction force was hypothesized to increase at walking speeds just below the preferred transition speed and to remain high at speeds above the preferred transition speed. For running, the variability in $VGRF_M$ was hypothesized to be high at speeds below the preferred transition speed and to decrease at the preferred transition speed.

1.3 Statistical Hypotheses

To test the primary research hypothesis, the following statistical hypothesis was tested:

H_0: $PTS_{weighted} - PTS_{unweighted} = 0$

$VGRF_{M\ weighted} - VGRF_{M\ unweighted} \neq 0$

H_1: 1. $PTS_{weighted} - PTS_{unweighted} \neq 0$

$VGRF_{M\ weighted} - VGRF_{M\ unweighted} = 0$

2. $PTS_{weighted} - PTS_{unweighted} \neq 0$

$VGRF_{M\ weighted} - VGRF_{M\ unweighted} \neq 0$

3. $PTS_{weighted} - PTS_{unweighted} = 0$

$VGRF_{M\ weighted} - VGRF_{M\ unweighted} = 0$

The null hypothesis is a starting point for this experiment based on the understanding that speed may have an influence on the human walk-run transition, while vertical ground reaction force has yet to be proven to have a controlling influence. The alternate hypotheses are options from the two-variable state specified in the null hypothesis. The first alternate is the most conclusive, requiring subjects to switch from walking to running at a consistent force, but at different speeds. If accepted, this hypothesis would indicate $VGRF_M$ to be a control parameter independent of speed. The second alternate hypothesis discounts the roles of both variables, with the preferred transition occurring at different speeds and force levels for the two conditions. If found to be true, this hypothesis would indicate that neither variable is a control parameter; perhaps another variable should be investigated. The third alternate hypothesis allows for both variables to be control parameters for the walk-run transition, but their roles and levels of influence may not be distinguished.

Using the force data to further test the hypothesis that musculoskeletal force, and not speed, is a control parameter in the human walk-run transition, the following statistical hypothesis was tested:

H_0: $\psi = 0$

H_1: $\psi \neq 0$

where ψ is the pair-wise comparison among the mean midstance vertical ground reaction forces while walking at the increments of speed directly above, below and at the preferred transition speed.

The ad hoc hypothesis of variability of $VGRF_M$ will be tested by the following statistical hypothesis:

H_0: $\psi = 0$

H_1: $\psi \neq 0$

where ψ is the pair-wise comparison among the pooled variances of midstance vertical ground reaction force at the increments of speed directly above, below and at the preferred transition speed. This hypothesis was tested for both walking and running.

1.4 Operational Definitions

Preferred Transition Speed (PTS): The velocity in a given weighted or unweighted condition at which a subject switched from walking to running.

The Difference in Preferred Transition Speeds (ΔPTS): The difference in transition velocities between the weighted and unweighted conditions.

Vertical Ground Reaction Force (VGRF): The vertical force exerted on the body by the ground. By isolating midstance when anterior-posterior ground reaction force is approximately zero and the lever arms of the foot are at a constant proportion, the VGRF is assumed to be proportional to the moment about the ankle joint. At midstance, there is very little angular acceleration of the foot and all active muscles are loading the Achilles tendon (Basmajian & De Luca, 1985). The combination of all of these factors yield proportional vertical ground reaction forces and Achilles tendon forces. For this experiment the vertical ground reaction forces at midstance were measured and compared. These will be denoted by the abbreviation: $VGRF_M$.

The Difference in Vertical Ground Reaction Force at Midstance ($\Delta VGRF_M$): The change in midstance forces between the preferred transition speeds of the weighted and unweighted conditions.

1.5 Assumptions

This research assumes that the controlled environment of laboratory treadmill walking and running will not adversely affect the gait of each subject, either by changing gait patterns or by causing gait transitions to occur at speeds which would not be characteristic of overground travel.

During a normal overground gait transition, the speed may change significantly when the gait changes. If a person is walking and begins to run, he or she will simultaneously make a substantial increase in velocity. This experiment isolated gait transitions and limited changes in speed to only 0.2 m/s every 30 s interval. It was assumed that this small change in velocity would provide a precise point of transition, the same point that would occur if the individual was able to accelerate or decelerate at his or her own preferred rate.

While the construction and evolution of a coordinative structure may be due to the environment, the individual, and/or the goal of the task, it was assumed that the experimental conditions were for a short enough duration to be insignificant in altering any coordinative structure which is assumed to be under study. Similarly, the amount of running and walking that each individual was to perform was assumed to have little or no affect on either the ground reaction forces or the force cutoff levels which may govern the transitions. Any other determining factors in gait selection were assumed to be unaffected by the experimental environment.

1.6 Limitations

The force-measurement system which was used in this experiment underwent extensive validation. These results, presented in Appendix C, indicate the vertical ground reaction forces measured by this system to be valid. In order to isolate midstance forces, each heelstrike and toe-off was identified from the force data. Midstance was then calculated as a percentage of stance time. The accuracy of the $VGRF_M$ values were affected by the characteristics of the force measurement system, the analysis algorithms, and the ability to accurately identify each heelstrike and toe-off.

The typical force measurements that were performed in this experiment are usually collected on a laboratory force plate, such as the Kistler 9281B11. Such plates are very rigid, resulting in a natural frequency of approximately 850 Hz and a high signal-to-noise ratio. Such a rigid structure will not alter the kinetics or kinematics of gait during measurement. The corresponding high natural frequency assures that resonance will have minimal effect on the measured ground reaction forces. As a result of these features, the force plate is able to produce very good measurements of ground reaction forces without affecting the gait of the subject.

In contrast, the measurement system created by mounting a treadmill on a set of transducers had a higher mass and less rigidity than a force plate. These two factors produced a lower natural frequency of ~275 Hz. Even with steps taken to stiffen the treadmill bed, the system was not as rigid as a force plate and may have caused variations in the expected VGRF signals. One example of such a difference may be a slightly elongated stance time. However, the human subject tests performed on the system indicate peak forces, rise rates, and contact times to be of reasonable accuracy.

1.7 Delimitations

While all people have a walk-run transition which may be studied, this investigation focused on physically-active university students between the ages of 18 and 25 years. The subjects were men and women who had not been diagnosed with diseases or injuries likely to affect walking or running movement patterns. The selected subjects had a level of fitness characterized by 1-4 h of exercise per week, of which running was a part. To maintain similar patterns of musculoskeletal loading, these subjects were required to be rearfoot-strikers when running.

1.8 Definitions

Some of the theoretical terminology associated with the present project follows:

Coordinative Structure: A system of muscles and joints which is constrained in its degrees of freedom by the nervous system for the completion of a certain function or task. An action performed by such a structure seems to be governed by fewer degrees of freedom than are mechanically involved (Magill, 1993).

All of the muscles and joints involved in walking may be considered to be a part of the coordinative structure governing locomotion. When changing speed, the individual does not independently increase the angular velocity of each joint, but instead all joints and muscles change their actions according to the desired change in speed. This illustrates how the system seems to behave more simply than the complexity of joints and muscles the coordinate structure constrains.

Collective Variable: The state of the system. In the case of a person moving from point A to point B, the collective variable is gait or locomotion with the attractor states of walking and running. This variable has also been called an order parameter (Clark, 1992, Thelen & Smith, 1994).

Attractor: A region of relative stability within a geometric model of all possible states of a system. The system tends toward these regions of stability. When someone is moving from point A to point B, they tend to choose walking or running as their method of locomotion. These are the attractor states or regions of stability for human gait (Clark, 1992, Thelen & Smith, 1994).

Bifurcation: A switch in the collective variable from one attractor state to another. Such a switch may occur when an instability occurs. The shift may behave differently depending on the direction in which the transition is occurring (Clark, 1992, Thelen & Smith, 1994). For gait, the walk-run and run-walk transitions together have been identified as a bifurcation (Beuter & Lefebvre, 1988).

Control Parameter: A parameter which is independent from the collective variable, but when scaled past a critical value will cause a bifurcation or shift in the collective variable (Clark, 1992, Thelen & Smith, 1994). For gait, possible control parameters include speed, metabolic cost, and musculoskeletal force.

In the case of a person moving from point A to point B, once the control parameter reaches a critical level, the person would shift from the attractor state of walking to that of running. Thus the collective variable would act through the coordinative structure to make the necessary changes to all involved muscles and joints.

Kinetic trigger: An hypothesized control parameter and its critical level of force at which the mode of locomotion becomes unstable, switching the gait between walking and running.

Ground Reaction Force: The force of the ground or treadmill pushing against the subject's foot as he or she steps on the device. This force is measured in three directions (vertical, anterior-posterior, and medial-lateral) by a force plate. The instrumented treadmill used in this study measured the vertical component.

Vertical ground reaction force (VGRF): The vertical component of the ground reaction force. Equal to the subject's weight plus any inertial forces due to accelerations of the subject's mass.

Impact Peak: On a typical walking or running vertical ground reaction force curve the first peak is due to the foot striking the ground. This high point, labeled the impact peak, may not be evident on all running trials, particularly if the subject landed on his or her mid- or forefoot.

Active Peak: For a vertical ground reaction force curve for walking or running, the second peak is due to the subject's body weight being over the foot and applying a force to accelerate the body vertically for the next step. This peak is called the active peak.

Center of Pressure (COP): If the vertical ground reaction force is calculated to be a load at a single point rather than over a foot-shaped loading area, the single point is called the center of pressure.

Inverse Dynamics: A method of using ground reaction forces, the center of pressure, and limb lengths to calculate the moments and forces higher in the body, such as at the ankle, knee, or hip. Limb lengths and lever arms are measured by video synchronized with the force measurements.

Stance phase: The period of time during walking or running in which the foot is touching the ground.

Midstance: The midpoint in the stance phase, as determined by percent of the stance phase. For this study, midstance was selected as the point of zero anterior-posterior ground reaction force.

Swing phase: The period of time during walking or running in which the foot is off the ground, swinging from a toe-off to the next heelstrike.

CHAPTER 2
REVIEW OF LITERATURE

The first purpose of this chapter is to introduce concepts and previous research concerning gait transitions, specifically the human walk-run transition. Therefore the first part of this chapter introduces research on gait transitions in humans, initially from the kinematic and physiological points of view, and then discussing the kinetics of gait. The latter section focuses on the transitions of animals and humans.

The secondary purpose of this chapter is to introduce previous tools and methods which will be used or built upon for use in this experiment. The evolution of the force-measuring treadmill is the primary focus of this section. Some attention is given to differences which exist between overground and treadmill running.

2.1 Locomotion: Kinematic and Physiological Factors

One of the first non-kinetic control parameters investigated for its relation to human locomotion was oxygen uptake. Margaria, Cerretelli, Aghemo, and Sassi (1963) reported a graph of metabolic cost vs. velocity for walking and running (see Figure 2.1). With increasing velocity the walking curve exhibited increasingly positive slope and at 8.5 kph (2.36 m/s) intersected the running curve which was a line with constant positive slope. From their experiment they interpreted that below the speed of 8.5 kph "walking is more economical than running; above it, running becomes more economical" (p. 367).

Menier and Pugh (1968) and Falls and Humphrey (1976) also compared oxygen uptake across velocities of walking and running (see Figure 2.2). Menier and Pugh's measurements of oxygen uptake of male Olympic walkers while walking and running over a range of speeds found an "upward concave curve" (p. 717) for oxygen uptake while walking at speeds up to 8 kph (2.2 m/s). When walking faster than 8 kph, oxygen uptake increased linearly with velocity. When the same athletes were measured while running

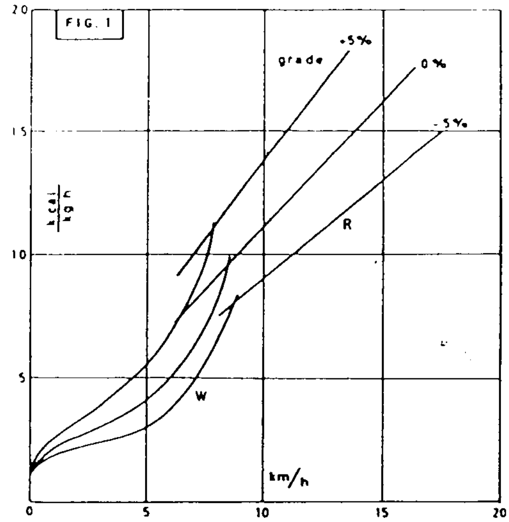

Figure 2.1
Metabolic Cost vs. Velocity:
Margaria, Cerretelli, Aghemo, and Sassi (1963), used with permission.
Energy expenditure in kcal/kg*hr as a function of speed in km/hr in walking (W, lower curves) and running (R, upper straight lines) on a treadmill on the level (0%), uphill, and downhill at a 5% grade.

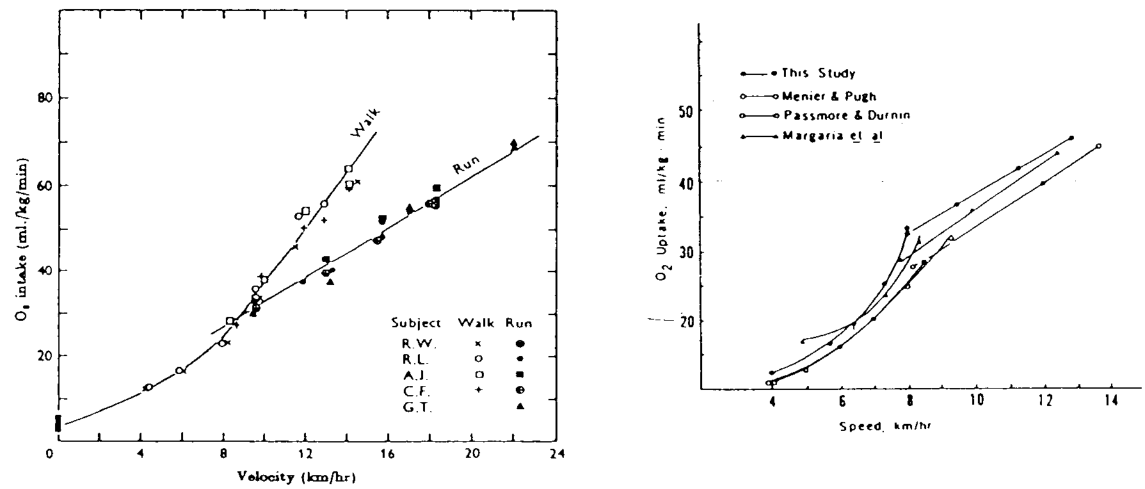

Figure 2.2
Metabolic Cost vs. Velocity:
L: Menier & Pugh (1968): Relation of oxygen intake and velocity in walking and running on a treadmill. Used with permission.
R: Falls & Humphrey (1976): Relationship between energy expenditure and speed of progression in walking and running. Used with permission.

between 8 and 21 kph, their oxygen uptake again increased linearly with velocity, but at half the rate of increase of walking. Above 8 kph, running was determined to have a lower metabolic cost for a given velocity. Similar to the results of Margaria, et al., Menier and Pugh's graph of these relations implies that walking has a lower metabolic cost up to 8 kph, above which running has the lower cost.

Falls and Humphrey reproduced these results with untrained female subjects. Walking followed a similar curvilinear metabolic cost-velocity relationship until approximately 8 kph. Above 8 kph the metabolic cost of running increased linearly. These results provide support for those of Margaria et al. and Menier and Pugh, although demonstrating a slightly lower speed of intersection of the two curves. It was hypothesized that the Olympic race-walkers had trained themselves to walk with a lower metabolic cost, thereby raising the speed at which their curves intersected. From these investigations it has been hypothesized that subjects choose the gait with the least metabolic cost. If true, persons will switch between walking and running at approximately 8 kph.

Thorstensson and Roberthson (1987) questioned the ability of subjects to sense the small variation in energetic demands which occur during slow changes in speed. The investigations described above (except for Margaria et al. for which the increments are not known) used increments of 0.8 kph (0.22 m/s) and the subjects were instructed which gait pattern to maintain. It remained unanswered if the metabolic cost-velocity relationship would result in the same intersection point if subjects were free to choose their own gait and velocity was changed in smaller increments or was changed continuously.

Hreljac (1993b) attempted to fill some of this void by determining the "energetically optimal transition speed" in addition to the preferred transition speed. By varying the treadmill speed by 0.1 to 0.2 m/s increments while subjects were free to choose their gait, the preferred walk-to-run transition (with increasing speed) and preferred run-to-walk transition (with decreasing speed) were measured. The average of these speeds was the preferred transition speed (PTS). Oxygen consumption was measured while walking at

70%, 80%, 90%, 100% and 110% of the PTS and while running at 90%, 100%, 110%, 120%, and 130% of the PTS. A curve was fit to the series of metabolic costs for each gait. The intersection of the two curves was labeled the "energetically optimal transition speed" or EOTS. The two speeds were significantly different (PTS = 2.244 ± 0.125 m/s vs. EOTS = 2.067 ± 0.102 m/s), calling into question the hypothesis that one switches gait patterns to maintain minimal metabolic cost.

Most research regarding the walk-run phase transition has used kinematic data. Nilsson, Thorstensson, and Halbertsma (1985) modeled how people adapt to increasing speed of locomotion. By recording lower body kinematics and EMG over a range of velocities for walking (0.4-3.0 m/s) and running (1.0-9.0 m/s), changes were observed as the subjects adapted to increased speeds. While maintaining a single gait and increasing speed, subjects would increase the frequency and amplitude of their leg movements. Another method of adaptation was to transition from walking to running, observed at approximately 2.0 m/s. Nilsson, Thorstensson, and Halbertsma used their findings to gain perspective on the control system of human gait. They did not identify any control parameters governing the gait transition, but they did summarize methods of adaptation which they related to those of other animals. Finding many similarities, they concluded that "the same basic structure of the stride cycle as in other animals suggests similarities in the underlying neural control" (p.457). With this point of view, knowledge of animal control systems should be related to and tested on human control systems, particularly those of gait.

Thorstensson and Roberthson (1987) measured plantar pressure and duration of the stance phase in 18 men while changing locomotion velocity at one of several constant accelerations (0.05, 0.08, and 0.11 m/s^2). Gait and leg length were analyzed with respect to the speed of the transition between walking and running. The mean transition speed for these subjects was observed to be 1.88 m/s with the range from 1.30 to 2.55 m/s. While they observed a tendency for subjects with longer legs to have higher speeds of transition, they believed this to be partly explained by natural frequency. From their observations they found it "unclear" why gait transitions occurred at speeds which may be easily maintained by either walking or running (p. 211). This discrepancy remains one of the leading reasons why research continues to pursue the identification of possible control parameters believed responsible for this otherwise unexplained transition speed.

Using stringent criteria, Hreljac (1995) attempted to identify kinematic factors which may determine the preferred transition speed. Three criteria tested by searching the literature were: (a) the value of the variable must change abruptly with a change in gait, (b) the value of the variable while running at the PTS must be equal to the value while walking at a lower speed, and (c) the variable must be able to affect proprioceptive receptors which may in turn transmit affective signals to the central nervous system. The fourth criterion was tested experimentally: the variable must be at the same level while walking at the PTS at three different inclinations. Of the velocities and accelerations studied, only maximum ankle angular velocity satisfied these criteria for being a possible control parameter. Maximum ankle angular acceleration did not completely satisfy these criteria. Hreljac hypothesized that "gait transitions are effected to prevent overexertion of the dorsiflexor muscles that perform at or near maximum capacity during fast walking" (p. 669) such as at the preferred transition speed. Unfortunately, he did not list the variables which did not meet all four of these criteria.

In modeling the walk-run and run-walk transitions, Beuter and Lefebvre (1988) suggested that the "mechanisms controlling these transitions can be described by a hysteresis cycle and a small number of parameters" (p.247). In testing trained runners, they observed that when speed is increased, then decreased, the walk-run transition occurs at a higher speed than the run-walk transition. With the addition of weight (14% of body weight), the run-walk transition speed increased substantially while the walk-run transition speed decreased slightly. The additional weight significantly decreased ($p < 0.025$) the difference between the two transition speeds from 0.235 m/s to 0.041 m/s. While two transitions were identified, this does not necessarily imply that the same control parameters govern both transitions. The few or single control parameters which Beuter and Lefebvre suggest to be influencing these transitions have not yet been properly identified using kinematic, kinetic, or oxygen uptake analysis.

2.2 Locomotion: Kinetics

To isolate a human control parameter, Nilsson and his colleagues probably would have suggested testing control parameters already identified in animals. By analyzing the kinetics and metabolic costs of the trot-gallop transition in horses, Farley and Taylor (1991) identified a critical musculoskeletal force level at which the transition took place, despite a corresponding jump in metabolic cost. By having the horses trot and gallop on a treadmill with a force plate mounted underneath the belt, ground reaction forces were measured for each hoof-fall. It was found that peak vertical ground reaction forces dropped by an average of 14% when the horses switched to a gallop. At the preferred transition speed, lever arm lengths and mechanical advantage were the same for the two gaits; thus during the transition, peak muscle and tendon forces dropped proportionately. At the same time that the horses switched gaits, they raised their metabolic cost by 13%. By adding weights (24% of body weight) to the horses, the transitions were reproduced at lower velocities (3.3 vs. 4.1 m/s). During trot, prior to the new transitions, the peak vertical ground reaction forces and corresponding musculoskeletal forces reached the same levels as in the unweighted transitions. While a higher speed would be metabolically optimal for the trot-gallop transition, musculoskeletal forces or peak vertical ground reaction forces were associated with the actual preferred transition speed. Therefore, musculoskeletal force levels may be the critical, and possibly controlling, factor in the transition and should be studied in humans.

Attempting to test whether a kinetic factor may be the control parameter for humans, Hreljac (1993a) measured five different kinetic variables: maximum loading rate, braking and propulsive impulse, and braking and propulsive force peaks. The preferred transition speed was measured for twenty subjects (10 male, 10 female) while unweighted and in two weighted conditions (+10% BW and + 20% BW). These speeds were measured on a motorized treadmill using the same procedure described above (1993b). On a subsequent day, the subjects walked and ran over a force plate to measure the ground reaction forces. Walking was performed at 70%, 80%, 90% and 100% of the PTS while running forces were measured only at the PTS. Of the variables tested, only maximum braking and propulsive ground reaction forces increased their levels up to the PTS, then dropped when running at the PTS. However, these levels increased with the weighted conditions; thus a critical level was not found, indicating that these variables are not control parameters in the walk-run transition. This study did not include any measurement of musculoskeletal forces, prompting further work in kinetics.

2.3 Dynamic Systems Theory

In their chapter "Dynamic Systems: Exploring Paradigms for Change," Thelen and Smith (1994) explain the fundamentals of dynamic systems theory as it relates to the physical sciences. The complexity observed in chemical reactions is discussed and related to the complexity of human movement. For dynamic systems, the behavior may be complex, but it is not random. While a control parameter is scaled, or increased, it may be hypothesized that the relative stability of the associated system will change. With enough perturbation, the collective variable may change from one attractor state to another, often more stable, attractor state. If human locomotion is considered under this theory, gait would be the collective variable with two stable regions, or attractors: running and walking. As possible control parameters (e.g. metabolic cost, speed or musculoskeletal force) are scaled up, the collective variable of gait becomes unstable or uncomfortable; perhaps enough to cause the subject to switch from walking to running. Thelen and Smith explain the method for change in the following passage:

> Systems shift into new forms only as the old forms get shaken up by internal perturbations; these are engendered by changes in the values of parameters to which the system is sensitive. As stable dynamic systems approach such transitions, their growing instability should be detectable by increased measures of variability; as they shift into new stable patterns, variability should again be reduced. (p. 64)

Based on this understanding, a perturbation of a system could be observed by increased variability. If an identifiable parameter is scaled and variability increases before a shift to a new attractor state, then decreases following the shift, it may be interpreted that the system is sensitive to the possible control parameter. Therefore, to assist the process of identifying a control parameter, variability should be measured.

2.4 Measurement Techniques

In humans it is difficult, although possible, to measure Achilles tendon forces directly. Komi (1990) described one *in vivo* method utilizing tendon buckles. Invasive techniques remain as the only direct measures of Achilles tendon force. Such techniques may not be feasible for all experiments measuring musculoskeletal forces. Farley and Taylor (1991) measured the lever arm lengths and vertical ground reaction forces of horses during trot and gallop while unweighted and during weighted trotting. The lever arms were the same length at the points of peak vertical ground reaction force; thus the muscle and tendon forces were also at peak levels. By simultaneously measuring the lever arm lengths and vertical ground reaction forces, Farley and Taylor were able to estimate muscle and tendon forces without invasive techniques.

The implied method of inverse dynamics estimation of tendon force has been highly correlated with tendon buckle measurements in the kangaroo rat (Biewener, Blickhan, Perry, Heglund, & Taylor, 1988). For steady-speed hopping and stationary jumping, inverse dynamics estimations were significantly correlated to directly measured tendon forces with $r = 0.95$ and $r = 0.93$, respectively.

Fukashiro, Komi, Järvinen and Miyashita (1993) compared these techniques in a human subject performing vertical jumps. Achilles tendon force was measured with a surgically implanted buckle transducer and estimated from ground reaction forces measured by a floor-mounted force plate and from joint locations determined from film. For a squatting jump, a counter-movement jump, and hopping, the estimated and measured forces correlated with coefficients of 0.95, 0.96 and 0.99, respectively. However, directly prior to and during early plantarflexion of the two jumps, inverse dynamics overestimated the Achilles tendon force by as much as 40%. For hopping, the estimated values exceeded the measured values by only 8% during the same phase. Overall, human Achilles tendon forces may be estimated with reasonable accuracy using indirect methods.

Previous work has used floor-mounted force plates to measure the ground reaction forces for inverse dynamics (e.g. Bresler & Frankel, 1950; Fukashiro et al., 1993; Wells, 1981; Winter, 1990, p. 90) while a limited number of studies have performed similar measures with a treadmill-mounted force plate (Farley & Taylor, 1991). To allow multiple trials and precise experimental control of locomotion speed, force plate methods are difficult. Therefore, a treadmill-based device should improve such experiments. Such a device would probably decrease the experimental time for each subject, thus decreasing any fatigue effect.

Previous treadmill-based force measurement systems have been built with various degrees of success and cost. Kram and Powell (1989) mounted a force-plate in a treadmill, requiring a whole new treadmill apparatus to be built from conveyor belt parts. A similar treadmill, also of Kram's design, was used by Farley and Taylor (1991).

Others have tried using existing treadmills and placing them on force measurement devices. E. Hennig placed a Kistler force plate under each corner of a Woodway treadmill (personal communication, July 8, 1994). Apparently the treadmill itself, designed to absorb shock while running, was flexing with each heel-strike. This resonance resulted in forces approximately one body-weight higher than expected, even when the sinusoidal flexing of the treadmill was subtracted.

Johnson (1992) tried a similar approach by mounting a treadmill on four strain-gage-based force measurement devices. He observed a substantial amount of noise in the signals, primarily due to mounting one transducer directly below the motor. He attempted to minimize this noise by integrating a 50 Hz low pass filter into his analog signal conditioning.

Kram and Powell (1989) attempted to minimize vibrations of the treadmill motor from degrading the force signals by placing the motor on a separate surface from the treadmill bed and connecting the two with a flexible coupling. Their treadmill had a signal-to-noise ratio of ~100:1 and a natural frequency of 160 Hz. While their signal-to-noise ratio is usable for most research, it is much worse than force plate noise which may be better than 750:1. Research treadmills must be powered; these drive mechanisms will generate vibrations which may easily introduce noise into the force signals. Thus steps must be taken to eliminate such vibration and the resulting noise.

While the present project has been in progress, a commercial treadmill has become available from Kistler Instrument Corporation (Fuglewicz & Klavoon, 1994). In making this system, Kistler stiffened an existing treadmill and mounted two force plates under the belt, allowing vertical ground reaction forces to be measured and enabling individual foot loadings to be separated. Even with these steps, the Kistler device measures VGRF's approximately 10% higher than expected--probably due to treadmill flex (R. Redd, personal communication, September 24, 1994). The device also is able to calculate the center of pressure accurate to 0.5 cm (R. Redd, personal communication, May 16, 1995).

2.5 Overground vs. Treadmill Locomotion

Because this experiment was conducted on a treadmill instead of overground, differences between the two must be discussed. Treadmill walking and running are known to be somewhat different from overground locomotion. To minimize kinematic differences when conducting an experiment on a treadmill, the usual procedure is to allow each subject an adjustment period of 15-30 minutes (Charteris, 1978; Schieb, 1986; Wall & Charteris, 1980; Wall & Charteris, 1981).

Nelson, Dillman, Lagasse, and Bickett (1972) compared overground and treadmill running at three speeds and three slopes for 16 experienced runners. Even after multiple practice sessions of treadmill locomotion, the researchers observed treadmill running to have longer foot contact time, lower vertical velocity, and less variability in both horizontal and vertical velocity.

Nigg, De Boer, and Fisher (1995) compared running on three treadmills of different sizes to running overground. Twenty-two subjects, evenly divided between runners and non-runners, ran on each treadmill "until they felt comfortable and did not require assistance of the railings" (p. 99). Kinematics were measured on each subject as he or she ran on each device at 3.0, 4.5, 5.0, and 6.0 m/s. They found that there were kinematic differences, such as landing with the foot in a flatter position while on a treadmill. There was a substantial amount of unexplained variability between treadmills and overground. It should be expected that each treadmill will be different from overground running, in part related to the stiffness and natural frequency of the treadmill. Possible kinetic differences have not been explored--in part due to the inability to do so reliably.

2.6 Midstance Forces

To quantify running characteristics across speeds, Munro, Miller, and Fuglevand (1987) measured ground reaction forces from 20 subjects as they ran across a force plate over a range of speeds. Speeds were between 2.5 and 5.5 m/s. Each subject performed 30 to 40 trials. From these data, normative information was calculated for each running speed. Variables quantified were impact peak, loading rate, maximum thrust, decay rate, average vertical ground reaction force, change in vertical velocity, braking impulse, propulsive impulse and stance time (p. 147). For the present study, midstance was defined as the moment of zero anterior-posterior shear or ground reaction force. At 3.0 m/s, the slowest speed they quantified, Munro, Miller, and Fuglevand measured zero fore-aft shear to be at 47.1 ± 3.1% of stance time (p. 149).

Similar variables were measured by Kinoshita and Bates (1983) in their study of walking with various loads on the body. Five subjects walked over a force plate 10 times for each of the conditions tested. For the weighted conditions, the subjects wore either a backpack system or a front-back system. The latter device had equal amounts of weight on the front and rear of the system. Three loadings were tested: zero load, plus 20% of body weight, and plus 40% of body weight. For the latter two conditions, both systems were tested. Means and standard deviations were calculated for many variables from the vertical, anterior-posterior, and medial-lateral ground reaction forces. These variables included forces of given landmarks, percent stance of these landmarks, and impulses. Zero anterior-posterior force was calculated as percent stance time. For normal unweighted walking, this point was at 49.97% of stance, with one standard deviation equal to 2.90%. With 20% of body weight carried in the front-back system zero anterior-posterior force was measured at 50.76 ± 2.07% of stance (p. 579). These points were later in the stance phase than the points of relative minimum vertical ground reaction force (44.70 ± 2.00% and 46.14 ± 2.36% respectively for the two conditions). These forces were measured in Newtons per kilogram of body weight. For the unweighted condition, the minimum force was measured to be 7.22 ± 0.26 N/kg. With an additional twenty percent of body weight in the front-back system, the minimum force was quantified as 8.30 ± 0.21 N/kg (p. 578).

2.7 Summary

Despite significant amounts of research on the human walk-run gait transition, control parameters have not been satisfactorily identified. To date, the majority of investigations have focused on kinematic analysis. Data from horses indicated a control parameter in musculoskeletal forces; once at a certain force level the horses switched from a trot to a gallop. Some recent work has focused on the kinetics of the walk-run transition in the hopes of identifying a control parameter. Hreljac (1993a) analyzed five aspects of ground reaction forces near the transition speeds, but did not identify a control parameter. Direct or indirect measurement of human musculoskeletal forces remains to be accomplished relative to the walk-run transition. With the understanding provided by dynamic systems theory, to identify a possible control parameter, variability should be measured in the collective variable and the control parameter while the parameter is scaled. If variability increases before a transition only to decrease following the transition, the manipulated parameter is indicated as a control parameter.

A treadmill-based vertical ground reaction force measurement system was required for this experiment. Previous efforts indicated that the new treadmill needed to be rigid to minimize bed flex during running and suggested that motor vibration be isolated to provide clean and accurate measurements.

CHAPTER 3
METHODS AND PROCEDURES

This chapter discusses the subjects, apparatus, experimental design and data analysis procedures for the study. Pilot studies conducted to investigate the validity of the apparatus and procedures are also outlined.

3.1 Subjects

Ten male and female subjects were recruited for this study. Subjects were from the general university population and were required to be between 18 and 25 years of age. Subjects were also required to have a fitness level capable of completing the moderate exercise involved in the study. More specifically, recreational runners involved in moderate frequency and duration of training (total 1-4 hours per week) were recruited. The subjects were free of any diseases or injuries which could have affected their walking or running gaits. In addition, to maintain similar patterns of musculoskeletal loading, these subjects were required to have a rearfoot-striking running gait.

3.1.1 Questionnaire

A questionnaire was used as a screening device to insure that subjects participated in 1-4 hours of physical activity per week and did not have any disabilities which could have affected gait or coordination. This questionnaire is in Appendix A.

3.1.2 Subject Characteristics

The heights and weights of the 10 subjects used in this study appear below.

Table 3.1
Subject Heights and Weights

Subject Number	Weight (kg)	Height (cm)
1	57.6	170
2	52.8	155
3	77.1	183
4	61.0	165
5	64.6	174
6	61.7	165
7	80.1	184
8	57.2	163
9	59.0	167
10	68.9	180
Average	64.0	171

3.2 Procedures

The procedures, risks, benefits, and the ability to withdraw without prejudice were explained to each subject prior to testing. Subjects were required to sign an informed consent form prior to beginning the experimental protocol. The Oregon State University Human Subjects Review, Consent Form, and Institutional Review Board approval are included in Appendix B.

On a day prior to testing, each subject signed the consent form and was given 30 min to adjust to walking and running on a treadmill under both the weighted and unweighted conditions. Thirty minutes was judged to be sufficient time to adjust to treadmill locomotion and to minimize kinematic variations compared to overground locomotion (Charteris & Taves, 1978; Schieb, 1986; Wall & Charteris, 1980, 1981). In addition, on test day, but prior to testing, each subject was allowed any needed practice time. The length of this practice time was determined by the subject, but was a minimum of 5 min.

On the testing day, before actual testing began, the subject's height and weight were measured in the Anthropometry Laboratory. Three tests were performed on each subject, with and without additional weight. The tests were (a) to determine the preferred transition speed (PTS), (b) to measure forces while walking over the experimental range of speeds, and (c) to measure forces while running over the range of speeds.

The first test was to determine the PTS for the subject. The speed of the treadmill was accelerated in increments of 0.2 m/s from 1 m/s to a maximum speed of 3 m/s. Each speed was held constant for 30 s before the speed was changed. See Figure 3.1 for a graphical representation of this test. During this test each subject was allowed to transition freely between walking and running. Subjects were instructed: "If it feels more comfortable to run, then run. If it feels more comfortable to walk, then walk."

Two trials of acceleration were performed for each subject. The PTS was defined as the speed at which the subject switched to a new gait and maintained the gait for the entire 30 s (Hreljac, 1993a). After the transition was complete according to the above requirement, three increments of increased velocity were performed. When the highest speed was recorded, the treadmill was stopped and the subject given the opportunity to rest. Once the subject was prepared to perform the next trial, the treadmill was started at the speed three increments below the PTS. From this level, the speed was increased through the PTS to the speed three increments faster. The preferred transition speeds were recorded for each acceleration (walk-run) trial with the average labeled the PTS for that condition. Before the next test each subject was given the opportunity to rest.

For the second test each subject was asked to maintain a walking gait through a single acceleration cycle (PTS - 0.6 m/s to PTS + 0.6 m/s). Force measurements were taken during this test to determine musculoskeletal forces while walking.

Test 3 was the same as Test 2, except in this case each subject was asked to maintain a running gait. Musculoskeletal forces were measured.

This set of three tests (total of four acceleration cycles) were repeated with each subject wearing a weight vest of approximately 15% of his or her body weight. This weight was chosen to be substantial enough to cause a change in the preferred transition

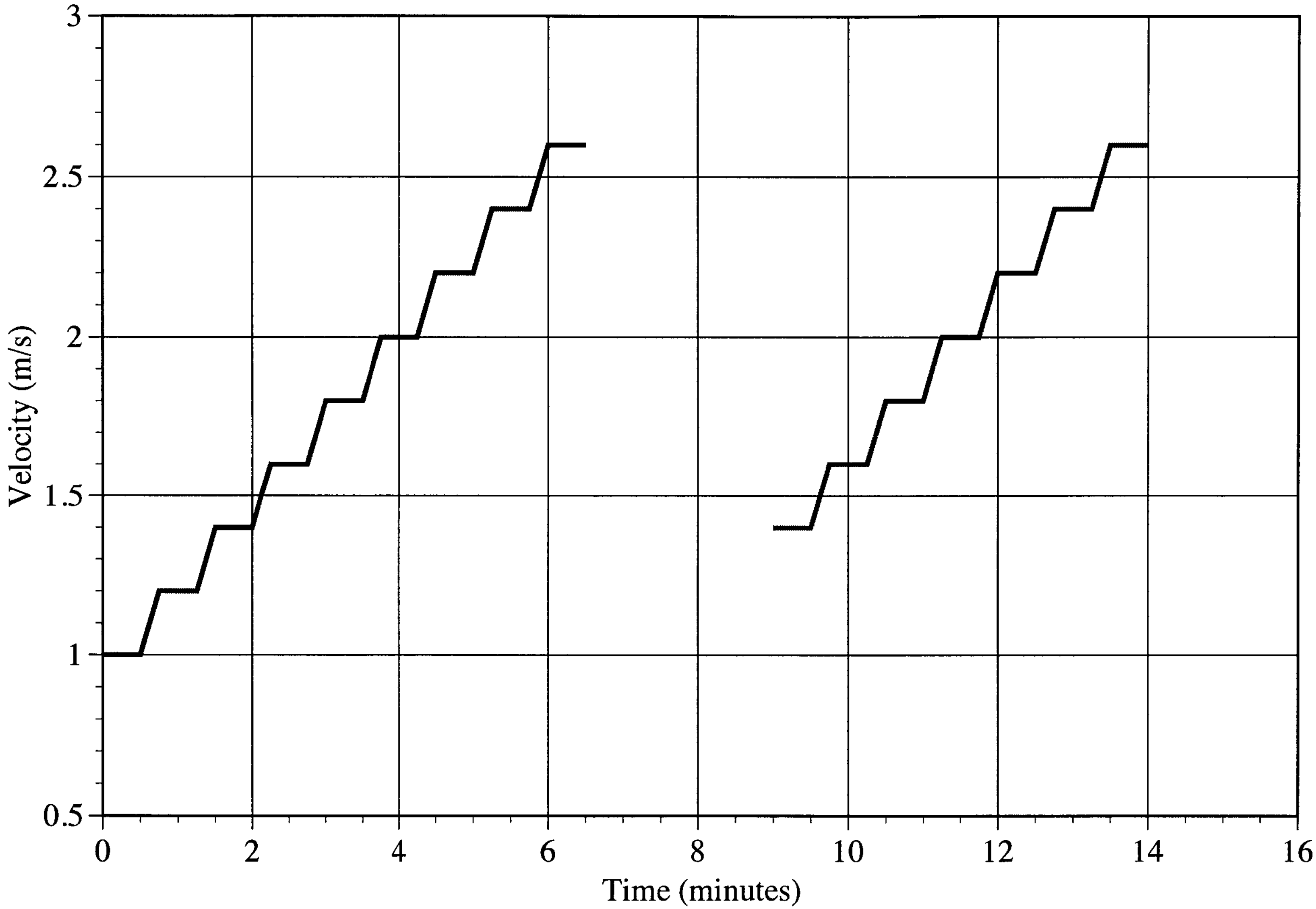

Figure 3.1 Test 1 Procedure: Velocity vs. Time

speed as observed by Beuter and Lefebvre (1988). The weight and its distribution on the body was intended to minimize changes to walking and running kinematics. Administration of the two testing conditions (weighted and unweighted) was counterbalanced to eliminate effects of order.

3.3 Apparatus

To measure an hypothesized control variable governing gait transitions, the subjects' vertical ground reaction forces were recorded while treadmill speed was systematically varied. This required an apparatus and several computer programs to perform the calculations.

3.3.1 Hardware

As discussed earlier, ground-reaction-force-measuring treadmills have been built for individual research projects and recently one has become available commercially. Problems encountered have included excessive noise, believed to be due to the drive motor, and higher-than-expected vertical ground reaction forces, possibly from the treadmill bed flexing with each foot-strike. Some projects have involved building a dedicated treadmill or using expensive hardware. The equipment needs of this project required building a VGRF-measuring treadmill using an existing treadmill and low-cost, non-dedicated hardware. The resulting natural frequency was to remain relatively high, the bed was required to not flex with each foot-strike and noise from the treadmill itself was to not degrade the signals.

To meet the stated goals, an existing treadmill (Quinton Q55) was modified. See Figure 3.2. The motor unit was disconnected from the treadmill bed except for a single drive belt. While the treadmill bed rested on the floor, the motor unit sat on a Kistler force plate, vibrationally isolated from the floor and the force-measuring bed. The motor was

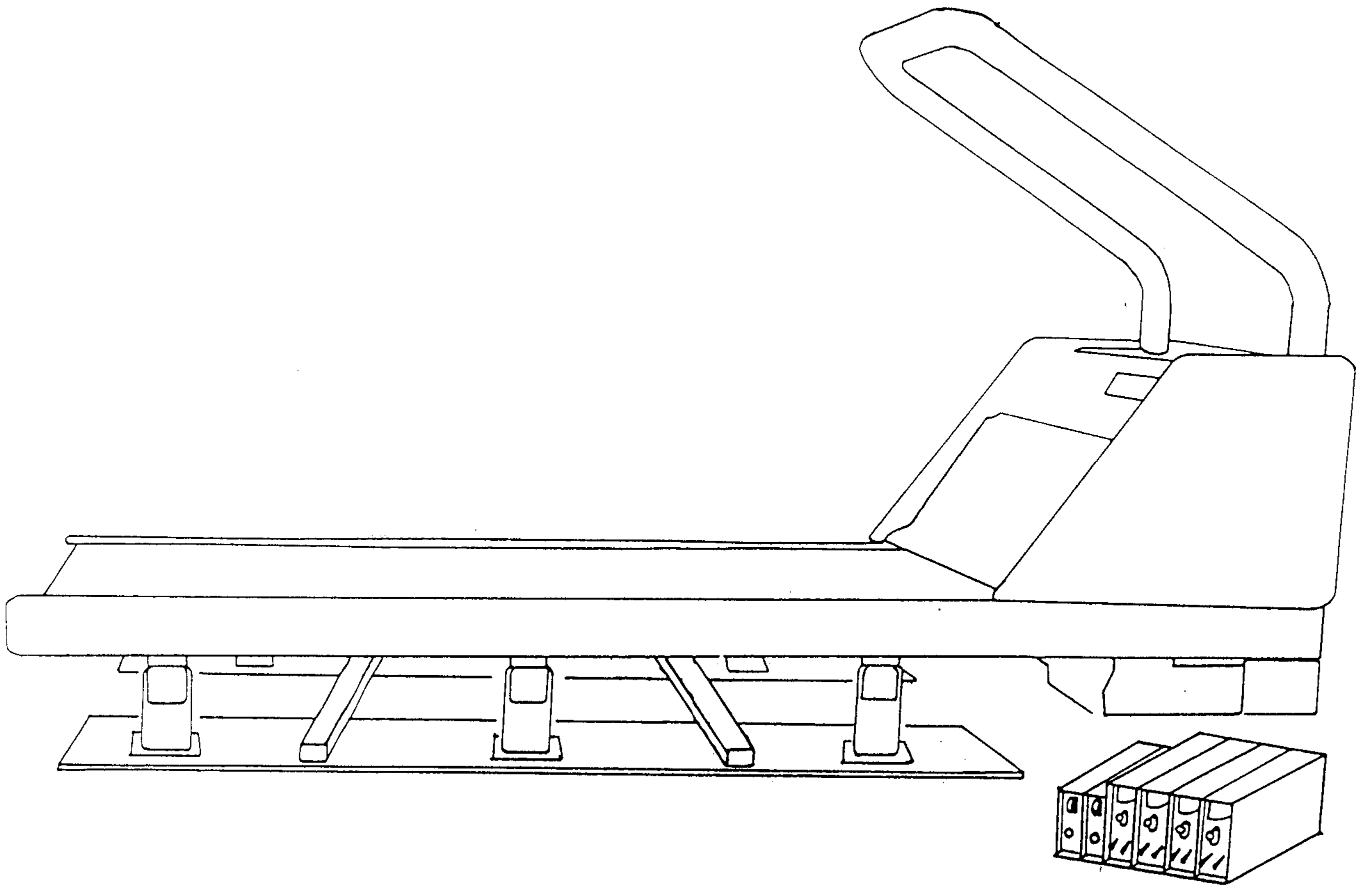

Figure 3.2
Treadmill Apparatus

aligned by pins to prevent motion in the horizontal plane, but by resting on a bed of foam on the force plate the motor unit was free to move within a limited vertical range.

The bed was supported by six rigid supports with a piezoelectric transducer (PCB Piezotronics 208A03 or 208A02) at each point of loading. The specifications for each of these transducers are in Table 3.2. To maximize rigidity and force transmission, the loading points were at the front, middle, and rear of the treadmill bed. The points of contact were at frame junctions in the treadmill bed, so as to minimize the distance along the frame that forces were transmitted. Each transducer was topped by a rounded impact head, assuring point loading. Lower-ranged, more sensitive sensors, were on the rear two supports because of the lower relative loads at these points.

Table 3.2
Transducer Specifications

Position	Left Front	Right Front	Left Middle	Right Middle	Left Rear	Right Rear
Channel	A	B	C	D	E	F
Model	208A03	208A03	208A03	208A03	208B02	208A02
S/N	11541	11551	11552	11553	12456	8766
Range (lbs)	0-500	0-500	0-500	0-500	0-100	0-100
Sensitivity (mV/lb)	10.41	10.19	10.33	10.22	50.89	53.3
Input TC (sec)	>2000	>2000	>2000	>2000	>500	500
Rise Time (μsec)	10	10	10	10	10	10
Natural Frequency (kHz)	70	70	70	70	70	70

Each support pillar was topped by a triangular plate to which a transducer was mounted. See Figure 3.3. The height of the plate above the pillar was adjustable using three bolts on the corners of each plate and a large, lockable center pin which supported the plate directly below the transducer. An alignment pin was welded on the upper corner of each of the four front pillars. These pins extended 2 in. (5.08 cm) above the transducers and were fitted with Delrin housings. The pins were aligned with corners of the treadmill frame so as to prevent movement of the bed in the horizontal plane. The Delrin housings were the contact points between the pins and the frame. The use of Delrin was to minimize any friction in the vertical direction, allowing all vertical loading to be measured by the transducers.

The pillars were attached to a steel frame in such a way as to allow adjustment in the horizontal plane. With the alignment pins in their appropriate corners and the transducers at the frame junctions, bolts were tightened to prevent any horizontal movement. See Figure 3.4 for an illustration of the frame.

Each transducer was powered by its own signal conditioning unit (a PCB Piezotronics 484B or 484B02). The specifications and settings for these units may be seen in Table 3.3. Each 484B was DC-coupled to utilize the 2000-s time constant of the 208A03 transducers. Zero-output voltage was clamped at approximately 0.5 V. The 484B02's, used with the lower range 208A02's, were AC-coupled, providing a 1000-s time constant which was longer than the constant of these transducers. The zero-output was held positive by a voltage-clamping circuit. The resulting unity gain signals were then sent to a 12-bit analog-to-digital conversion board (Keithley-Metrabyte DAS-16) and microcomputer for data collection.

The design of the treadmill modifications addressed concerns of bed flex and signal noise reported in the literature. The modifications prevented bed flexing and motor noise from degrading the force signals, demonstrated by a signal-to-noise ratio as good as 225:1. To calculate this ratio, signal was measured from the peak forces during running

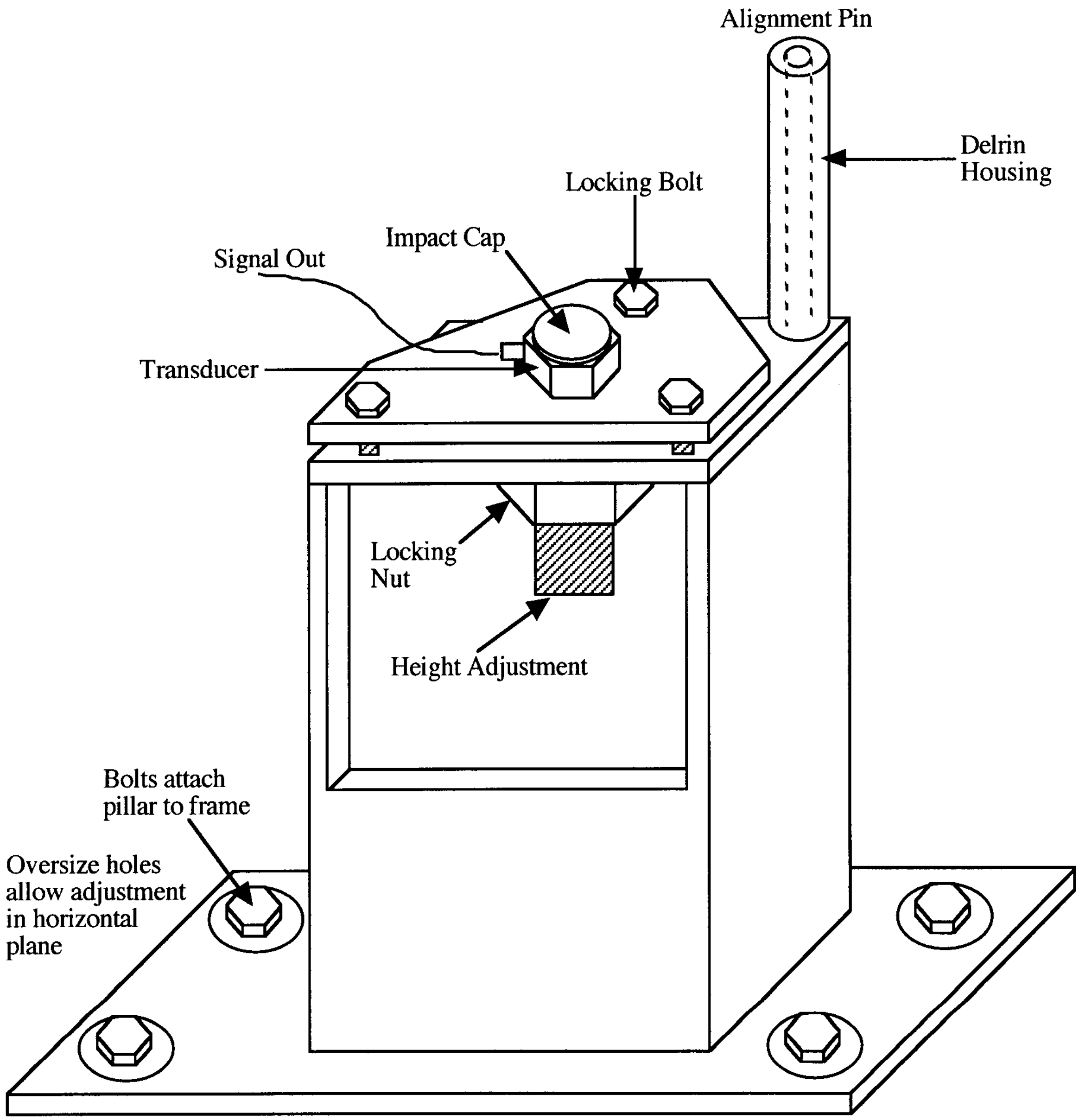

Figure 3.3
Support Pillar

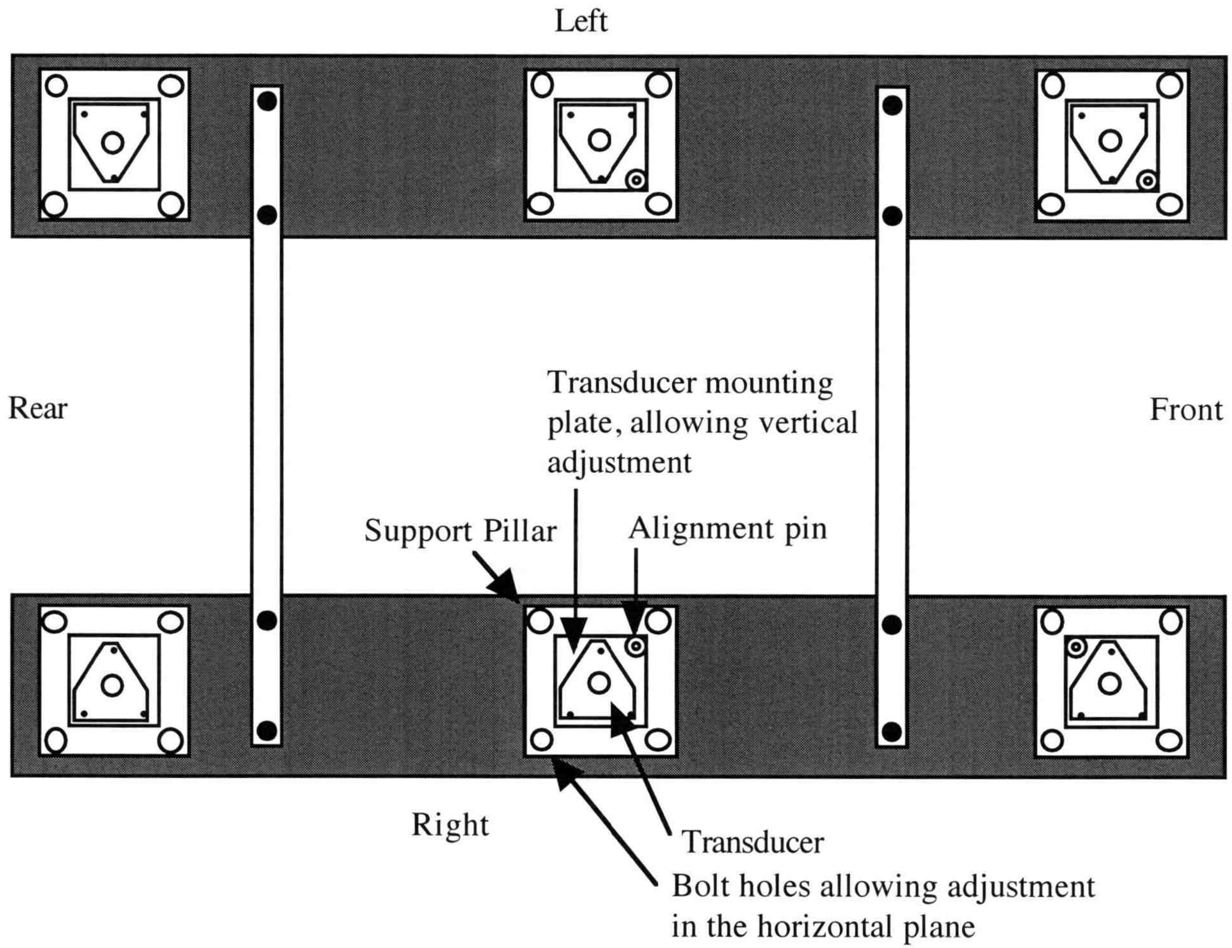

Figure 3.4 Support Frame

Table 3.3
Signal Conditioner Specifications and Settings

Position	Left Front	Right Front	Left Middle	Right Middle	Left Rear	Right Rear
Channel	A	B	C	D	E	F
Signal Conditioner	484B	484B	484B	484B	484B02	484B02
Coupling	DC	DC	DC	DC	AC	AC
Bias (V)	11	11	11	11	11	11
Time Constant (sec)	>2000	>2000	>2000	>2000	1000	1000
Clamping	On	On	On	On	Active	Active

(~1800 N), and noise was measured during the flight phase (±8 N). In addition, a "drift" was noted in the flight phase of running during a pilot study. Drift was quantified as the drop in total force during flight (~35 N), resulting in a 50:1 signal-to-drift ratio. Initially this drift was thought to be due to the characteristics of the transducers and signal conditioning chosen. If true, longer time constants should lower any drift. An alternate hypothesis was that this drift was actually the result of alternating tension in the drive belt. In some graphs the "drift" appeared to be low frequency noise, perhaps from improper drive belt tension. In follow-up tests, particular care was taken to achieve the proper alignment of the drive belt. The result was clean flight phases, free of drift. While some improvements may have been made to the time constants of the transducers, to eliminate or minimize this "drift" it was more important to achieve proper tension in the drive belt. The six-support structure further stiffened a relatively rigid bed. When the bed was hit with a mallet, the ringing of the system was measured, resulting in a relatively high natural frequency of ~275 Hz. Results of pilot studies demonstrated great similarity to expected

VGRF curves. For these data see the individual channels and summed signals in Figure 3.5. The total VGRF curves exhibit the quickly increasing force of the impact peak, the subsequent dip in force, and then the rounded active peak. Between strides the flight phases may be observed with approximately zero force.

A speedometer developed by Sports Medicine Industries (St. Cloud, MN) monitored the treadmill speed. A small photo-electric diode mounted under the rear roller detected each of 9 retro-reflective markers evenly spaced on the 56.1 cm of roller as they passed by. By knowing the bed length, the number of markers, and by counting the number of markers passed in front of the sensor during a five-second period, speed was calculated with a resolution of 0.02 kph (0.0056 m/s). See Figure 3.6 for the diagram of the signal paths.

3.3.2 Computer Programs

Separate programs collected, converted, and filtered the data from the 6 force channels. The sum of the 6 force channels was the total vertical ground reaction force (VGRF). A separate program to be discussed in Section 3.4.2 was used to analyze the resulting force curves. The general flow of data through the computer programs is diagrammed below in Figure 3.7.

Collection was completed using a program written in Microsoft Visual Basic for DOS Version 1.00. The program "VBDTMDMA.EXE" collected data from the treadmill at 600 Hz per scan across all 6 transducers (effective frequency of 3600 Hz). Ten seconds of data were recorded in digital form to a data file for subsequent analysis.

After collection, the conversion program "CNVTVBD.MAK," written in Microsoft Visual Basic 3.0, batch processed the files by subject. The program converted digital units to force units (Newtons) using each transducer's calibration factor, applied the appropriate second-order Butterworth filter, and wrote the resulting data to a text file. For the data collected in this experiment, the cutoff frequency was set at 60 Hz. Over a set of trials, the

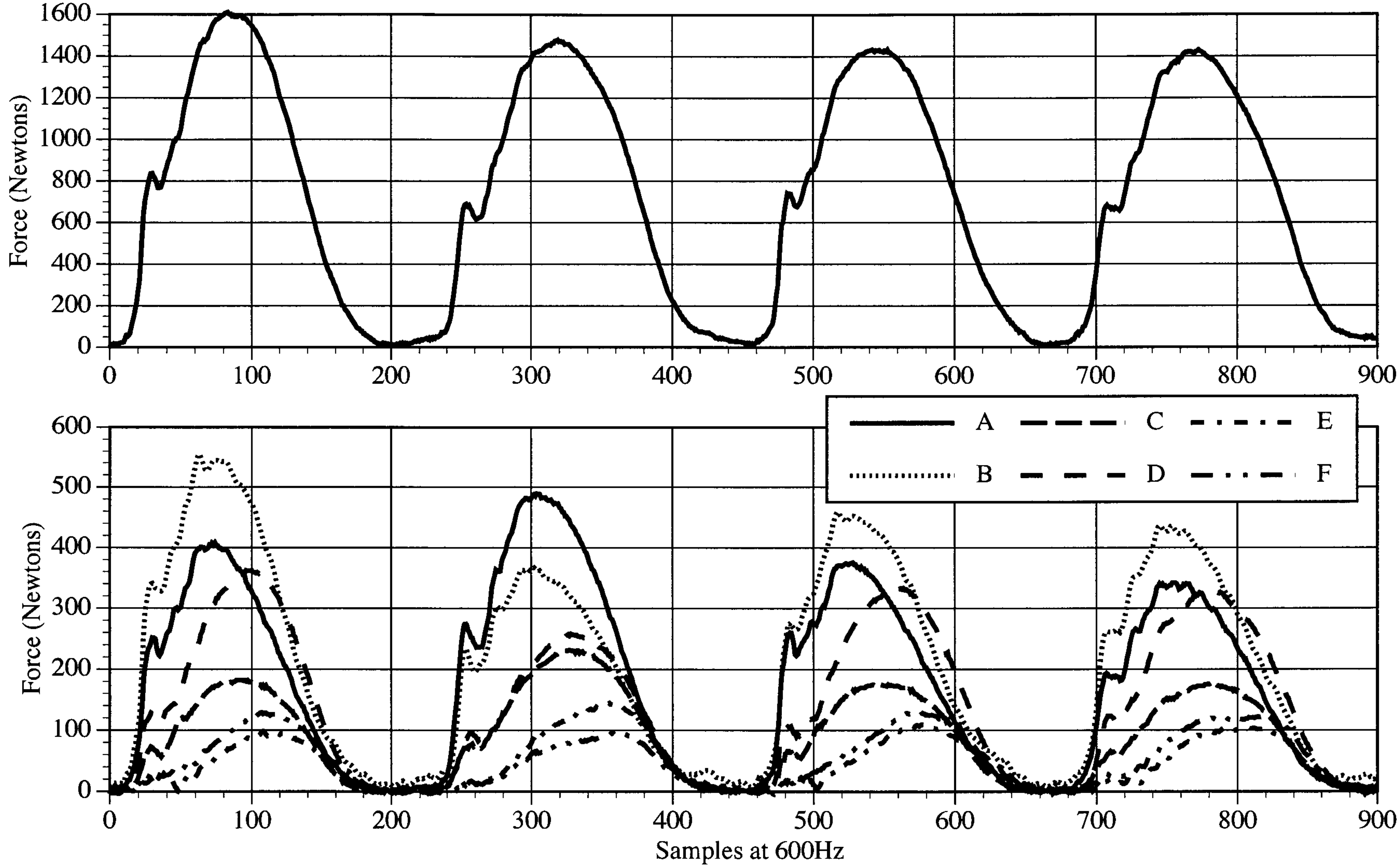

Figure 3.5 Summed (top) and Single Channel (bottom) Vertical Ground Reaction Force Signals from Running

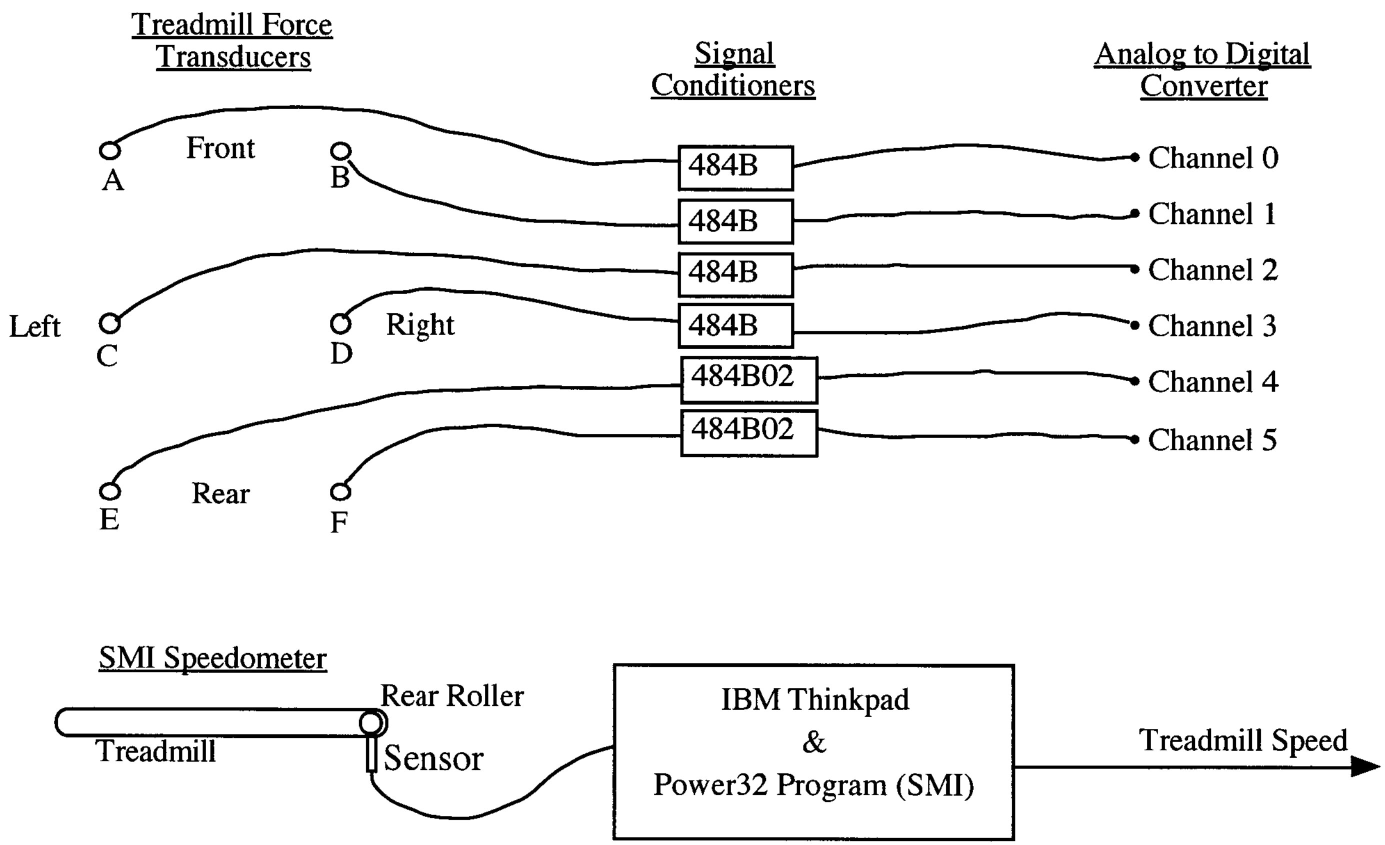

Figure 3.6 Signal Paths

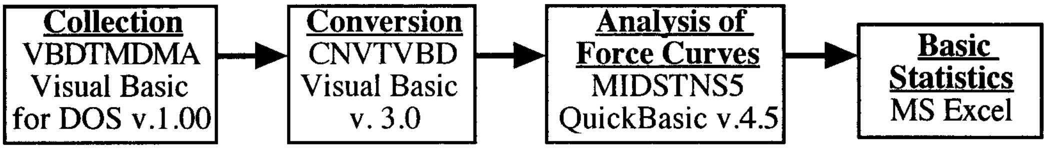

Figure 3.7
Data Flow

force output from the piezoelectric transducers would drift, or decline slightly in magnitude. In order to calculate a correction factor for each trial, the conversion program also created a file of the times at which the original data were collected. These times were then multiplied by the drift rate to determine the amount of force to be added to each file to correct for any drift.

3.4 Data Analysis

3.4.1 Force Analysis Software

The analysis and associated software had to perform the tasks of providing an accurate zero-force level, selecting the midstance force for each stride, and compensating for any signal drift due to the piezoelectric crystals. To achieve an accurate zero level, a "pre-zero" file was calculated and subtracted from each file. In addition, the flight phases of running were forced to zero. Midstance was selected as a percent of stance (50.37% and 47.10% for walking and running, respectively), after having marked the heelstrike and toe-off points for each stride. Drift compensation provided a correction factor which counteracted any loss of signal output from the piezoelectric transducers. After zero-levels were achieved and any drift compensated for, the forces at the selected midstances were considered accurate.

Once a subject's warm-up was complete, the treadmill was vacated, the belt stopped, and a file sampled with no load on the bed of the treadmill. This "pre-zero" file was later averaged over the 10 s to determine the initial offset for each of the six channels. Once the pre-zero was taken, testing for that condition began. A new pre-zero file was recorded immediately preceding the start of each condition or series of tests.

The initial offset values were recorded and input into the analysis program prior to analyzing one of the text files. Proper removal of signal offset was necessary to provide an accurate measure of medial-lateral center of pressure as well as to yield accurate vertical ground reaction forces.

Each running trial was checked for its force during flight. Knowing that flight phases should be measured as zero force, two flight phases were averaged for each of the highest speeds for a set of trials. These trials were chosen due to the distinct flight phase recorded at high speeds. For the running trials this calculation corrected the flight phases to within 5 N of baseline. If the flight-phase calculation proved to be different from the pre-zero value, the flight-phase calculation was used to obtain zero force during flight.

The analysis program, "MIDSTNS5.BAS," was written in Microsoft QuickBASIC v.4.5 to graphically display the converted data, allowing events to be selected. The screen had two graphical displays: one for total vertical ground reaction force and one for medial-lateral center of pressure. A cursor could be moved simultaneously on the two displays. At the base of the screen were displayed the name of the file being analyzed, the sample which was selected by the cursor, and the force and medial-lateral center of pressure of that sample. From the two screens and the numerical displays, each heelstrike and toe-off was marked. The program displayed the label for the most recently picked point.

For running, the right and left stance forces could be seen separately with a flight phase in between. In walking, however, the double support phase made picking events, especially toe-off, more difficult. To confirm that the stances were properly marked, several files were analyzed using a spreadsheet. Allowing greater flexibility than the analysis program, this environment allowed the forces for right and left feet to be separated, verifying the endpoints of stance (Davis & Cavanagh, 1993).

To correlate vertical ground reaction force with musculoskeletal force, the vast majority of force exerted on the body by the ground was required to be in the vertical direction. To meet this requirement, anterior-posterior force was to be at zero. While the treadmill could measure the vertical component of ground reaction force, the anterior-posterior component could not be measured. Thus zero anterior-posterior force was estimated as a percent of stance time, based on previous studies.

The analysis program used the endpoints of stance to calculate stance time and then select midstance. Kinoshita and Bates's (1983) points of zero AP force in walking were averaged for the weighted and unweighted conditions to pick midstance at 50.37% of stance time. For running, Munro, Miller, and Fuglevand (1987) found zero AP force at 47.1% of stance. Using these values, midstance was calculated, and the corresponding force labeled midstance vertical ground reaction force ($VGRF_M$). The program wrote these data to a results file. For each trial the end results contained the file name, number of cycles, sample numbers for each right and left heelstrike, midstance, and toe-off, as well as the vertical ground reaction force at each midstance.

When a series of files was recorded over several minutes, it was expected that the transducers would register some drift. The amount of drift was quantified by calculating any erroneous change in subject body-weight from the first to the last trials of a series. With initial transducer offsets removed, an integral number of body weights were marked in the first file. Picking obvious landmarks such as maximum VGRF in running or heelstrike in walking, the maximum possible number of cycles was selected. The same procedure was completed for the last file of the series. Knowing that an average of an integral number of cycles should be equal to body weight, and that body weight should not change within a single session, the program calculated amount of drift, or signal loss, between the first and last files. A drift rate, or slope, was calculated as the signal change divided by the number of seconds elapsed between the creation of the first and last files. The time of collection of each file was known from the file created by the conversion program. Using the drift rate and the time of collection of each file, a compensation factor was calculated and subtracted from each of the midstance forces for that file.

To confirm that the analysis program had correctly picked off the midstance forces, the output files were checked by hand. For a given trial, the forces were expected to be similar across cycles, while sample numbers were required to increase with heelstrike, midstance, and toe-off of each consecutive stride. Some erroneous points were found, marked, and checked using the analysis program. If these points were indeed in error, they were eliminated from the analysis. These program errors did not substantially affect the number of midstance forces used in the statistical analyses. While most files had more than nine strides for each side per file, a few had less than seven. These small numbers were due to the subject switching between walking and running at the highest walking speeds.

3.4.2 Force Analysis

It was assumed that the anterior-posterior ground reaction forces would be close to zero at midstance. Knowing that at midstance all force is vertical and that muscles are only active in pulling on the Achilles tendon (Basmajian & De Luca, 1985, p. 350), a direct relationship may be made between the vertical ground reaction force and the Achilles tendon force. Because the invasive techniques required to directly measure Achilles tendon force were not possible for this study, the midstance vertical ground reaction forces were measured and subsequently analyzed.

The data from Tests 2 and 3 were used to plot $VGRF_M$ versus speed of locomotion. By having data for both walking and running over the whole range of speeds, the dynamics of force could be observed for gaits which were otherwise not preferred. The resulting graph for each subject illustrated the $VGRF_M$ curves for walking and running for each condition and over the whole range of speeds.

To compare these curves between subjects, the force and velocity axes were normalized. For both the weighted and unweighted conditions, force was normalized to body weight, and velocity was translated to the preferred transition speed. The two PTS's were placed at the same point on the x-axis and the other speeds plotted at increments of speed above and below the PTS (i.e. PTS, PTS ± 0.2 m/s, PTS ± 0.4 m/s, and PTS ± 0.6 m/s). In the case of a subject whose PTS was calculated to be between two of the measured speeds, the lower of the two adjacent speeds was used for this normalized graph and the associated statistical analyses. Data points for all subjects were averaged for each gait, condition, and speed, resulting in a generalized description of the force dynamics before and after the preferred transition speed in both weighted and unweighted conditions.

Individual variability in the force measures was quantified using the same normalization procedure. In this case, the standard deviations of the $VGRF_M$ for each subject, condition, and speed were calculated and aligned at the preferred transition speed. When averaged across subjects, the mean variabilities were plotted and statistically analyzed.

To test if there were significant differences in force or variability at speeds below, above, or at the preferred transition speed, paired *t*-tests were performed on the forces and standard deviations for each side. The values at PTS - 0.2 m/s, PTS, and PTS + 0.2 m/s were compared in three pair-wise comparisons. The level of significance was adjusted using a Bonferroni *t*-Procedure.

For each individual, the preferred transition speeds (Test 1) and the corresponding midstance vertical ground reaction forces (Tests 2 and 3) were used to calculate the differences between the weighted and unweighted conditions (ΔPTS and $\Delta VGRF_M$). Hreljac (1993a) found it appropriate to pool results for both genders. After an initial check of the means, pooled results were used in this study as well. The ΔPTS and $\Delta VGRF_M$ values were plotted on a force-versus-velocity graph. After overlaying all subjects' data the general spread was analyzed. If the data were scattered randomly about the $\Delta VGRF_M = 0$ line, but at various levels of ΔPTS, then the hypothesis would be correct: midstance vertical ground reaction force is probably a control variable in the walk-run transition. The spread of the data about the $\Delta VGRF_M = 0$ line compared to the $\Delta PTS = 0$ line demonstrated the relative influence on the transition by midstance vertical ground reaction force and velocity.

Independent *t*-tests compared the velocity and force differences for the subjects tested. One-at-a-time confidence intervals were calculated to quantify the spread of the data. If the walk-run transition occurred at a critical level of $VGRF_M$, but not at a critical speed, the zero value would lie within the confidence interval for $VGRF_M$, but not within that for ΔPTS. The levels of influence were also determined by transforming the values for both variables to *z*-scores and calculating a confidence ellipse around the data (Johnson & Wichern, 1992). This method of integrating the two variables into a simultaneous calculation of a confidence area resulted in a more specific space than the intersection of the two independently calculated confidence intervals.

3.5 Pilot Studies

The pilot studies which were performed have been divided into the categories of treadmill, speedometer, and estimation of sample size.

3.5.1 Treadmill

To establish the validity of the treadmill apparatus, many individual pilot studies were performed and are summarized in Table 3.4. The results from these studies appear in Appendix C: Treadmill Validation. These pilot studies utilized the collection and conversion programs designed for the research study. The analysis program written to pick midstance forces was used in the last pilot test which quantified repeatability between trials. Thus the treadmill pilot studies validated both the hardware and software.

Table 3.4
Treadmill Pilot Studies

Pilot Study	Quantity(ies) Measured	Belt: On or Off	Repeated after moving treadmill?
Static loads	Force	Off	Yes
Comparison to force plate	Peak impact force, active peak force, rise rate, contact time, total impulse	On	No
Subject reliability	Peak impact force, active peak force, rise rate, contact time, total impulse	On	Repeat without moving TM
Point loads	COP	Off and On	Yes
Dynamic point loads (force hammer)	Force	Off and On	No
Constant moving force (1 BW)	Force and COP	On	No
Experimental variability and repeatability with subjects	Midstance force, difference between trials	On	No

3.5.2 Speedometer

The computerized speedometer was validated by videotaping a marker on the treadmill belt while at constant speed. By counting frames and knowing the length of the treadmill belt, the velocity was calculated. The video-calculated velocity was then compared to the value obtained by the speedometer sensor. Systematic deviation was accounted for in the speedometer program by changing the value for the circumference of the roller. After changes were made, the validation was repeated. With a subject walking on the treadmill, the speedometer displayed speeds of 6.42, 3.68-70, and 8.80-8.82 kph.

The corresponding velocities calculated from video were 6.44, 3.70, and 8.81 kph. This test confirmed the speedometer to be valid when a subject was walking on the treadmill.

3.5.3 Estimation of Sample Size

Using data from Kinoshita and Bates (1983), estimates were made of the number of subjects necessary to achieve statistical power. Two methods were used: Barcikowski and Robey (1985) and the program Stat-Power v. 2.2 (Scientific Software, Inc., Chicago, IL, 1993). Kinoshita and Bates listed vertical ground reaction forces at midstance to be 7.22 N/kg of body weight for the unweighted condition, and 8.30 N/kg of BW for the weighted condition. The standard deviations for these two conditions were 0.264 and 0.210 N/kg, respectively.

Using these values in the program Stat-Power, power was calculated with the following levels: significance = 0.025, correlation between unweighted and weighted conditions = 0.3, power = 0.8 yielding a sample size of 5. This calculation was performed for a paired 2-tailed *t*-test.

A second estimation was made based on the writings of Barcikowski and Robey. With a comparison alpha-level of 0.05, large effect size, correlation between unweighted and weighted conditions estimated to be 0.5, and power set at 0.8, sample size was estimated at 14. Due to the limitations of the tables used in this calculation, the outcome was expected to be a conservative measure or to overestimate of the sample size. Based on these two tests, ten subjects were recruited.

CHAPTER 4
RESULTS AND DISCUSSION

This chapter reports the results of the study. The data and the analyses are presented and discussed in relation to the goals of the study.

4.1 Testing

Each of the 10 subjects reported to the Oregon State University Sports Medicine and Disabilities Research Lab on a day prior to testing for familiarization to treadmill walking and running. On a second day they reported and testing was performed. During these testing sessions, the subjects exhibited some unexpected behavior: In the determination of preferred transition speed, some subjects transitioned at different speeds for the two trials for a given condition.

Tests two and three were supposed to measure forces while walking and running at speeds from 0.6 m/s below the PTS to 0.6 m/s above the PTS. Because of the unexpected results from test one, the same range of speeds was not tested for all subjects. For subjects 3 and 7 the range of speeds tested started at 0.6 m/s below the upper value of PTS. Subject 6 was tested over the range calculated about the lower value of PTS. Subject 8 was tested at three speeds below the lower PTS and three speeds above the upper PTS. These data are discussed further in section 4.3.

During the walking trials, none of the subjects were able to walk at all of the speeds prescribed above their preferred transition speed. For the unweighted condition, all subjects were able to walk at PTS + 0.2 m/s, but the not all subjects were able to continue above this speed, with only two subjects able to complete the speed 0.6 m/s above their PTS. This problem was more acute for weighted walking, during which two subjects were unable to walk at any speed greater than their PTS. None of the subjects were able to complete the weighted walking condition at 0.6 m/s above PTS. At the highest walking speeds some of the subjects broke into a run. When this occurred during a force measurement, only the walking strides were analyzed. For a complete record of the number of strides analyzed for each trial, see Tables 4.1 through 4.8.

During testing, one subject was observed to run with a midfoot or forefoot striking gait. Because this subject did not run with the required rearfoot striking pattern, the subject was removed from analysis of midstance vertical ground reaction forces across the range of speeds. This subject was included in the analysis of change in preferred transition speed and change in $VGRF_M$ between the weighted and unweighted conditions.

4.2 Preferred Transition Speed

When preferred transition speed was measured for the unweighted condition (UW), five subjects transitioned at different speeds for the two trials. For the weighted condition (Wt), one subject transitioned differently between the trials. See Table 4.9. These inconsistencies were unforeseen. In such cases it was hypothesized that the true preferred transition speed was between the two observed values. Thus, for the specific tests and graphs which used ΔPTS, the two speeds were averaged and the force values pooled. For the analyses of the force curves and the variabilities, a single analysis group with the maximum number of subjects was desired. For these analyses, the lower of the two values was used for alignment at the PTS.

The subjects tested did not show a consistent change in preferred transition speed with the addition of weight. Beuter & Lefebvre (1988) did not observe any significant change in the walk-run PTS with 14% of body weight added to each of the trained runners they tested. By adding 24% of body weight to horses, Farley and Taylor (1991) observed a significant decrease in the preferred trot-gallop transition speed. For the subjects tested in this study, the preferred transition speeds increased, decreased, or remained the same with the addition of 15% of body weight. A 90% confidence interval

Table 4.1
Number of Strides Analyzed
Unweighted Walking, Left Leg

Speed	Subject Number									Mean
(m/s)	1	2	3	4	5	6	7	8	10	
1.2				8						8.0
1.4	9		9	9					8	8.8
1.6	9	10	8	9	9	10	8	10	8	9.0
1.8	10	11	9	10	10	10	9	11	9	9.9
2.0	9	12	10	12	10	11	9	8	10	10.1
2.2	12	12	6	13	11	12	10	11	7	10.4
2.4	12	7		13	10	12	11	12		11.0
2.6						12	3	12		8.3

Table 4.2
Number of Strides Analyzed
Unweighted Walking, Right Leg

Speed	Subject Number									Mean
(m/s)	1	2	3	4	5	6	7	8	10	
1.2				7						7.0
1.4	8		9	8					7	8.0
1.6	9	10	9	8	9	10	8	9	8	8.9
1.8	10	11	9	10	9	10	9	10	9	9.7
2.0	11	12	7	12	10	10	9	10	10	10.0
2.2	12	12	6	13	10	11	10	11	7	10.2
2.4	12	8		13	13	12	10	12		11.4
2.6						14	5	12		10.3

Table 4.3
Number of Strides Analyzed
Weighted Walking, Left Leg

Speed (m/s)	Subject Number 1	2	3	4	5	6	7	8	10	Mean
1.2									8	8.0
1.4	9	10	9	9				9	8	9.0
1.6	10	10	9	10			9	10	9	9.6
1.8	10	11	10	10	9	10	10	10	9	9.9
2.0	11	11	11	11	10	10	10	11	10	10.6
2.2	12	7		12	11	12	11	12		11.0
2.4	13			12	12	10	2	13		10.3
2.6						13				13.0

Table 4.4
Number of Strides Analyzed
Weighted Walking, Right Leg

Speed (m/s)	Subject Number 1	2	3	4	5	6	7	8	10	Mean
1.2									7	7.0
1.4	8	9	8	8				10	8	8.5
1.6	9	10	8	7			8	10	8	8.6
1.8	10	11	9	10	9	9	9	10	9	9.6
2.0	11	11	10	11	9	10	9	11	10	10.2
2.2	12	7		11	9	12	11	12		10.6
2.4	12			13	12	13	3	13		11.0
2.6						13				13.0

Table 4.5
Number of Strides Analyzed
Unweighted Running, Left Leg

Speed (m/s)	Subject Number 1	2	3	4	5	6	7	8	10	Mean
1.2				11						11.0
1.4	11		11	11					11	11.0
1.6	12	12	10	12	12	11	11	11	11	11.3
1.8	12	12	11	12	12	11	12	11	12	11.7
2.0	12	13	11	12	12	12	12	12	12	12.0
2.2	13	13	11	13	12	11	12	12	12	12.1
2.4	12	13	12	13	12	12	12	13	12	12.3
2.6	13	13	11		13	12	12	13	12	12.4
2.8		13			12	13	12	13		12.6
3.0								14		14.0

Table 4.6
Number of Strides Analyzed
Unweighted Running, Right Leg

Speed (m/s)	Subject Number 1	2	3	4	5	6	7	8	10	Mean
1.2				12						12.0
1.4	12		11	12					11	11.5
1.6	12	13	11	12	12	11	12	11	12	11.8
1.8	13	13	11	12	11	12	11	12	12	11.9
2.0	13	12	11	13	12	12	12	12	13	12.2
2.2	13	13	11	12	12	12	12	13	12	12.2
2.4	12	13	11	12	12	12	12	13	12	12.1
2.6	13	13	12		13	12	11	13	13	12.5
2.8		13			13	11	12	13		12.4
3.0								13		13.0

Table 4.7
Number of Strides Analyzed
Weighted Running, Left Leg

Speed (m/s)	Subject Number 1	2	3	4	5	6	7	8	10	Mean
1.2									9	9.0
1.4	11	12	10	12				11	11	11.0
1.6	12	12	10	11			11	11	11	11.1
1.8	12	12	10	12	12	10	12	12	11	11.4
2.0	12	12	11	12	12	11	12	12	13	11.9
2.2	13	12	12	13	12	12	12	12	12	12.2
2.4	13	13	12	13	13	12	12	13	13	12.7
2.6	13	13	11	13	13	12	12	13		12.5
2.8					13	12	11			12.0
3.0					13					13.0

Table 4.8
Number of Strides Analyzed
Weighted Running, Right Leg

Speed (m/s)	Subject Number 1	2	3	4	5	6	7	8	10	Mean
1.2									9	9.0
1.4	11	11	10	12				10	10	10.7
1.6	11	12	11	11			11	11	10	11.0
1.8	11	12	11	12	12	11	11	12	11	11.4
2.0	12	12	11	12	12	11	11	12	13	11.8
2.2	12	13	11	13	12	11	12	12	12	12.0
2.4	13	13	11	12	12	12	11	12	12	12.0
2.6	13	13	12	13	13	12	12	13		12.6
2.8					12	13	12			12.3
3.0					13					13.0

Table 4.9
Determination of Preferred Transition Speed
Speed in meters per second

	Unweighted			Weighted			Difference
Subject	Trial 1	Trial 2	Average	Trial 1	Trial 2	Average	(Wt - UW)
1	2.00	2.00	2.00	2.00	2.00	2.00	0.00
2	2.20	2.20	2.20	2.00	2.00	2.00	-0.20
3	2.00	1.80	1.90	2.00	2.00	2.00	0.10
4	1.80	1.80	1.80	2.00	2.00	2.00	0.20
5	2.20	2.20	2.20	2.40	2.40	2.40	0.20
6	2.60	2.40	2.50	2.20	2.40	2.30	-0.20
7	2.00	2.20	2.10	2.20	2.20	2.20	0.10
8	2.40	2.20	2.30	2.00	2.00	2.00	-0.30
9	2.00	2.20	2.10	2.00	2.00	2.00	-0.10
10	2.00	2.00	2.00	1.80	1.80	1.80	-0.20
Average			2.11			2.07	-0.04
SD			0.19			0.17	0.17

for the change in preferred transition speed with the addition of weight was between 0.061 to -0.141 m/s.

For the following tables and graphs, the labels for the four conditions will be abbreviated: Unweighted and weighted walking are UW W and Wt W, while the respective running conditions are UW R and Wt R.

4.3 Force Curves

The second and third tests measured vertical ground reaction force for a range of speeds. When midstance forces were picked off and averaged for a given speed and condition, the relationship between midstance force and speed could be mapped. For each individual the forces were plotted against speed. See Figures 4.1 through 4.18. The data for these figures are in Tables 4.10 though 4.17. The curves for a given condition were graphed with all of the subjects overlaid (Figures 4.19 through 4.26). For greater generalization, the preferred transition speeds were aligned and the forces averaged across subjects. See Figures 4.27 and 4.28. In these two figures, the error bars represent

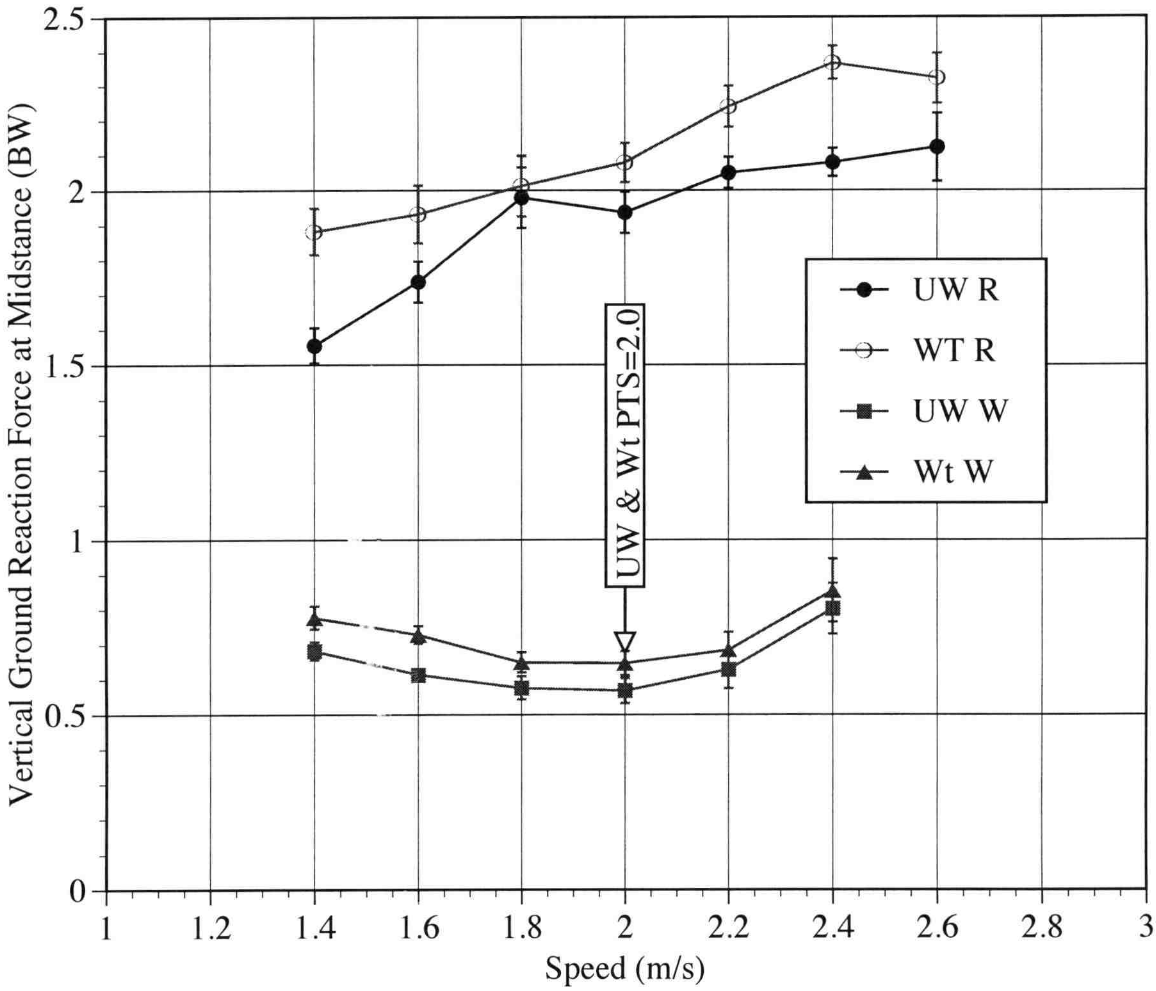

Figure 4.2
$VGRF_M$ vs. Speed: Subject 1, Right Leg
All Conditions

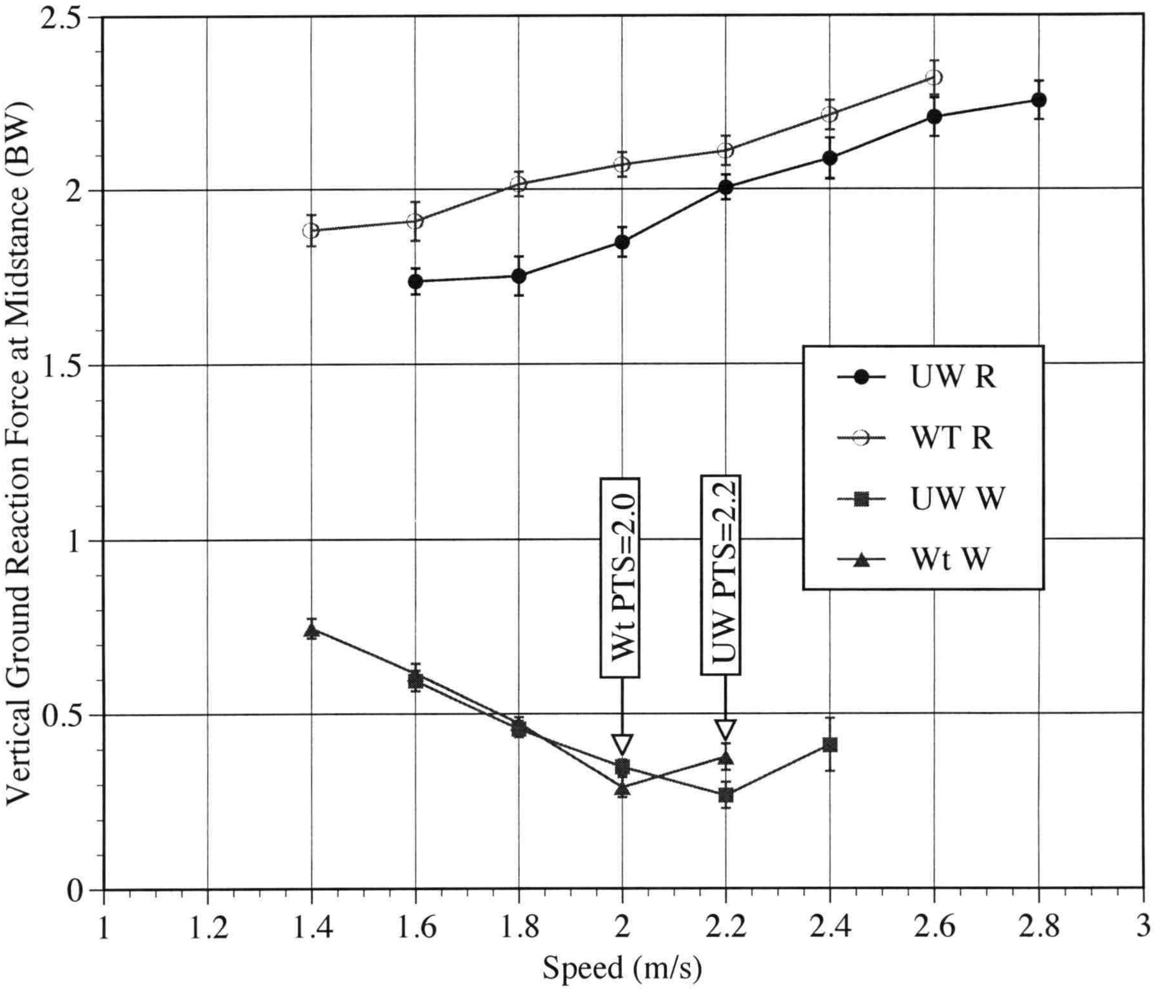

Figure 4.3
$VGRF_M$ vs. Speed: Subject 2, Left Leg
All Conditions

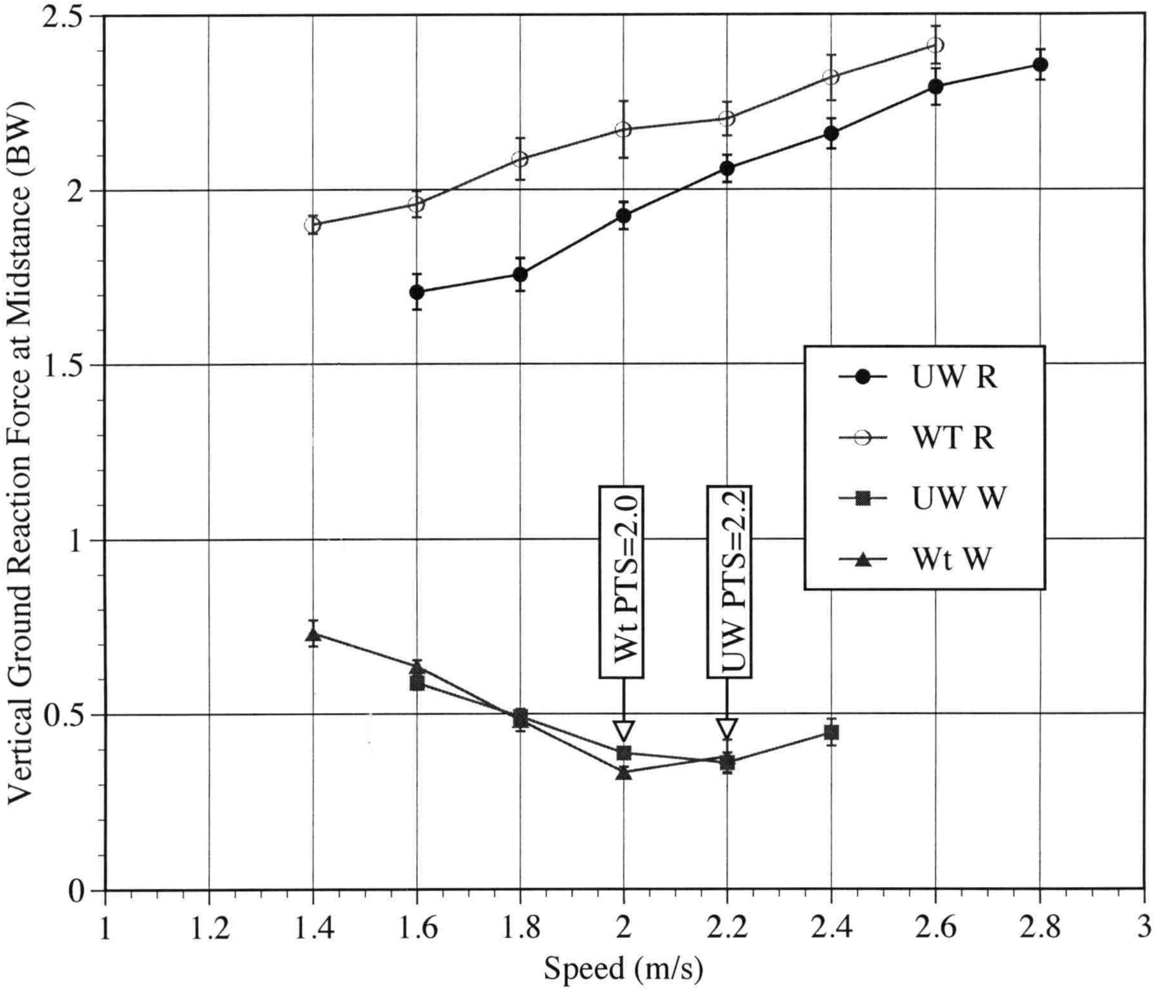

Figure 4.4
$VGRF_M$ vs. Speed: Subject 2, Right Leg
All Conditions

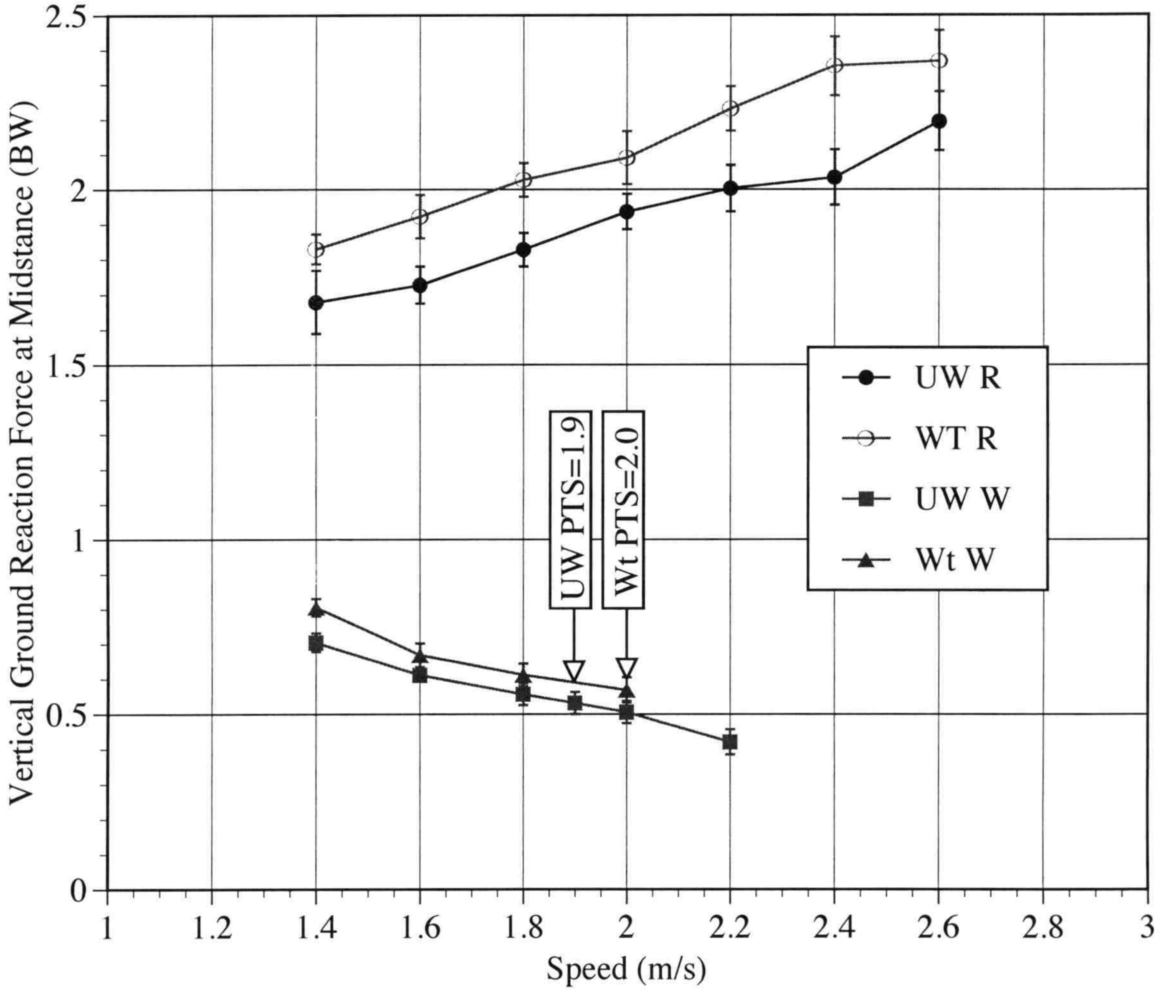

Figure 4.5
$VGRF_M$ vs. Speed: Subject 3, Left Leg
All Conditions

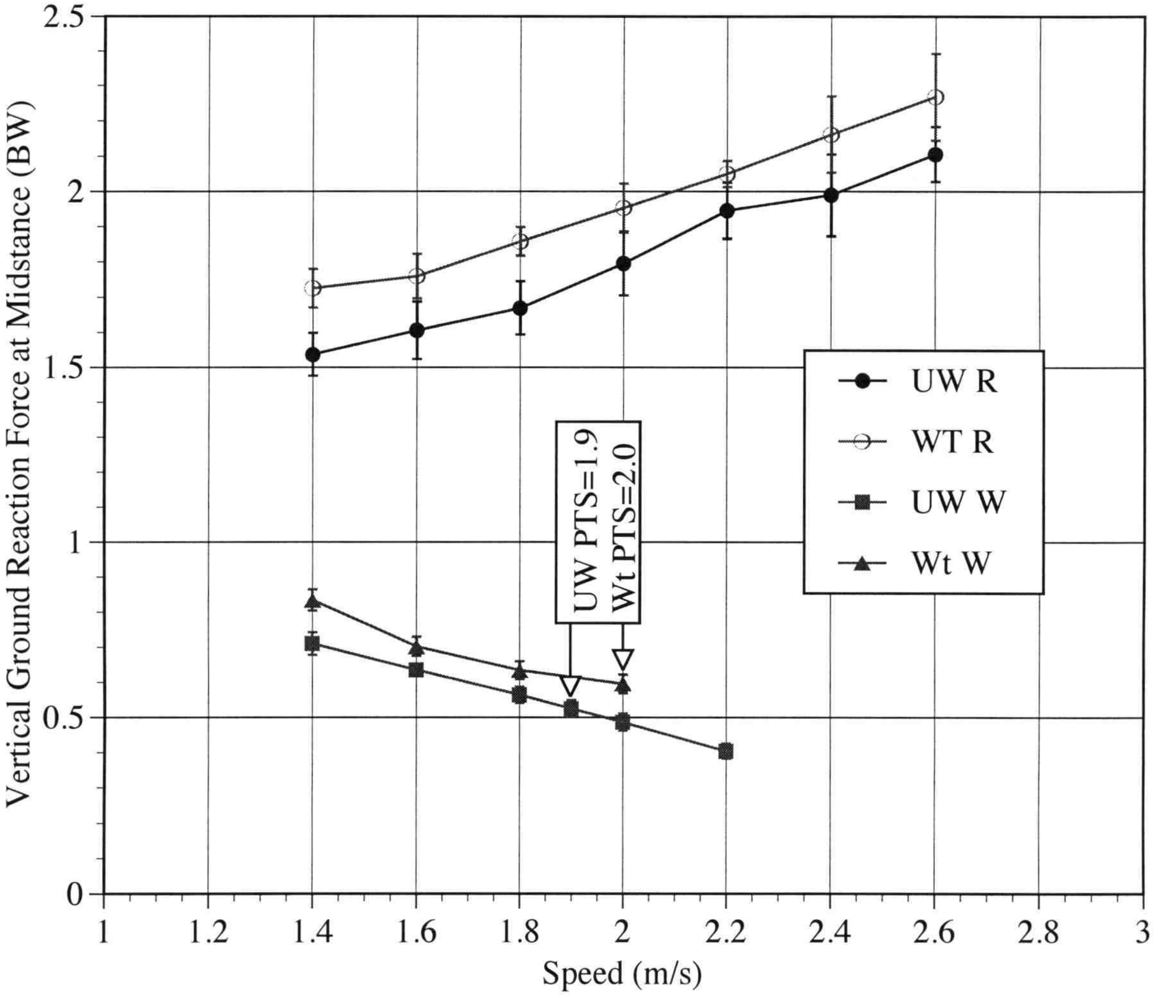

Figure 4.6
$VGRF_M$ vs. Speed: Subject 3, Right Leg
All Conditions

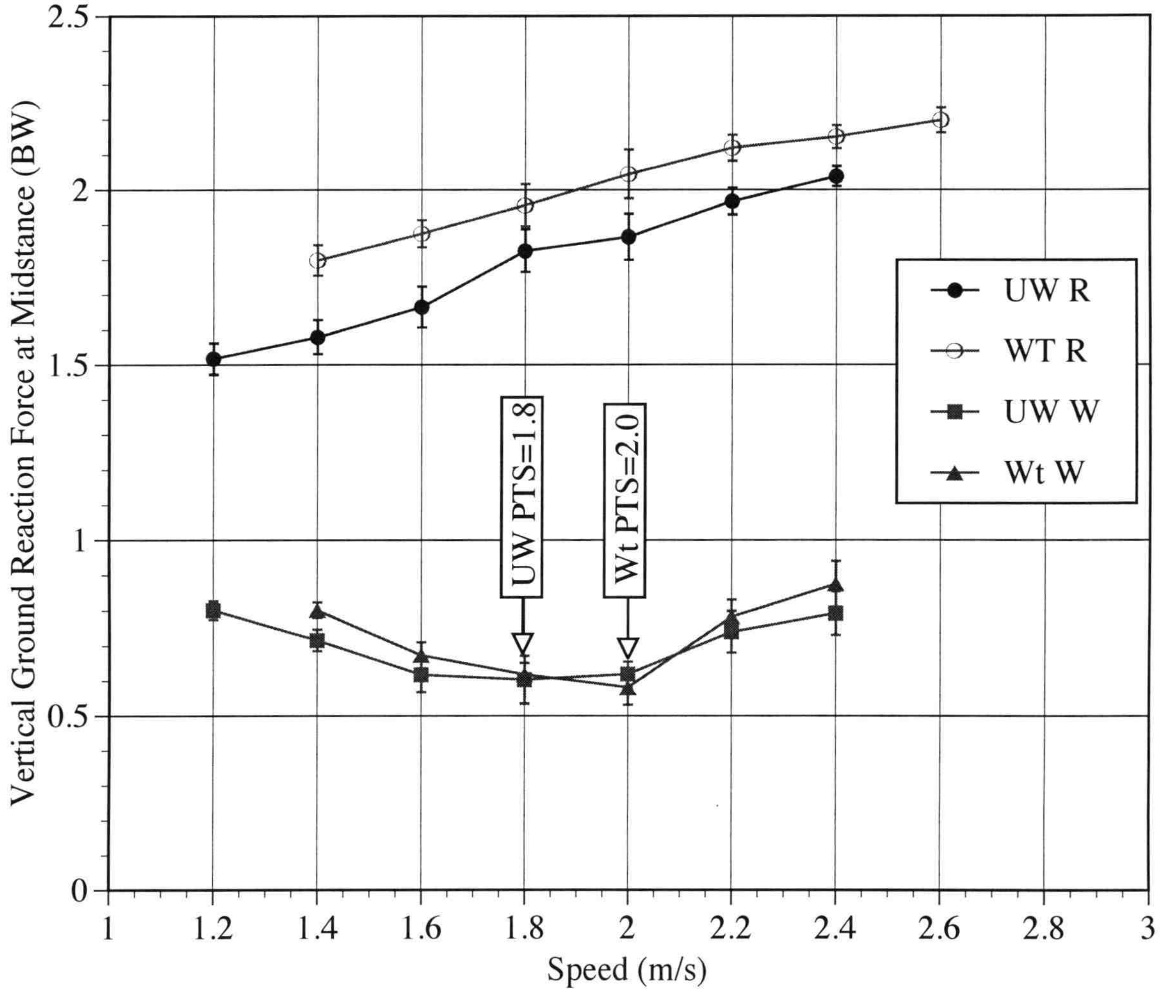

Figure 4.7
$VGRF_M$ vs. Speed: Subject 4, Left Leg
All Conditions

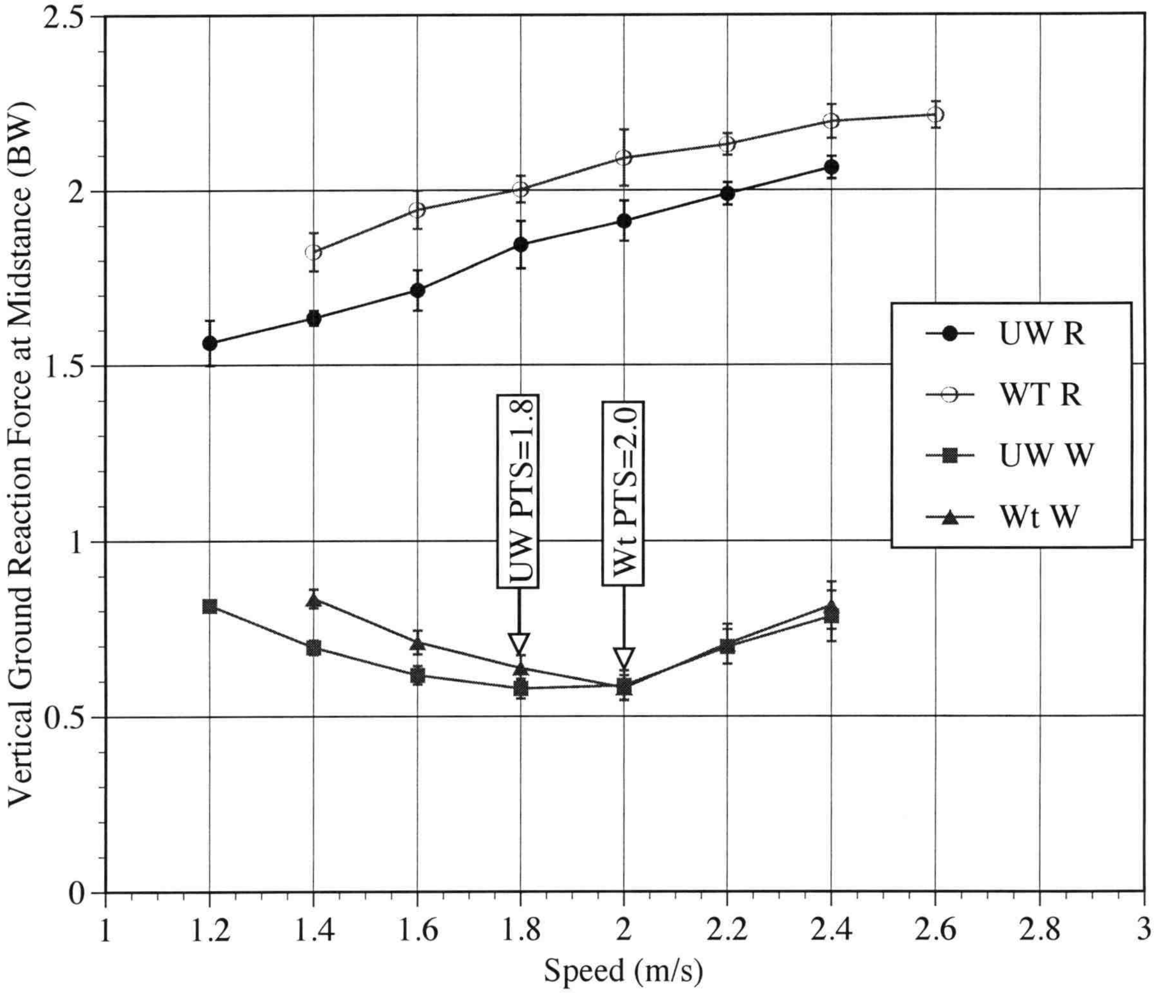

Figure 4.8
$VGRF_M$ vs. Speed: Subject 4, Right Leg
All Conditions

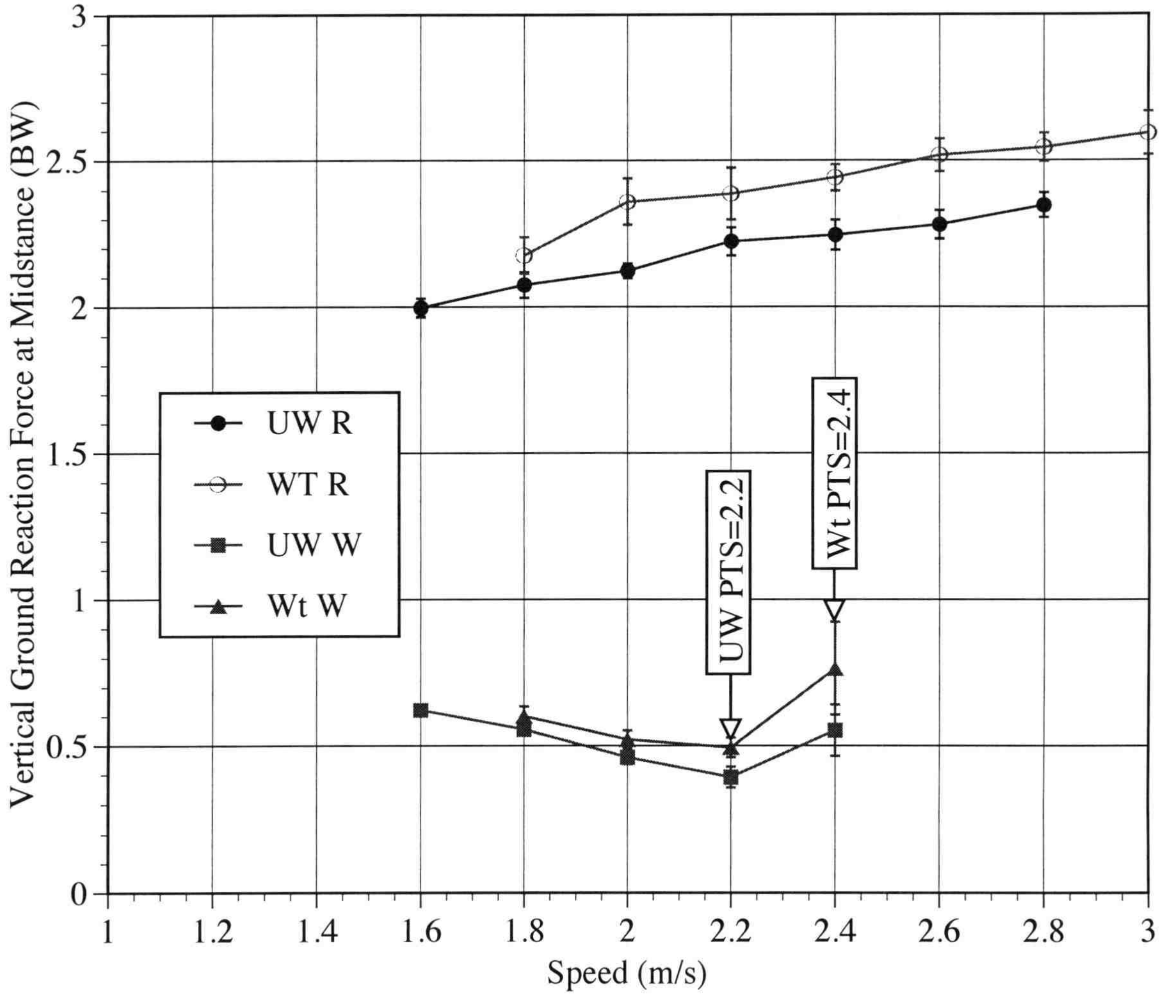

Figure 4.9
$VGRF_M$ vs. Speed: Subject 5, Left Leg
All Conditions

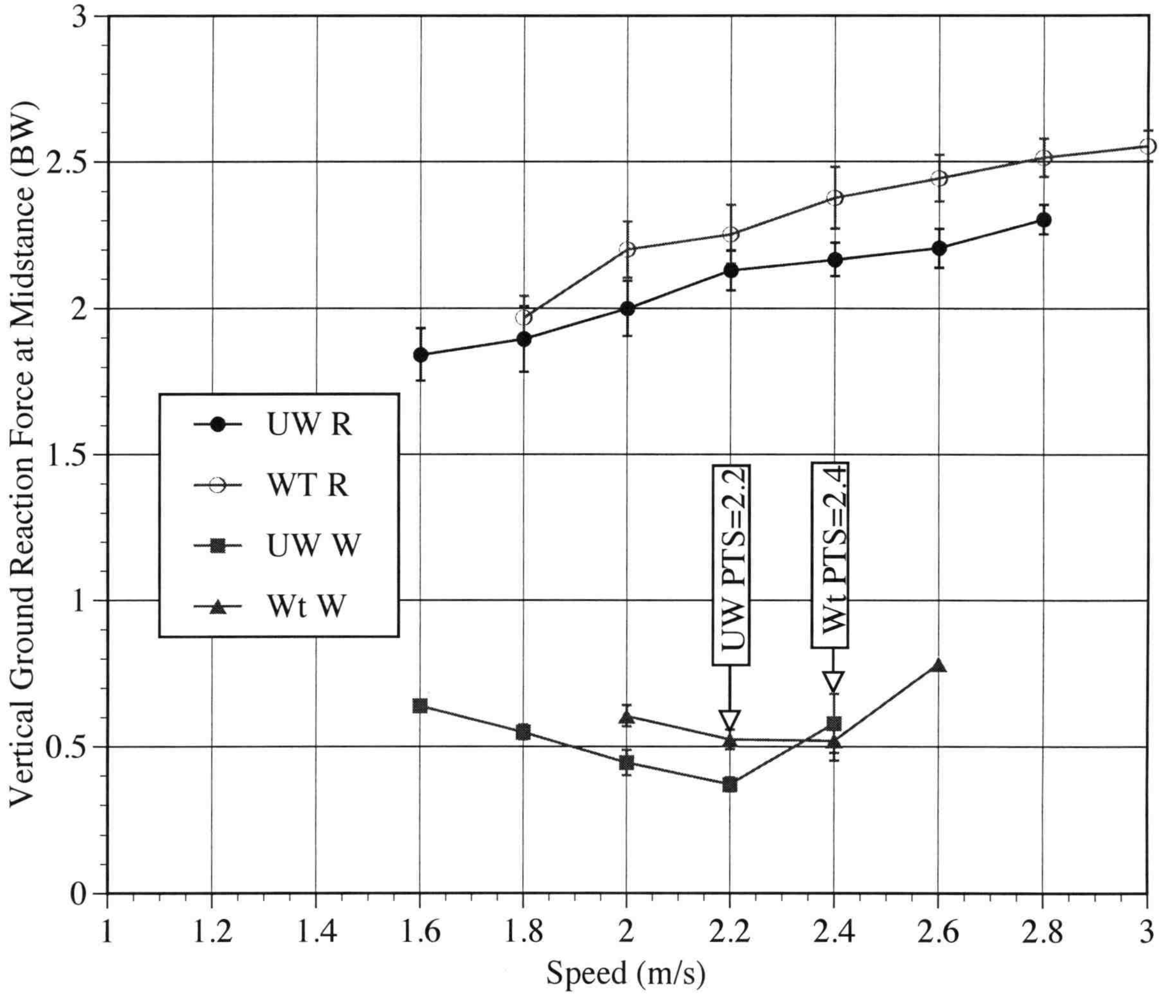

Figure 4.10
$VGRF_M$ vs. Speed: Subject 5, Right Leg
All Conditions

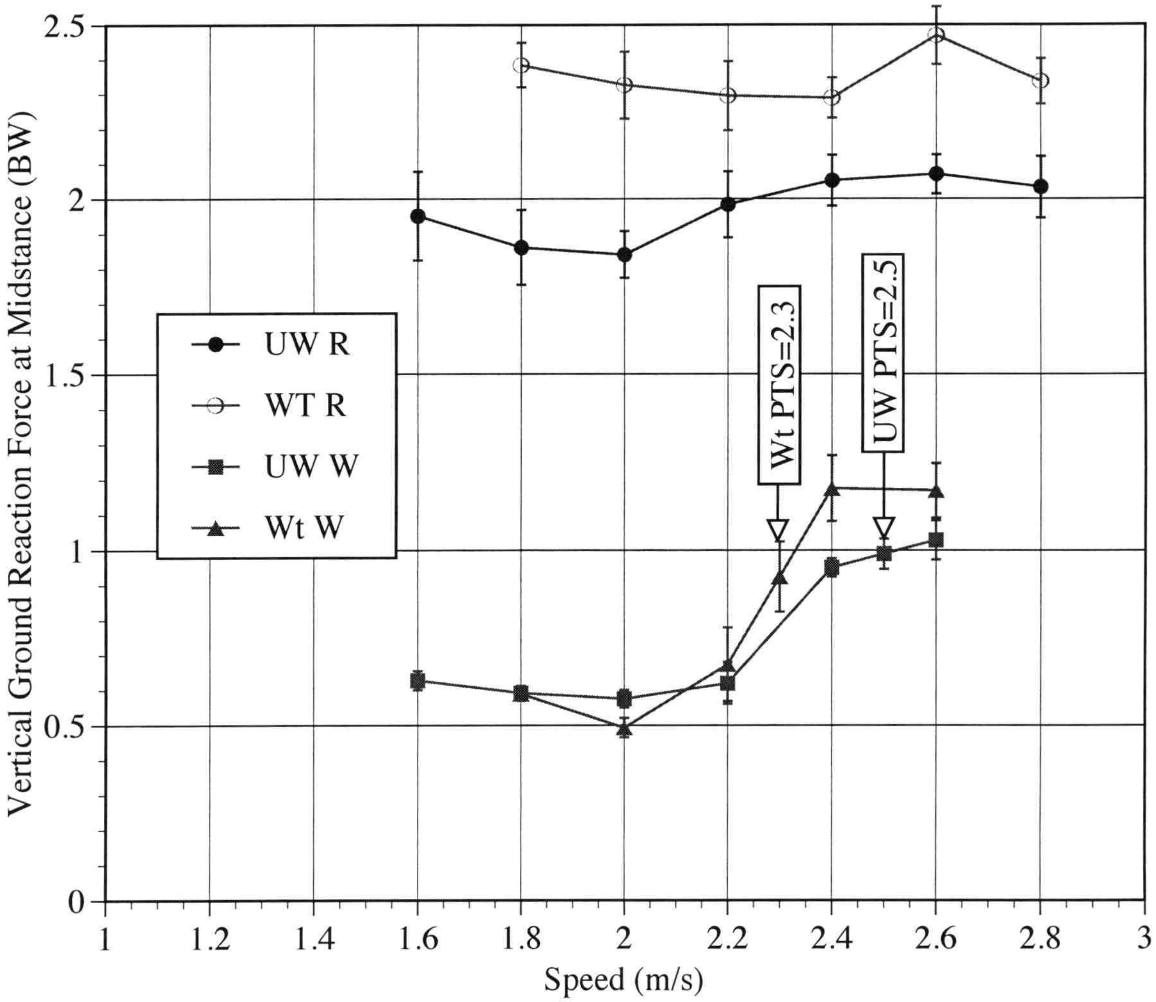

Figure 4.11
$VGRF_M$ vs. Speed: Subject 6, Left Leg
All Conditions

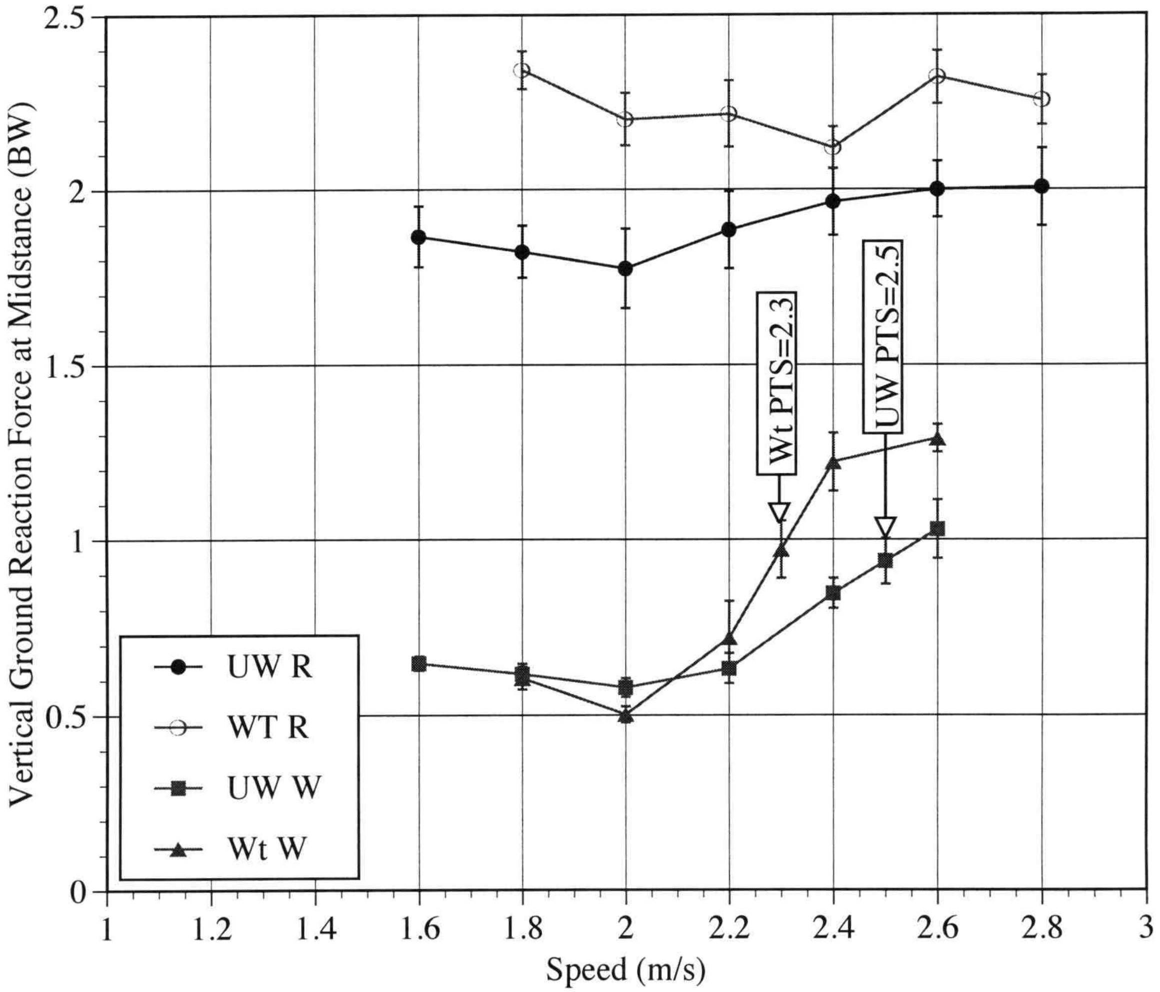

Figure 4.12
$VGRF_M$ vs. Speed: Subject 6, Right Leg
All Conditions

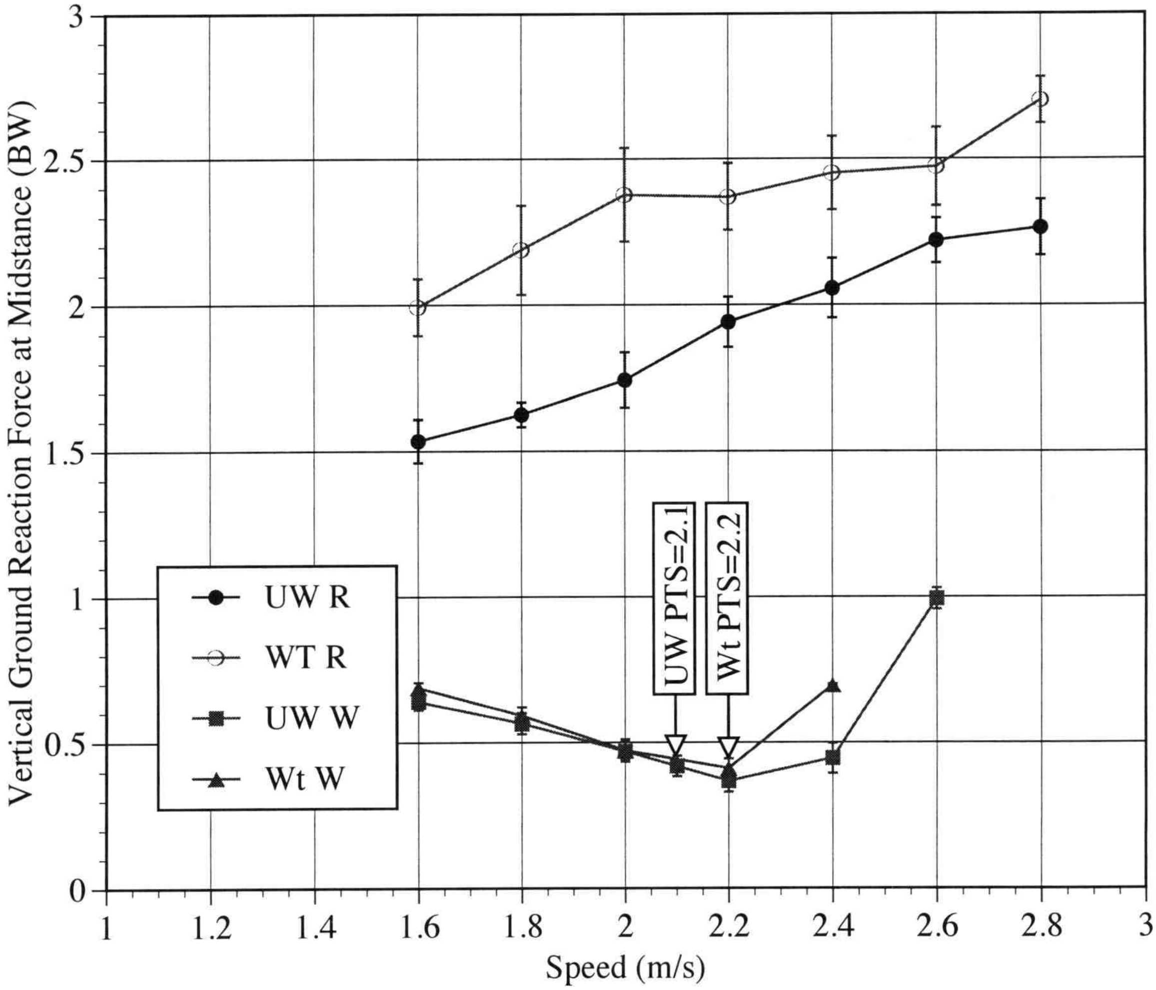

Figure 4.13
$VGRF_M$ vs. Speed: Subject 7, Left Leg
All Conditions

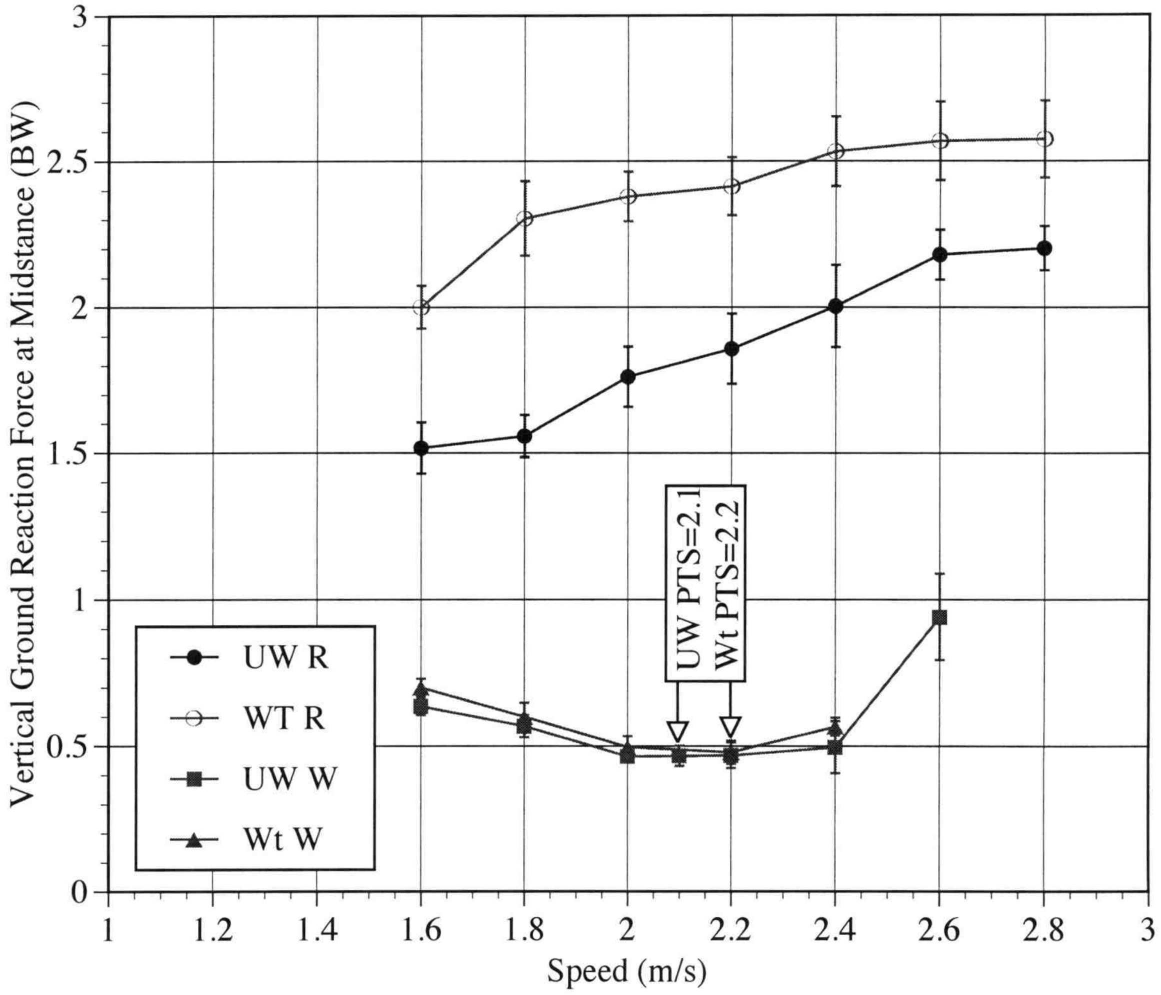

Figure 4.14
$VGRF_M$ vs. Speed: Subject 7, Right Leg
All Conditions

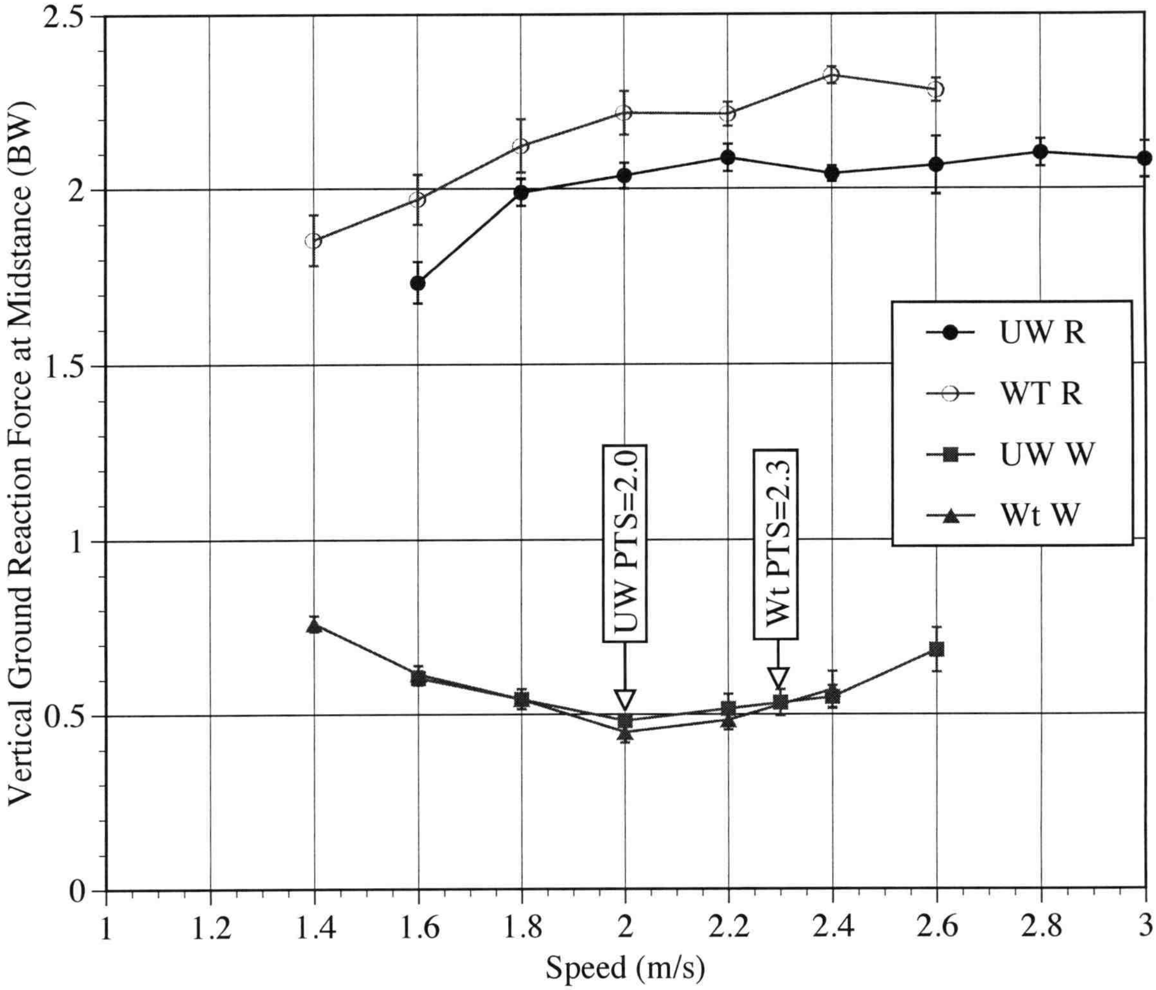

Figure 4.15
$VGRF_M$ vs. Speed: Subject 8, Left Leg
All Conditions

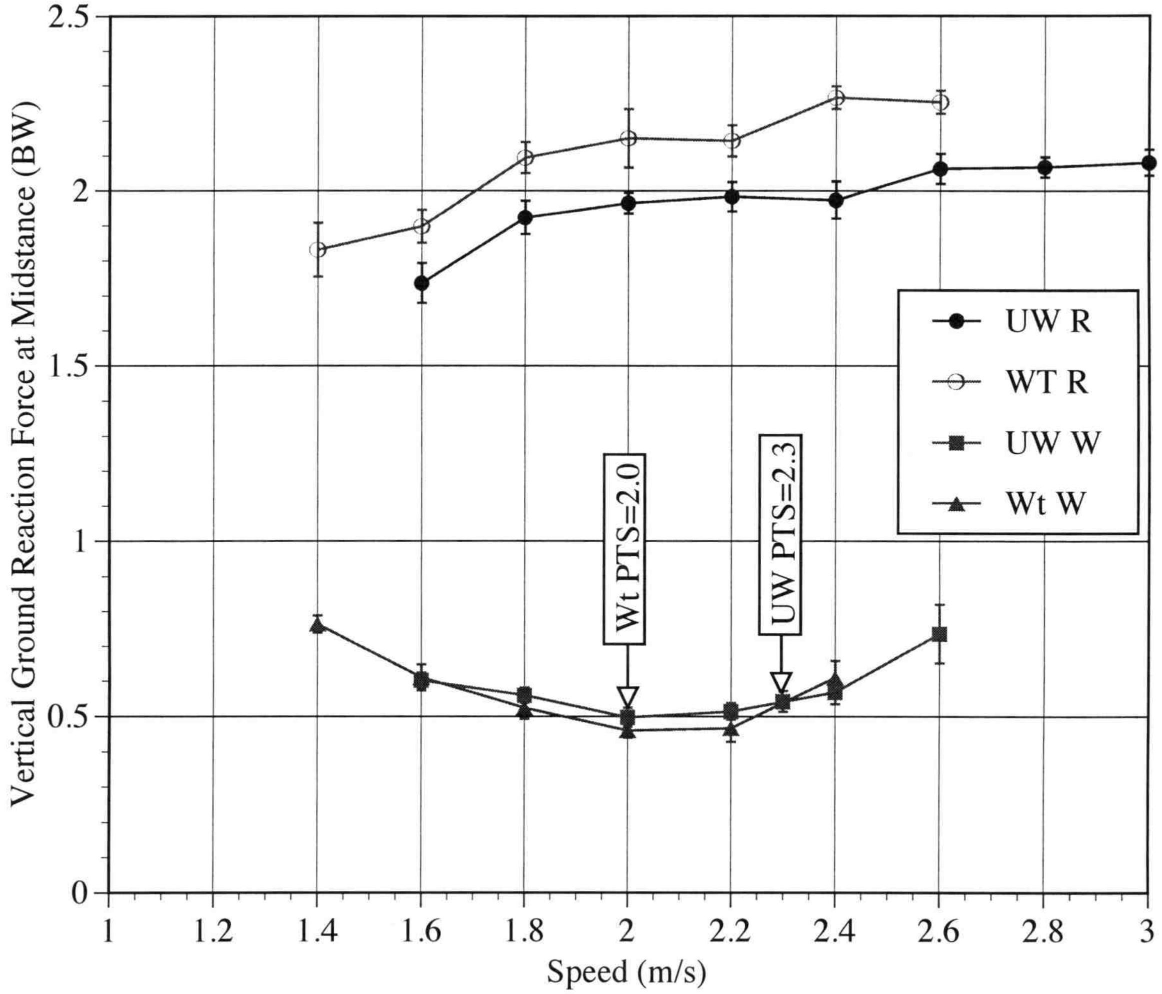

Figure 4.16
$VGRF_M$ vs. Speed: Subject 8, Right Leg
All Conditions

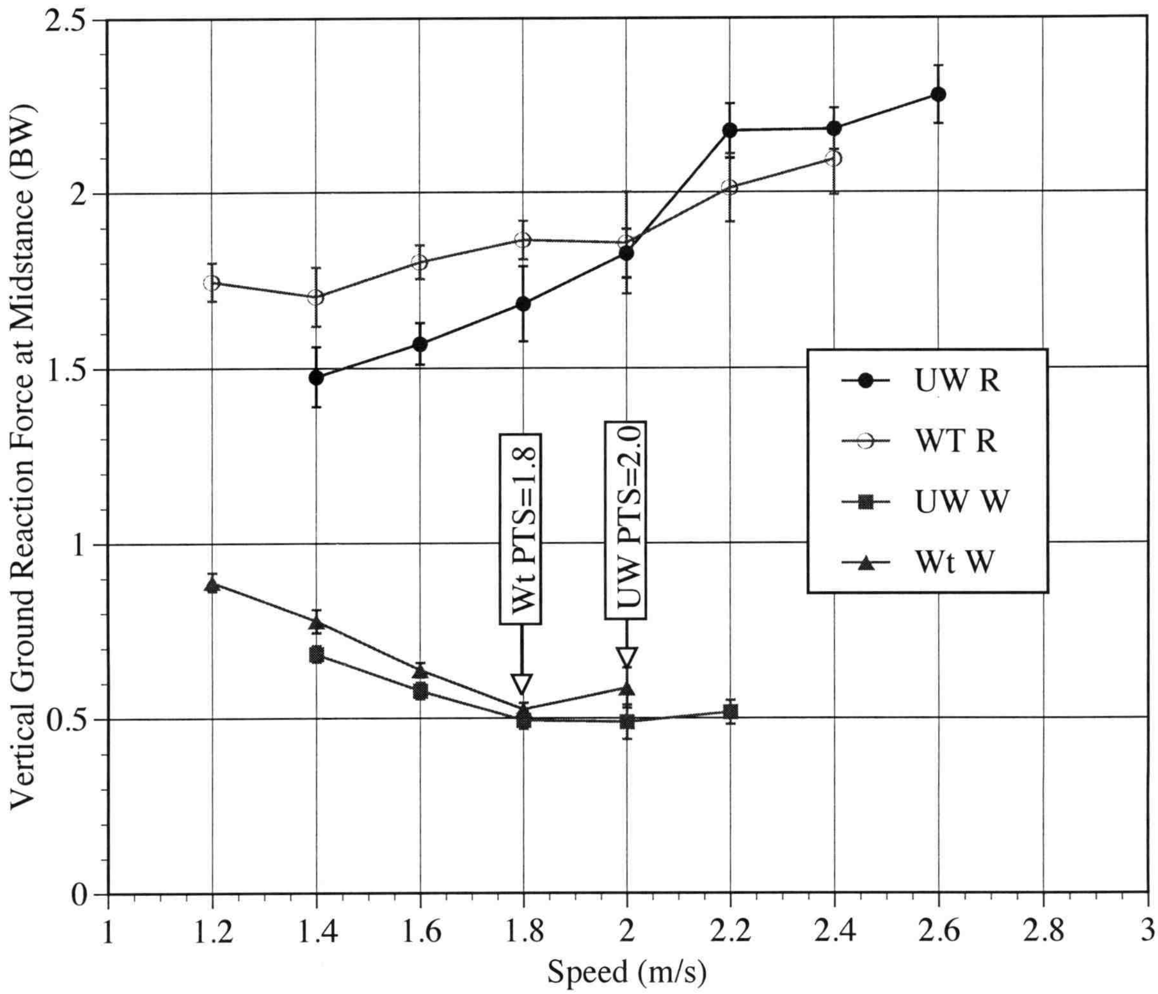

Figure 4.17
$VGRF_M$ vs. Speed: Subject 10, Left Leg
All Conditions

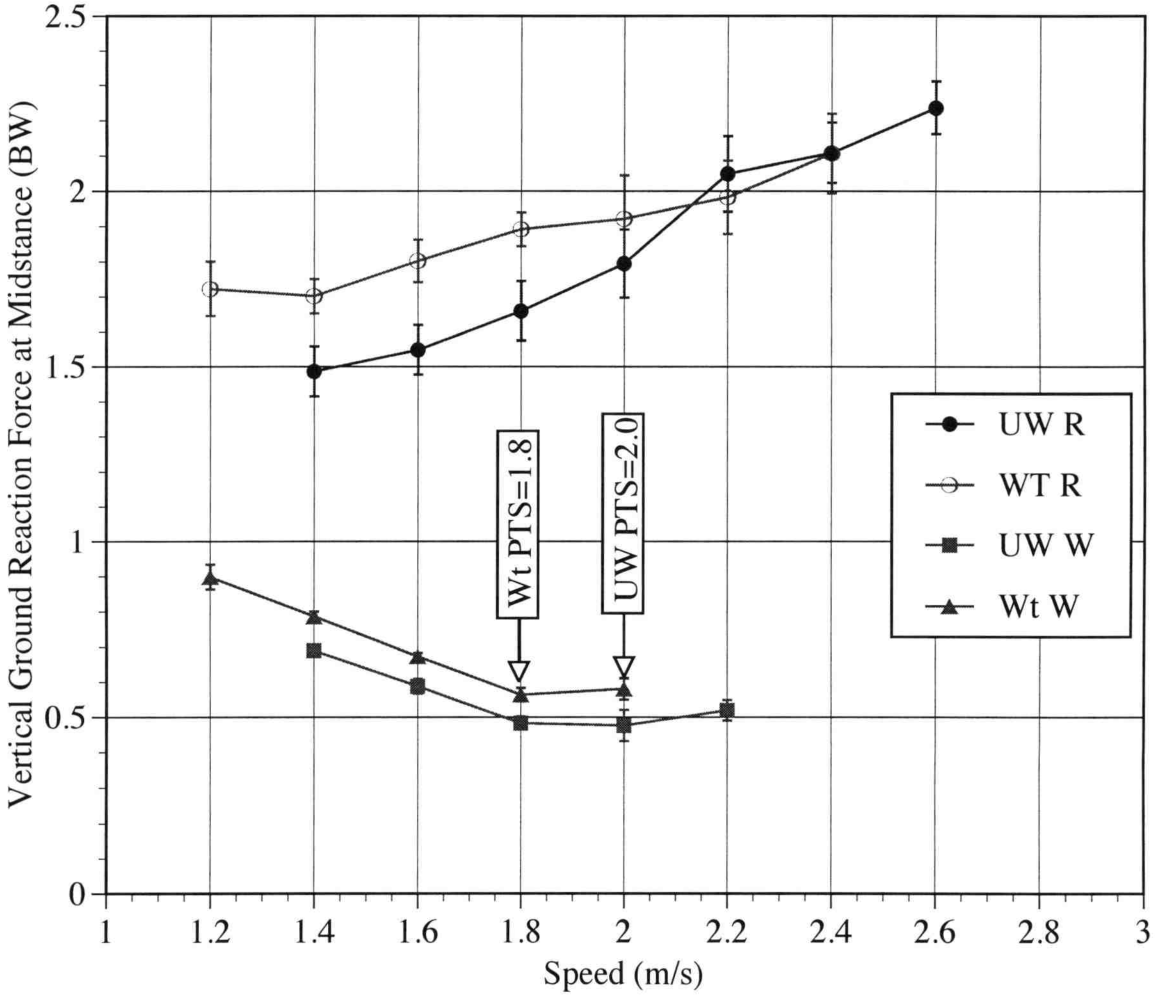

Figure 4.18
$VGRF_M$ vs. Speed: Subject 10, Right Leg
All Conditions

Table 4.10
Force vs. Speed: Unweighted Walking, Left Leg
$VGRF_M$ in Body Weights (Standard Deviation)

Speed (m/s)	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Subject 6	Subject 7	Subject 8	Subject 10	Average
1.2				0.800 (0.027)						0.800
1.4	0.679 (0.028)		0.704 (0.027)	0.715 (0.030)					0.682 (0.024)	0.695 (0.017)
1.6	0.624 (0.035)	0.595 (0.029)	0.611 (0.018)	0.617 (0.050)	0.623 (0.016)	0.628 (0.027)	0.639 (0.030)	0.603 (0.020)	0.577 (0.023)	0.613 (0.019)
1.8	0.575 (0.024)	0.456 (0.023)	0.557 (0.032)	0.603 (0.068)	0.556 (0.017)	0.592 (0.020)	0.563 (0.037)	0.542 (0.017)	0.493 (0.026)	0.549 (0.047)
2.0	0.564 (0.031)	0.348 (0.021)	0.506 (0.032)	0.619 (0.035)	0.460 (0.025)	0.575 (0.025)	0.469 (0.032)	0.481 (0.017)	0.489 (0.049)	0.501 (0.079)
2.2	0.627 (0.043)	0.267 (0.038)	0.420 (0.036)	0.739 (0.059)	0.392 (0.036)	0.620 (0.060)	0.367 (0.037)	0.515 (0.042)	0.516 (0.035)	0.496 (0.149)
2.4	0.759 (0.042)	0.410 (0.076)		0.792 (0.062)	0.552 (0.088)	0.951 (0.026)	0.445 (0.052)	0.548 (0.033)		0.637 (0.200)
2.6						1.029 (0.055)	0.993 (0.037)	0.682 (0.063)		0.901 (0.191)

Table 4.11
Force vs. Speed: Unweighted Walking, Right Leg
$VGRF_M$ in Body Weights (Standard Deviation)

Speed (m/s)	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Subject 6	Subject 7	Subject 8	Subject 10	Average
1.2				0.814 (0.013)						0.814
1.4	0.682 (0.026)		0.710 (0.032)	0.695 (0.021)					0.690 (0.013)	0.694 (0.012)
1.6	0.613 (0.010)	0.588 (0.020)	0.636 (0.011)	0.617 (0.027)	0.639 (0.019)	0.648 (0.021)	0.635 (0.030)	0.601 (0.021)	0.588 (0.021)	0.618 (0.023)
1.8	0.576 (0.033)	0.492 (0.011)	0.564 (0.023)	0.578 (0.028)	0.549 (0.025)	0.618 (0.029)	0.567 (0.037)	0.559 (0.022)	0.483 (0.019)	0.554 (0.042)
2.0	0.568 (0.036)	0.387 (0.013)	0.486 (0.024)	0.587 (0.043)	0.445 (0.043)	0.579 (0.027)	0.464 (0.021)	0.497 (0.026)	0.476 (0.045)	0.499 (0.067)
2.2	0.628 (0.054)	0.360 (0.028)	0.404 (0.021)	0.697 (0.049)	0.371 (0.024)	0.634 (0.043)	0.468 (0.044)	0.513 (0.023)	0.520 (0.029)	0.510 (0.122)
2.4	0.803 (0.072)	0.446 (0.039)		0.783 (0.072)	0.579 (0.101)	0.846 (0.042)	0.495 (0.089)	0.567 (0.034)		0.646 (0.162)
2.6						1.029 (0.084)	0.940 (0.147)	0.735 (0.084)		0.901 (0.151)

Table 4.12
Force vs. Speed: Weighted Walking, Left Leg
$VGRF_M$ in Body Weights (Standard Deviation)

Speed (m/s)	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Subject 6	Subject 7	Subject 8	Subject 10	Average
1.2									0.888 (0.027)	0.888
1.4	0.779 (0.018)	0.745 (0.028)	0.804 (0.025)	0.801 (0.022)				0.758 (0.023)	0.776 (0.033)	0.777 (0.023)
1.6	0.682 (0.027)	0.616 (0.028)	0.668 (0.034)	0.673 (0.037)			0.686 (0.017)	0.614 (0.023)	0.636 (0.021)	0.654 (0.031)
1.8	0.641 (0.027)	0.473 (0.017)	0.612 (0.032)	0.618 (0.032)	0.601 (0.033)	0.589 (0.019)	0.591 (0.029)	0.543 (0.029)	0.524 (0.019)	0.577 (0.053)
2.0	0.628 (0.048)	0.290 (0.029)	0.569 (0.036)	0.581 (0.050)	0.521 (0.030)	0.493 (0.028)	0.471 (0.039)	0.448 (0.029)	0.586 (0.058)	0.510 (0.101)
2.2	0.678 (0.102)	0.376 (0.038)		0.782 (0.047)	0.494 (0.033)	0.674 (0.105)	0.410 (0.033)	0.483 (0.028)		0.557 (0.154)
2.4	0.911 (0.162)			0.875 (0.064)	0.765 (0.159)	1.175 (0.094)	0.695 (0.004)	0.570 (0.052)		0.832 (0.209)
2.6						1.169 (0.077)				1.169

Table 4.13
Force vs. Speed: Weighted Walking, Right Leg
$VGRF_M$ in Body Weights (Standard Deviation)

Speed (m/s)	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Subject 6	Subject 7	Subject 8	Subject 10	Average
1.2									0.898 (0.035)	0.898
1.4	0.777 (0.033)	0.730 (0.037)	0.834 (0.030)	0.834 (0.027)				0.763 (0.025)	0.787 (0.013)	0.787 (0.041)
1.6	0.728 (0.025)	0.635 (0.017)	0.702 (0.027)	0.710 (0.034)			0.699 (0.030)	0.610 (0.037)	0.674 (0.010)	0.680 (0.043)
1.8	0.649 (0.029)	0.481 (0.032)	0.634 (0.026)	0.637 (0.036)	0.605 (0.031)	0.605 (0.032)	0.599 (0.048)	0.524 (0.032)	0.564 (0.019)	0.589 (0.056)
2.0	0.646 (0.035)	0.332 (0.015)	0.595 (0.026)	0.581 (0.035)	0.524 (0.036)	0.502 (0.023)	0.496 (0.036)	0.460 (0.021)	0.581 (0.031)	0.524 (0.093)
2.2	0.685 (0.051)	0.377 (0.048)		0.705 (0.056)	0.520 (0.034)	0.721 (0.103)	0.478 (0.039)	0.466 (0.039)		0.565 (0.137)
2.4	0.854 (0.090)			0.814 (0.068)	0.783 (0.069)	1.221 (0.083)	0.566 (0.030)	0.609 (0.048)		0.808 (0.233)
2.6						1.289 (0.040)				1.289

Table 4.14
Force vs. Speed: Unweighted Running, Left Leg
$VGRF_M$ in Body Weights (Standard Deviation)

Speed (m/s)	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Subject 6	Subject 7	Subject 8	Subject 10	Average
1.2				1.517 (0.045)						1.517
1.4	1.540 (0.055)		1.678 (0.090)	1.578 (0.049)					1.475 (0.085)	1.568 (0.085)
1.6	1.771 (0.058)	1.736 (0.037)	1.727 (0.052)	1.664 (0.058)	1.997 (0.032)	1.951 (0.127)	1.534 (0.074)	1.732 (0.059)	1.568 (0.059)	1.742 (0.154)
1.8	2.004 (0.079)	1.750 (0.055)	1.828 (0.048)	1.825 (0.060)	2.074 (0.044)	1.861 (0.107)	1.624 (0.042)	1.989 (0.039)	1.681 (0.107)	1.848 (0.152)
2.0	1.989 (0.038)	1.846 (0.042)	1.936 (0.051)	1.864 (0.066)	2.122 (0.025)	1.841 (0.067)	1.742 (0.095)	2.036 (0.036)	1.825 (0.070)	1.911 (0.119)
2.2	2.132 (0.047)	2.004 (0.036)	2.003 (0.067)	1.966 (0.038)	2.222 (0.048)	1.983 (0.095)	1.941 (0.085)	2.086 (0.039)	2.172 (0.077)	2.057 (0.100)
2.4	2.155 (0.046)	2.087 (0.058)	2.035 (0.079)	2.038 (0.029)	2.244 (0.052)	2.052 (0.073)	2.055 (0.101)	2.040 (0.022)	2.178 (0.059)	2.098 (0.076)
2.6	2.194 (0.049)	2.205 (0.055)	2.195 (0.084)		2.280 (0.049)	2.070 (0.056)	2.218 (0.076)	2.065 (0.082)	2.274 (0.083)	2.188 (0.081)
2.8		2.252 (0.055)			2.347 (0.043)	2.033 (0.088)	2.263 (0.096)	2.100 (0.039)		2.199 (0.129)
3.0								2.080 (0.052)		2.080

Table 4.15
Force vs. Speed: Unweighted Running, Right Leg
$VGRF_M$ in Body Weights (Standard Deviation)

Speed (m/s)	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Subject 6	Subject 7	Subject 8	Subject 10	Average
1.2				1.564 (0.064)						1.564
1.4	1.554 (0.050)		1.537 (0.061)	1.633 (0.021)					1.486 (0.072)	1.553 (0.061)
1.6	1.736 (0.059)	1.707 (0.051)	1.605 (0.082)	1.713 (0.058)	1.842 (0.090)	1.866 (0.087)	1.516 (0.087)	1.735 (0.057)	1.548 (0.071)	1.697 (0.121)
1.8	1.978 (0.087)	1.756 (0.048)	1.669 (0.075)	1.844 (0.067)	1.895 (0.112)	1.822 (0.073)	1.557 (0.072)	1.923 (0.048)	1.660 (0.085)	1.789 (0.139)
2.0	1.936 (0.059)	1.924 (0.039)	1.795 (0.090)	1.912 (0.058)	1.999 (0.093)	1.775 (0.113)	1.761 (0.104)	1.964 (0.029)	1.794 (0.097)	1.873 (0.091)
2.2	2.049 (0.045)	2.059 (0.039)	1.946 (0.079)	1.989 (0.032)	2.128 (0.068)	1.884 (0.109)	1.857 (0.120)	1.983 (0.042)	2.049 (0.107)	1.994 (0.088)
2.4	2.078 (0.040)	2.158 (0.043)	1.990 (0.116)	2.063 (0.031)	2.166 (0.057)	1.964 (0.095)	2.003 (0.140)	1.973 (0.053)	2.109 (0.086)	2.056 (0.078)
2.6	2.122 (0.097)	2.291 (0.052)	2.106 (0.078)		2.205 (0.066)	1.999 (0.080)	2.178 (0.085)	2.062 (0.043)	2.237 (0.074)	2.150 (0.096)
2.8		2.353 (0.044)			2.303 (0.050)	2.006 (0.111)	2.201 (0.075)	2.067 (0.029)		2.186 (0.149)
3.0								2.080 (0.037)		2.080

Table 4.16
Force vs. Speed: Weighted Running, Left Leg
$VGRF_M$ in Body Weights (Standard Deviation)

Speed (m/s)	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Subject 6	Subject 7	Subject 8	Subject 10	Average
1.2									1.744 (0.054)	1.744
1.4	1.926 (0.060)	1.881 (0.045)	1.828 (0.043)	1.798 (0.043)				1.854 (0.072)	1.701 (0.083)	1.831 (0.077)
1.6	1.980 (0.057)	1.908 (0.056)	1.922 (0.062)	1.873 (0.039)			1.992 (0.097)	1.969 (0.071)	1.799 (0.048)	1.920 (0.068)
1.8	2.030 (0.109)	2.014 (0.035)	2.028 (0.049)	1.955 (0.060)	2.175 (0.062)	2.384 (0.064)	2.187 (0.152)	2.121 (0.077)	1.862 (0.055)	2.084 (0.153)
2.0	2.118 (0.072)	2.070 (0.036)	2.090 (0.076)	2.044 (0.070)	2.359 (0.080)	2.326 (0.096)	2.376 (0.160)	2.215 (0.063)	1.855 (0.145)	2.161 (0.173)
2.2	2.278 (0.060)	2.109 (0.042)	2.231 (0.063)	2.119 (0.036)	2.386 (0.089)	2.296 (0.099)	2.368 (0.115)	2.212 (0.034)	2.011 (0.097)	2.223 (0.125)
2.4	2.404 (0.050)	2.211 (0.042)	2.353 (0.084)	2.150 (0.033)	2.441 (0.044)	2.289 (0.058)	2.450 (0.125)	2.322 (0.025)	2.092 (0.101)	2.301 (0.127)
2.6	2.394 (0.050)	2.317 (0.049)	2.367 (0.088)	2.197 (0.036)	2.516 (0.055)	2.468 (0.083)	2.472 (0.133)	2.279 (0.033)		2.376 (0.109)
2.8					2.543 (0.048)	2.336 (0.065)	2.698 (0.079)			2.526 (0.181)
3.0					2.591 (0.073)					2.591

Table 4.17
Force vs. Speed: Weighted Running, Right Leg
$VGRF_M$ in Body Weights (Standard Deviation)

Speed (m/s)	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Subject 6	Subject 7	Subject 8	Subject 10	Average
1.2									1.723 (0.078)	1.723
1.4	1.881 (0.066)	1.900 (0.026)	1.725 (0.055)	1.823 (0.054)				1.831 (0.077)	1.702 (0.048)	1.810 (0.081)
1.6	1.931 (0.083)	1.958 (0.038)	1.759 (0.063)	1.944 (0.055)			2.000 (0.073)	1.898 (0.047)	1.802 (0.060)	1.899 (0.087)
1.8	2.011 (0.087)	2.086 (0.060)	1.858 (0.040)	2.002 (0.038)	1.968 (0.073)	2.341 (0.055)	2.303 (0.129)	2.094 (0.045)	1.891 (0.048)	2.062 (0.167)
2.0	2.078 (0.056)	2.171 (0.081)	1.952 (0.070)	2.091 (0.080)	2.200 (0.096)	2.200 (0.075)	2.378 (0.085)	2.150 (0.083)	1.921 (0.123)	2.127 (0.138)
2.2	2.239 (0.059)	2.201 (0.048)	2.050 (0.037)	2.129 (0.031)	2.253 (0.100)	2.215 (0.094)	2.413 (0.099)	2.142 (0.044)	1.982 (0.104)	2.180 (0.125)
2.4	2.367 (0.049)	2.318 (0.065)	2.161 (0.108)	2.194 (0.048)	2.377 (0.104)	2.117 (0.060)	2.532 (0.118)	2.265 (0.032)	2.106 (0.113)	2.271 (0.141)
2.6	2.321 (0.073)	2.410 (0.054)	2.268 (0.124)	2.211 (0.038)	2.443 (0.079)	2.319 (0.076)	2.567 (0.133)	2.252 (0.033)		2.349 (0.117)
2.8					2.513 (0.065)	2.253 (0.070)	2.573 (0.131)			2.446 (0.170)
3.0					2.552 (0.053)					2.552

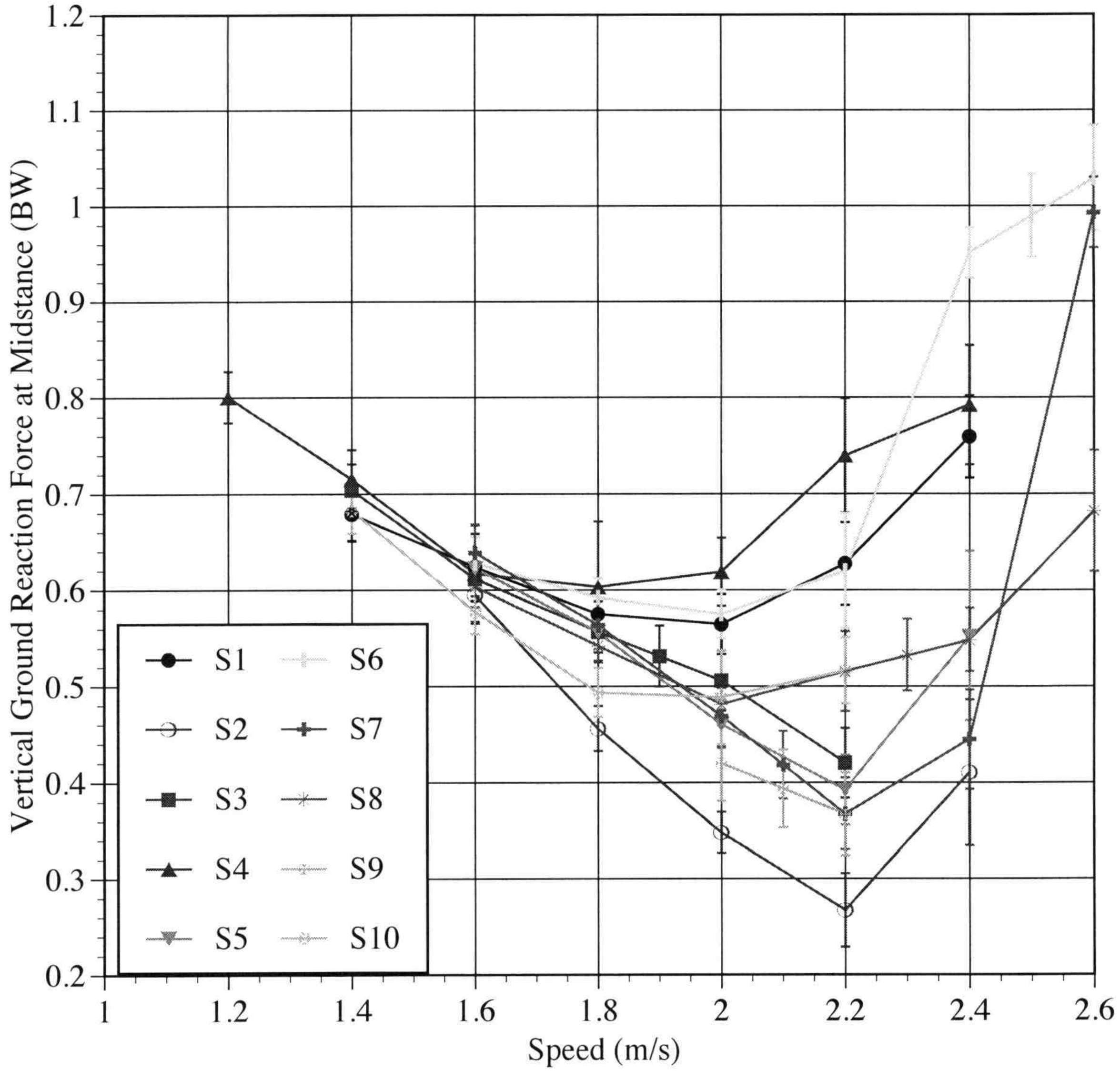

Figure 4.19
Overlaid Curves: Unweighted Walking, Left Leg

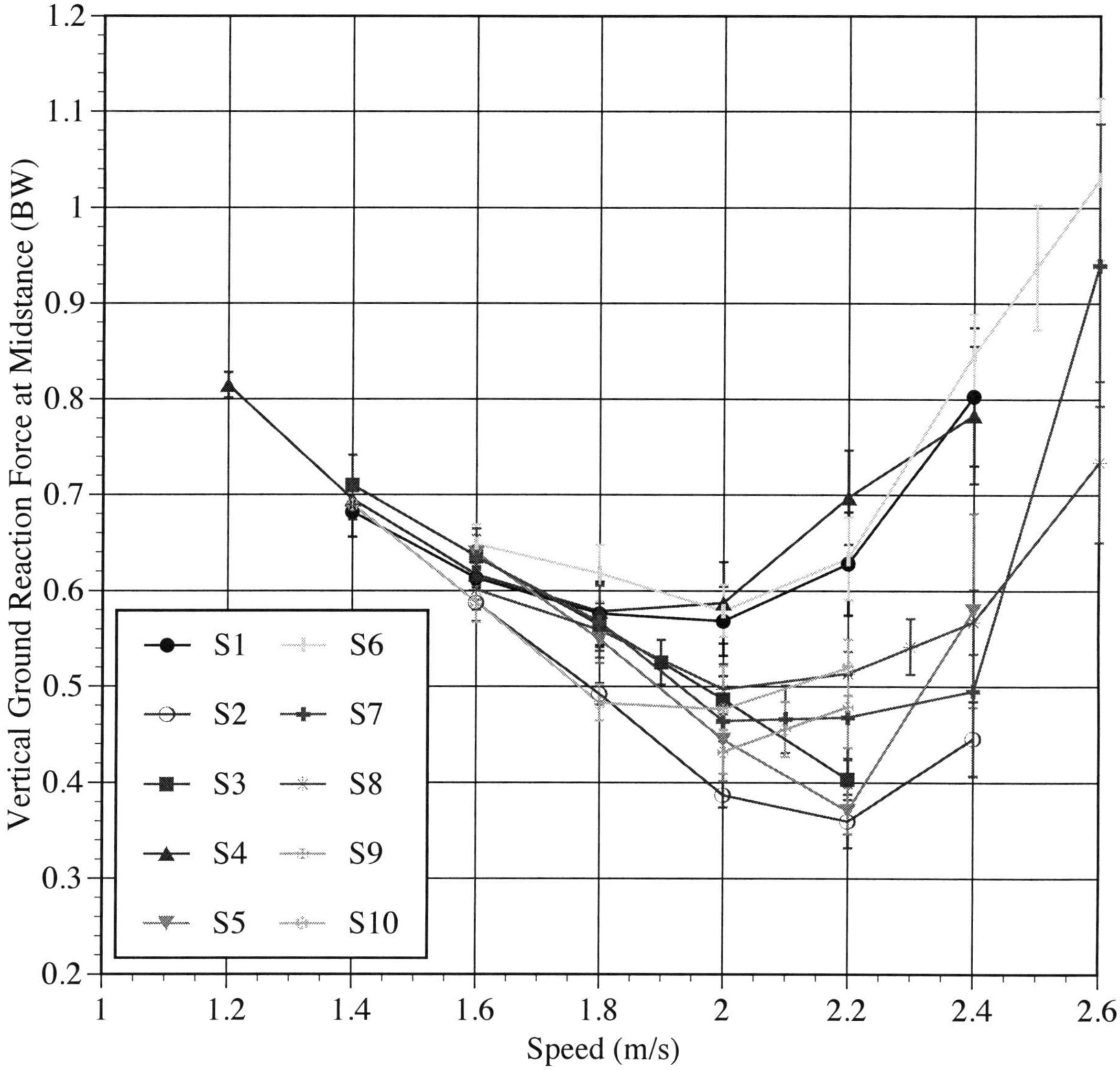

Figure 4.20
Overlaid Curves: Unweighted Walking, Right Leg

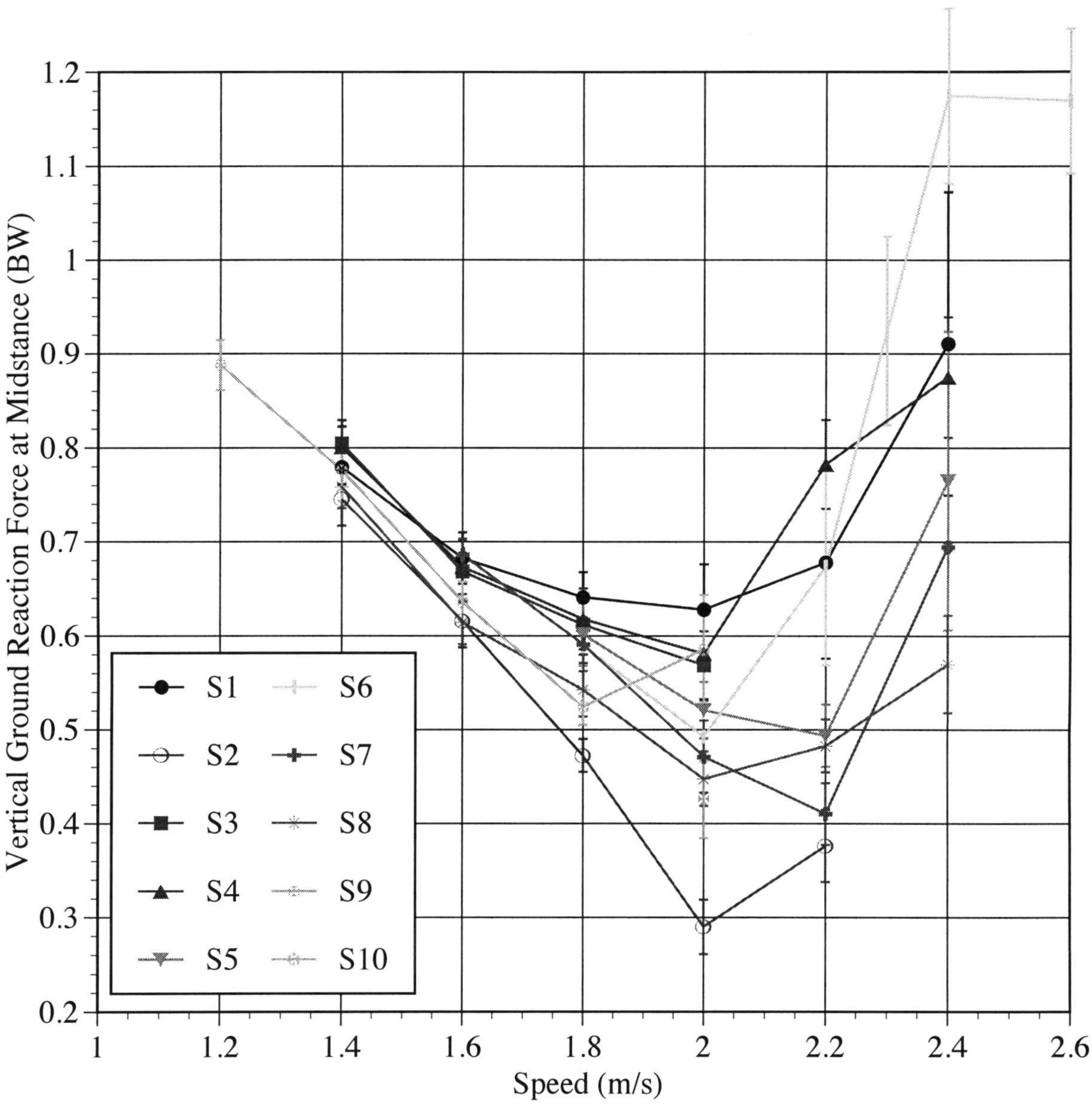

Figure 4.21
Overlaid Curves: Weighted Walking, Left Leg

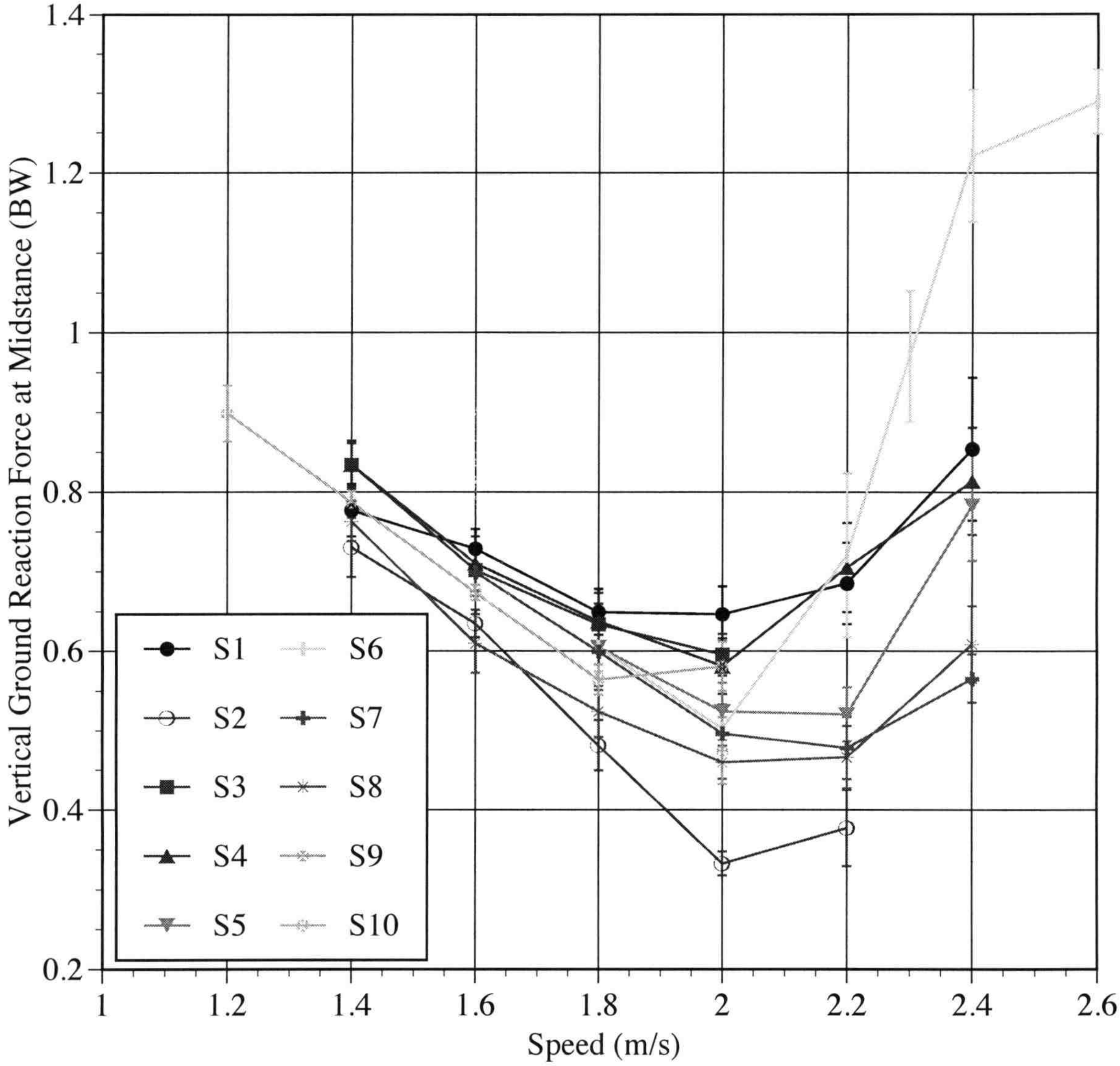

Figure 4.22
Overlaid Curves: Weighted Walking, Right Leg

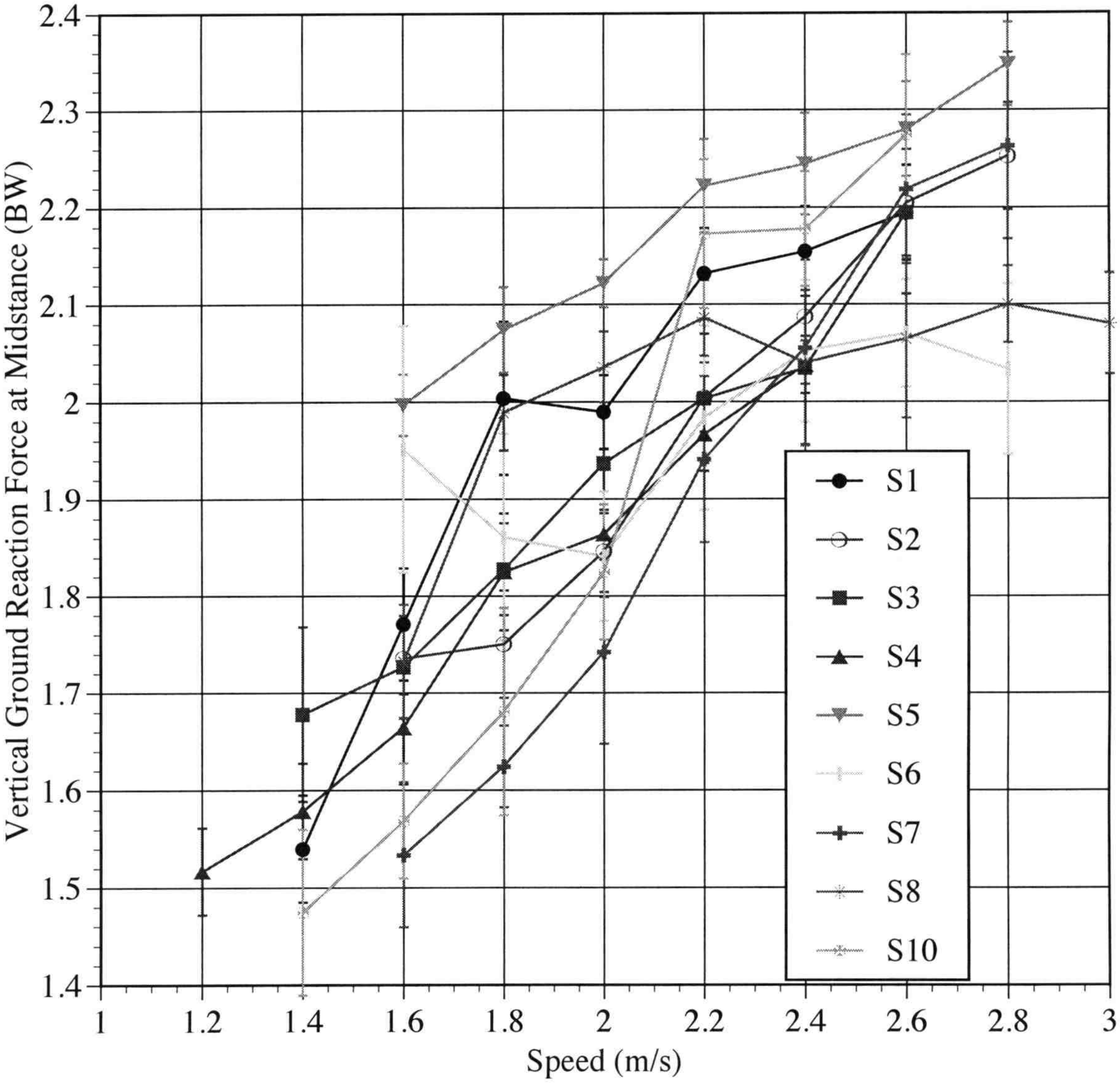

Figure 4.23
Overlaid Curves: Unweighted Running, Left Leg

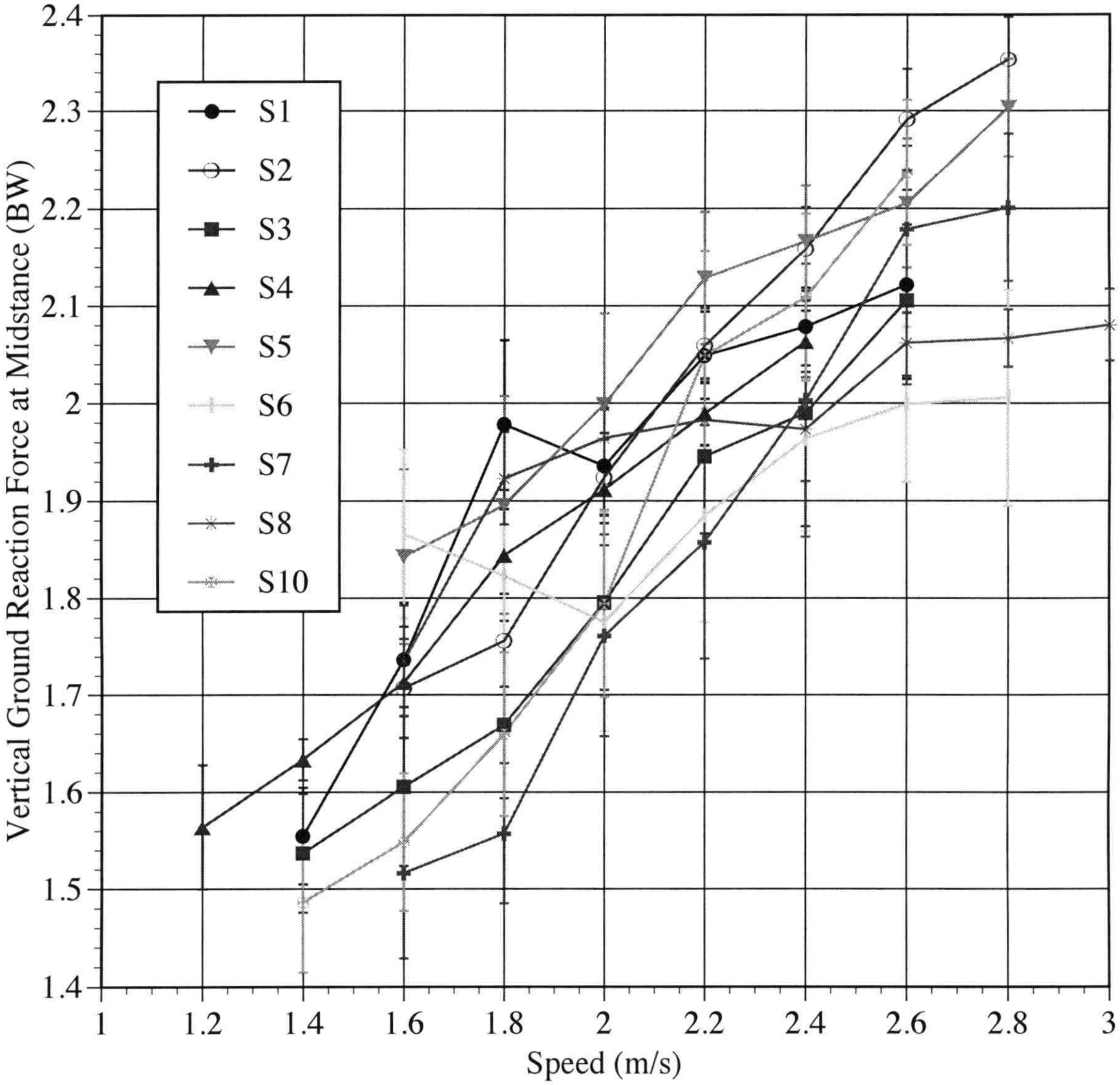

Figure 4.24
Overlaid Curves: Unweighted Running, Right Leg

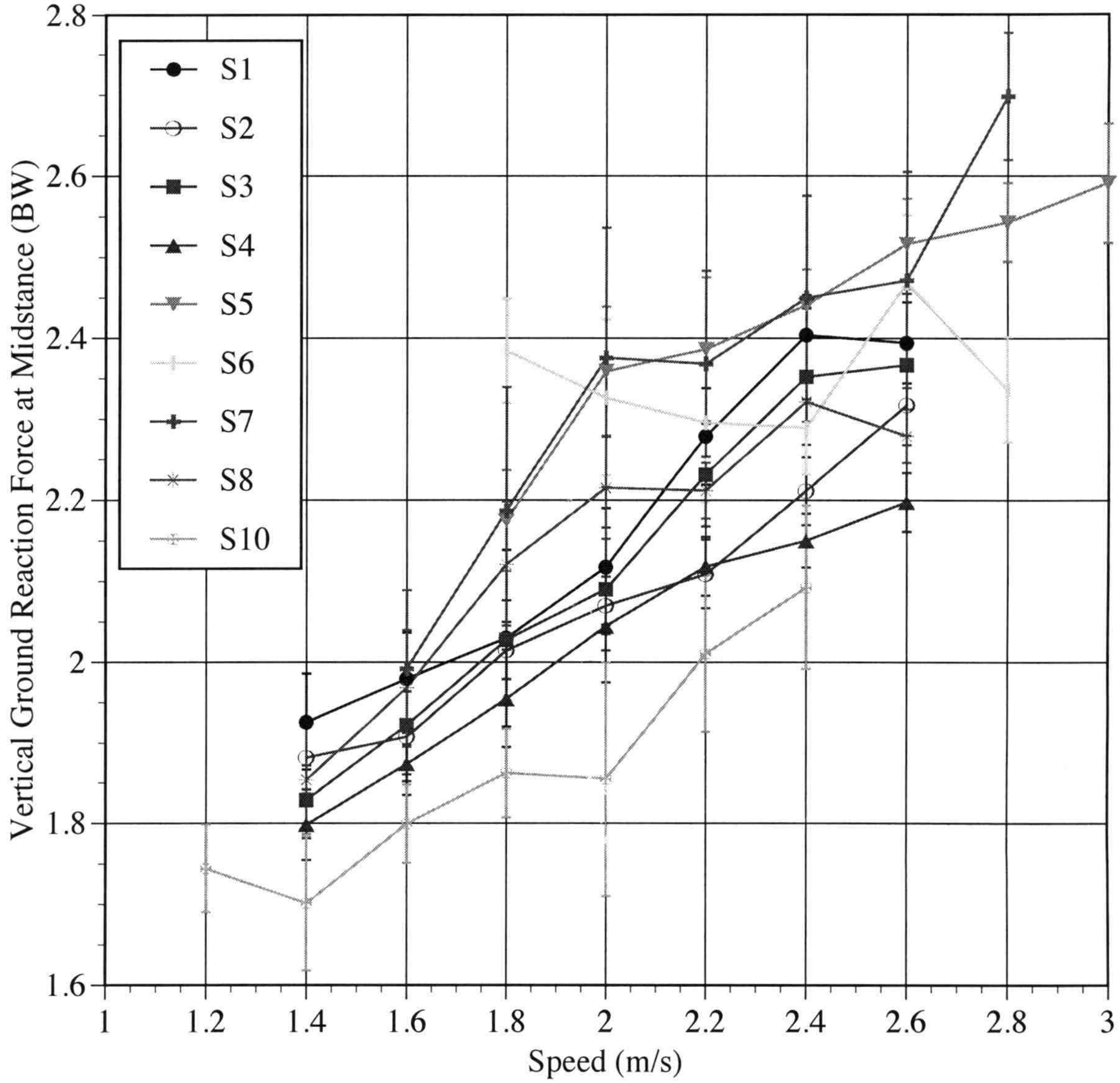

Figure 4.25
Overlaid Curves: Weighted Running, Left Leg

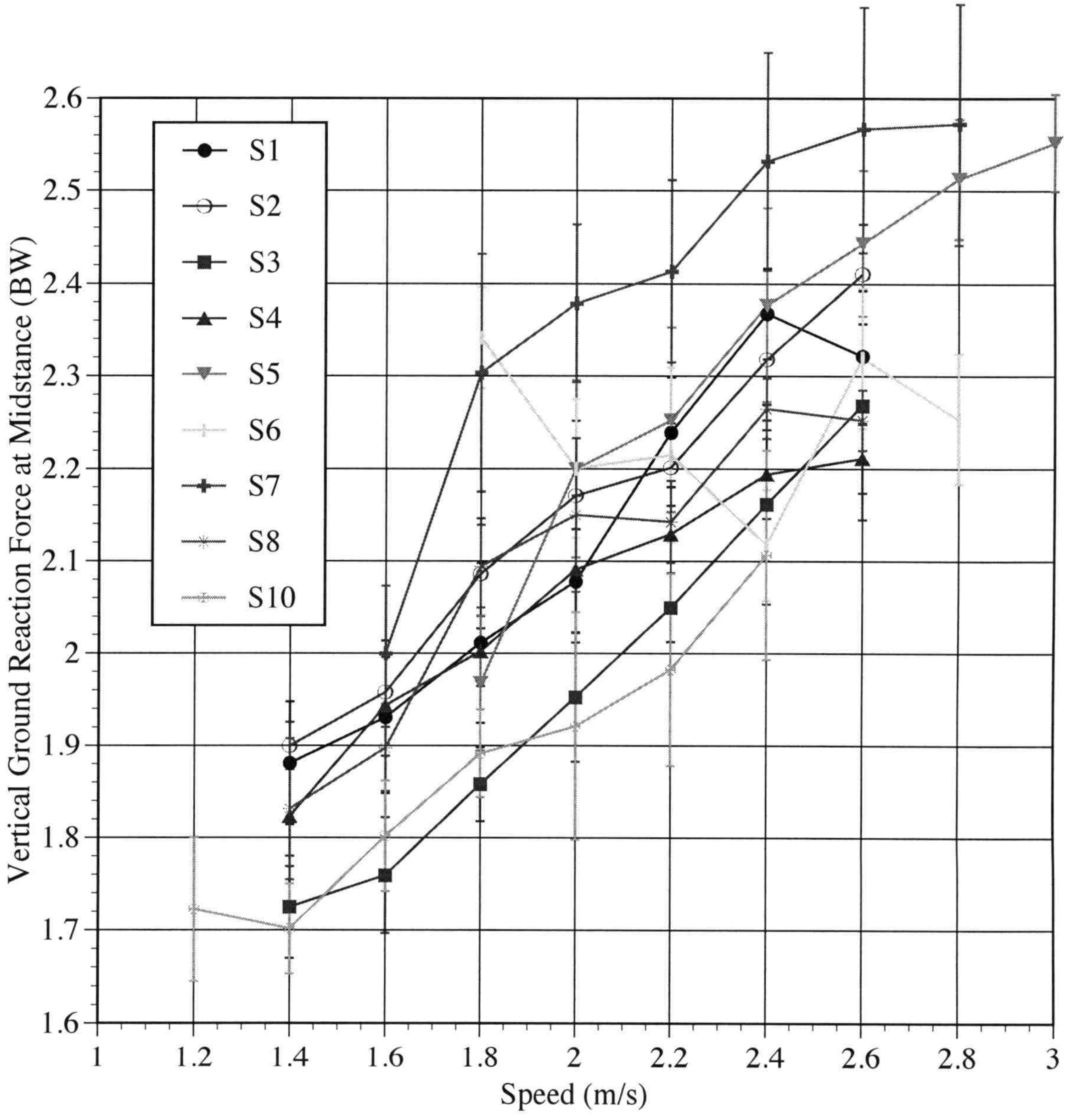

Figure 4.26
Overlaid Curves: Weighted Running, Right Leg

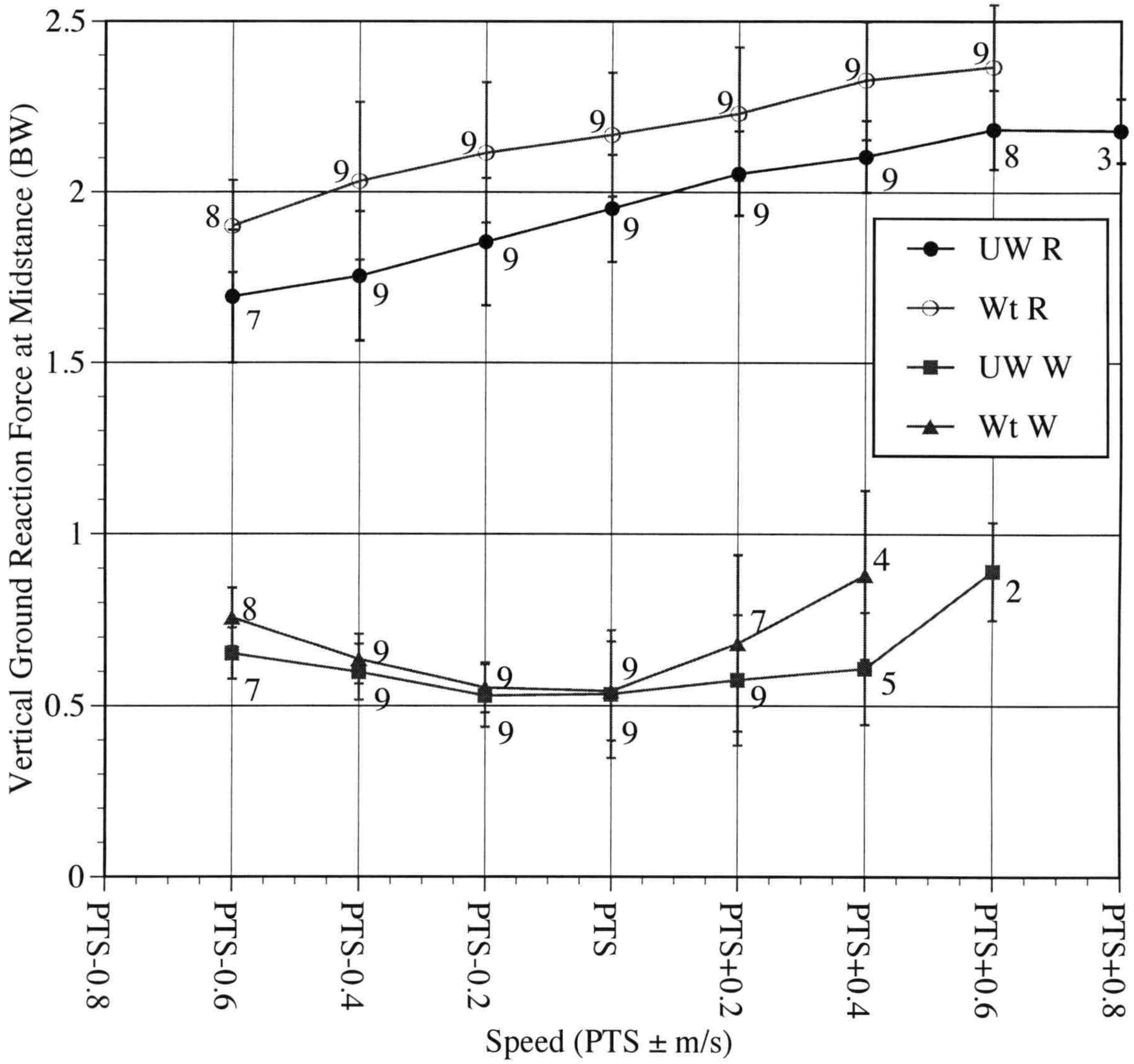

Figure 4.27
Average $VGRF_M$ vs. Speed, Left Leg
Translated to PTS, Averaged Across Subjects
Numbers next to points indicate number of subjects represented in the value. Maximum possible = 9.

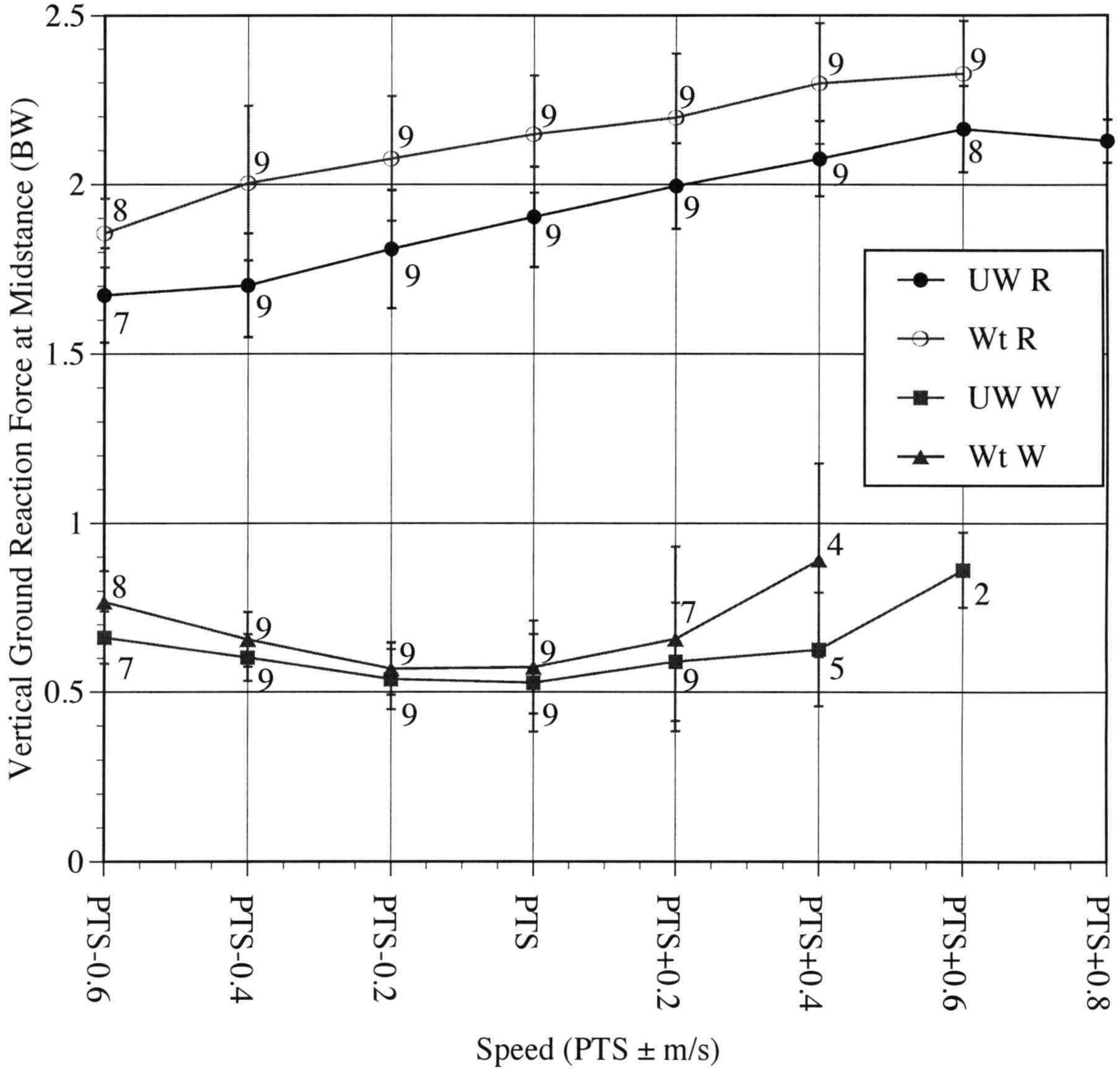

Figure 4.28
Average $VGRF_M$ vs. Speed, Right Leg
Translated to PTS, Averaged Across Subjects
Numbers next to points indicate number of subjects represented in the value. Maximum possible = 9.

the standard deviation among the force values of the subjects averaged. These aligned data may be seen in Tables 4.18 though 4.25. These tables also indicate the trials which were completed.

As mentioned earlier, some subjects transitioned at two different speeds during the determination of preferred transition speed. In order to average across speeds and to retain all subjects in one analysis, the lower value of preferred transition speed was selected for alignment. For example, in the unweighted condition, subject 3 transitioned at 1.80 and 2.00 m/s. For this analysis and the associated statistical analyses, 1.80 m/s was used as the PTS. For the cluster graphs and ΔPTS vs. $\Delta VGRF_M$ graphs, the preferred transition speed was the average of the two trials. Similarly, for the statistical analyses completed on the differences in PTS and $VGRF_M$, the transition speeds were the average of the two trials, and the forces at transition were the result of pooling the forces of the two adjacent speeds.

Once the forces were aligned and averaged (Figures 4.27 and 4.28), the kinetics could be analyzed qualitatively . The forces for both the weighted and unweighted walking conditions seemed to reach a minimum value at or slightly before the preferred transition speed. As speed increased to the PTS, the midstance vertical ground reaction forces declined slightly. At the PTS and higher speeds, the forces increased, most notably in the weighted condition. This general pattern may be seen, but the Bonferroni-adjusted *t*-tests failed to indicate any significant difference between the force levels at PTS - 0.2 m/s, PTS, and PTS + 0.2 m/s. See Tables 4.26 and 4.27 for these comparisons. The described pattern was visible in most of the individual subject data.

For the paired *t*-tests the critical value of Bonferroni's *t* was interpolated for an alpha-level of 0.10, eight degrees of freedom and twelve comparisons, yielding $t(B) = 3.4658$ (Myers & Well, 1991, p. 629). To have statistical significance, the following relationship was required to be true:

Table 4.18
Force Aligned at PTS: Unweighted Walking, Left Leg
$VGRF_M$ in Body Weights (Standard Deviation)

Speed (m/s)	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Subject 6	Subject 7	Subject 8	Subject 10	Average
PTS - 0.8						0.628 (0.027)				0.628
PTS - 0.6	0.679 (0.028)	0.595 (0.029)		0.800 (0.027)	0.623 (0.016)	0.592 (0.020)		0.603 (0.020)	0.682 (0.024)	0.653 (0.075)
PTS - 0.4	0.624 (0.035)	0.456 (0.023)	0.704 (0.027)	0.715 (0.030)	0.556 (0.017)	0.575 (0.025)	0.639 (0.030)	0.542 (0.017)	0.577 (0.023)	0.599 (0.082)
PTS - 0.2	0.575 (0.024)	0.348 (0.021)	0.611 (0.018)	0.617 (0.050)	0.460 (0.025)	0.620 (0.060)	0.563 (0.037)	0.481 (0.017)	0.493 (0.026)	0.530 (0.092)
PTS	0.564 (0.031)	0.267 (0.038)	0.557 (0.032)	0.603 (0.068)	0.392 (0.036)	0.951 (0.026)	0.469 (0.032)	0.515 (0.042)	0.489 (0.049)	0.534 (0.186)
PTS + 0.2	0.627 (0.043)	0.410 (0.076)	0.506 (0.032)	0.619 (0.035)	0.552 (0.088)	1.029 (0.055)	0.367 (0.037)	0.548 (0.033)	0.516 (0.035)	0.575 (0.190)
PTS + 0.4	0.759 (0.042)		0.420 (0.036)	0.739 (0.059)			0.445 (0.052)	0.682 (0.063)		0.609 (0.164)
PTS + 0.6				0.792 (0.062)			0.993 (0.037)			0.892 (0.142)

Table 4.19
Force Aligned at PTS: Unweighted Walking, Right Leg
$VGRF_M$ in Body Weights (Standard Deviation)

Speed (m/s)	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Subject 6	Subject 7	Subject 8	Subject 10	Average
PTS - 0.8						0.648 (0.021)				0.648
PTS - 0.6	0.682 (0.026)	0.588 (0.020)		0.814 (0.013)	0.639 (0.019)	0.618 (0.029)		0.601 (0.021)	0.690 (0.013)	0.662 (0.078)
PTS - 0.4	0.613 (0.010)	0.492 (0.011)	0.710 (0.032)	0.695 (0.021)	0.549 (0.025)	0.579 (0.027)	0.635 (0.030)	0.559 (0.022)	0.588 (0.021)	0.602 (0.070)
PTS - 0.2	0.576 (0.033)	0.387 (0.013)	0.636 (0.011)	0.617 (0.027)	0.445 (0.043)	0.634 (0.043)	0.567 (0.037)	0.497 (0.026)	0.483 (0.019)	0.538 (0.089)
PTS	0.568 (0.036)	0.360 (0.028)	0.564 (0.023)	0.578 (0.028)	0.371 (0.024)	0.846 (0.042)	0.464 (0.021)	0.513 (0.023)	0.476 (0.045)	0.527 (0.144)
PTS + 0.2	0.628 (0.054)	0.446 (0.039)	0.486 (0.024)	0.587 (0.043)	0.579 (0.101)	1.029 (0.084)	0.468 (0.044)	0.567 (0.034)	0.520 (0.029)	0.590 (0.176)
PTS + 0.4	0.803 (0.072)		0.404 (0.021)	0.697 (0.049)			0.495 (0.089)	0.735 (0.084)		0.627 (0.169)
PTS + 0.6				0.783 (0.072)			0.940 (0.147)			0.862 (0.111)

Table 4.20
Force Aligned at PTS: Weighted Walking, Left Leg
$VGRF_M$ in Body Weights (Standard Deviation)

Speed (m/s)	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Subject 6	Subject 7	Subject 8	Subject 10	Average
PTS - 0.6	0.779 (0.018)	0.745 (0.028)	0.804 (0.025)	0.801 (0.022)	0.601 (0.033)		0.686 (0.017)	0.758 (0.023)	0.888 (0.027)	0.758 (0.085)
PTS - 0.4	0.682 (0.027)	0.616 (0.028)	0.668 (0.034)	0.673 (0.037)	0.521 (0.030)	0.589 (0.019)	0.591 (0.029)	0.614 (0.023)	0.776 (0.033)	0.637 (0.073)
PTS - 0.2	0.641 (0.027)	0.473 (0.017)	0.612 (0.032)	0.618 (0.032)	0.494 (0.033)	0.493 (0.028)	0.471 (0.039)	0.543 (0.029)	0.636 (0.021)	0.553 (0.073)
PTS	0.628 (0.048)	0.290 (0.029)	0.569 (0.036)	0.581 (0.050)	0.765 (0.159)	0.674 (0.105)	0.410 (0.033)	0.448 (0.029)	0.524 (0.019)	0.543 (0.144)
PTS + 0.2	0.678 (0.102)	0.376 (0.038)		0.782 (0.047)		1.175 (0.094)	0.695 (0.004)	0.483 (0.028)	0.586 (0.058)	0.682 (0.257)
PTS + 0.4	0.911 (0.162)			0.875 (0.064)		1.169 (0.077)		0.570 (0.052)		0.881 (0.246)

Table 4.21
Force Aligned at PTS: Weighted Walking, Right Leg
$VGRF_M$ in Body Weights (Standard Deviation)

Speed (m/s)	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Subject 6	Subject 7	Subject 8	Subject 10	Average
PTS - 0.6	0.777 (0.033)	0.730 (0.037)	0.834 (0.030)	0.834 (0.027)	0.605 (0.031)		0.699 (0.030)	0.763 (0.025)	0.898 (0.035)	0.768 (0.091)
PTS - 0.4	0.728 (0.025)	0.635 (0.017)	0.702 (0.027)	0.710 (0.034)	0.524 (0.036)	0.605 (0.032)	0.599 (0.048)	0.610 (0.037)	0.787 (0.013)	0.656 (0.082)
PTS - 0.2	0.649 (0.029)	0.481 (0.032)	0.634 (0.026)	0.637 (0.036)	0.520 (0.034)	0.502 (0.023)	0.496 (0.036)	0.524 (0.032)	0.674 (0.010)	0.569 (0.078)
PTS	0.646 (0.035)	0.332 (0.015)	0.595 (0.026)	0.581 (0.035)	0.783 (0.069)	0.721 (0.103)	0.478 (0.039)	0.460 (0.021)	0.564 (0.019)	0.573 (0.138)
PTS + 0.2	0.685 (0.051)	0.377 (0.048)		0.705 (0.056)		1.221 (0.083)	0.566 (0.030)	0.466 (0.039)	0.581 (0.031)	0.657 (0.274)
PTS + 0.4	0.854 (0.090)			0.814 (0.068)		1.289 (0.040)		0.609 (0.048)		0.891 (0.286)

Table 4.22
Force Aligned at PTS: Unweighted Running, Left Leg
$VGRF_M$ in Body Weights (Standard Deviation)

Speed (m/s)	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Subject 6	Subject 7	Subject 8	Subject 10	Average
PTS - 0.8						1.951 (0.127)				1.951
PTS - 0.6	1.540 (0.055)	1.736 (0.037)		1.517 (0.045)	1.997 (0.032)	1.861 (0.107)		1.732 (0.059)	1.475 (0.085)	1.694 (0.194)
PTS - 0.4	1.771 (0.058)	1.750 (0.055)	1.678 (0.090)	1.578 (0.049)	2.074 (0.044)	1.841 (0.067)	1.534 (0.074)	1.989 (0.039)	1.568 (0.059)	1.754 (0.189)
PTS - 0.2	2.004 (0.079)	1.846 (0.042)	1.727 (0.052)	1.664 (0.058)	2.122 (0.025)	1.983 (0.095)	1.624 (0.042)	2.036 (0.036)	1.681 (0.107)	1.854 (0.187)
PTS	1.989 (0.038)	2.004 (0.036)	1.828 (0.048)	1.825 (0.060)	2.222 (0.048)	2.052 (0.073)	1.742 (0.095)	2.086 (0.039)	1.825 (0.070)	1.953 (0.157)
PTS + 0.2	2.132 (0.047)	2.087 (0.058)	1.936 (0.051)	1.864 (0.066)	2.244 (0.052)	2.070 (0.056)	1.941 (0.085)	2.040 (0.022)	2.172 (0.077)	2.054 (0.123)
PTS + 0.4	2.155 (0.046)	2.205 (0.055)	2.003 (0.067)	1.966 (0.038)	2.280 (0.049)	2.033 (0.088)	2.055 (0.101)	2.065 (0.082)	2.178 (0.059)	2.104 (0.104)
PTS + 0.6	2.194 (0.049)	2.252 (0.055)	2.035 (0.079)	2.038 (0.029)	2.347 (0.043)		2.218 (0.076)	2.100 (0.039)	2.274 (0.083)	2.182 (0.114)
PTS + 0.8			2.195 (0.084)				2.263 (0.096)	2.080 (0.052)		2.179 (0.092)

Table 4.23
Force Aligned at PTS: Unweighted Running, Right Leg
$VGRF_M$ in Body Weights (Standard Deviation)

Speed (m/s)	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Subject 6	Subject 7	Subject 8	Subject 10	Average
PTS - 0.8						1.866 (0.087)				1.866
PTS - 0.6	1.554 (0.050)	1.707 (0.051)		1.564 (0.064)	1.842 (0.090)	1.822 (0.073)		1.735 (0.057)	1.486 (0.072)	1.673 (0.140)
PTS - 0.4	1.736 (0.059)	1.756 (0.048)	1.537 (0.061)	1.633 (0.021)	1.895 (0.112)	1.775 (0.113)	1.516 (0.087)	1.923 (0.048)	1.548 (0.071)	1.702 (0.152)
PTS - 0.2	1.978 (0.087)	1.924 (0.039)	1.605 (0.082)	1.713 (0.058)	1.999 (0.093)	1.884 (0.109)	1.557 (0.072)	1.964 (0.029)	1.660 (0.085)	1.809 (0.175)
PTS	1.936 (0.059)	2.059 (0.039)	1.669 (0.075)	1.844 (0.067)	2.128 (0.068)	1.964 (0.095)	1.761 (0.104)	1.983 (0.042)	1.794 (0.097)	1.904 (0.149)
PTS + 0.2	2.049 (0.045)	2.158 (0.043)	1.795 (0.090)	1.912 (0.058)	2.166 (0.057)	1.999 (0.080)	1.857 (0.120)	1.973 (0.053)	2.049 (0.107)	1.995 (0.126)
PTS + 0.4	2.078 (0.040)	2.291 (0.052)	1.946 (0.079)	1.989 (0.032)	2.205 (0.066)	2.006 (0.111)	2.003 (0.140)	2.062 (0.043)	2.109 (0.086)	2.076 (0.111)
PTS + 0.6	2.122 (0.097)	2.353 (0.044)	1.990 (0.116)	2.063 (0.031)	2.303 (0.050)		2.178 (0.085)	2.067 (0.029)	2.237 (0.074)	2.164 (0.127)
PTS + 0.8			2.106 (0.078)				2.201 (0.075)	2.080 (0.037)		2.129 (0.063)

Table 4.24
Force Aligned at PTS: Weighted Running, Left Leg
$VGRF_M$ in Body Weights (Standard Deviation)

Speed (m/s)	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Subject 6	Subject 7	Subject 8	Subject 10	Average
PTS - 0.6	1.926 (0.060)	1.881 (0.045)	1.828 (0.043)	1.798 (0.043)	2.175 (0.062)		1.992 (0.097)	1.854 (0.072)	1.744 (0.054)	1.900 (0.135)
PTS - 0.4	1.980 (0.057)	1.908 (0.056)	1.922 (0.062)	1.873 (0.039)	2.359 (0.080)	2.384 (0.064)	2.187 (0.152)	1.969 (0.071)	1.701 (0.083)	2.031 (0.230)
PTS - 0.2	2.030 (0.109)	2.014 (0.035)	2.028 (0.049)	1.955 (0.060)	2.386 (0.089)	2.326 (0.096)	2.376 (0.160)	2.121 (0.077)	1.799 (0.048)	2.115 (0.205)
PTS	2.118 (0.072)	2.070 (0.036)	2.090 (0.076)	2.044 (0.070)	2.441 (0.044)	2.296 (0.099)	2.368 (0.115)	2.215 (0.063)	1.862 (0.055)	2.167 (0.180)
PTS + 0.2	2.278 (0.060)	2.109 (0.042)	2.231 (0.063)	2.119 (0.036)	2.516 (0.055)	2.289 (0.058)	2.450 (0.125)	2.212 (0.034)	1.855 (0.145)	2.229 (0.195)
PTS + 0.4	2.404 (0.050)	2.211 (0.042)	2.353 (0.084)	2.150 (0.033)	2.543 (0.048)	2.468 (0.083)	2.472 (0.133)	2.322 (0.025)	2.011 (0.097)	2.326 (0.173)
PTS + 0.6	2.394 (0.050)	2.317 (0.049)	2.367 (0.088)	2.197 (0.036)	2.591 (0.073)	2.336 (0.065)	2.698 (0.079)	2.279 (0.033)	2.092 (0.101)	2.364 (0.186)

Table 4.25
Force Aligned at PTS: Weighted Running, Right Leg
$VGRF_M$ in Body Weights (Standard Deviation)

Speed (m/s)	Subject 1	Subject 2	Subject 3	Subject 4	Subject 5	Subject 6	Subject 7	Subject 8	Subject 10	Average
PTS - 0.6	1.881 (0.066)	1.900 (0.026)	1.725 (0.055)	1.823 (0.054)	1.968 (0.073)		2.000 (0.073)	1.831 (0.077)	1.723 (0.078)	1.856 (0.102)
PTS - 0.4	1.931 (0.083)	1.958 (0.038)	1.759 (0.063)	1.944 (0.055)	2.200 (0.096)	2.341 (0.055)	2.303 (0.129)	1.898 (0.047)	1.702 (0.048)	2.004 (0.228)
PTS - 0.2	2.011 (0.087)	2.086 (0.060)	1.858 (0.040)	2.002 (0.038)	2.253 (0.100)	2.200 (0.075)	2.378 (0.085)	2.094 (0.045)	1.802 (0.060)	2.076 (0.184)
PTS	2.078 (0.056)	2.171 (0.081)	1.952 (0.070)	2.091 (0.080)	2.377 (0.104)	2.215 (0.094)	2.413 (0.099)	2.150 (0.083)	1.891 (0.048)	2.149 (0.173)
PTS + 0.2	2.239 (0.059)	2.201 (0.048)	2.050 (0.037)	2.129 (0.031)	2.443 (0.079)	2.117 (0.060)	2.532 (0.118)	2.142 (0.044)	1.921 (0.123)	2.197 (0.189)
PTS + 0.4	2.367 (0.049)	2.318 (0.065)	2.161 (0.108)	2.194 (0.048)	2.513 (0.065)	2.319 (0.076)	2.567 (0.133)	2.265 (0.032)	1.982 (0.104)	2.299 (0.178)
PTS + 0.6	2.321 (0.073)	2.410 (0.054)	2.268 (0.124)	2.211 (0.038)	2.552 (0.053)	2.253 (0.070)	2.573 (0.131)	2.252 (0.033)	2.106 (0.113)	2.328 (0.156)

Table 4.26
Significant Differences of Forces, Left Leg
Means of Forces in Body Weights

Condition	Pair Tested	$\overline{X_1}$	$\overline{X_2}$	Mean Squared Error	$\lvert\overline{X_1}-\overline{X_2}\rvert$	$t(B)\sqrt{\frac{2MS_e}{n}}$	Significance
UW R	1 to 2	1.854	1.953	0.00163	0.098	0.066	Yes
UW R	1 to 3	1.854	2.054	0.01016	0.200	0.165	Yes
UW R	2 to 3	1.953	2.054	0.00693	0.101	0.136	No
Wt R	1 to 2	2.115	2.167	0.00094	0.052	0.050	Yes
Wt R	1 to 3	2.115	2.229	0.00358	0.114	0.098	Yes
Wt R	2 to 3	2.167	2.229	0.00193	0.062	0.072	No
UW W	1 to 2	0.530	0.534	0.00836	0.004	0.149	No
UW W	1 to 3	0.530	0.575	0.01370	0.045	0.191	No
UW W	2 to 3	0.534	0.575	0.00352	0.041	0.097	No
Wt W	1 to 2	0.533	0.543	0.01046	0.010	0.167	No
Wt W	1 to 3	0.533	0.531	0.07331	0.003	0.442	No
Wt W	2 to 3	0.543	0.531	0.08095	0.013	0.465	No

Table 4.27
Significant Differences of Forces, Right Leg
Means of Forces in Body Weights

Condition	Pair Tested	$\overline{X_1}$	$\overline{X_2}$	Mean Squared Error	$\lvert\overline{X_1}-\overline{X_2}\rvert$	$t(B)\sqrt{\frac{2MS_e}{n}}$	Significance
UW R	1 to 2	1.809	1.904	0.00270	0.095	0.085	Yes
UW R	1 to 3	1.809	1.995	0.00667	0.186	0.133	Yes
UW R	2 to 3	1.904	1.995	0.00284	0.091	0.087	Yes
Wt R	1 to 2	2.076	2.149	0.00055	0.073	0.038	Yes
Wt R	1 to 3	2.076	2.197	0.00434	0.121	0.108	Yes
Wt R	2 to 3	2.149	2.197	0.00287	0.048	0.087	No
UW W	1 to 2	0.538	0.527	0.00423	0.011	0.106	No
UW W	1 to 3	0.538	0.590	0.01221	0.052	0.181	No
UW W	2 to 3	0.527	0.590	0.00392	0.063	0.102	No
Wt W	1 to 2	0.569	0.573	0.01002	0.005	0.164	No
Wt W	1 to 3	0.569	0.511	0.07459	0.057	0.446	No
Wt W	2 to 3	0.573	0.511	0.07559	0.062	0.449	No

$$\left|\overline{X}_1 - \overline{X}_2\right| \geq t(B)\sqrt{\frac{2MS_e}{n}}$$

In this equation, X_1 and X_2 were the means being compared, MS_e was the mean squared error for the comparison, and n was 9 (Portney & Watkins, 1993, p. 407).

The dynamics of the $VGRF_M$ vs. speed relationship for walking were indicative of the changes which were observed to occur to the vertical ground reaction force curve for each stride as speed was increased. The normal walking curve, formed by two peaks flanking a trough at approximately midstance, changed with speed. As speed was increased, the peaks increased in height and the trough deepened, actually getting closer to zero-force. Once the preferred transition speed was passed, the curve changed. The first peak increased in height and became broader while the second peak shrank and the trough was absorbed between these two peaks. The force level at the bottom of the trough increased, and the trough moved later in the stance. At high walking speeds, the trough was only visible as a slight dip on the decline from the initial peak to zero. After PTS there was a change, not only in the midstance force, but also in the general shape of the vertical ground reaction force curve.

With increasing speed, midstance vertical ground reaction forces for running increased, as indicated by Figures 4.27 and 4.28. For the left side, the forces at PTS and PTS + 0.2 m/s were significantly different from those at PTS - 0.2 m/s for both weighted and unweighted conditions. On the right side, the forces during unweighted running also significantly increased between PTS and PTS + 0.2 m/s. These Bonferroni-adjusted comparisons are listed in Tables 4.26 and 4.27.

In the walking trials conducted for this study, a trend was observed for the $VGRF_M$ levels to increase at and above the PTS. When subjects walked at speeds higher than the PTS, the forces stopped declining and began to increase. When tested statistically, the conservative statistical procedures performed were unable to confirm this trend with significant differences between the forces above and below the PTS. While the results failed to reject the null hypothesis, this trend supports the hypothesis that musculoskeletal forces are a control parameter in the walk-run transition. Significant increases were observed in $VGRF_M$ with increased running speed; however these differences do not influence the hypothesis under study.

The dynamics of kinetic control parameters have been observed by Farley and Taylor (1991) and hypothesized by Hreljac (1993a). For a musculoskeletal force to be a control parameter, the force would show a consistent increase or decrease with walking speed (or trotting speed, in the case of horses) up to the PTS. At the PTS the gait would switch to running (or galloping) and the force level should change substantially. Ideally this level would decrease with the change in gaits, then increase past the critical level with increased running speed. Alternately it would be possible for the force level of a control parameter to increase substantially with the change in gait and continue to increase at higher speeds. Farley and Taylor found the musculoskeletal forces of horses to increase with trotting speed, then after reaching a critical level at the PTS, to decrease with the switch to galloping. At higher speeds of galloping, the forces increased past the critical level.

The data from the present study demonstrated a similar pattern in the force curves, except with the opposite direction of change. With increased walking speed, the midstance vertical ground reaction forces declined to a minimum level near the PTS. After the switch to running, the force levels made a substantial jump. With higher speeds of running, the forces increased. In comparison to the findings of Farley and Taylor (1991), these dynamics indicate a critical minimum level of force rather than a critical maximum value. The opposite direction of change of force still agrees with the hypothesized dynamics of a control parameter.

4.4 Variation in Midstance Force

The curves of average walking force across subjects seemed to indicate an increase in variability at and above the preferred transition speed. This was obvious at the highest walking speeds when the subjects were spontaneously switching to running. While the variability in choice of gait could not be quantified, the variability in $VGRF_M$ could be summarized. To properly quantify this variability, the variances for each subject were pooled for each speed. See Tables 4.28 and 4.29. When plotted versus speed, a clear increase in variability occurs at the PTS for both walking curves. The variability in the running curves does not follow a clear pattern of increase or decrease after the preferred transition speed. See Figures 4.29 and 4.30. Using an alpha-level of 0.10, n = 9, and correlations calculated from the average force values, the t-statistics for the paired comparisons were calculated using the following equation:

$$t = \frac{s_1^2 - s_2^2}{\sqrt{\frac{4s_1^2 s_2^2}{n-2}(1 - r_{12}^2)}}$$

When compared to the critical Bonferroni-adjusted *t*-value for twelve comparisons, only one significant difference was found. See Tables 4.30 and 4.31 for these comparisons. On the right side, the variability in unweighted walking was significantly different between PTS and PTS + 0.2 m/s. Even this single significant comparison supports the pattern evident for walking. The lack of statistical significance for more tests and the failure to reject the null hypothesis does not mean that differences do not exist. By having a small sample size (*n*=9) and 12 comparisons (3 speeds * 2 conditions * 2 sides) the adjusted level of significance became very difficult to satisfy. If more subjects were tested or fewer comparisons had been made, it may have been possible to find statistical significance in addition to the pattern of change.

Table 4.28
Variance Pooled Across Subjects, Left Leg
Aligned at Preferred Transition Speed
(Body Weight^2)

Speed	Unweighted Walking	Weighted Walking	Unweighted Running	Weighted Running
PTS - 0.6	0.000584	0.000620	0.004248	0.003874
PTS - 0.4	0.000645	0.000857	0.003723	0.006538
PTS - 0.2	0.001218	0.000856	0.004201	0.007950
PTS	0.001683	0.005444	0.003513	0.005429
PTS + 0.2	0.002552	0.004747	0.003520	0.006149
PTS + 0.4	0.003077	0.011006	0.004595	0.005302
PTS + 0.6	0.006989		0.003552	0.004471
PTS + 0.8			0.007373	

Table 4.29
Variance Pooled Across Subjects, Right Leg
Aligned at Preferred Transition Speed
(Body Weight^2)

Speed	Unweighted Walking	Weighted Walking	Unweighted Running	Weighted Running
PTS - 0.6	0.000481	0.000987	0.004544	0.004214
PTS - 0.4	0.000527	0.001004	0.005482	0.005418
PTS - 0.2	0.000923	0.000898	0.005899	0.004727
PTS	0.000993	0.002535	0.005575	0.006680
PTS + 0.2	0.003666	0.003106	0.005791	0.005439
PTS + 0.4	0.005398	0.004523	0.006180	0.006571
PTS + 0.6	0.016482		0.005134	0.006851
PTS + 0.8			0.005178	

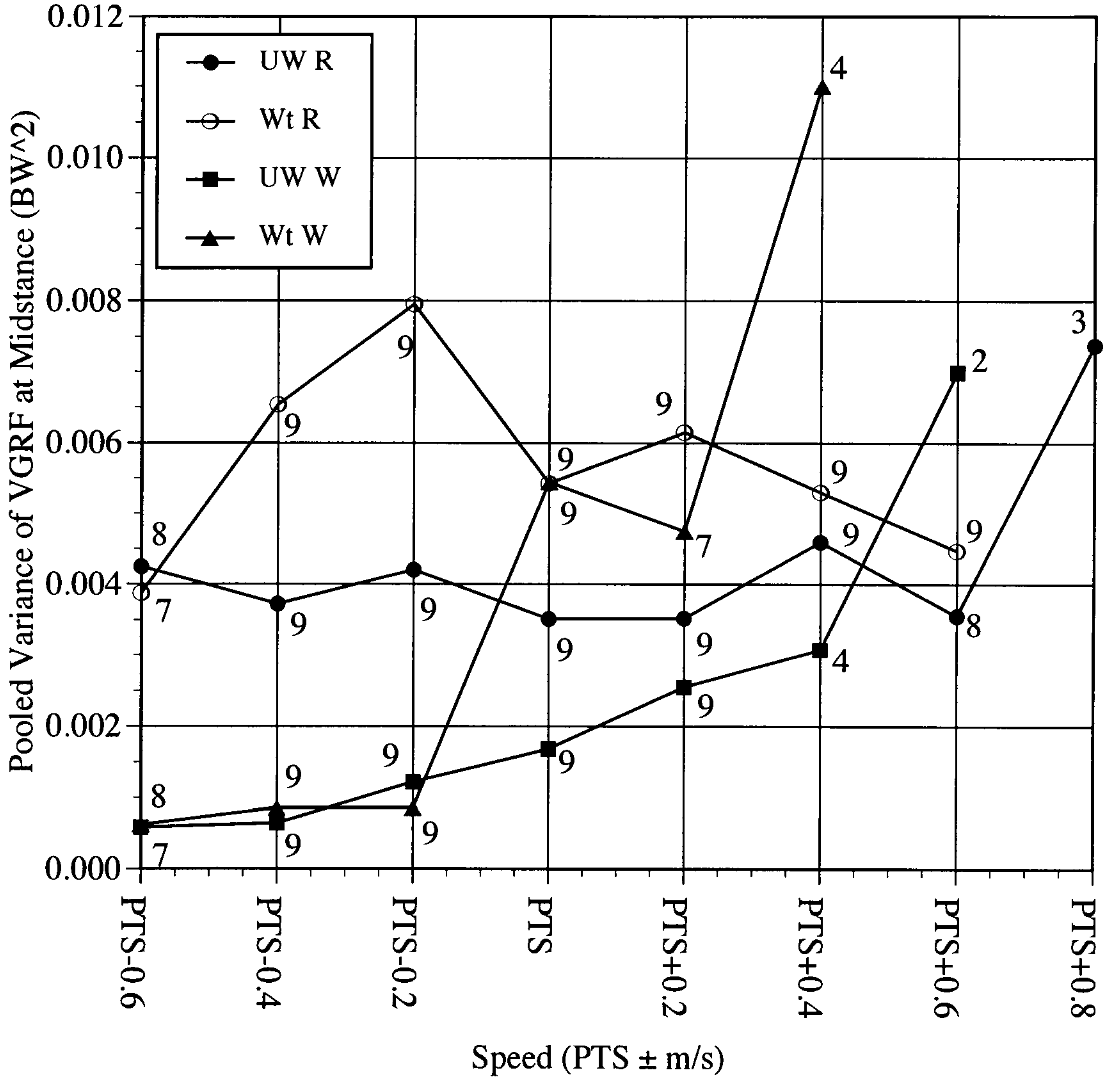

Figure 4.29
Variability of Force: Left Leg
Pooled variances of midstance vertical ground reaction force vs. speed.
Numbers next to points indicate number of subjects
represented in the value. Maximum possible = 9.

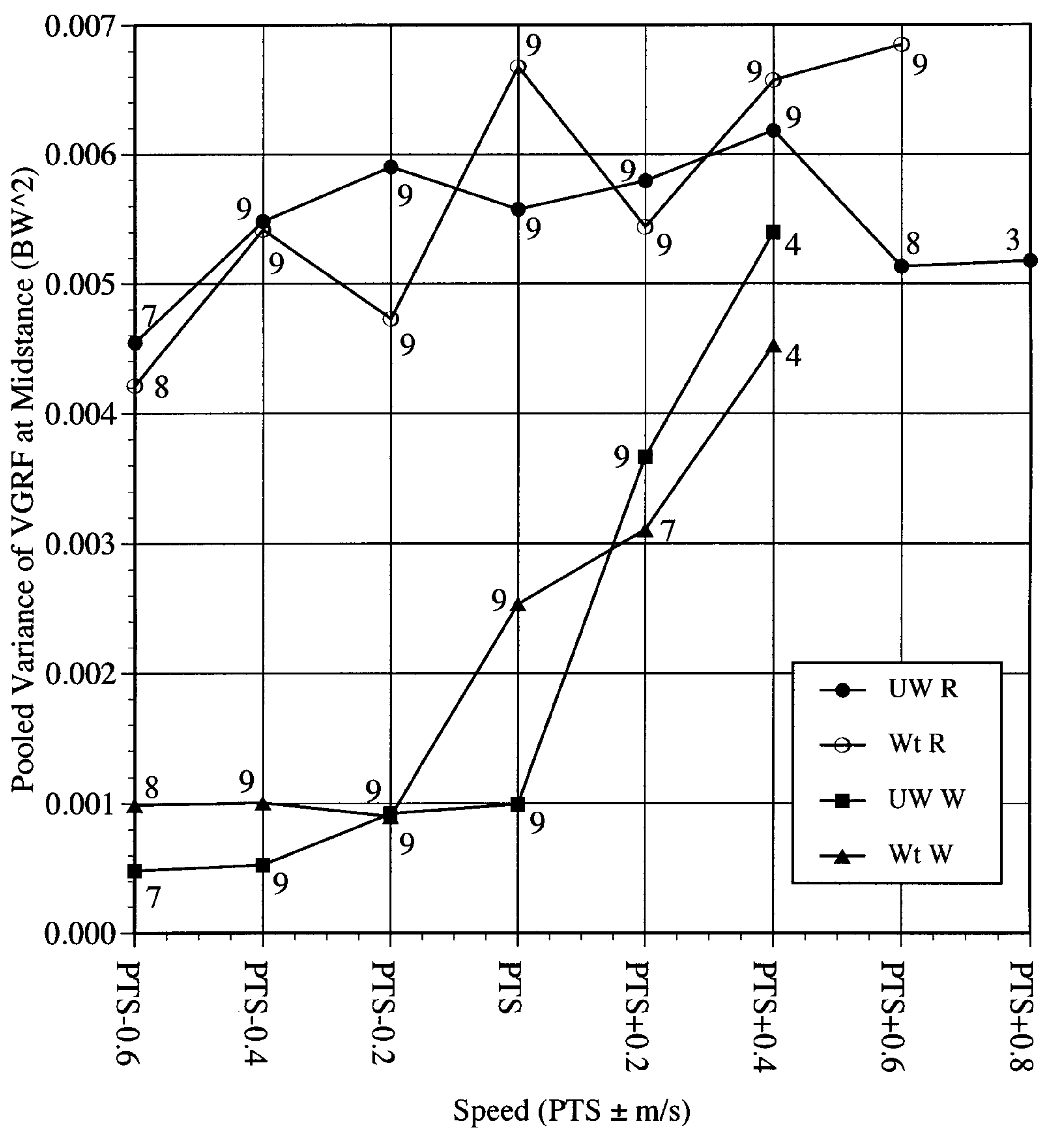

Figure 4.30
Variability of Force: Right Leg
Pooled variances of midstance vertical ground reaction force vs. speed.
Numbers next to points indicate number of subjects
represented in the value. Maximum possible = 9.

Table 4.30
Significant Differences of Variances, Left Leg
Pooled Variances in Body Weights[2]

Condition	Pair Tested	$\hat{\sigma}^2_{pooled1}$	$\hat{\sigma}^2_{pooled2}$	t	$t(B)$	Significance
UW R	1 to 2	0.004201	0.003513	0.8482	3.6218	No
UW R	1 to 3	0.004201	0.003520	0.3071	3.6218	No
UW R	2 to 3	0.003513	0.003520	0.0032	3.6218	No
Wt R	1 to 2	0.007950	0.005429	2.7546	3.6218	No
Wt R	1 to 3	0.007950	0.006149	0.8272	3.6218	No
Wt R	2 to 3	0.005429	0.006149	0.5185	3.6218	No
UW W	1 to 2	0.001218	0.001683	0.6765	3.6218	No
UW W	1 to 3	0.001218	0.002552	1.1542	3.6218	No
UW W	2 to 3	0.001683	0.002552	1.2837	3.6218	No
Wt W	1 to 2	0.000856	0.005444	2.9043	3.6218	No
Wt W	1 to 3	0.000856	0.004583	2.4935	3.6218	No
Wt W	2 to 3	0.005444	0.004583	0.3756	3.6218	No

Table 4.31
Significant Differences of Variances, Right Leg
Pooled Variances in Body Weights[2]

Condition	Pair Tested	$\hat{\sigma}^2_{pooled1}$	$\hat{\sigma}^2_{pooled2}$	t	$t(B)$	Significance
UW R	1 to 2	0.005899	0.005575	0.1783	3.6218	No
UW R	1 to 3	0.005899	0.005791	0.0369	3.6218	No
UW R	2 to 3	0.005575	0.005791	0.0993	3.6218	No
Wt R	1 to 2	0.004727	0.006680	2.6052	3.6218	No
Wt R	1 to 3	0.004727	0.005439	0.3851	3.6218	No
Wt R	2 to 3	0.006680	0.005439	0.6830	3.6218	No
UW W	1 to 2	0.000923	0.000993	0.1578	3.6218	No
UW W	1 to 3	0.000923	0.003666	2.2184	3.6218	No
UW W	2 to 3	0.000993	0.003666	3.6817	3.6218	Yes
Wt W	1 to 2	0.000898	0.002535	1.4767	3.6218	No
Wt W	1 to 3	0.000898	0.003004	1.6964	3.6218	No
Wt W	2 to 3	0.002535	0.003004	0.4548	3.6218	No

Thelen and Smith (1994) argue that an instability or perturbation will result from a control parameter being scaled to a critical level. The instability, indicated by increased variability, will then cause a switch in the collective variable from one attractor state to another. After the switch has occurred, variability should decrease. In the case under study, a perturbation would cause a switch from walking to running. The possible control parameter being investigated, $VGRF_M$, was measured for variability near PTS. The resulting pattern of variability increasing at and above the PTS for walking may be indicative of a perturbation to the collective variable of gait. This suggests a relationship between midstance vertical ground reaction force and the walk-run transition. If the variation in the running forces decreased at and above the preferred transition speed, this relationship would be further strengthened. The observed variability changes suggest that musculoskeletal or vertical ground reaction force may be a variable for further study.

4.5 Cluster Graphs

This experiment was designed to differentiate between the influences of speed and vertical ground reaction force on the walk-run transition. If force had a greater role in the transition than speed, then the subjects would transition at different speeds, but at approximately the same force for the two conditions. The first test quantified the preferred transition speeds for each condition. The corresponding force was then picked from the walking values measured in test two. For this analysis, if a PTS was between two of the measurement speeds, the average speed was used. Likewise, the vertical ground reaction force at midstance was interpolated from the forces at the adjacent speeds and the standard deviation was pooled. These data appear in Tables 4.32 and 4.33.

These values of midstance force and preferred transition speed were plotted for each subject and condition (see Figures 4.31 & 4.32). The average values for the weighted and unweighted conditions were overlaid. On these graphs the two conditions are mixed together; no clear clusters were formed for the two conditions. Thus the

Table 4.32
$VGRF_M$ at the Preferred Transition Speed, Left Leg
Forces in Body Weights

	Unweighted			Weighted			Difference	Standard
Subject	Force	SD	n	Force	SD	n	(Wt - UW)	Error
1	0.564	0.031	9	0.628	0.048	11	0.064	0.019
2	0.267	0.038	12	0.290	0.029	11	0.023	0.014
3	0.531	0.032	19	0.569	0.036	11	0.038	0.013
4	0.603	0.068	10	0.581	0.050	12	-0.022	0.025
5	0.392	0.036	11	0.765	0.159	12	0.373	0.049
6	0.990	0.043	24	0.924	0.100	22	-0.065	0.022
7	0.418	0.035	19	0.410	0.033	11	-0.008	0.013
8	0.533	0.037	23	0.448	0.029	11	-0.085	0.013
9	0.393	0.040	15	0.426	0.042	12	0.033	0.016
10	0.489	0.049	10	0.524	0.019	9	0.036	0.018
Average	0.518			0.558			0.039	0.020
SD	0.194			0.191			0.127	0.011

Table 4.33
$VGRF_M$ at the Preferred Transition Speed, Right Leg
Forces in Body Weights

	Unweighted			Weighted			Difference	Standard
Subject	Force	SD	n	Force	SD	n	(Wt - UW)	Error
1	0.568	0.036	11	0.646	0.035	11	0.078	0.015
2	0.360	0.028	12	0.332	0.015	11	-0.027	0.009
3	0.525	0.023	16	0.595	0.026	10	0.070	0.010
4	0.578	0.028	10	0.581	0.035	12	0.003	0.014
5	0.371	0.024	10	0.783	0.069	12	0.413	0.023
6	0.938	0.065	26	0.971	0.082	25	0.033	0.021
7	0.466	0.035	19	0.478	0.039	11	0.012	0.014
8	0.542	0.029	23	0.460	0.021	11	-0.082	0.010
9	0.455	0.029	15	0.475	0.042	11	0.019	0.014
10	0.476	0.045	10	0.564	0.019	9	0.088	0.016
Average	0.527			0.591			0.061	0.015
SD	0.166			0.192			0.133	0.005

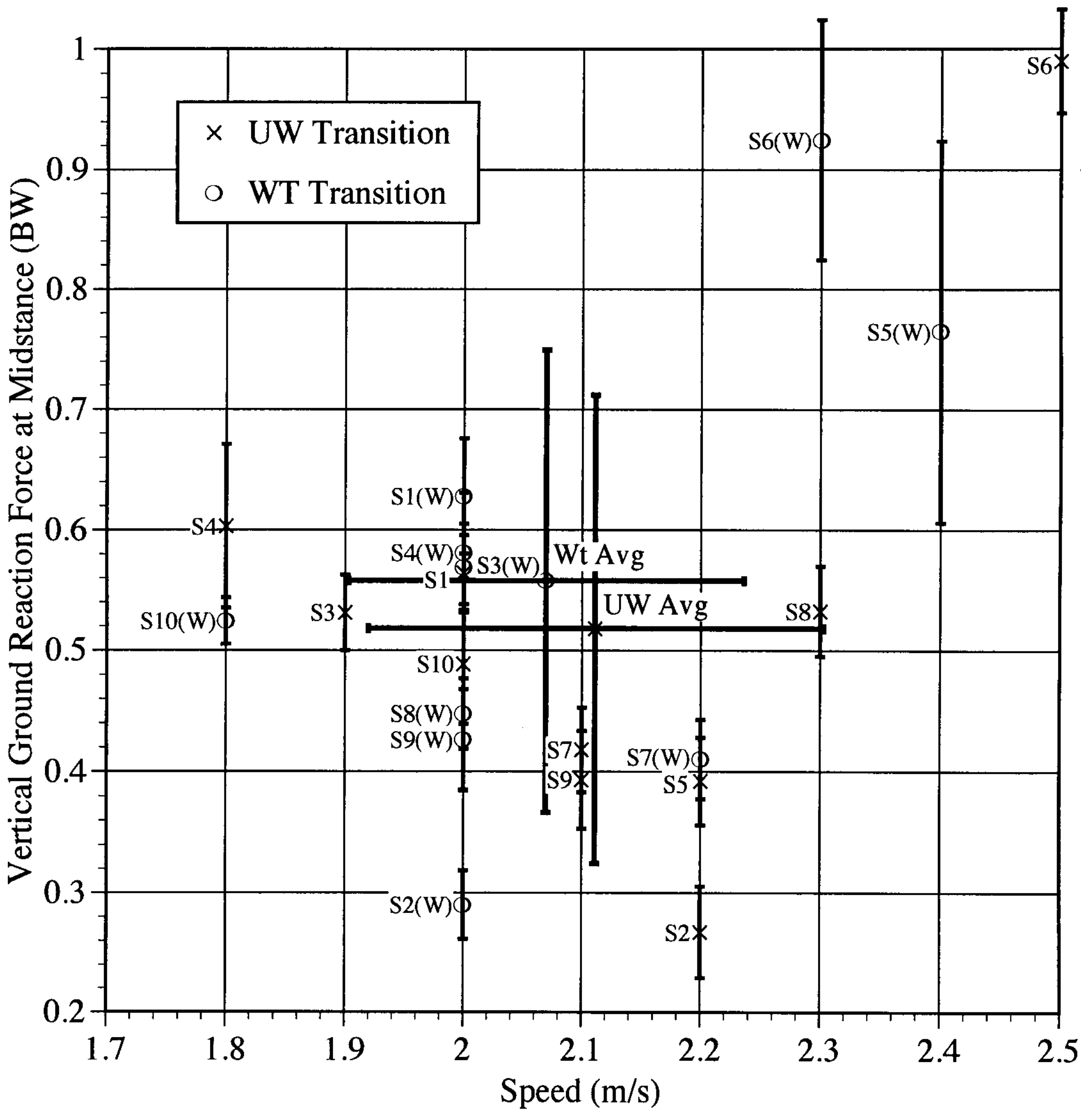

Figure 4.31
Cluster Graphs: Left Leg
Vertical Ground Reaction Force at Midstance
vs. Speed of Locomotion
All Subjects, Weighted & Unweighted Transitions

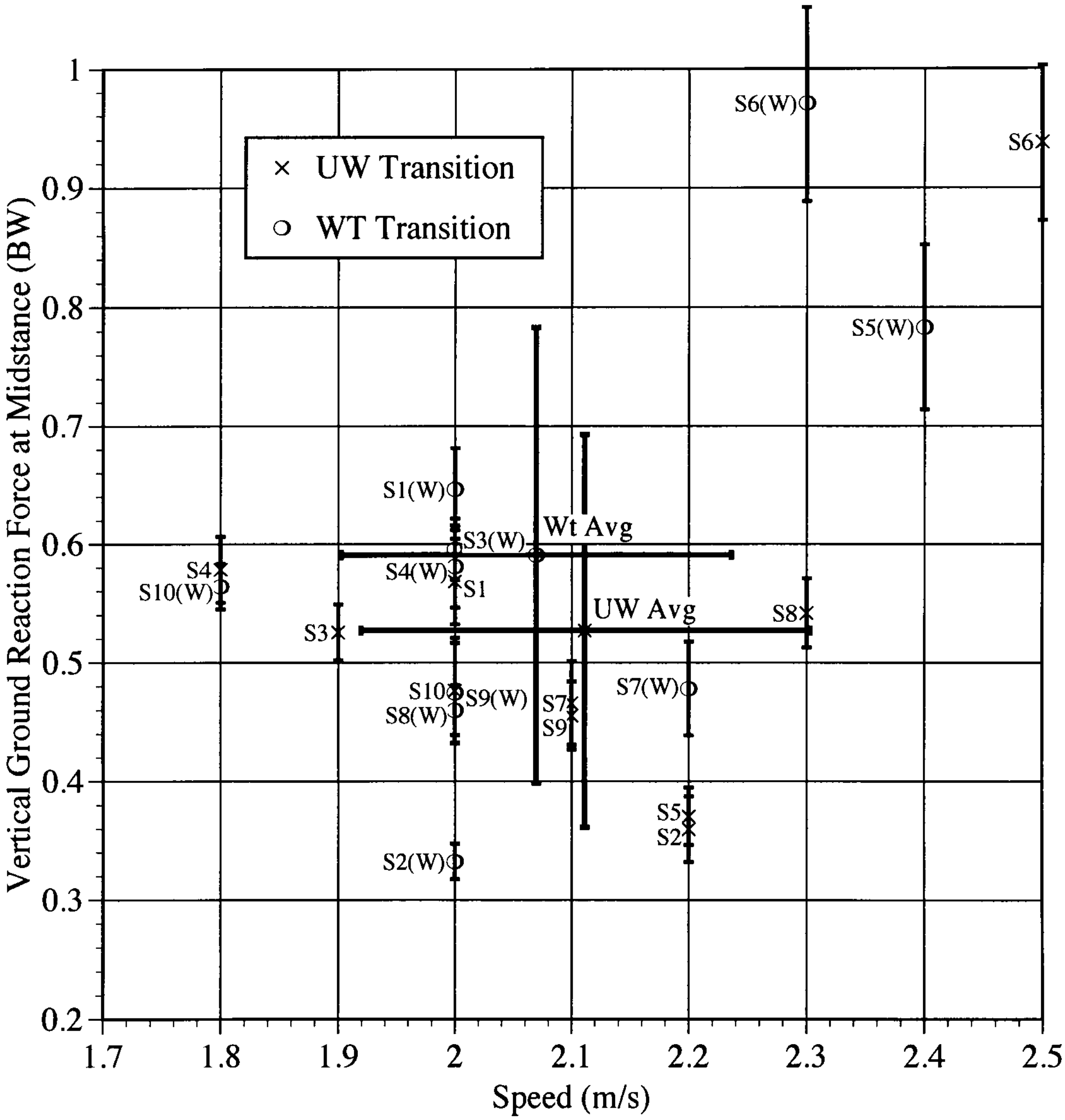

Figure 4.32
Cluster Graphs: Right Leg
Vertical Ground Reaction Force at Midstance
vs. Speed of Locomotion
All Subjects, Weighted & Unweighted

influences of speed and midstance vertical ground reaction force could not be differentiated. If force was a control parameter and speed was not, two clusters would have formed at approximately the same force level, but separated along the speed axis.

4.6 ΔPTS vs. $\Delta VGRF_M$ Graphs

The next step was to calculate the differences between conditions. The change in speed between the weighted and unweighted preferred transition speeds was labeled ΔPTS and was equal to PTS (weighted) - PTS (unweighted). The difference in force, $\Delta VGRF_M$, was quantified by the relationship: $VGRF_M$(weighted) - $VGRF_M$(unweighted) and the standard error calculated from the two conditions. These differences were plotted against each other in Figures 4.33 and 4.34. Confidence intervals were calculated for each variable. For the subjects tested, the 90% confidence interval for the difference between the two preferred transition speeds (ΔPTS) was 0.061 to -0.141 m/s. The 90% confidence interval for change in midstance vertical ground reaction force ($\Delta VGRF_M$) was from 0.112 to -0.035 BW for the left leg and from 0.138 to -0.017 BW for the right leg. Because the VGRFM confidence interval includes zero, the null hypothesis may be rejected and alternative hypothesis number 3 ($PTS_{weighted} - PTS_{unweighted} = 0$; $VGRF_{M\ weighted} - VGRF_{M\ unweighted} = 0$) adopted. However, hypothesis number 1 ($PTS_{weighted} - PTS_{unweighted} \neq 0$; $VGRF_{M\ weighted} - VGRF_{M\ unweighted} = 0$), with the most definitive interpretation for a control parameter, could not be adopted. As a result of both of these confidence intervals including zero, both speed and $VGRF_M$ potentially could be control parameters, but they cannot be distinguished using these data.

An alternative method of assessing the ΔPTS-$\Delta VGRF_M$ relationship was to determine a confidence ellipse. This isobar-type design determined an area of probability, taking into account the two variables (Johnson & Wichern, 1992, pp. 188-91). For this calculation, both variables had to be on the same scale. Thus, the ΔPTS and

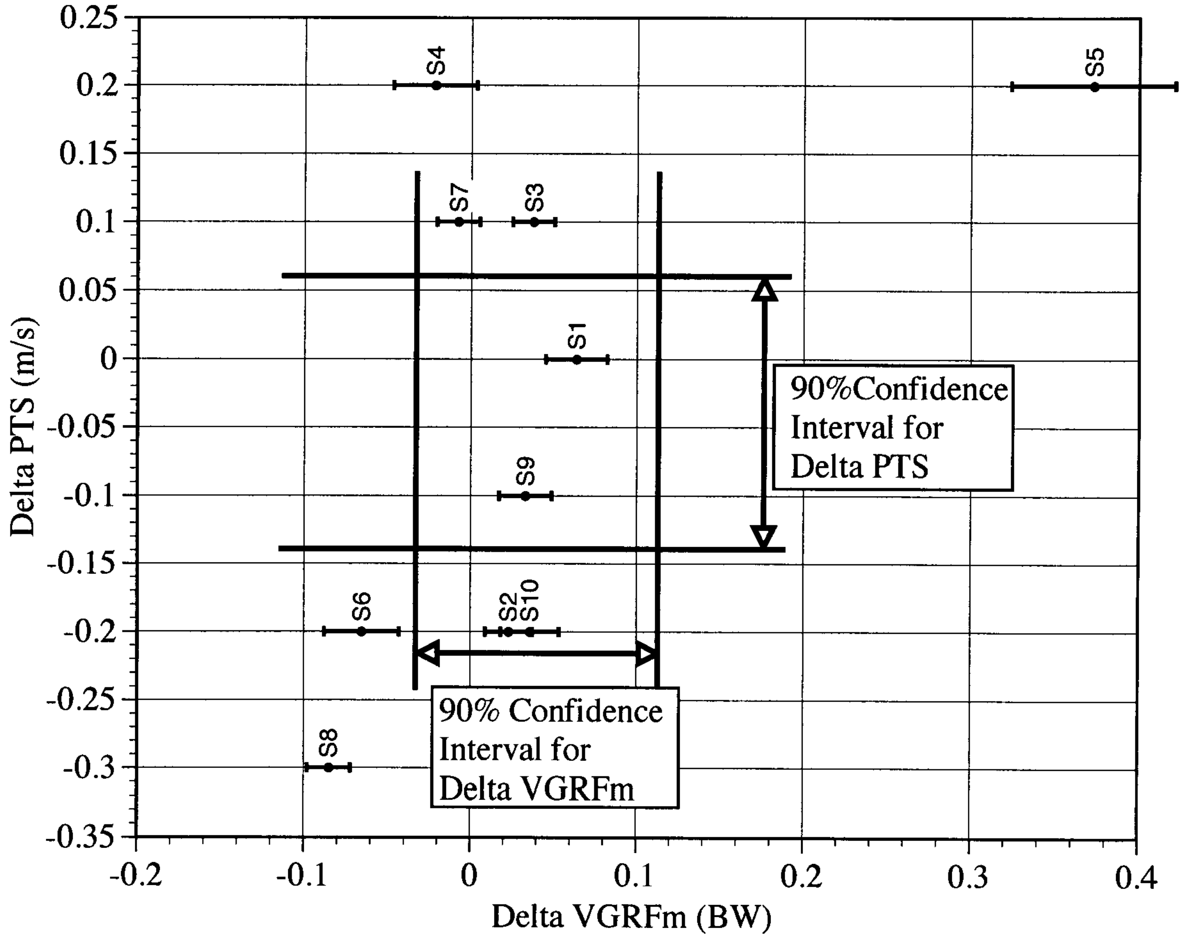

Figure 4.33
Delta PTS vs. Delta $VGRF_M$
Left Leg

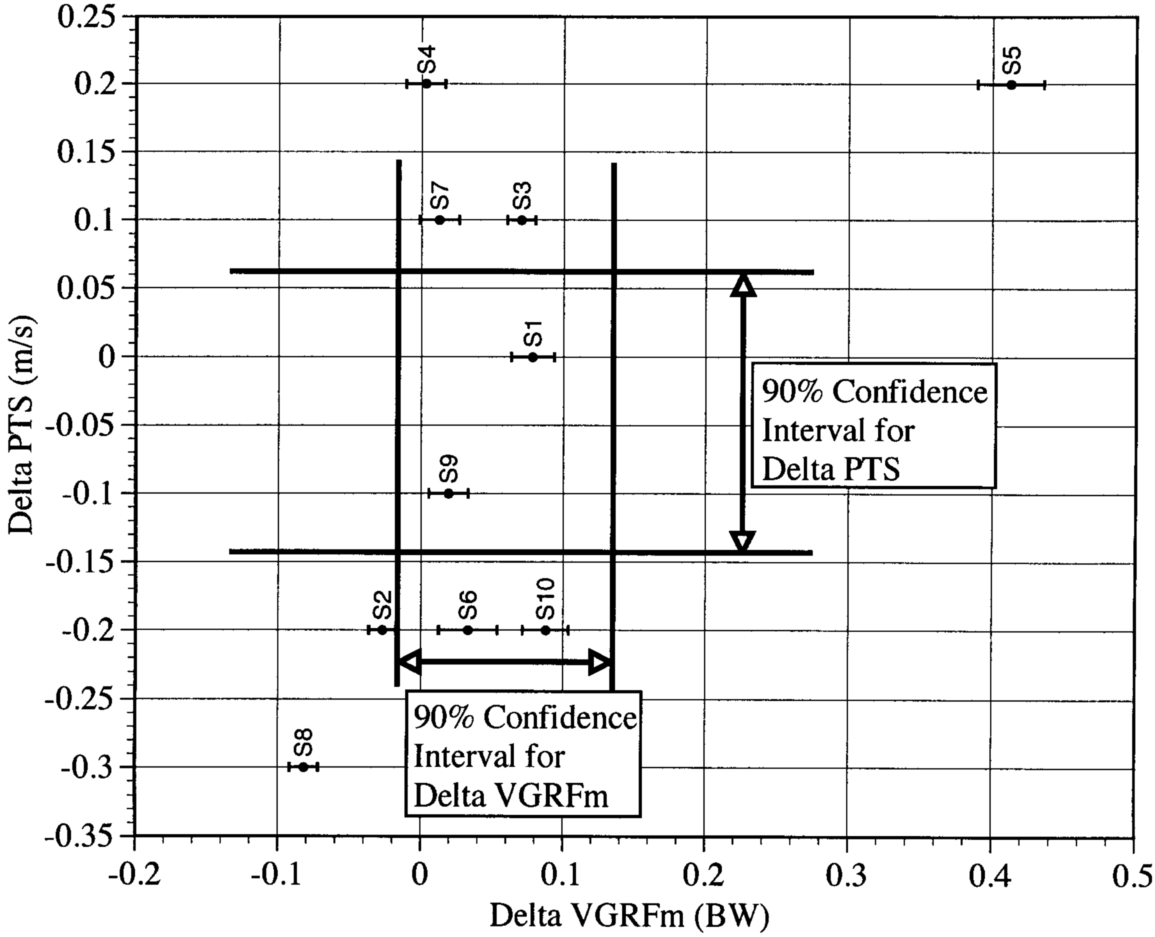

Figure 4.34
Delta PTS vs. Delta $VGRF_M$
Right Leg

$\Delta VGRF_M$ data were converted to *z*-scores and the resulting variance and covariance values were entered into the equation for the 90% confidence ellipse:

$$n\left[\bar{x} - \mu_1, \bar{x}' - \mu_2\right]\left[S\right]^{-1}\begin{bmatrix} \bar{x} - \mu_1 \\ \bar{x}' - \mu_2 \end{bmatrix} \leq \frac{p(n-1)}{(n-p)} F_{p,n-p}(\alpha)$$

In this equation μ_1 and μ_2 were the values to be tested, the *z*-scores of ΔPTS and $\Delta VGRF_M$, $[S]^{-1}$ was the inverse of the 2-by-2 variance/covariance matrix, and p was the number of dimensions, or variables compared in ellipse. This reduced to the equation:

$$10(1.4008)(0 - \mu_1)^2 + 10(1.4008)(0 - \mu_2)^2 - 20(0.7998)(0 - \mu_1)(0 - \mu_2) \leq 6.9975$$

The coordinates for the point to be tested, $\Delta PTS = 0$ m/s and $\Delta VGRF_M = 0$ BW, were converted to *z*-scores and entered into μ_1 and μ_2, respectively. For vertical ground reaction force at midstance to be a possible control parameter, the $\Delta VGRF_M$ value of zero must have been contained within the ellipse. For speed to be discounted as a possible control parameter, $\Delta PTS = 0$ must not be contained within the ellipse. When these two values were tested, they were both found to lie within the ellipse. Therefore, the two possible control parameters could not be differentiated.

Farley and Taylor (1991) identified a control parameter in the trot-gallop transition of horses by differentiating between the roles of speed and musculoskeletal force. This was accomplished by adding weight to the horses and determining the preferred transition speed and force for each condition. Finding that with added weight the horses transitioned at lower speeds ($\Delta PTS < 0$) but at the same force level ($\Delta VGRF_M = 0$), musculoskeletal force was isolated as the control parameter for the trot-gallop transition. The current experiment followed a similar design, but did not produce the same results for the human subjects tested. For these subjects the added weight did not significantly change the preferred transition speed ($\Delta PTS = 0$) but the midstance force also remained at the same level at the PTS for each condition ($\Delta VGRF_M = 0$). Thus, from the data from these subjects, $VGRF_M$ or musculoskeletal force and speed cannot be differentiated. This analysis of musculoskeletal force level and the differences between conditions does not discount musculoskeletal force as a possible control parameter; however, neither can these results promote its role in the human walk-run transition.

4.7 Order Effect

Independent t-tests were performed to evaluate whether or not the order of the weighted conditions significantly affected the measures of change in midstance vertical ground reaction force ($\Delta VGRF_M$) or change in preferred transition speed (ΔPTS). When $\Delta VGRF_M$ was compared between the groups (5 subjects in each), there was no significant order effect for the left or right leg (one-sided *p*-values: $p > 0.15$ and $p > 0.1$, respectively). The change in preferred transition speed also did not exhibit a significant order effect (one-sided *p*-value: $p > 0.15$). If subject 5, having a much larger than normal change in $VGRF_M$, was removed from the analysis, the probability of a significant order effect was decreased ($p > 0.4$, $p > 0.2$ and $p > 0.2$ for the left and right force measurements and the change in preferred transition speed, respectively). The relatively low *p*-values were largely an effect of the low sample size. With 10 subjects in this analysis, a single subject may substantially affect the group means. For the number of subjects used, the counterbalancing of order seems to have minimized order effects.

4.8 Limitations and Assumptions

In order to best correlate vertical ground reaction force with possible musculoskeletal forces, the vertical force at midstance was isolated temporally. The calculation of midstance, or zero anterior-posterior ground reaction force, was based on previous research conducted at speeds close to the subjects' preferred speed of running or walking. It was assumed that zero AP force would remain close to this percentage of stance at the extreme speeds of locomotion which were tested in this study. If, however, zero AP force changed relative location in the stance time at the very high or low speeds tested, midstance forces would be inappropriately selected.

At high speeds of walking, such an error would probably result in an overestimate of the midstance forces. This is due to midstance erroneously being selected out of the well or trough of minimum vertical ground reaction force near midstance. For running, an error could either over- or underestimate the forces, depending on the direction of the temporal error. Midstance is usually at a point at which the vertical ground reaction forces are declining from the active peak. If midstance is selected later than the actual zero AP force, the recorded force will be an underestimate. The opposite is true in the case of a premature selection of midstance.

The accuracy of this assumption was not tested in this experiment. Any errors from this assumption may have influenced the changes in force or variability with increasing speed. The change in force between the weighted and unweighted preferred transition speeds probably would have been affected less, due to the similar transition speeds for most subjects.

4.9 Final Discussion

Many variables have been proposed as possible control parameters governing the human walk-run transition. These include speed of locomotion, metabolic cost, and several kinetic and kinematic variables. A limited number of kinetic variables were tested by Hreljac (1993a). None of the variables tested were found to explain the human gait transition. In contrast, Farley and Taylor (1991) found the trot-gallop transition in horses to be controlled by musculoskeletal forces proportional to the peak vertical ground reaction force. Such musculoskeletal variables had not been subsequently tested in humans.

This experiment was designed to test if musculoskeletal forces act as control parameters in the transition between walking and running. To evaluate this hypothesis, a treadmill was constructed and validated which was able to measure the vertical ground reaction forces of subjects while they walked and ran. With valid and reliable results, the treadmill allowed for speed to be controlled by the experimenter while the forces of multiple footstrikes were measured by a computer. Midstance vertical ground reaction force was picked off from these measurements because at this point anterior-posterior ground reaction forces are approximately zero and the midstance forces may be proportional to musculoskeletal loads, such as in the Achilles tendon. Using this treadmill, speed was manipulated to test the kinetic influence on the walk-run transition. If such an influence could have been discerned by the experiment, it would have appeared as weighted and unweighted transitions occurring at the same midstance VGRF level, but at different speeds. The summarizing tests, such as the cluster graphs, the ΔPTS vs. $\Delta VGRF_M$ graphs, and their associated statistical analyses, failed to distinguish between speed and midstance vertical ground reaction force. From the results presented, both remain as possible control parameters.

Farley and Taylor (1991) added 23% of body weight to horses and observed the trot-gallop transition to occur at a lower speed, but at the same level of musculoskeletal force. They concluded from these findings that a critical level of musculoskeletal force was the control parameter governing the trot-gallop transition. In a study on humans, Beuter and Lefebvre (1988) added weights to trained runners and found the preferred walk-run transition speed to decrease only slightly. The present study hypothesized that with added weight untrained runners would exhibit a change in the preferred walk-run transition speed. To test this hypothesis, 15% of body weight was added to untrained runners and the effects on the walk-run transition observed. In accordance with the findings of Farley and Taylor, the transitions for the two conditions occurred at the same force level; however, no significant changes were found in transition speed. While the addition of weight to horses substantially lowered their trot-gallop transition speed, the addition of 15% of body weight to the untrained human runners of this study failed to produce significant changes. With no significant change in force or speed of transition, the two factors could not be differentiated.

However, the mapping of forces across speeds and conditions indicated a possible influence of force on the transition. With increased speed, significant increases were found in the midstance VGRF's for running, but no significant changes were found for the midstance forces in walking. While the variation in the choice of gait could not be quantified, the variation in midstance VGRF for walking demonstrated a substantial, although statistically significant in only one case, increase at and above the preferred transition speed. As described by Thelen and Smith (1994), such a change in variability may be indicative of a perturbation capable of producing a bifurcation, or switch in the collective variable of gait from walking to running. This understanding suggests that vertical ground reaction force may have some influence on the human walk-run transition.

Many variables have been hypothesized to control the human walk-run transition. This experiment focused on the kinetic variable of midstance vertical ground reaction force, indicative of musculoskeletal forces in the lower leg. While the results demonstrate that $VGRF_M$ may be a control parameter in the transition, the findings are insufficient to factor out the effects of speed of locomotion. Further experiments with greater experimental control and more subjects may be able to discern such differences, if they exist.

CHAPTER 5
CONCLUSIONS AND RECOMMENDATIONS FOR FUTURE WORK

5.1 Summary

With increasing speed, a person's choice of a walking gait will eventually become uncomfortable, prompting a switch to running. The speed at which this switch occurs is called the preferred transition speed. While it is associated with increasing speed, earlier work shows that the transition is not related just to the speed. This study was designed to test the influence of forces on the human body at midstance on the walk-run transition.

Ten college-aged males and females walked and ran on a treadmill with and without a weight vest approximating 15% of body weight. These subjects performed three tests: to determine the speed at which they preferred to transition from walking to running, and to measure the vertical loads experienced by the body while walking and running over a range of speeds (1.0 to 3.0 m/s). It was hypothesized that vertical force would have an influence or control over the transition between walking and running while speed, commonly thought to influence the transition, would not. If true, the subjects would transition at different speeds, but at approximately the same force for the weighted and unweighted conditions. When measured, these subjects tended to transition at the same force levels for both conditions. However, on average, these subjects also switched gaits at approximately the same speed for the two conditions. Thus for the population tested, this experiment was not able to distinguish between the roles of speed and musculoskeletal force on the walk-run transition.

When each subject walked or ran under each condition and speed, many strides were analyzed. No two strides were exactly the same, even within a single condition and speed. For each speed and condition, the average force and the variability of these measures were determined. The results demonstrated that both the force and its variability increased when the subjects attempted to walk at speeds greater than their preferred transition speed. These changes may be influencing the control system of human gait. When an individual experiences such increases, he or she may reach a force level which is not preferred, or the increased variability in force may cause a perturbation, both of which may result in the subject switching from walking to running. Thus the results of this study indicate that the musculoskeletal forces associated with midstance vertical ground reaction force may have an influence on the human walk-run transition.

5.2 Recommendations for Future Work

5.2.1 Walk-Run Transition

To better distinguish the preferred transition speed, smaller increments of speed could be used (0.1 m/s). This still would not get around the fact that with added weight some subjects transitioned at slower speeds, some at higher speeds and some the same. By using smaller increments of speed, the resolution for this variable would be increased.

For even greater resolution the forces could be measured while speed was changing at a slow rate. It is still desired to be able to average force for multiple strides at a given speed. Without this ability it would not be possible to quantify the variability of force for a given speed. Such continuous data collection would require improvements to the hardware and software of the treadmill system. The rate of change of speed could be analyzed and taken into account during the analysis. Perhaps the protocol could be administered with computer control of treadmill speed. At the least, a continuous measurement of treadmill speed should be input along with the force data. It should be noted that the stride-to-stride variability of treadmill speed would be noticed in such a measurement.

Following the indications that a perturbation in midstance vertical ground reaction force has a role in the walk-run transition, future work could attempt to create or increase this perturbation. If force could be spontaneously increased, by adding weight to a subject while he or she walks, or by some other method, it may be possible to create enough variability in the midstance VGRF's to trigger a transition from walking to running. This would strengthen the argument of midstance VGRF being a control parameter in the walk-run transition.

The variability of the collective variable of gait should be measured or monitored at the same time as possible control parameters are both measured and manipulated. It is desired to manipulate a single possible control parameter and observe subsequent variability in the collective variable prior to a bifurcation or transition. Following the bifurcation the variability of the collective variable should decrease.

This experiment did not measure force while the subjects were free to transition between walking and running. For greater description and to gain further insight into the force changes during the transition itself, the forces could be collected when the subject is free to transition. Such a protocol would require continuous, or nearly continuous, data collection.

During this study it was observed that the rate of change of speed may have an influence on the preferred transition speed. If a subject is walking on a treadmill near, but below their preferred transition speed and speed is increased a given increment, a fast rate of increase may be expected to cause a transition to running to occur at a lower speed than if a slow rate of increase was used. Perhaps the subject would feel "thrown" into the next attractor state or he or she would anticipate the quickly changing speed and transition at a speed lower than that otherwise chosen. Further investigations should study the influence of rate of change of speed on the walk-run transition.

5.2.2 Other Research

The instrumented treadmill developed for this study may be utilized for experiments which would be impossible with a force plate. Providing a great amount of experimenter control and simultaneous measurement of left and right vertical ground reactions forces, the treadmill makes more involved experiments possible. A simple experiment would be to compare left and right sides with any subject population. This may be of particular interest with injuries, such as runners who are having pain in one leg. It would be interesting to correlate changes to ground reaction forces with such injuries. During pilot tests, two runners who were having chronic leg pain recorded increased vertical ground reaction forces for the effected side. Such an evaluation tool could be used clinically or to biomechanically analyze the causes or results of injuries. Perhaps a biofeedback technique could be used to minimize injuries.

A biofeedback mechanism would open more research opportunities, such as balance training. Such abilities could be utilized in the line of balance research presently conducted by the Motor Behavior Laboratory of Oregon State University, under the guidance of Debra Rose, Ph.D. Variables such as contact time, maximum vertical ground reaction force, and perhaps other variables of interest could be displayed for the subject to see. If the treadmill or the possible biofeedback device proved beneficial, they could be a part of a dynamic, or walking, balance training program.

5.3 Conclusions

For this study a treadmill was developed to measure vertical ground reaction forces while subjects walked and ran. Once validated, this tool was used to measure changes in midstance vertical ground reaction force with increased speed and load. By manipulating these two variables it was hoped to differentiate between their roles in the walk-run transition. From the results, however, it was impossible to distinguish between the influences on the transition; both speed and musculoskeletal force remain as possible control parameters. After mapping the forces and the variability of force for each speed and condition, there were indications that midstance vertical ground reaction force may have had a contribution to perturbing the system from a walking to running gait.

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APPENDICES

APPENDIX A

Musculoskeletal Forces in Human Locomotion
Jonathan B. Fewster and Gerald A. Smith

Potential Subject: Medical Questionnaire

Please fill in the following information as completely as possible.

Name:

Age:

Sex: M F

Present exercise program: Please list activities, relative intensity and weekly hours of participation. Be sure to mention any amount of running in terms of weekly mileage.

Activity	Time	Mileage (Walking or Running)

Injuries or medical conditions: Please list any injuries occurring within the last year which may affect walking or running. Be sure to include any achilles tendon injuries incurred at any time. List any heart or other health conditions which may be cause to avoid light to moderate exercise.

Chronic Diseases or Disabilities: Please list any neuromuscular conditions which may affect your movement or locomotion. These may include, but are not limited to, spinal injuries, MS and CP.

APPENDIX B

Oregon State University
Human Subjects Review:

Musculoskeletal Forces in Human Locomotion
Jonathan B. Fewster and Gerald A. Smith

Significance:
The understanding of human movement control is presently changing. The differences between the old and new schools of thought may be manifest and thus tested at transition points in human movement. In the study of gait transitions in human locomotion many influencing factors have been identified. However, a new school of thought in motor behavior, called Dynamical Systems, would hold that a single factor "pushes" the transition over the edge from one gait to another.

For the walk-run transition it is our hypothesis that tendon forces are the deciding variable for determining which gait is used at a given velocity. If this hypothesis is supported, it will provide strong support for a new understanding of motor control.

Methods:
Each subject will be asked to perform several trials either walking or running while treadmill speed is systematically varied. With force measurement devices measuring the vertical ground reaction forces and a video camera recording body position, the treadmill will be accelerated from 1 m/s to 3 m/s in 0.2 m/s increments. The subject will have 30 seconds at each speed before the velocity is changed. Two acceleration cycles will be performed with the subject choosing to walk or run at each speed. In separate trials the subject will be asked to walk through the whole range of speeds, and then run through the whole range. These four trials will be repeated with each subject wearing a pre-determined amount of weight (approximating 15% of body weight) in a vest on his or her upper body. The total activity time for this procedure will be between 90 minutes and two hours. In subsequent analysis, motion characteristics (position, length of lever arms) of the lower limbs will be determined from video analysis at each speed. The combination of video and force data will be used to calculate tendon forces at mid-stance, impact forces and loading rate.

On a day prior to the experiment each subject will be required to walk and run on the treadmill at experimental speeds under the both the weighted and unweighted conditions. The total time for this adjustment period will be thirty minutes, unless more time is desired by the subject.

Subjects:
Ten subjects (male and female) will be recruited for this project. Subjects will be from the general university population and will range between the ages of 18 and 25. Subject requirements will include a fitness level capable of completing the moderate exercise involved in the study. More specifically, subjects will be recreational runners who run at least one hour but no more than four hours per week, and do not have a physical disability or have a chronic illness (i.e. MS, CP, or spinal injuries).

Risks and Benefits:
Participation in this study carries minimal risk. Each subject will be asked to walk or run at slow to moderate paces on a treadmill for trials of short duration. They will be given opportunity to familiarize themselves to walking and running on the treadmill before the data collection. The subjects will be given the opportunity to rest following any of the

trials. The relatively low intensity of the exercise involved makes the probability of a serious, health affecting event very low.

Each subject will be paid $10 for his or her participation and completion of the study.

Informed Consent:
The subjects in this study will be recruited from the large population of healthy individuals at the Oregon State University. Prior to testing, each subject will be orally informed of the purposes of the research, the protocol to be followed during the testing and the approximate time involved. In addition, each subject will be asked to read and sign the Informed Consent Form (attached) which will reinforce the nature of the research, describe the minimal risks associated with the study and emphasize the subject's freedom to end participation in the study at any time.

Confidentiality:
Any information obtained from me will be kept confidential. A code number will be used to identify any test results or other information that I provide. The only persons who will have access to this information will be the investigators and no names will be used in any data summaries or publications.

Informed Consent Form

Oregon State University
Corvallis, Oregon

Title of Investigation: Musculoskeletal Forces in Human Locomotion

Investigators: Jonathan B. Fewster, M.S. Candidate and Gerald A. Smith, Ph.D.

Purpose: To determine the type of musculoskeletal forces that influence walking and running at different speeds.

I have received an oral explanation of the study procedures and understand that they entail the following:

All testing will be conducted in the Sports Medicine and Anthropometry Laboratories in the Women's Building at Oregon State University. As a subject, I will report to the laboratories one time for the following procedures:

1. Body measurements. My height and weight will be measured in the Anthropometry Laboratory.

2. Biomechanical data collection. My lower legs will be videotaped from the side view and the impact forces on my feet will be measured while I am walking and jogging on a flat motorized treadmill at a speed between 1 m/s and 3 m/s (2.2 mph - 6.7 mph). During each trial the treadmill will be slowly accelerated, requiring five minutes to increase from 1 to 3 m/s. For two trials I will choose to either walk or run at each speed. In two subsequent trials I will walk and then run over the whole range of speeds. These four trials will be performed normally and four more trials will be performed while carrying additional weight approximately equal to 15% of my body weight.

In addition to these experimental procedures I will report to the laboratories on one previous day to practice walking and running on the treadmill as outlined above. For thirty minutes I will walk and run at the experimental speeds while under both the weighted and unweighted conditions.

I understand that my risks associated with participation are minimal. Running may be associated with muscle soreness; however, at the moderate speed of walking and running used in this study, such effects should be mild. Coronary complications such as chest pain, irregular heart beats or even death have occasionally been associated with vigorous exercise. However, based on my relatively good fitness and the relatively low exercise intensity, it is unlikely that such problems will be encountered.

I understand the University does not provide a research subject with compensation or medical treatment in the event that the subject is injured as a result of participation in the research project.

Any information obtained from me will be kept confidential. A code number will be used to identify any test results or other information that I provide. The only persons who will have access to this information will be the investigators and no names will be used in any data summaries or publications.

The benefits of my participation in this study include contributing to the scientific study of exercise science and human locomotion. I will also receive a $10 award in return for my completion of this experiment.

I understand that my participation in the project will entail one practice session and one laboratory session requiring a total of approximately 2 hours.

I have been completely informed about and understand the nature and purpose of this research. The researchers have offered to answer any further questions that I may have. I understand that my participation in this study is completely voluntary and that I may withdraw from the study at any time without prejudice or loss of benefits to which my participation entitles me.

If any questions arise during my participation in this research project, I am to call Jon Fewster at (503) 737-5933 or Gerald Smith at (503) 737-5928.

Any other questions that I have should be directed to Mary Nunn, Sponsored Programs Officer, OSU Research Office, (503) 737-0670.

My signature below indicates that I have read and that I understand the procedures described above and give my informed and voluntary consent to participate in this study. I understand that I will receive a signed copy of this consent form.

____________________________ ________________

Subject's Signature Date

____________________________ ________________

Subject's Name (Printed) Date

____

Subject's Address

____

____________________________ ________________

Investigator's Signature Date

OREGON STATE UNIVERSITY

Committee for the Protection of Human Subjects

Chair's Summary of Review

Title: Musculoskeletal forces in the walk-run transition of human locomotion

Program Director: Gerald Smith

Recommendation:

_X_ Approval* ____ Provisional Approval ____ Disapproval ____ No Action

*The informed consent forms obtained from each subject need to be retained for the long term. Archives Division of the OSU Department of Budgets and Personnel Service is willing to receive and archive these on Microfilm. At present at least, this can be done without charge to the research project. Please have the forms retained in archives as well as in your files.

Remarks:

All concerns of the IRB have been suitably addressed and necessary changes made.

Redacted for privacy

Date: 21 April 1994 Signature:

If the recommendation of the committee is for provisional approval or disapproval, the program director should resubmit the application with the necessary corrections within one month.

INSTITUTIONAL REVIEW BOARD FOR THE PROTECTION OF HUMAN SUBJECTS

OREGON STATE UNIVERSITY

Report of Review

TITLE: Musculoskeletal forces in human locomotion

PRINCIPAL INVESTIGATOR: Gerald A. Smith, ExSS

STUDENT: Jonathan B. Fewster

COMMITTEE DECISION: Approved

COMMENTS:

1. The informed consent form obtained from each subject should be retained in program/project's files for three years beyond the end date of the project.

2. Any proposed change to the protocol or informed consent form that is not included in the approved application must be submitted to the IRB for review and must be approved by the committee before it can be implemented.

Redacted for privacy

Date: August 3, 1995

Warren N. Suzuki, Chair
Committee for the Protection of Human Subjects
(Education, 7-6393, suzukiw@ccmail.orst.edu)

APPENDIX C
TREADMILL VALIDATION

C.1 Static Validation

The first stage of validation for the treadmill used static loads. Known weights were stacked at various locations on the treadmill belt. For a given position one measurement was taken while stacking the weights, another while removing them. After the treadmill was moved and set up again, the whole procedure was repeated. In the second trial the exact positions were not repeated, although the same range of locations were used.

After the data were converted and filtered at 60 Hz, the summed VGRF's were compared against the actual weight applied. The values for each weight are reported in Table C.1. The total force values appear in Table C.2. The total forces were plotted for the two trials and appear in Figure C.1. A least-squares-fit was performed for each of the two trials. For trial one, TM = 0.9609 * Wt + 1.4412 N, R^2 = 0.9961, R = 0.9980, and the Average Error was 40.32 N. For trial two, TM = 0.9715 * Wt + 8.7461 N, R^2 = 0.9966, R = 0.9983, and the Average Error was 26.56 N (TM = Treadmill measured force value, Wt = Known force applied, both in Newtons). This test satisfactorily evaluated the treadmill system's ability to accurately and precisely measure applied forces.

C.2 Treadmill vs. Force Plate

The second pilot test validated the treadmill's ability to measure VGRF's against the criterion measure of a floor-mounted force plate. Five subjects ran over a Kistler force plate at three self-selected speeds (slow, medium, and fast). For each trial the speed was measured by head-height infrared timing lights centered about the force plate. Each self-selected speed was repeated until three trials were performed within ± 0.02 kph

Table C.1
Static Load Errors

	Trial 1				Trial 2			
Known Force (N)	*n*	Mean Force (N)	SD (±N)	Average Error (N)	*n*	Mean Force (N)	SD (±N)	Average Error (N)
219.96	21	211.55	18.03	19.98	23	221.47	21.32	21.37
219.04	19	213.37	6.12	8.44	24	215.44	6.30	7.29
201.83	19	195.72	6.02	8.69	24	195.20	5.41	8.67
199.24	19	191.03	8.52	11.99	23	192.44	8.90	11.29
135.29	5	129.48	8.50	10.70	6	134.50	6.00	6.06
136.40	5	134.66	5.86	6.17	6	132.26	6.72	8.11
136.67	4	129.14	3.91	9.54	6	134.22	3.96	4.79
133.77	3	127.09	7.07	10.81	6	126.16	6.57	10.62

Table C.2
Total Static Load Errors

	Trial 1				Trial 2			
Known Force (N)	*n*	Mean Force (N)	SD (±N)	Average Error (N)	*n*	Mean Force (N)	SD (±N)	Average Error (N)
219.96	21	211.55	18.03	19.98	23	221.47	21.32	21.37
439.00	19	213.37	6.12	8.44	24	436.65	20.69	20.37
640.83	19	195.72	6.02	8.69	24	631.39	20.93	22.54
840.07	19	191.03	8.52	11.99	23	823.87	21.98	26.90
975.36	5	129.48	8.50	10.70	6	957.01	11.41	23.12
1111.76	4	134.66	5.86	6.17	6	1089.27	13.77	28.23
1248.43	3	129.14	3.91	9.54	6	1223.49	14.02	30.71
1382.20	2	127.09	7.07	10.81	6	1349.64	16.28	39.21

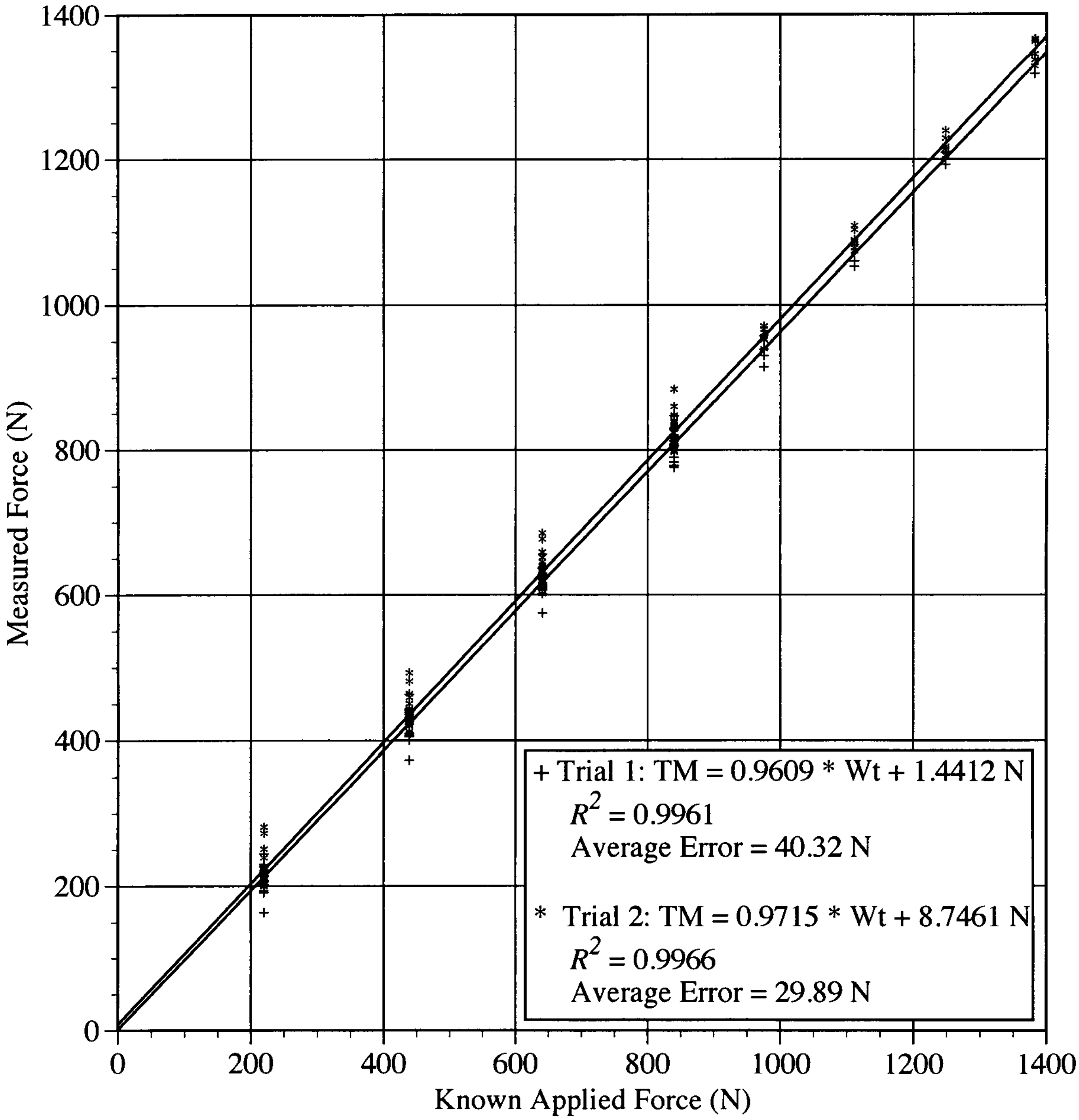

Figure C.1
Static Loading: Comparison of Trials
Measured vs. Known Forces

(± 0.0056 m/s). Within a given speed, each subject was consistent with the choice of foot to land on the force plate. The actual velocities were different between subjects.

With force plate data collection complete, the treadmill recorded VGRF data at the chosen speeds for each subject. The treadmill speed was adjusted to each individual's slow, medium, and fast speed. The collection program recorded ten seconds of data at each speed.

The right or left footstrikes (chosen during the force plate trials) from the treadmill were analyzed along with the corresponding force plate data. These data were analyzed for the characteristics of peak impact force, active peak force, loading rate, contact time, and total impulse. See Figure C.2. The same number of footstrikes were analyzed for each device. In the case of the treadmill, the footstrikes were chosen at random from those recorded. From a given subject's trials at a given speed using a single device, the mean value and standard deviation were calculated for each characteristic. These treadmill vs. force plate values were plotted to generate a correlation factor, a slope, and an average error.

One of the five subjects exhibited a change in footstrike pattern between the force plate and the treadmill conditions. She seemed to be a midfoot striker on the force plate, but was a rearfoot striker on the treadmill. This changed the impact peak and rise rate. For these two correlations the subject was removed. The observed difference in striking demonstrated the possibility for difference between force plate and treadmill running or as described in the literature "overground vs. treadmill running." While a subject may attain constant speed over the force plate, he or she may not get into a pace during the short run-up. In contrast, while on the treadmill, the subject must maintain the measured speed for a minimum of thirty seconds. This may influence the subject getting into a pace or a gait which may be different from the striding which was observed over the force plate. This should influence force plate procedures and in the future may be developed into a exploration of kinetic differences between force plate and treadmill running.

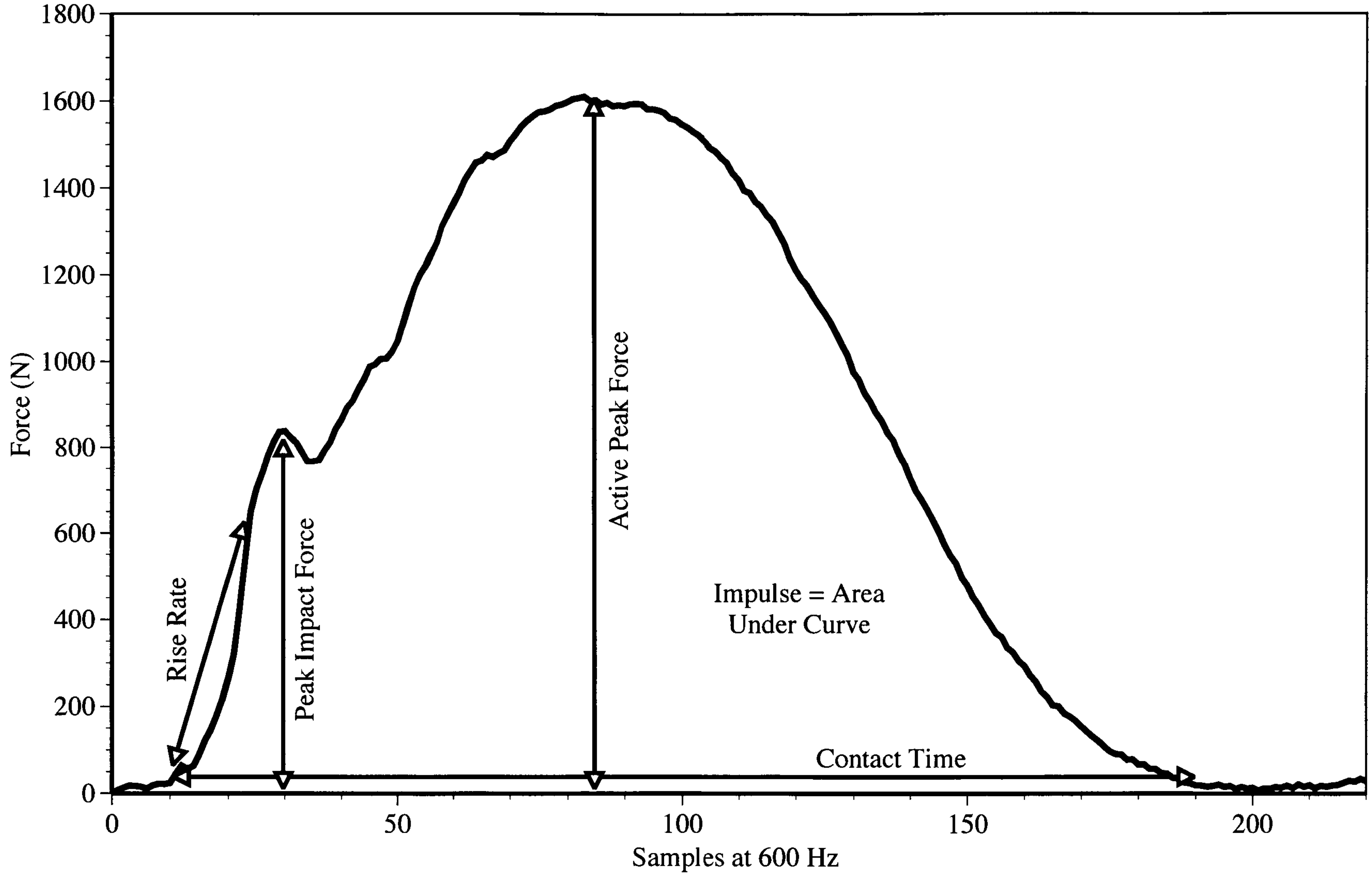

Figure C.2. Vertical Ground Reaction Force Characteristics

The correlation graphs may be seen in Figures C.3 - C.7 (TM = treadmill, FP = force plate). Impact force yielded a fit of TM = 1.1915 * FP - 301.4 N with $R^2 = 0.8319$, $R = 0.9121$ and Average Error = 58.8 N. The active peak forces are expected to be less affected by changes in striking patterns. The active peaks were fit with the line: TM = 0.7993 * FP + 235.9 N, $R^2 = 0.6932$, $R = 0.8326$ and Average Error = 95.3 N. Rise rates calculated from the points of 20% and 80% of peak impact force were correlated with the equation TM = 0.9092 * FP - 4110.1 N/sec, $R^2 = 0.5267$, $R = 0.7257$ and Average Error = 13409.6 N/sec. This correlation demonstrated the responsiveness of the treadmill, but was limited by any differences in gait caused by the two devices. Contact time was measured to check for time of contact differences. These proved to be very similar: TM = 1.0673 * FP - 0.0045 sec, $R^2 = 0.8158$, $R = 0.9032$ and Average Error = 0.0092 seconds. The last measure was total impulse. This proved to be a very good correlation with TM = 0.9479 * FP + 7.7597, $R^2 = 0.9678$ N*sec, $R = 0.9838$ and Average Error = 2.07 N*sec.

These results demonstrated some deviation from the criterion measure of the force plate. This deviation may have been due to the accuracy of the treadmill, subject trial-to-trial variability, and differences in running between the force plate and the treadmill. The results presented above demonstrate errors much more reasonable than the large errors in peak forces experienced by E. Hennig, with these results being in the correct range. Some of the remaining differences in measurements should be attributed to the differences between laboratory running and true overground running. There are known kinematic differences between treadmill and force plate running (e.g. Nigg, De Boer, & Fisher, 1995). A trial of a subject running over a force plate in a laboratory is not a true measure of overground running when compared to that normally performed outside for long distances and times. Therefore, it was hypothesized that running on this treadmill, and the measurements taken with it, were actually more similar to true overground

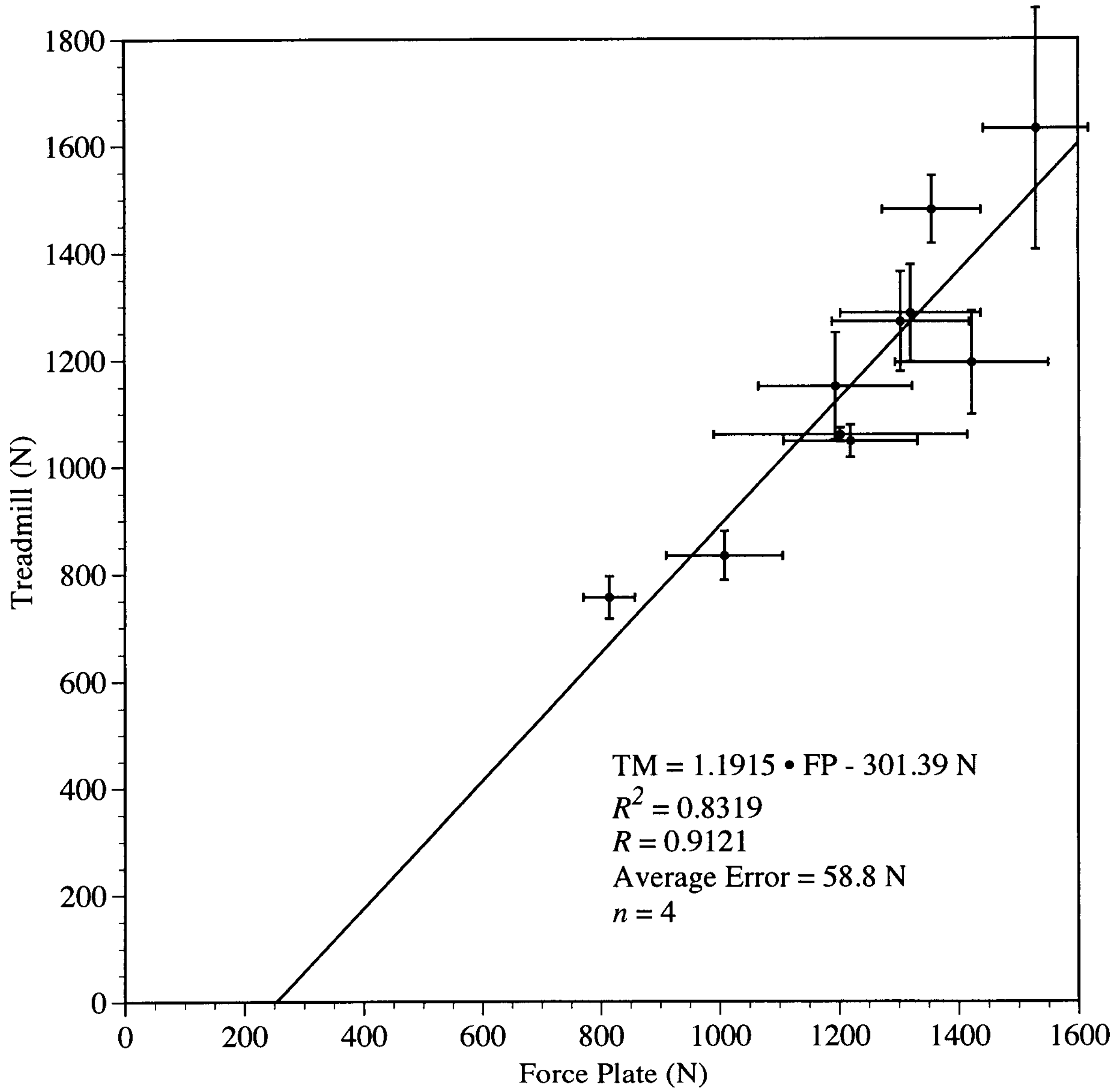

Figure C.3
Treadmill vs. Force Plate:
Peak Impact Force Correlation

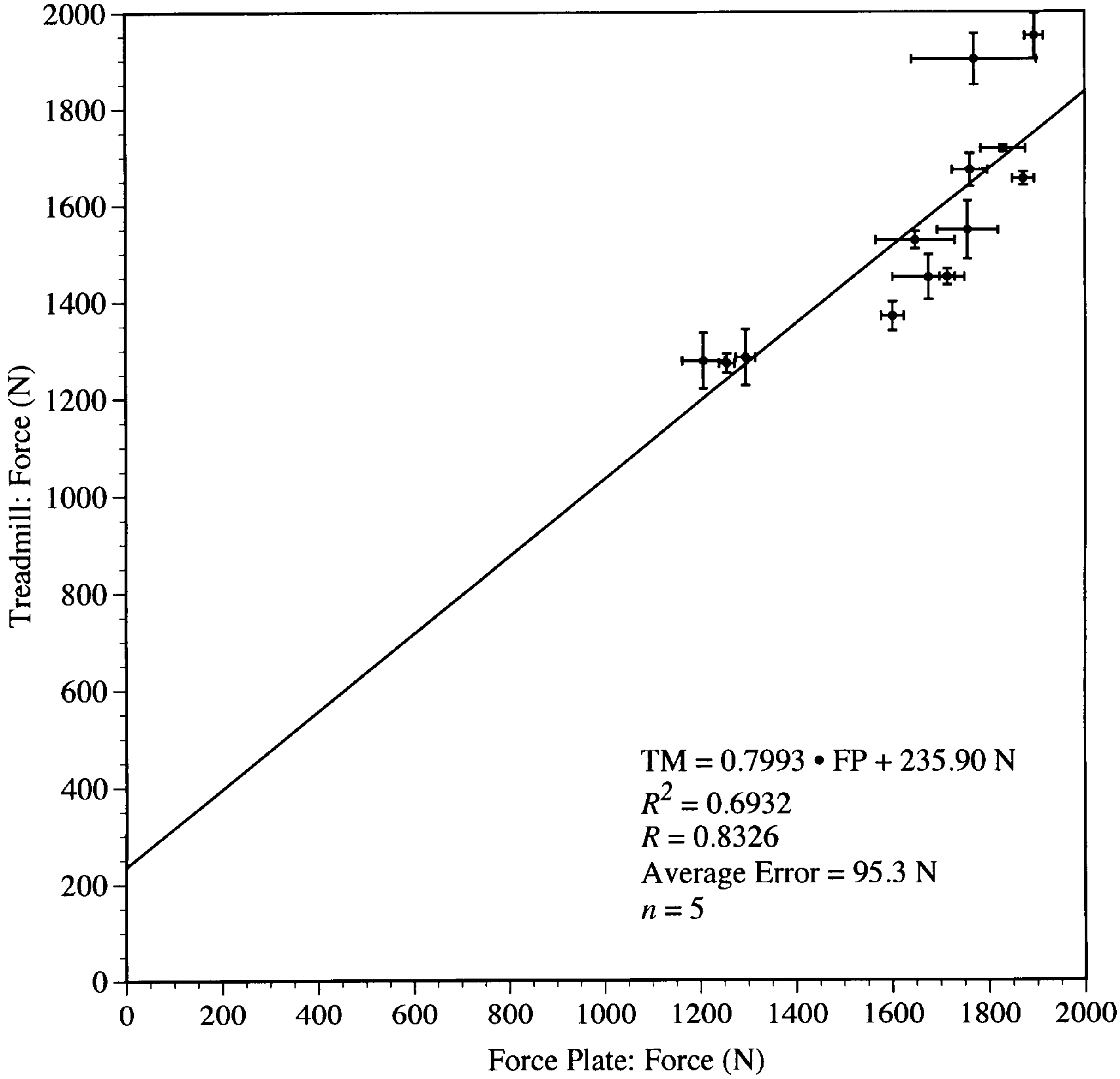

Figure C.4
Treadmill vs. Force Plate:
Active Peak Force Correlation

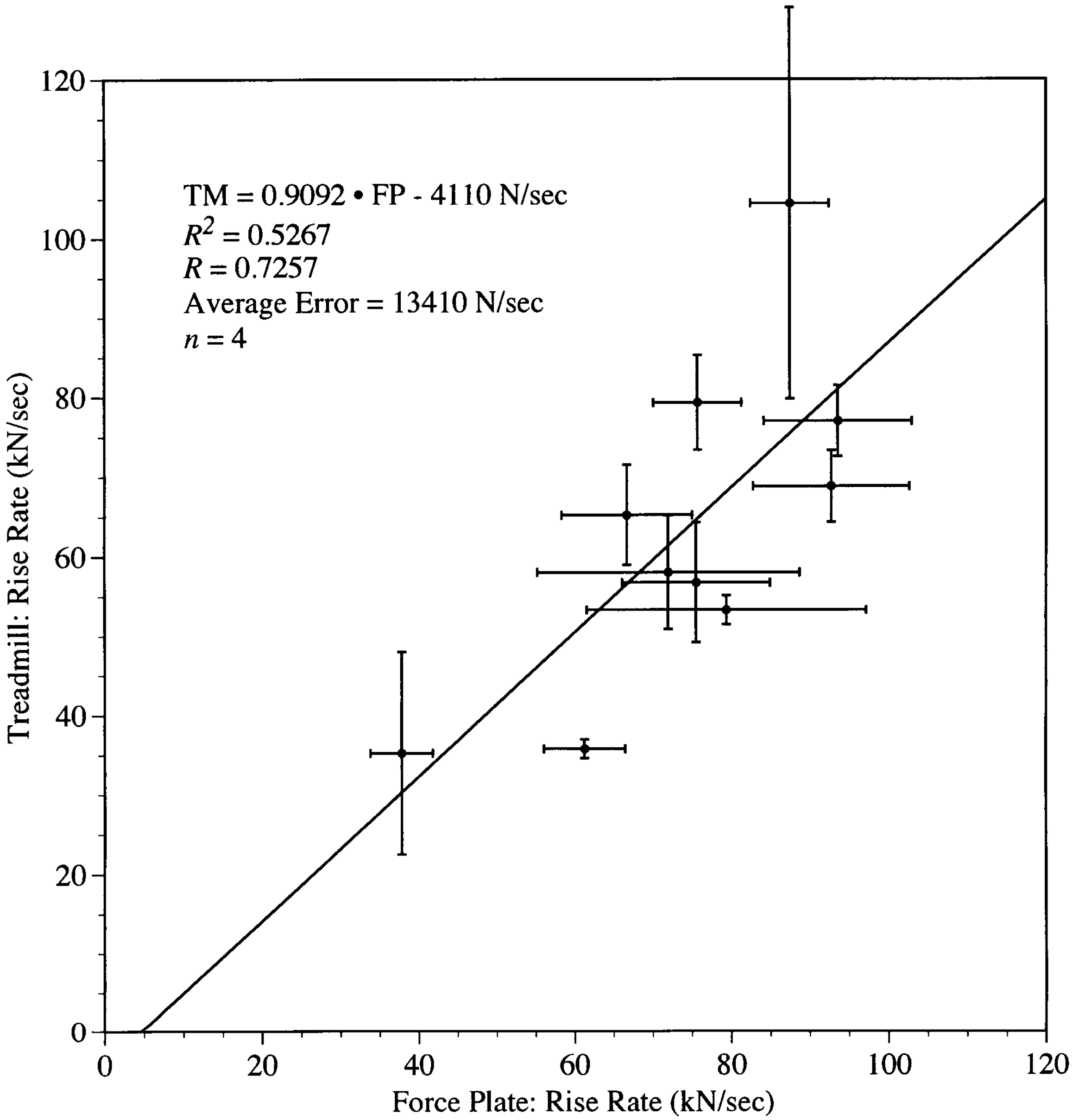

Figure C.5
Treadmill vs. Force Plate:
Rise Rate Correlation

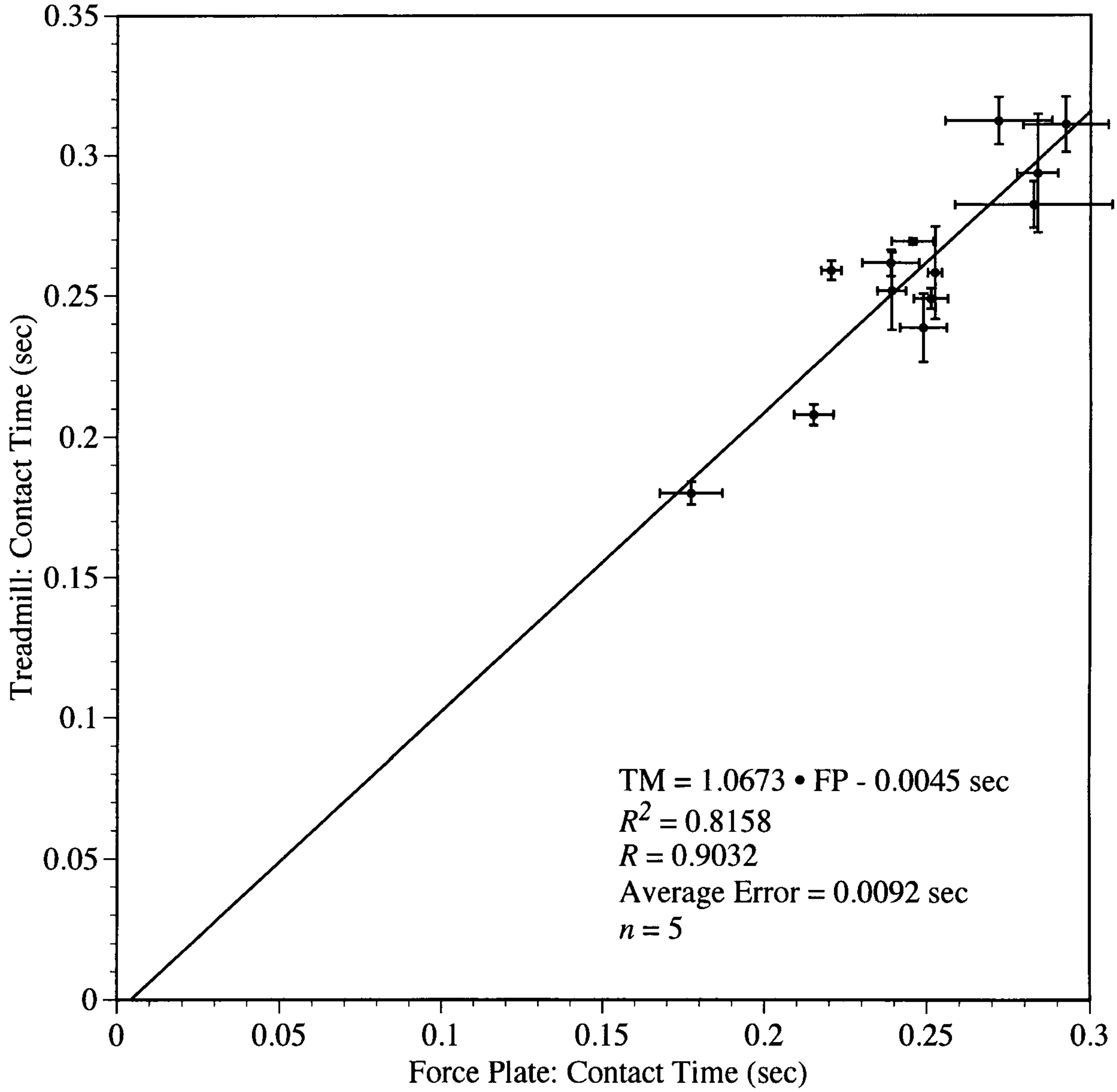

Figure C.6
Treadmill vs. Force Plate:
Contact Time Correlation

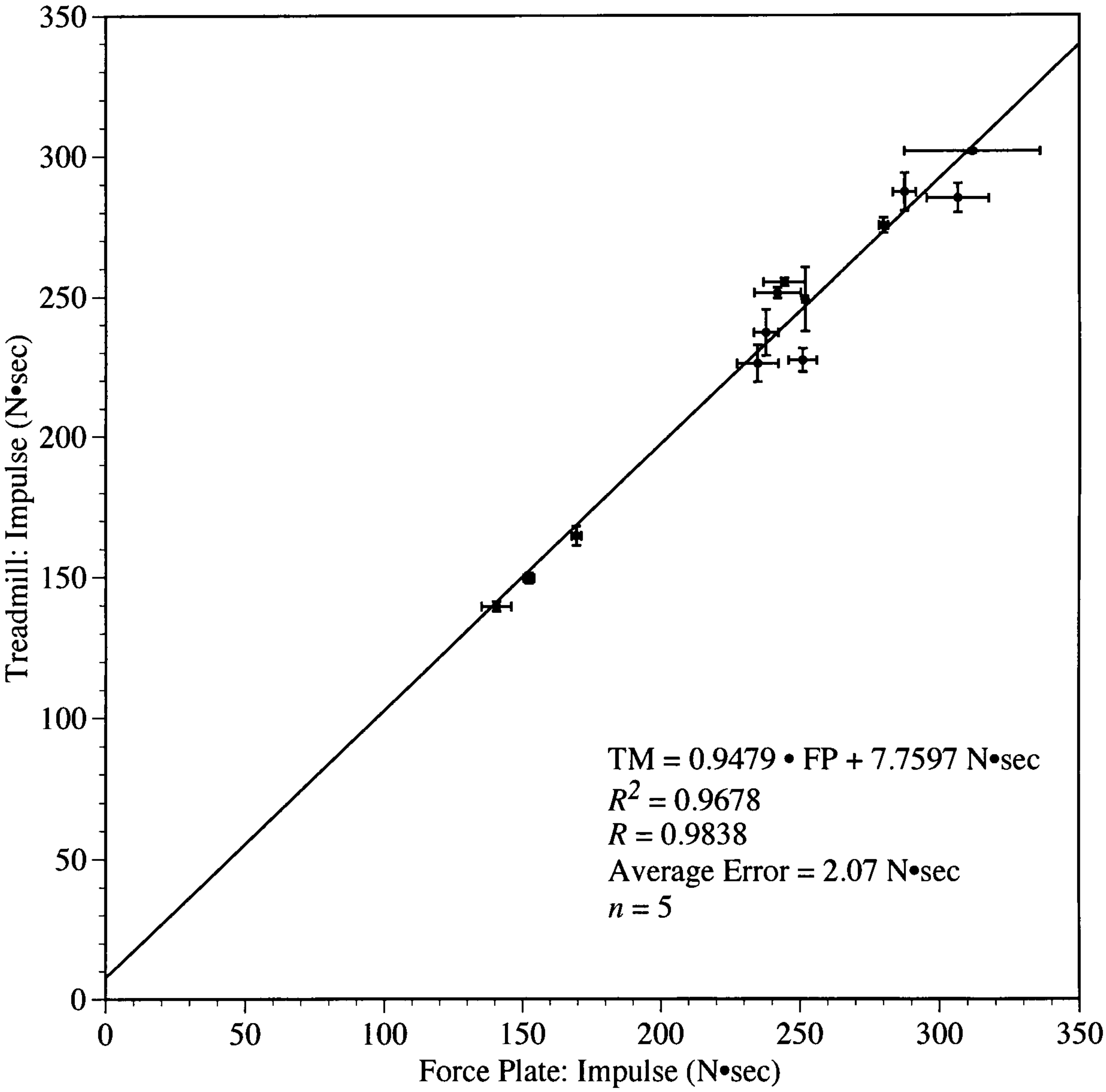

Figure C.7
Treadmill vs. Force Plate:
Total Impulse Correlation

running than was the force plate running. Thus some of the quantified differences should be discounted as deviations from the true activity which was simulated.

C.3 Reliability

Once the treadmill was validated against the force plate, an attempt was made to test the reliability of human measures. Following the experiment the numbers which follow were found to contain a systematic error due to an improperly grounded transducer. Transducer number five (left rear) was not offset from zero, thus ground appeared to be off-scale. Thus the signal loss consisted of the small amount of signal required to achieve a mearureable force level. The loss was at most 50 N, and was probably much less due to the small fraction of the range which was lost and the low range (100 lb or 444 N) of the transducer. Reliability may evaluated from these data with the understanding that this was a systematic error. Further discussion of this error follows the reporting of the reliability data.

Using the standard warm-up procedure for the treadmill, the apparatus was prepared for testing. A single subject ran on the treadmill at three self-selected speeds (7.35, 11.90, and 15.33 kph or 2.04, 3.31, and 4.25 m/s). Following one trial at the three speeds, the subject got off the treadmill, and the motor was turned off. Several minutes of break occurred before the subject performed another series of the three speeds. During this time, the 484B02 conditioners were grounded and the treadmill warm-up procedure repeated. For each speed, the unfiltered VGRF curves were analyzed for the characteristics described earlier. For each ten-second trial, all right footstrikes, numbering 13 or 14, were analyzed. See Table C.3. Comparisons were made between trials to determine test-retest reliability. Within trial analysis summarized the intra-subject variability.

For peak impact force, the slow trials were not significantly different from each other ($p > 0.09$), while the trials at each of the medium and fast speeds were significantly

Table C.3
Subject Reliability Data
(Standard Deviations)

Speed	Trial	Peak Impact Force (N)	Active Peak Force (N)	Contact Time (sec)	Rise Rate (N/sec)	Impulse (N*sec)
Slow (2.04 m/s)	1	747.74 (102.03)	1187.48 (55.05)	0.3524 (0.0116)	44203 (8493)	232.02 (8.86)
	2	671.94 (90.86)	1275.52 (49.89)	0.3335 (0.0117)	40814 (9326)	236.72 (5.46)
	3	690.84 (72.31)	1334.13 (42.60)	0.3236 (0.0110)	53586 (12407)	237.23 (3.40)
Medium (3.31 m/s)	1	1032.88 (126.52)	1593.67 (34.42)	0.2555 (0.0093)	88794 (24054)	241.11 (2.64)
	2	1045.07 (106.34)	1614.31 (22.57)	0.2538 (0.0103)	109601 (19658)	239.64 (2.87)
	3	1230.17 (113.29)	1587.48 (35.30)	0.2541 (0.0067)	141303 (32236)	236.75 (3.98)
Fast (4.26 m/s)	1	1441.74 (77.57)	1839.70 (26.24)	0.2070 (0.0122)	141826 (14724)	235.57 (2.72)
	2	1442.63 (123.41)	1832.83 (48.35)	0.2012 (0.0056)	140805 (17607)	235.64 (2.24)
	3	1574.66 (80.91)	1779.87 (41.14)	0.2100 (0.0120)	171596 (19131)	230.62 (3.80)

different ($p < 0.0001$ and $p < 0.0008$, respectively). Active Peak forces were significantly different for slow ($p < 0.0001$) and fast ($p < 0.0005$) trials while only medium speed trials were not significantly different ($p > 0.075$). Contact time was significantly different for slow trials ($p < 0.0001$) while medium and fast speeds were repeatable ($p > 0.87$ and $p > 0.086$, respectively). Rise rate was variable between trials at all three speeds ($p < 0.009$ for slow and $p < 0.0001$ for medium and fast). Total impulse proved to be significantly different between repeated trials of both medium ($p < 0.005$) and fast ($p < 0.0001$) speeds but slow speed trials were not significantly different ($p > 0.08$).

For these characteristics, the variability of both the subject and the device were small enough that a change in velocity caused a significant difference in many of the measured characteristics. Significance was expected to vary with the subject, the speeds chosen, and the difference between speeds.

Due to the improper grounding of a transducer, these reliability trials were flawed. With the understanding that the error was small and systematic, these results may be analyzed. They compare the landmarks of the vertical ground reaction force curve of running. For studies of impact, or those requiring absolute zero-values, this experiment should be repeated with proper grounding and more subjects. Furthermore, filtering should be performed on the data to minimize high frequency noise in the signals, such as was visible on channels 5 and 6. An additional repeatability pilot experiment specific to the walk-run transition study was conducted and is discussed in section C.10.

C.4 Center of Pressure

The next pilot study was to test and validate the treadmill's ability to determine center of pressure, both statically and dynamically. For static testing, the center of pressure was calculated while a stationary point load was placed on the treadmill bed. A ski pole tip was placed at ten locations on the bed similar to the loading positions which occur during walking and running. Loads were varied during a trial by the experimenter applying more weight to the ski pole. Before center of pressure calculations were made, the data were filtered at 60 Hz, and zero-values were calculated for each transducer. Correlations and errors were calculated from the known positions (KP). The resulting fit may be seen in Figure C.8, Center-of-Pressure Validation. The least-squares-fit followed the relationship: TM = 0.9722 * KP + 0.0239 m with $R^2 = 0.9955$, $R = 0.9977$, and the Average Error equal to 0.0277 m.

These measures were repeated with the treadmill running. Six point loads were placed on the bed of the treadmill, to the side of the moving belt. The resulting linear fit followed the equation: TM = 0.9429 * KP + 0.0245 m with $R^2 = 0.9988$, $R = 0.9994$, and the Average Error = 0.0199 m. See Figure C.9. Despite filtering at 60 Hz, low frequency vibration from the motor and belt resulted in some variation in each of the center of pressure readings. The standard deviations of position ranged from 0.0095 to 0.0183 m.

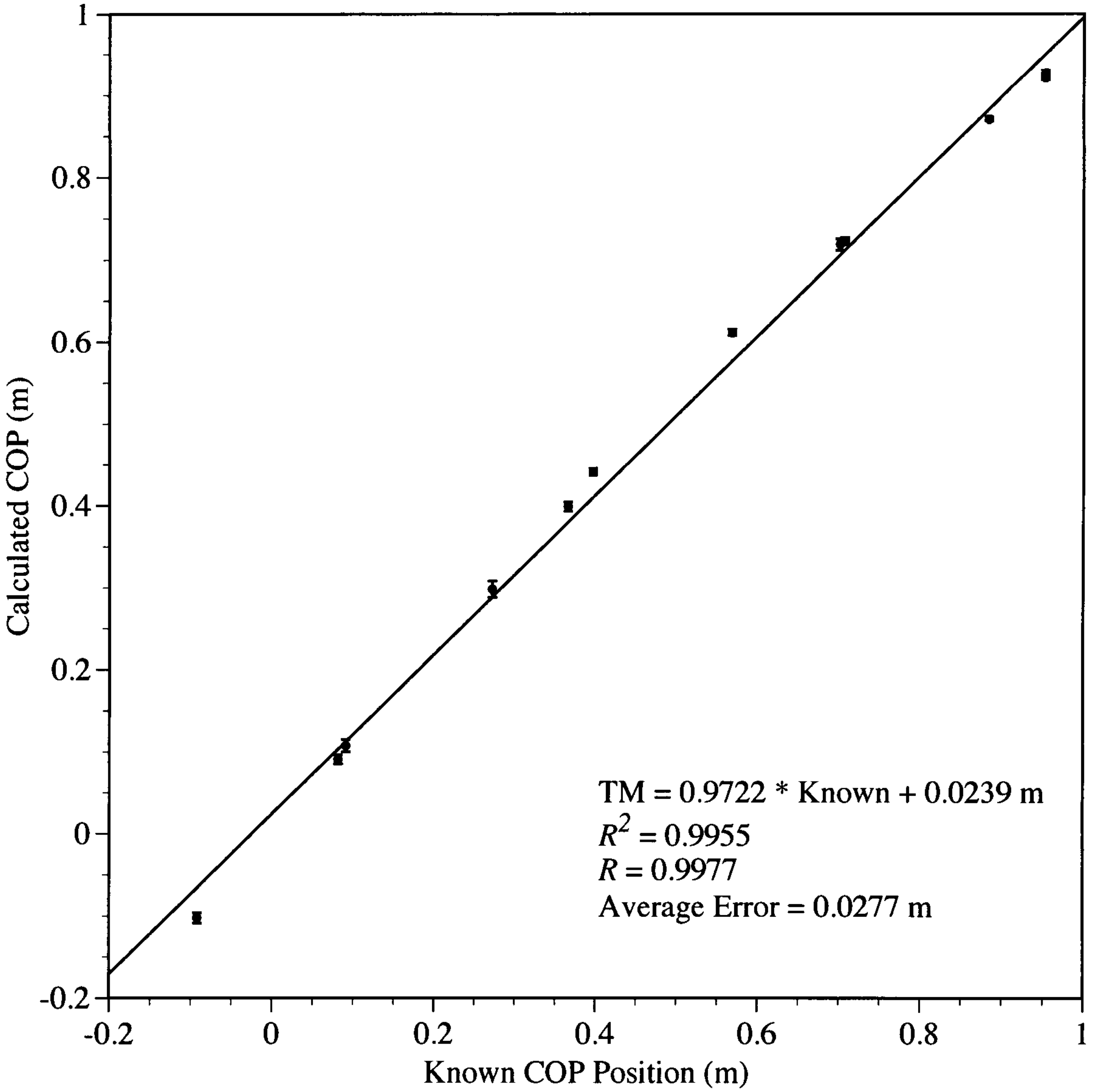

Figure C.8
Center of Pressure Validation: Treadmill Off
Calculated values vs. Known Positions
(±SD, ±Error)

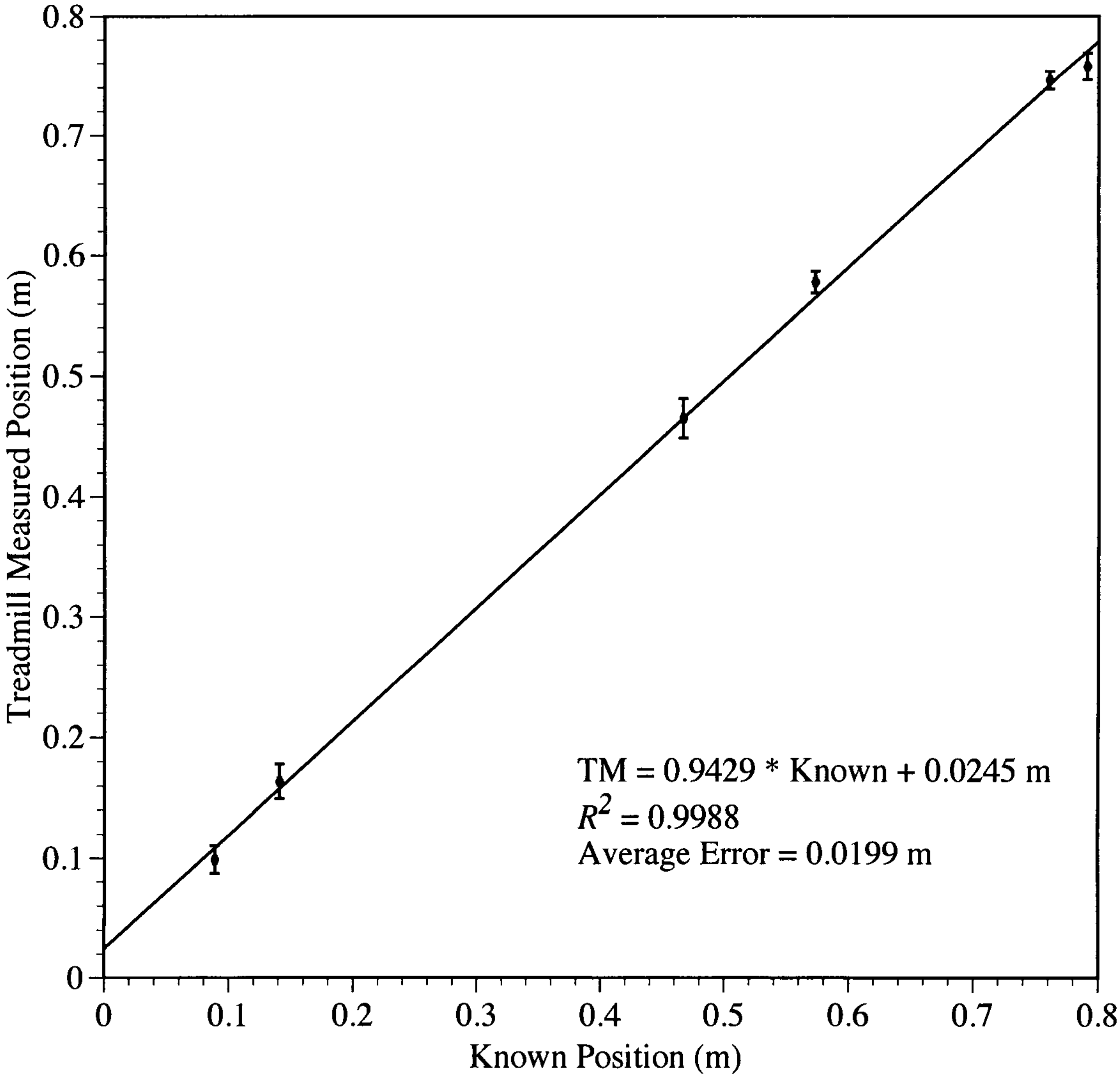

Figure C.9
Center of Pressure Validation: Treadmill On
Treadmill Measured Values vs. Known Positions

For both tests initial loading and unloading caused gross variation in the COP. Besides these conditions, variation in applied force did not change the calculated center of pressure location.

The data points for these two tests were combined to yield the equation: TM = 0.9657 * KP + 0.0227 m with $R^2 = 0.9957$, $R = 0.9978$ and the Average Error was equal to 0.0307 m (See Figure C.10). This fit did not support a higher degree equation for the trend line, as had been indicated by a fit performed on the data with the treadmill off.

The calculated values for center of pressure were reasonably good, but were not sufficiently accurate to allow inverse dynamics calculations to be performed. The variation from known values and the variation introduced with the belt on would have caused significant errors in calculated joint moments.

C.5 Constant Moving Force: Body Weight

To test the ability of the treadmill to measure force along the length of the belt, a constant force was applied to the moving treadmill belt. A subject lowered himself onto the front of the treadmill belt with a minimum of vibration. Once on the moving belt, the subject stood still, allowing the belt to move him rearward at a constant velocity. At the rear of the treadmill, the subject dropped off the belt and the treadmill.

Four trials were performed. The force values were analyzed for average force and relative difference along the length of the bed. While the weight was being applied there was a period of adjustment before the force measurements became stable. At the end of the measurements, when the subject was over the rearmost transducers, the signals from transducers E and F reached maximum. Data were evaluated from the end of the loading period to the point of signal overload for the rear transducers. For overall measures see Table C.4.

For differentiation between the regions of the bed, front, middle, and rear values are presented in Table C.5. These were roughly divided into regions by time. Because

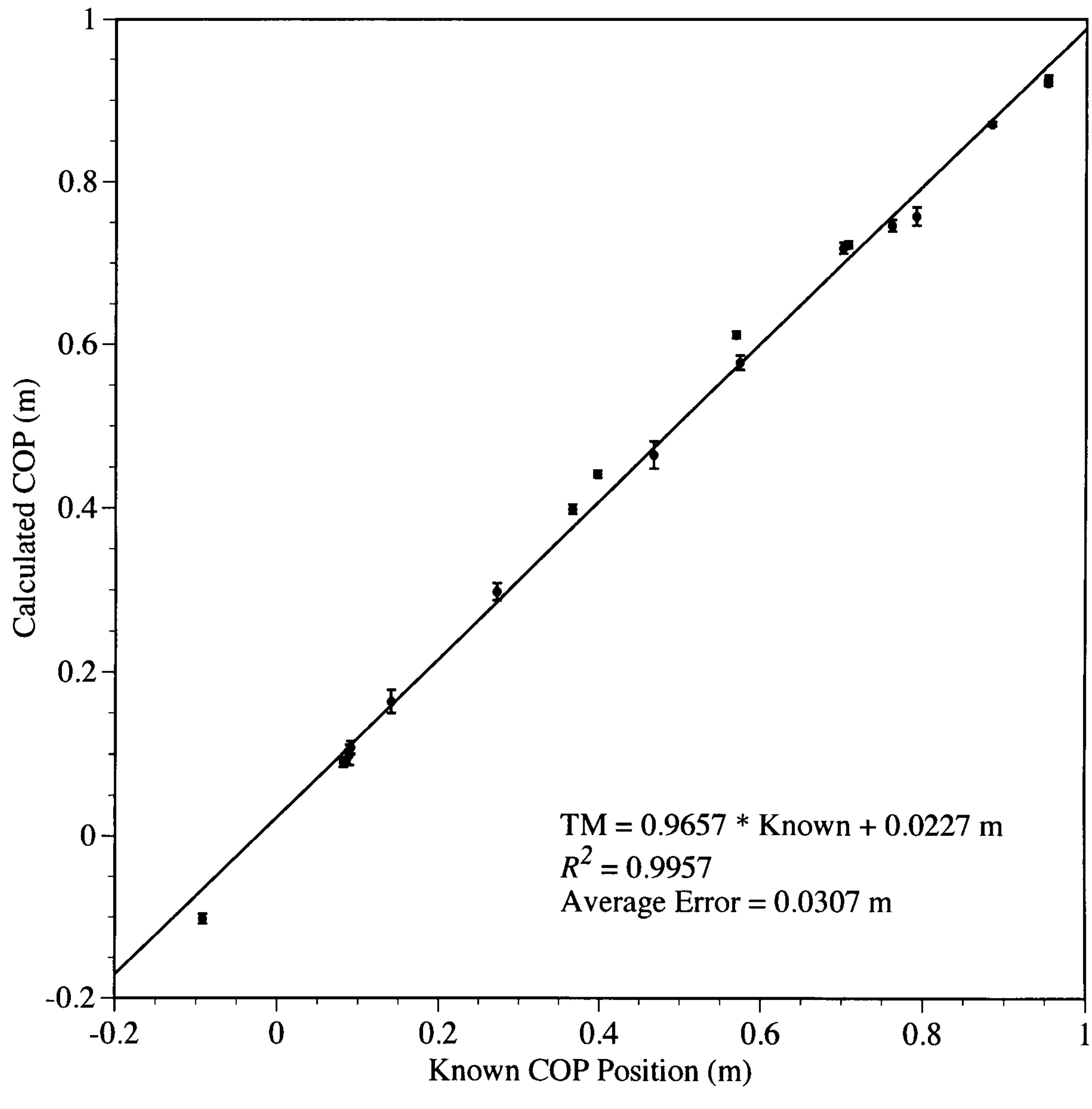

Figure C.10
Center of Pressure Validation:
Combined Measurements
Calculated values vs. Known Positions

Table C.4: Total Body Weight Errors

			Actual BW=792.90 N
Trial	Average BW (N)	Absolute Error (N)	% Error
1	805.59	12.70	1.60
2	813.76	20.87	2.63
3	829.82	36.93	4.66
4	820.66	27.73	3.50
Average	817.46	24.56	3.10

Table C.5: Body Weight Along the Bed

Trial	Average Force	Front	% Deviation	Middle	% Deviation	Rear	% Deviation
1	805.59	805.60	0.00	804.24	-0.17	800.31	-0.66
2	813.76	807.74	-0.74	815.72	0.24	813.19	-0.07
3	829.82	828.40	-0.17	825.41	-0.53	833.06	0.39
4	820.66	816.61	-0.49	820.23	-0.05	823.25	0.32
Average	817.46	814.59	-0.35	816.40	-0.13	817.45	0.00

the divisions were neither exact nor consistent in position or time, the percent deviations do not sum to 0%.

From this experiment, it was indicated that the treadmill was precise in its measurements although accuracy could have been improved. At the same time, however, the absolute errors for force were within the average errors calculated for the static measurements. For the study which was undertaken, this level of validity was sufficient. It should be noted that the rear two transducers could have been maximized if subjects walked or stood on the rearmost portion of the belt. The remaining usable area of the belt was sufficient for all experiments.

C.6 Body Weight Calculated From Gait

The treadmill was tested for its ability to calculate body weight with a subject either standing or walking on the treadmill. Two sets of four static trials were performed, each with a "pre-zero" file taken immediately preceding the data collection. See Tables C.6 and C.7.

From these static body weight measures, the treadmill proved to be more precise than accurate. The errors between treadmill measurements were smaller than those

Table C.6: Static Body Weight Measures, Set 1
Actual Body weight = 782.88 N

Trial	Measured BW	Absolute Error	Percent Error
1	737.04	-45.84	-5.86
2	727.60	-55.28	-7.06
3	729.47	-53.41	-6.82
4	735.92	-46.97	-6.00
Average	732.51	-50.38	-6.43
Standard Deviation	4.67	4.67	0.60

Table C.7: Static Body Weight Measures, Set 2
Actual Body Weight = 775.10

Trial	Measured BW	Absolute Error	Percent Error
1	715.15	-59.95	-7.73
2	710.17	-64.93	-8.38
3	715.41	-59.69	-7.70
4	699.51	-75.59	-9.75
Average	710.06	-65.04	-8.39
Standard Deviation	7.43	7.43	0.96

between treadmill and criterion measures; note the low standard deviations, but high absolute errors.

Three tests were performed to calculate body weight from an integral number of walking strides. During a ten-second measurement session, the subject either got on or off the treadmill, providing both a zero-value and walking strides within the same file. Body weight was calculated as the difference between the average force over an integral number of cycles and the zero-value from the file. Walking was performed at 4 kph, 6 kph and 8 kph, with the results reported in Tables C.8, C.9, and C.10, respectively.

When the average force calculated from each of these walking trials was compared to the average static measures for the same body weight, the differences were found to be -14.36 N (-1.96%), -53.02 N (-7.47%), and -39.34 N (-5.54%), respectively. These tests of body weight demonstrated that absolute force values from the treadmill contained errors which were consistent, but relatively large. Therefore body weight measures could be compared to each other with little error, but when compared to measures taken with criterion devices, errors were large.

C.8. Repeat of Body Weight Measurements

Following the body weight calculations, one of the transducers was contaminated with moisture, resulting in fast signal reductions following a loading, effectively reducing the time constant. This transducer (E) was replaced with a new 208B02, "B" indicating hermetically sealed. Otherwise this transducer was identical to the 208A02 which it replaced. Following replacement, the new sensitivity value was entered into the conversion program. To verify that the system had not changed with this new transducer, the last two validation experiments were repeated: measurement of a body weight moving rearward on the treadmill and calculation of body weight from stationary and walking trials.

Table C.8: Body Weight Calculation from Walking at 4 kph
Actual Body Weight = 782.88

Trial	Measured BW	Absolute Error	Percent Error
1	746.97	-35.92	-4.59
2	746.39	-36.49	-4.66
3	744.65	-38.23	-4.88
4	749.44	-33.44	-4.27
Average	746.86	-36.02	-4.60
Standard Deviation	1.98	1.98	0.25

Table C.9: Body Weight Calculation from Walking at 6 kph
Actual Body Weight = 775.10

Trial	Measured BW	Absolute Error	Percent Error
1	758.44	-16.66	-2.15
2	763.38	-11.72	-1.51
3	770.52	-4.58	-0.59
4	759.97	-15.13	-1.95
Average	763.08	-12.02	-1.55
Standard Deviation	5.37	5.37	0.69

Table C.10: Body Weight Calculation from Walking at 8 kph
Actual Body Weight = 775.10

Trial	Measured BW	Absolute Error	Percent Error
1	737.66	-37.44	-4.83
2	757.00	-18.10	-2.34
3	750.06	-25.04	-3.23
4	752.87	-22.23	-2.87
Average	749.40	-25.70	-3.32
Standard Deviation	8.33	8.33	1.07

C.8.1 Body Weight Moving Rearward

The same method was used as above: a subject gently lowered himself onto the front of the treadmill belt and rode the slow-moving belt rearward. Two sets of such trials were conducted while force was measured. Zero was calculated from the force record prior to the weight application. For the first set of trials, body weight was calculated at the relative positions along the length of the bed. The results from these trials, demonstrating similar errors to the previous trials, appear in Tables C.11, C.12, and C.13.

C.8.2 Calculation of Body Weight from Static and Locomotion Trials

Several trials were performed in which body weight was calculated from static body weight or an integral number of walking or running strides. These results appear in Table C.14.

From these trials it was apparent that the system had not changed with the replacement of one of the transducers. The treadmill apparatus remained more precise

Table C.11: Total Body Weight Errors
Trial 1

			Actual BW=789.56 N
Trial	Average BW (N)	Absolute Error (N)	% Error
1	773.67	-15.89	-2.01
2	782.40	-7.16	-0.91
3	807.61	18.05	2.29
4	809.22	19.66	2.49
5	800.49	10.93	1.38
Average	794.68	5.12	0.65

Table C.12: BW Along the Bed
Trial 1

Trial	Average Force	Front	% Deviation	Middle	% Deviation	Rear	% Deviation
1	773.67	770.42	-0.42	773.94	0.03	775.64	0.25
2	782.40	767.87	-1.86	784.52	0.27	785.96	0.46
3	807.61	812.25	0.57	805.95	-0.21	809.32	0.21
4	809.22	798.25	-1.36	811.67	0.30	807.21	-0.25
5	800.49	783.64	-2.10	805.94	0.68	798.34	-0.27
Average	794.68	786.49	-1.03	796.40	0.22	795.29	0.08

Table C.13: Total Body Weight Errors
Trial 2

			Actual BW=784.00 N
Trial	Average BW (N)	Absolute Error (N)	% Error
1	794.69	10.69	1.36
2	765.85	-18.15	-2.31
3	799.85	15.85	2.02
4	795.58	11.58	1.48
Average	798.99	4.99	0.64

than accurate, with a lower measurement of force than actual. Therefore the other validation tests were not repeated.

C.9 Validation with Dynamic Loads

To validate the ability of the treadmill to measure dynamic loads, a force hammer was employed. This device had a transducer at its head to measure force. The transducer was a 208A03 crystal from PCB Piezotronics, the same kind of transducer used in the

Table C.14: Calculation of Body Weight
Body Weight = 800.68 N

Method	Measured BW (N)	Absolute Error (N)	% Error
Walk at 1.5 kph	793.67	-7.00	-0.88
Walk at 1.5 kph	768.53	-32.15	-4.01
Walk at 1.5 kph	767.76	-32.92	-4.11
Run at 9 kph	761.73	-38.95	-4.86
Stationary BW	762.68	-38.00	-4.75
Stationary BW	750.85	-49.83	-6.22

treadmill apparatus. A 408B signal conditioner set at DC-coupling provided comparable, if not equal, conditioning for the criterion measure of the hammer as compared to the to-be-proven treadmill.

The hammer was pressed into the treadmill bed along the midline of the belt. Rather than striking the treadmill, load was applied to the hammer for about a second. While this was not an impact normally associated with a hammer, it was more realistic for the loads which were applied to the treadmill during gait.

The data for these loads were collected using a version of the normal collection program modified for faster data collection (VBDHMDMA.EXE). The original program was written to collect from seven channels; the seventh intended for synchronization with video. For this procedure the force hammer was connected to the seventh channel and only the scan rate was changed. To provide greater time-base resolution, a Keithley-Metrabyte DAS-1402 data acquisition board was used. This enabled data collection at 2000 Hz, although page boundaries were a problem, limiting the length of the data collections. The universal drivers (Computer Boards Universal Library) allowed an upgrade to this board with the change of a single number in the code.

Conversion of the data was performed using the same Visual Basic program which converted all of the subject data. For these data no filtering was used. Again the program was written for seven channels of data. In this case the conversion factor for channel seven was changed from unity to 11.23 mV/lb (2.52 mV/N) corresponding to the hammer transducer.

Following conversion, a spreadsheet was utilized to subtract the offset values, sum across transducers, and correlate the treadmill and hammer forces. When plotted as force versus time, the curves were very similar (See Figures C.11 and C.12). The only differences arose in the first tenth of a second when resonance was evident on the treadmill force curve. The low amplitude vibrations were superimposed on the general shape of the force curves from the hammer. A gross measurement of this vibration indicated the resonance to be at approximately 100 Hz. This was much lower than the value of 275 Hz measured previously .

When the hammer (HM) and treadmill (TM) force curves were plotted against each other, the slopes were close to one as was the R^2-value. See Figures C.13 and C.14. One loading provided a trend line at TM = 1.0043 * HM + 0.1665 N, R^2 = 0.9995. Another curve yielded a trend line of TM = 1.018 * HM + 0.176 N, R^2 = 0.9996. When impact was removed from these plots, the trendlines switched to TM = 0.9992 * HM + 1.1411 N, R^2 = 0.9998 and TM = 1.0122 * HM + 1.211 N, R^2 = 0.9998, respectively. These curves were made with the load increasing to 250 N.

More than two impacts were originally recorded. Because of page boundary limits, each of the data files was a second shorter then expected, cutting off some impacts. Another file was incorrectly triggered, missing the impacts altogether. Despite having only two trials, the data analyzed indicated validity of dynamic force measurements. While resonance may have been a concern if heelstrike was to be analyzed, filtering could have removed the bulk of the vibration from the signal. For the changes in force studied in gait, the validation with the force hammer proved they were accurate.

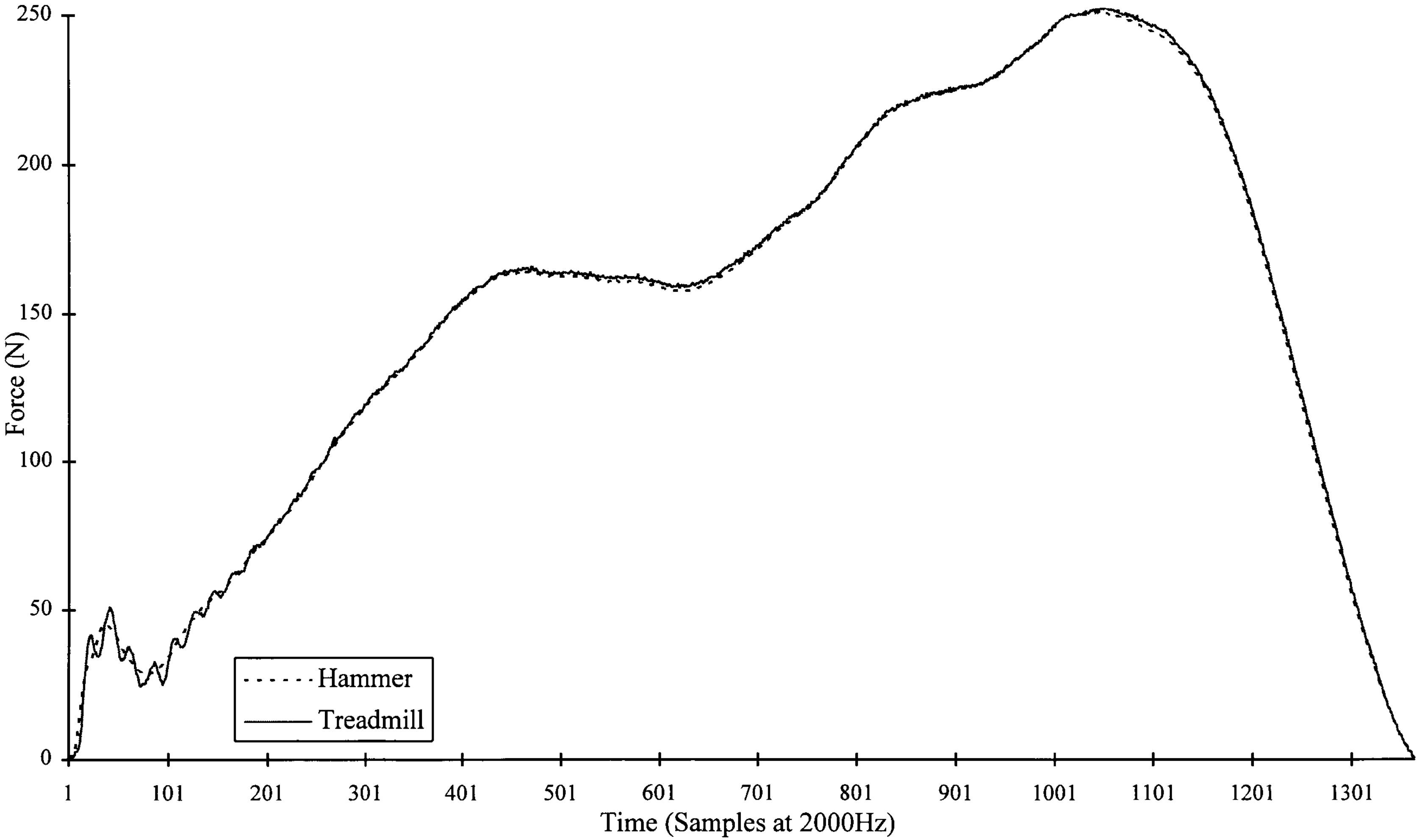

Figure C.11 Force vs. Time: Treadmill and Hammer

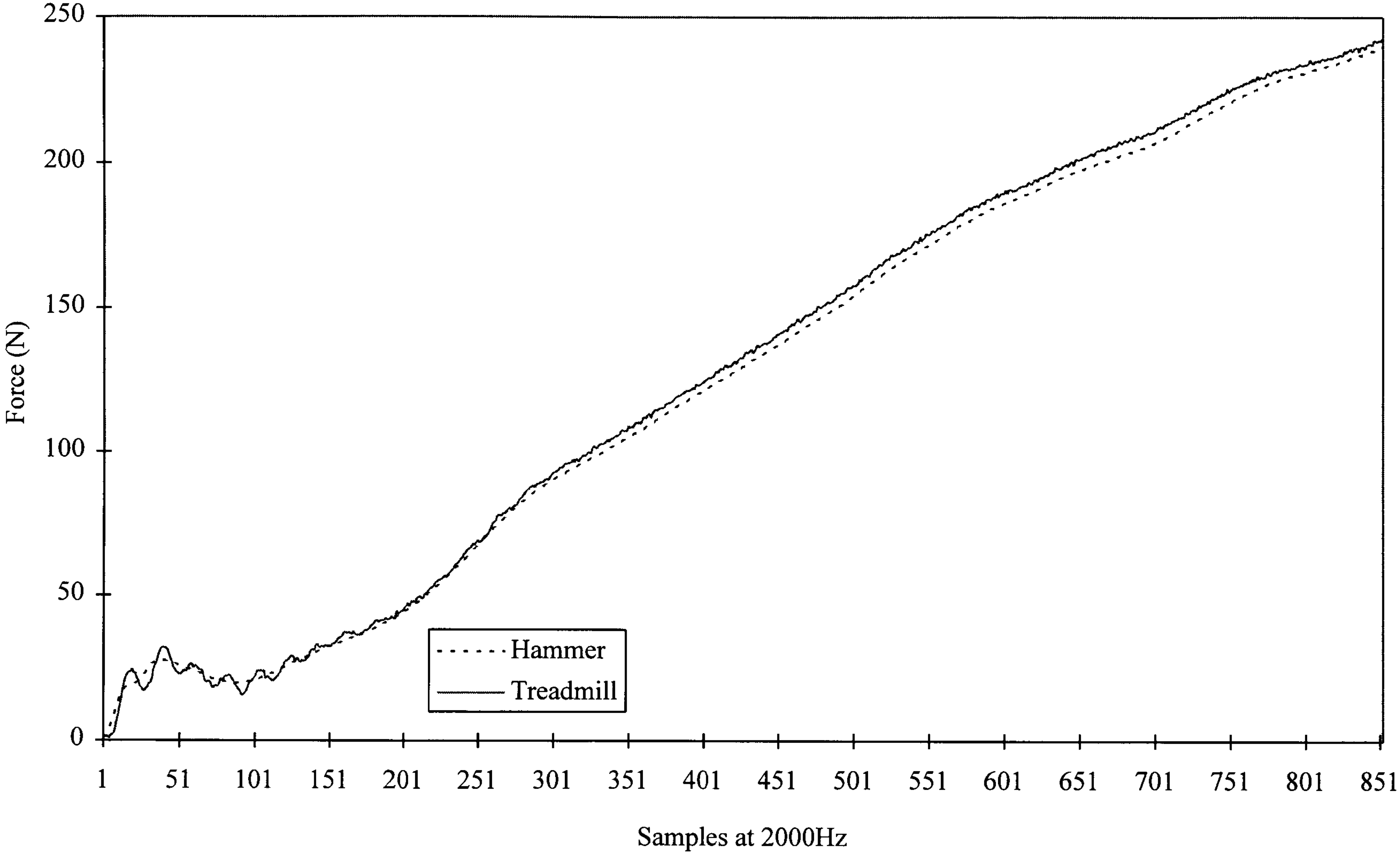

Figure C.12 Force vs. Time: Treadmill and Hammer

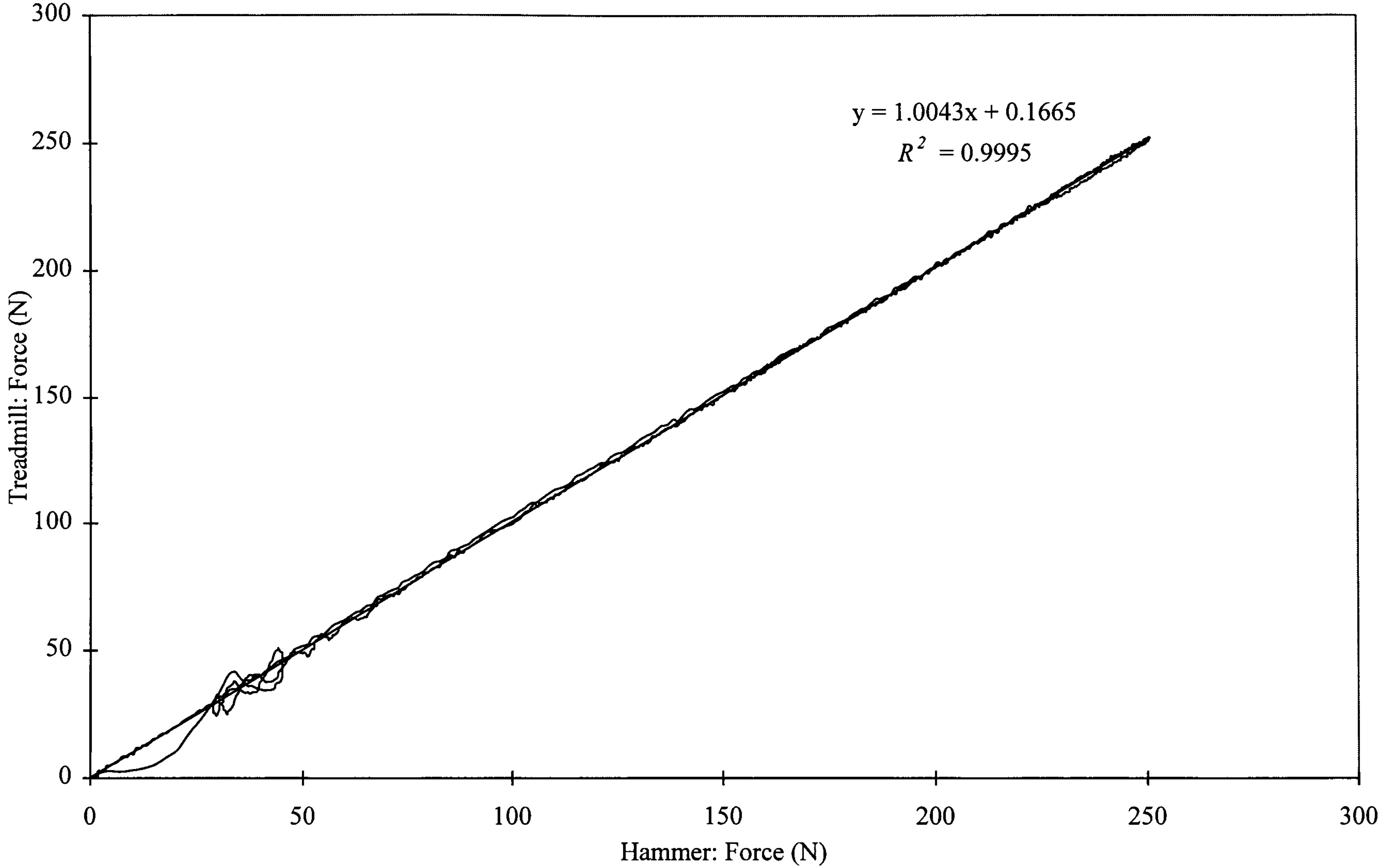

Figure C.13 Treadmill vs. Hammer: Force Correlation

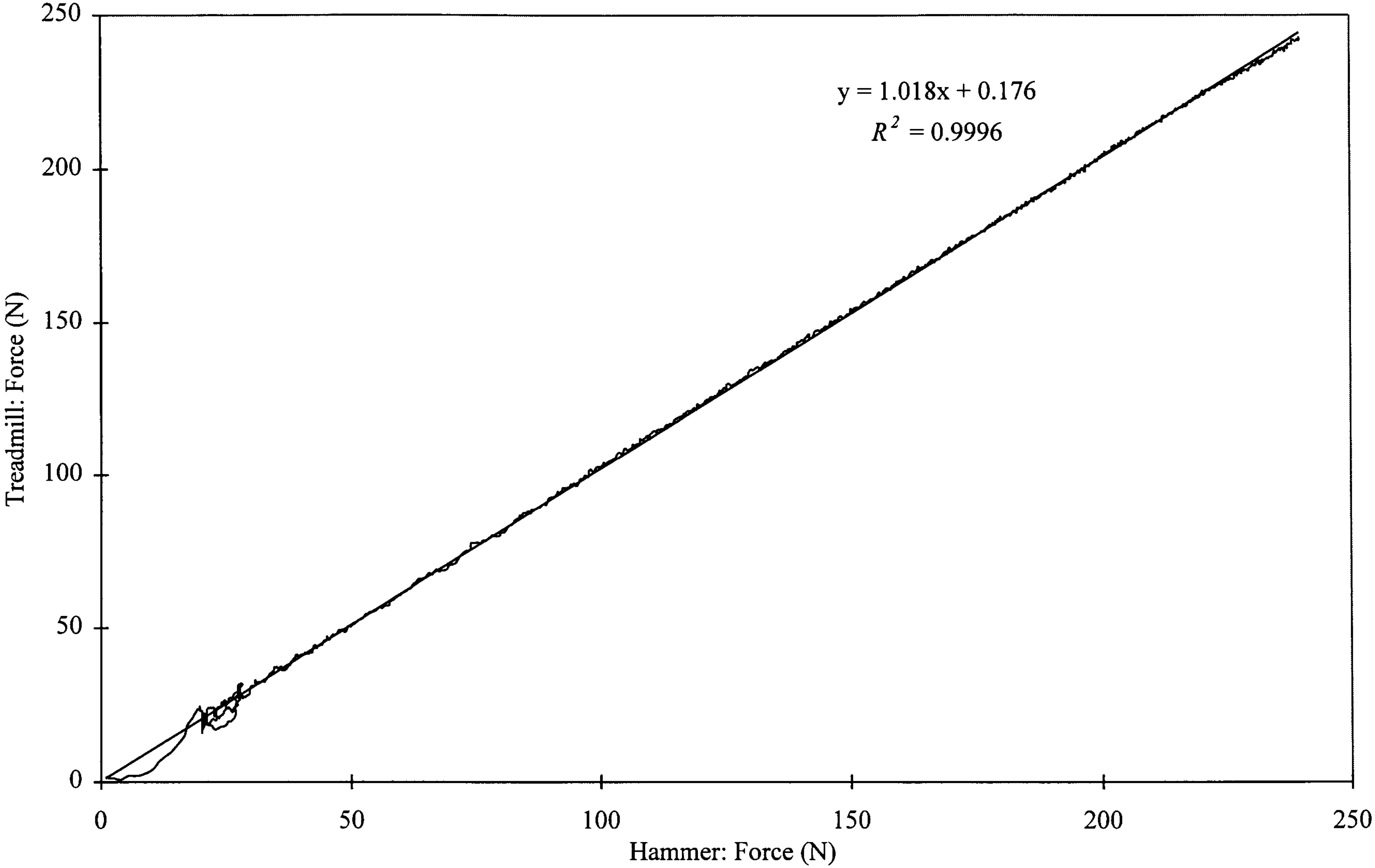

Figure C.14 Treadmill vs. Hammer: Force Correlation

C.10 Repeatability

To test the repeatability of the force measurements taken by the treadmill, specific tests were performed. These tests were designed to be very similar to actual testing conditions for the study which was undertaken. Two subjects walked and ran on the treadmill at four speeds: 1.4, 1.6, 1.8, and 2.0 m/s. The order of testing was to walk over the range of speeds, then run over the range of speeds. This series of trials was repeated three times within a day. Between each trial, the transducers were turned off, then reset.

Because the experiment to be undertaken was to measure midstance vertical ground reaction force, these repeatability tests evaluated the same variable. Following testing, the analysis programs were used to pick off midstance and the corresponding vertical ground reaction force. For each speed and trial, the mean, standard deviation, coefficient of variation, and range of forces were calculated. These were summarized for each foot. See Tables C.15, C.16, C.17 and C.18: Repeatability Measures.

With only two subjects an R-value could not be calculated. Instead two error terms were calculated: the mean of the differences between sessions and the mean squared error. The differences were calculated between sessions (1-2, 1-3, 2-3). The error score was the sum of the absolute values of the differences divided by the number of differences (three). The mean squared error was calculated as the square root of the sum of squares.

For walking, the error scores varied from a low of 4.58 N (1.5% of average force) to 51.05 N (9% of average force). The corresponding mean squared errors were 3.45 N (1.1% of average force) and 38.60 N (7.0% of average force). When calculated for running, the lowest error score was 46.96 N (4.8% of average force) while the highest was 225.05 N (22.2% of average force). The corresponding mean squared errors were 38.00 N (3.9% of average force) and 175.77 N (17.4% of average force). The lower bound of these errors were in the range of the average error calculated from static force measurements. While the largest errors were alarming, these were from two particular

Table C.15 Repeatability Measures: Left Foot, Walking
All Values are Midstance Vertical Ground Reaction Forces (N)

	Gait:	Walk	Walk	Walk	Walk	Walk	Walk	Walk	Walk
	Speed:	Slow	Slow	Med	Med	Med-Fast	Med-Fast	Fast	Fast
Trial	Subject:	1	2	1	2	1	2	1	2
1	Mean	447.5	589.0	372.5	492.3	342.8	473.3	305.5	398.6
	SD	23.9	15.2	13.7	35.3	24.8	44.3	17.8	22.2
	CV	18.7	38.7	27.1	13.9	13.8	10.7	17.2	17.9
	Max	486.1	616.0	389.4	561.4	404.5	556.8	334.0	436.6
	Min	399.1	572.0	351.9	442.0	313.9	434.7	272.4	367.4
2	Mean	457.1	544.9	398.6	498.6	336.4	462.3	286.6	414.7
	SD	10.3	10.6	17.2	18.6	16.1	24.1	24.4	32.8
	CV	44.3	51.3	23.2	26.9	20.9	19.2	11.7	12.6
	Max	472.1	556.4	428.1	518.5	359.7	495.8	319.4	458.3
	Min	442.2	527.7	374.2	472.9	308.1	420.2	239.0	364.3
3	Mean	502.3	516.0	399.4	450.2	330.8	409.7	306.0	371.5
	SD	17.4	13.8	13.5	21.6	14.8	18.3	12.0	41.5
	CV	28.8	37.4	29.7	20.8	22.4	22.4	25.4	9.0
	Max	518.3	537.2	420.2	480.1	349.6	437.0	322.3	424.8
	Min	466.8	500.9	387.0	410.7	305.5	380.2	283.0	305.5
Mean Diff.	x1-x2	-9.6	44.1	-26.1	-6.3	6.4	11.0	18.9	-16.1
	x1-x3	-54.8	73.0	-27.0	42.1	12.0	63.6	-0.6	27.2
	x2-x3	-45.2	28.9	-0.8	48.4	5.6	52.6	-19.4	43.3
Error:	(N)	36.5	48.7	18.0	32.3	8.0	42.4	13.0	28.9
MSe:	(N)	29.3	36.8	15.3	26.3	6.0	34.0	11.1	21.9

Table C.16 Repeatability Measures: Right Foot, Walking
All Values are Midstance Vertical Ground Reaction Forces (N)

	Gait:	Walk	Walk	Walk	Walk	Walk	Walk	Walk	Walk
	Speed:	Slow	Slow	Med	Med	Med-Fast	Med-Fast	Fast	Fast
Trial	Subject:	1	2	1	2	1	2	1	2
1	Mean	450.3	590.8	381.2	492.6	334.4	469.2	312.2	419.7
	SD	11.0	15.8	11.8	6.8	14.8	36.0	19.6	30.3
	CV	41.0	37.4	32.3	72.0	22.6	13.0	15.9	13.8
	Max	467.1	605.8	401.4	501.5	357.4	544.0	340.2	478.1
	Min	434.4	561.4	364.0	482.0	319.1	433.3	268.7	377.7
2	Mean	450.0	544.0	404.1	502.1	348.0	460.0	308.1	427.3
	SD	14.2	16.2	15.5	13.2	16.0	19.0	15.5	34.7
	CV	31.6	33.7	26.1	38.0	21.7	24.2	19.8	12.3
	Max	477.0	555.2	423.7	520.6	373.9	489.6	337.0	483.1
	Min	429.8	506.0	377.2	483.1	324.7	433.5	287.6	390.8
3	Mean	479.5	514.2	405.1	470.1	339.6	409.6	315.0	402.3
	SD	14.4	17.4	16.0	15.2	12.4	19.8	21.3	29.9
	CV	33.4	29.6	25.4	30.8	27.3	20.7	14.8	13.4
	Max	498.0	541.8	429.3	490.8	352.7	440.9	362.6	440.9
	Min	460.5	487.7	378.1	445.3	318.9	373.7	289.8	335.8
Mean Diff.	x1-x2	0.2	46.8	-22.9	-9.5	-13.6	9.1	4.0	-7.6
	x1-x3	-29.2	76.6	-23.8	22.5	-5.2	59.5	-2.8	17.4
	x2-x3	-29.5	29.7	-1.0	32.0	8.4	50.4	-6.9	25.0
Error:	(N)	19.6	51.0	15.9	21.3	9.1	39.7	4.6	16.7
MSe:	(N)	16.9	38.6	13.5	16.4	6.9	32.1	3.5	12.8

Table C.17 Repeatability Measures: Left Foot, Runing
All Values are Midstance Vertical Ground Reaction Forces (N)

	Gait:	Run	Run	Run	Run	Run	Run	Run	Run
	Speed:	Slow	Slow	Med	Med	Med-Fast	Med-Fast	Fast	Fast
Trial	Subject:	1	2	1	2	1	2	1	2
1	Mean	1000.4	1389.9	1153.7	1418.9	1155.3	1401.1	1257.3	1475.7
	SD	50.1	32.8	51.7	27.4	51.8	30.7	26.4	34.4
	CV	20.0	42.4	22.3	51.8	22.3	45.6	47.7	42.9
	Max	1114.0	1435.5	1240.0	1476.1	1234.8	1429.3	1321.1	1528.1
	Min	943.2	1311.7	1071.0	1377.4	1066.5	1318.4	1207.3	1411.0
2	Mean	952.9	1264.4	816.2	1263.0	1081.4	1315.0	1202.0	1367.3
	SD	47.2	48.2	330.2	20.8	45.4	33.4	33.5	31.2
	CV	20.2	26.2	2.5	60.7	23.8	39.4	35.9	43.8
	Max	1049.6	1369.5	997.0	1300.2	1169.3	1372.6	1272.5	1415.5
	Min	864.7	1184.0	61.7	1222.9	1019.2	1256.9	1157.2	1297.7
3	Mean	1032.3	1387.1	1069.9	1445.7	1189.0	1437.5	1281.1	1466.8
	SD	56.4	41.6	56.6	21.5	40.1	31.2	31.9	51.1
	CV	18.3	33.3	18.9	67.4	29.7	46.1	40.2	28.7
	Max	1138.8	1454.5	1190.9	1490.4	1276.1	1483.7	1347.0	1539.9
	Min	963.3	1313.6	994.9	1409.4	1106.0	1385.9	1240.4	1371.4
Mean Diff.	x1-x2	47.5	125.6	337.6	155.9	73.8	86.1	55.3	108.4
	x1-x3	-32.0	2.8	83.8	-26.9	-33.8	-36.4	-23.7	8.9
	x2-x3	-79.4	-122.8	-253.7	-182.7	-107.6	-122.5	-79.0	-99.4
Error:	(N)	52.9	83.7	225.1	121.8	71.7	81.7	52.7	72.3
MSe:	(N)	40.0	71.7	175.8	98.7	55.0	62.9	40.6	60.2

Table C.18 Repeatability Measures: Right Foot, Running
All Values are Midstance Vertical Ground Reaction Forces (N)

Trial	Gait:	Run	Run	Run	Run	Run	Run	Run	Run
	Speed:	Slow	Slow	Med	Med	Med-Fast	Med-Fast	Fast	Fast
	Subject:	1	2	1	2	1	2	1	2
1	Mean	1013.0	1381.9	1212.6	1447.5	1203.1	1421.8	1304.6	1496.7
	SD	36.2	34.2	61.7	20.7	76.8	27.7	42.5	30.2
	CV	28.0	40.3	19.6	70.0	15.7	51.4	30.7	49.6
	Max	1081.0	1433.9	1317.0	1475.7	1286.4	1499.4	1354.3	1537.6
	Min	966.4	1322.6	1115.9	1414.1	1026.9	1389.8	1200.2	1442.7
2	Mean	942.6	1266.9	966.8	1288.2	1106.8	1343.2	1251.7	1365.0
	SD	47.6	38.8	30.8	26.6	33.2	19.1	25.4	31.2
	CV	19.8	32.7	31.4	48.3	33.3	70.3	49.3	43.8
	Max	989.0	1304.0	1015.5	1340.2	1157.7	1372.7	1291.6	1413.0
	Min	841.8	1188.4	927.8	1255.2	1043.0	1296.8	1208.7	1318.6
3	Mean	1002.5	1388.3	1067.0	1417.8	1186.6	1433.7	1326.5	1461.0
	SD	37.4	30.5	54.0	16.6	46.1	32.3	19.0	52.5
	CV	26.8	45.6	19.8	85.5	25.7	44.3	69.8	27.8
	Max	1088.2	1446.9	1155.0	1444.1	1238.9	1486.2	1351.1	1562.7
	Min	947.6	1353.6	971.4	1379.7	1088.9	1373.7	1290.9	1386.9
Mean Diff.	x1-x2	70.4	115.0	245.8	159.4	96.3	78.6	52.9	131.7
	x1-x3	10.5	-6.4	145.6	29.7	16.5	-11.9	-21.9	35.8
	x2-x3	-59.9	-121.4	-100.2	-129.6	-79.7	-90.5	-74.8	-95.9
Error:	(N)	47.0	81.0	163.9	106.2	64.2	60.3	49.9	87.8
MSe:	(N)	38.0	68.3	123.6	84.7	51.5	49.2	38.4	68.1

speeds, with the other errors being substantially smaller. Furthermore, these large errors were from the running trials, which were used in only two of the statistical analyses.

C.11 Discussion

Many pilot experiments were conducted on the treadmill apparatus. From these tests some procedures for operating the treadmill were established. These procedures and other comments regarding the treadmill and possible future work are included in this discussion section.

C.11.1 Apparatus

Variability was noted in treadmill force values due to different pre-calibration procedures. To prevent such variation and prepare the transducers for measurement they were warmed up for at least two hours prior to testing. After this period the 484B02 conditioners were grounded. When initially turned on, the 484B02's output values were above the collection range; grounding lowered the signals to measurable units. Directly prior to any force measurement a subject walked or ran on the treadmill for several minutes. This generated any offset that would occur in any of the transducers, particularly for channels 5 & 6. With the treadmill vacant, a zero-reading was taken. In running, zero-values were calculated during flight phases. If a long period of time passed between zero-value measurements, drift occured. This drift was due to the type of transducers and the length of their time constants.

Two separate pilot experiments measured the natural frequency of the treadmill to be either ~275 Hz or ~100 Hz. While neither values should have caused a problem with the measurements conducted in this study, the large discrepancy between the two values is not easily explained. Perhaps two different sections of treadmill bed were tested in each experiment. More specifically, perhaps one set of impacts was directly over a frame member, while the other set was over an area without close support by the frame. The resulting difference in bed stiffness could have attenuated the force transmission of the initial impact, perhaps lowering the natural frequency measured by the six force transducers. Alternately, it is possible that different treadmill locations and set-up procedures caused the different results. Perhaps the vertical adjustment of the transducers yielded improper support of the bed in one case, resulting in a lower stiffness and diminished natural frequency.

For further experiments, especially those involving impacts of running, the treadmill should be mounted permanently on a solid floor, leveled with extreme care, and then tested for natural frequency over a range of locations on the bed. These values are expected to be at least 100 Hz.

C.11.2 Treadmill Versus Force Plate

A pilot study was performed to demonstrate the similarity in measurement ability between the criterion of the floor-mounted force plate and the force-measuring treadmill. Knowing that the treadmill was an accurate device, the majority of the variability was intrinsic to the subjects.

If better validation and correlation is desired, the force plate vs. treadmill experiment should be repeated with more subjects with a wide range of body weights and running abilities. A greater number of trials, particularly on the force plate, would also help control for intra-individual variability. This experiment would quantify the kinetic differences between treadmill locomotion and overground or in-laboratory locomotion. Much research has been performed on the kinematic differences between treadmill and force plate running (Nigg, De Boer, & Fisher, 1995), but no literature has addressed the kinetic differences.

C.11.3 Possible Improvements to the Treadmill Apparatus

The most important pieces of the treadmill measurement system were the transducers themselves. While the piezoelectric crystals used did not require balancing, they did drift over a trial. Kistler has produced a ground reaction force measuring treadmill which utilizes piezoelectric crystals. To allow approximately 30 min of data collection without more than 5 N drift they made a special amplifier which does not use time constants (R. Redd, personal communication, August 25, 1995). Such signal conditioning would be optimal if it could be made for the existing transducers. PCB Piezotronics should be offered this challenge.

If such improvements are not possible and hardware improvements are desired, the sensors may be changed to minimize drift. Strain-gage devices or load cells should be considered as the replacements. Such devices may be much larger than the existing transducers, causing design changes to the supports and motor mounts. In addition, the load cells would require a new set of signal conditioning.

When making such improvements, the ability and benefits of engineering the treadmill to measure all three (A-P, M-L and vertical) ground reaction forces should be considered. Both anterior-posterior and medial-lateral forces would be difficult because of the drive belt and the movement of the belt on which the subject walks. On the other hand, such a system should be feasible and may be able to provide qualitative, if not quantitative, information in these other two directions.

Software is the key to collecting and analyzing data from the treadmill. Any improvements which would facilitate data collection and decrease the amount of time necessary to analyze the data should be considered. All of the existing software for collection, filtering, conversion, and analysis could be integrated into one package. Further modifications could easily be made to minimize analysis time. These changes would depend on the research directions and variables to be analyzed.

If desired, the A/D board and software could be upgraded to allow for biofeedback to the subject. For gait analysis or training, such variables could include stance time, peak vertical ground reaction force, average force, or impulse.

C.11.4 Summary

The treadmill built for this experiment was valid for both static and dynamic measures. Extensive testing was performed to quantify the variability of the device. These results indicate that when the zero-value for the transducer was properly calculated, average errors could be expected to be approximately 30 N. The treadmill was also valid for dynamic loading. Several software steps forced baseline to zero force and compensated for drift. The overall result was a valid measuring device which was economical and rivals the commercially available force measuring treadmill. This device will remain at Oregon State University for the use of researchers in the Department of Exercise and Sport Science.