SAGITTAL PLANE CHARACTERIZATION OF NORMAL HUMAN ANKLE FUNCTION ACROSS A RANGE OF WALKING GAIT SPEEDS

by

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ABSTRACT

The function of the human ankle during the stance phase of walking is characterized in terms of simple mechanical elements that can reproduce the sagittal plane dynamics of a normal ankle. The dynamics of the ankle are taken from the analysis of the slow, normal, and fast gait of 10 healthy young subjects. Thus gait-speed-dependent changes in ankle function are evaluated.

Characterization of the ankle is divided into three phases of gait: controlled plantarflexion, controlled dorsiflexion, and powered plantarflexion. Ankle function during controlled plantarflexion is characterized by a linear, torsional spring. The work done by the spring increases with gait speed, and while gait speed is constant, the work is kept constant by modulating the stiffness from stride to stride. Ankle function during controlled dorsiflexion is characterized by a nonlinear, torsional spring that becomes more stiff as it rotates. Ankle function during powered plantarflexion is characterized by a torque actuator that assists the nonlinear, torsional spring in plantarflexing the foot. The combined work done by the actuator and spring increases with gait speed as does the portion of that work that is done by the torque actuator.

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To my parents
Martin and Bonnie Palmer
Who helped me get here in the first place

and

To my wife
n!na
Who gave me every reason to stay

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As this thesis nears completion, I feel that it is more than appropriate to take a moment, after a long sigh of relief, to acknowledge and thank those who have made this brief year and a half at MIT possible as well as enjoyable. First of all, thank you, Hugh, for placing your trust in me and putting up with me during times when my research was near the bottom of my priorities list. Thank you most of all for pushing me into the deep water and helping me to remember what can be gained by continually expanding the limits of my ability. Thank you, Dr. Woodie Flowers, for being quick to take time out of your day to help and encourage me. A sincere thank you goes to Dr. Pat Riley and the gang at the Spaulding Gait Laboratory. I doubt you gained anything from answering my endless questions and doing so many favors for me, but I'm glad you always made the time for me and answered my questions without so much as rolling your eyes. And then there's my Leg Lab family. Dan, thank you for your good natured humor and your willingness to help. Max, thanks for the chance to get familiar with the ice (probably a little too familiar), but thanks even more for always asking questions. Olaf, thanks for sharing your space heater. I wouldn't have made it through January without it. Joaquin, thanks for your contagious enthusiasm, for being quick to laugh, and for giving me reason to hope, by the success of your project, that I'm not just another MIT theorist. And finally, thank you, Greg, for befriending me, for being a mentor to me, for listening to me, and for teaching me new moves on the court. All for no good reason that I know of other than that your desk sits next to mine.

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CHAPTER 1
INTRODUCTION

1.1 SYSTEM CHARACTERIZATION IN BIOMECHANICS

Developing a relatively simple model of a complex biomechanical system in order to characterize the gross function of that system is common practice. The literature is replete with examples of how characterizing a biomechanical system with a simple model can be a tool for understanding the function of that system in a given situation. In legged locomotion, one of the most pervasive cases is that of characterizing the legs of vertebrate animals as linear, translational springs during bouncing gaits such as running, trotting, and hopping. (Blickhan, 1989; McMahon and Cheng, 1990; Farley et al., 1993; Farley and Morgenroth, 1999; Herr and McMahon, 2000; Herr and McMahon, 2001). Alexander (1990) characterized certain tendons and ligaments of running animals as springs and the foot pads as damped spring systems. The human leg has been characterized as a damped, linear, translational spring to understand its function during the stance phase of walking (Siegler et al., 1982; Pandy and Berme, 1988). As a final example, Gilchrist and Winter (1996) characterized the function of the human foot during stance using several independent damped spring systems.

1.2 MOTIVATION FOR CHARACTERIZING ANKLE FUNCTION

The goal of this research was to characterize the function of the human ankle in the sagittal plane during the stance phase of a normal walking gait. Achieving that goal promises several benefits. First, a simple model of the ankle could be a tool for artificial ankle design. Serving as the standard for ankle performance, a simple model of the ankle would provide a means by which the performance of any artificial ankle could be measured against that of a biological ankle. In this way, compromises between performance and other design considerations can be identified early in the design process thereby reducing the cost of developing an artificial ankle by reducing the number of prototypes required. A model of normal ankle function could also provide direction in correcting pathological gait since the model's structure would suggest the type and function of corrective orthoses. Additionally, the process of developing an ankle model shows how the model's parameters may change from stride to stride or from one gait speed to another. By observing when and to what extent the model's parameters change, insight can be gained about the control strategy that governs normal ankle function.

1.3 What Can Be Learned from Previous Work

Previous research on the energy flow into and out of the system representing the human ankle during the stance phase of a walking gait provided a starting point for this research. At normal walking speeds, the energy output at the ankle during a single gait cycle is approximately 3 times greater than the energy input (Inman et al., 1981; Winter, 1983). Winter (1983) also observed that the energy output monotonically increases with gait speed while the energy input remains relatively constant across gait speeds. Since the energy input to the ankle system is always less than the energy output, it was expected that passive elements alone could not characterize ankle function. Active elements that could supply energy would have to be included in the model.

The net energy flow into and out of the ankle system, or work done at the ankle, also indicated that modeling the ankle using only sagittal plane kinematics and kinetics was a good starting point for capturing ankle function. While 74% of the work done at the hip during a single gait cycle and 85%

of the work done at the knee are done in the sagittal plane, 93% of the work done at the ankle is done in the sagittal plane (Eng and Winter, 1995). The simple fact that the human ankle has the ability to move in the transverse and frontal planes suggests that kinematics and kinetics of the ankle in those planes are important. However, since the vast majority of ankle work is done in the sagittal plane, it would appear that the function of the ankle in the transverse and frontal planes is overshadowed by the sagittal plane function. Thus it seemed reasonable to assume that a model based solely on sagittal plane kinematics and kinetics would capture important features of ankle function during stance.

The dynamics of the human foot-ankle system have been characterized before in *stationary* conditions. It was shown that the dynamic response of the passive mechanics of the foot-ankle system to a position disturbance is characterized by a mass connected to ground through a linear, torsional spring in parallel with a linear, torsional damper where the spring stiffness as well as the damping coefficient are dependent on the angular position of the ankle (Weiss et al., 1986a). It was also found that the basic structure of the model used to describe the dynamic response of the foot-ankle system did not change from the passive to the active case. Adding dorsiflexor or plantarflexor torque simply changed the model's parameters, namely the spring stiffness and damping coefficient, at a given ankle position (Weiss et al., 1986b; Weiss et al., 1988). Thus it was shown that the human body has the ability to "tune" the dynamic response of the foot-ankle system by controlling the level of muscle activity.

Given these results from stationary conditions, it appeared logical to ask how the human body uses this ability to tune the dynamic response of the foot-ankle system during the conditions of normal walking. Or in other words, it seemed reasonable to ask what is the control strategy that the human body employs to tune the dynamic response of the foot-ankle system while walking. It was expected that knowledge about that control strategy would come by considering trends in the variation of the ankle model parameters.

Observing when and to what extent a parameter varied can give insight into the control strategy that dictated the variation. For example, if the ankle where to be characterized by a linear, damped spring system during walking as it was in stationary conditions, it might be observed that the damping ratio of the combined foot-ankle system decreased as gait speed increased. It might then be hypothesized that the control strategy was to decrease the damping ratio in order to get a quicker dynamic response from the foot-ankle system as gait speed increased.

In at least one instance, characterizing a complex biomechanical system with a simple model has led to insight about the control strategy governing the function of that system by observing when and to what extent the model's parameters varied. Consider again the example of characterizing the legs of vertebrate animals as linear, translational springs during bouncing gaits. By observing variation in the stiffness of the spring characterizing leg function, it was learned that humans and other mammals do not change leg spring stiffness as the speed of forward locomotion changes (He et al., 1991; Farley et al., 1993). It has also been learned that human runners do adjust leg stiffness to change stride frequency (Farley and Gonzalez, 1996) and to adapt to changes in the stiffness of the running surface (Ferris et al., 1998). Thus through this fairly simple model of leg function, insight has been gained about the human body's control strategy for adapting to situations that require a change in speed, stride frequency, or surface stiffness while running.

1.4 The Approach

The objective of this study was to develop a simple model in order to characterize the function of the human ankle in the sagittal plane during the stance phase of normal walking. The model structure and values of the model parameters were determined using gait data collected from healthy, young subjects. The model structure was taken to be the simplest combination of mechanical elements that could reproduce the kinematic and kinetic patterns observed in the ankle data. The data for each subject included trials from a range of gait speeds, making it possible to observe how the model parameters changed with gait speed. Thus the human ankle was not only characterized in terms of the mechanical elements that described its function but also in terms of how those mechanical elements were tuned in order to adapt to gait speed variations.

CHAPTER 2
METHODS

2.1 DATA COLLECTION AND PROCESSING

Subjects for this research were 10 healthy, young volunteers–6 female, 4 male (Table 2.1). The subjects had a mean age of 29 years (range 21 - 42 years), a mean body mass of 69.1 kg (range 50.0 - 107.1 kg), and a mean height of 1.71 m (range 1.58 - 1.83 m).

Table 2.1 The subjects' sex, age, and anthropometric data.

Subject Label	Sex	Age [yr]	Mass [kg]	Height [m]	Mean Leg Length[a] [m]
DEF	M	24	72.7	1.80	0.94
DMG	M	26	70.0	1.83	1.02
EAS	F	31	107.7	1.74	0.86
EEB	F	21	62.1	1.70	0.86
FJI	M	27	84.5	1.82	0.97
JLM	F	28	53.8	1.63	0.85
MJK	F	25	53.0	1.58	0.85
MKK	F	24	50.0	1.62	0.87
RAH	F	42	52.1	1.58	0.80
RWW	M	42	85.5	1.77	0.87
Mean		29	69.1	1.71	0.89
SD		7.0	17.91	0.093	0.612

[a] Measured between the ASIS and the lateral malleolus.

The data were collected as part of an independent study (Riley et al., 2001), and only a brief summary of the procedures and setup are given here. Subjects walked barefoot at their self-selected normal speed on a 10 m walkway. The subjects were then asked to walk "faster than you would normally walk". Finally they were asked to walk "slower than you would normally walk". Kinematic and kinetic data were acquired for both the left and right lower limbs using a six-camera VICON 512 system (Oxford Metrics, Oxford, UK) and two AMTI forceplates (AMTI, Newton, MA). The data were processed at 120 Hz with VICON BodyBuilder (Oxford Metrics, Oxford, UK) using the standard model of the lower limbs included with the software (Ramakrishnan et al., 1987; Kadaba et al., 1990; Davis et al., 1991). The data derived using the standard BodyBuilder model, including ankle angular position and ankle torque, were then analyzed using MATLAB (MathWorks, Natick, MA).

The following sign conventions were used: positive ankle position for dorsiflexion and positive ankle torque for dorsiflexor torque. Zero ankle position was arbitrarily assigned to be the position at which the segment representing the foot was perpendicular to the segment representing the shank.

Ankle angular velocity was calculated by first filtering the ankle position data and then numerically differentiating. The position was filtered using a zero-lag, fourth-order, low-pass Butterworth filter with a cutoff frequency of 6 Hz (Winter, 1990). Numerical differentiation of the filtered position to get the velocity was done using high-accuracy, divided-difference formulas (Chapra and Canale, 1998).

For the purposes of this study, power at the ankle was calculated by taking the product of the magnitude of ankle torque and the magnitude of ankle velocity *in the sagittal plane*. Contributions to power at the ankle by any out-of-plane components of either torque or velocity were not accounted for. The work done at the ankle during a certain period of time was given by the area under the power versus time curve for that period.

The gait speed for each trial was taken to be the slope of the line given by

$$x = vt + x_0 \tag{2.1}$$

where v is the slope, t is time, x is the absolute position of a point fixed in the pelvis along an axis parallel to the direction of forward motion, and x_0 is the initial position of that point. The slope was calculated from the data by simple linear regression.

2.2 CREATING AND COMPARING GAIT SPEED GROUPS

Since it was desired to study natural locomotion, no quantitative constraint was placed on gait speed. After being given the qualitative instruction to walk slow, normal, or fast, the subject freely selected the gait speed. The inevitable result was that the actual gait speed of a particular trial was not necessarily the same as the gait speed for the other trials when the subject had been instructed to walk at that same self-selected speed (e.g. not all "fast" trials were actually fast). Therefore it was necessary to establish a procedure by which trials would be grouped, or not grouped, into one of the three self-selected speed categories.

The procedure for grouping trials into slow, normal, and fast categories according to the actual gait speed was designed to ensure that each self-selected speed group was homogeneous and that the three self-selected speed groups for each subject were distinct. First, each trial from a single subject was grouped according to the instruction that the subject had received prior to that trial making a slow, normal, and fast group for each subject. The calculated gait speed of each trial was then compared to the gait speeds calculated for the other trials within the same group, and by this comparison trials with outlying gait speeds were identified and eliminated from the group (Grubbs, 1969). After outliers had been removed, the mean gait speed for the group was calculated. Finally, trials with gait speeds greater than or less than 0.1 m/s of the group mean were also removed from the group. As a result of this procedure, there was always separation between the gait speed ranges of the slow, normal, and fast groups for each subject.

Once the trials for each subject had been placed into three homogeneous and distinct self-selected speed groups, the mean of certain gait parameters as well as parameters characterizing ankle function were calculated for each group. The mean of a parameter for each self-selected speed group was determined by first calculating that parameter for each trial individually and then by taking the mean of that parameter across each trial in the group (i.e. the calculated values of a parameter for all trials in a group were ensemble averaged to obtain the group mean). Trials from the left side were pooled together with trials from the right side in each self-selected group. Thus the mean of a parameter represented the mean of the left and right side.

Significant changes in the mean of a parameter from one self-selected speed group to another were tested for using one-way analysis of variance requiring $p < 0.05$ for significance. The results of these significance tests were the basis for conclusions made about trends in the variation of any given parameter with gait speed.

In order to maximize the probability of observing gait speed dependent differences in ankle function, a procedure was developed to identify subjects who had a substantial difference, relative to a pool of 18 subjects, between the mean gait speed at the self-selected slow speed and the mean gait speed at the self-selected fast speed. To make a "fair" comparison across subjects, it was decided to normalize the mean gait speed at each self-selected speed according to the equation

$$v_N = \frac{\bar{v}}{(gL_{leg})^{1/2}} \tag{2.2}$$

where v_N is the normalized gait speed, $\bar{v}$ is the mean gait speed, g is the acceleration due to gravity, and L_{leg} is the subject's leg length. The change in the normalized mean gait speed from slow to fast was then used as the metric to identify subjects with a substantial difference between the mean gait speed at the slow speed and the mean gait speed at the fast speed. The 10 subjects selected for this study were chosen because the change in the normalized mean gait speed from the slow to the fast speed for these subjects (ranging from 0.27 to 0.37) was greater than that for the other 8 subjects in the pool.

Depending on the interpretation of the quantity $(gL_{leg})^{1/2}$, looking for subjects with a certain amount of change in the normalized mean gait speed from slow to fast can be understood in two different ways. The quantity $(gL_{leg})^{1/2}$ can either be seen as the gait speed at which the Froude number is equal to unity (Alexander, 1989), or as a theoretical maximum walking speed predicted by an inverted pendulum model of legged locomotion (Alexander, 1990). Thus comparing changes in mean gait speeds normalized by $(gL_{leg})^{1/2}$ across subjects can either be seen as looking at how much the gait speeds vary relative to gait speeds at which the Froude numbers for each subject are equal[1] or as looking at how much the gait speeds vary relative to a theoretical maximum gait speed for each subject. Either way, the net effect is to establish a more objective standard by which changes in gait speed can be compared across subjects.

2.3 The Gait Cycle and Three Periods of Stance

The gait cycle for each trial was defined as beginning with heel strike (HS) on a forceplate and as ending at the next HS of the same foot. The instant of HS on a forceplate was defined as being coincident with the first of five consecutive samples from the measured vertical ground reaction force data where the absolute difference between each sample and the subsequent sample was greater than 1 N. Since the subsequent HS of the same foot did not fall on a forceplate, the duration of the gait cycle was calculated from the ankle position data. Local minimums in the ankle angular position

[1] Having equal Froude numbers is one of the requirements for dynamic similarity in legged locomotion (Alexander, 1989).

versus time curve occur at foot flat (FF) and at toe off (TO) and can be reliably located. The duration of the gait cycle was taken to be the time elapsed between these local minimums (Figure 2.1), or in other words the duration of the gait cycle was taken to be the time elapsed between FF and the subsequent FF of the same foot or as the time elapsed between TO and the subsequent TO of the same foot. The instant when the gait cycle ended was then estimated by adding the duration of the gait cycle to the time when the initial HS occurred on the forceplate.

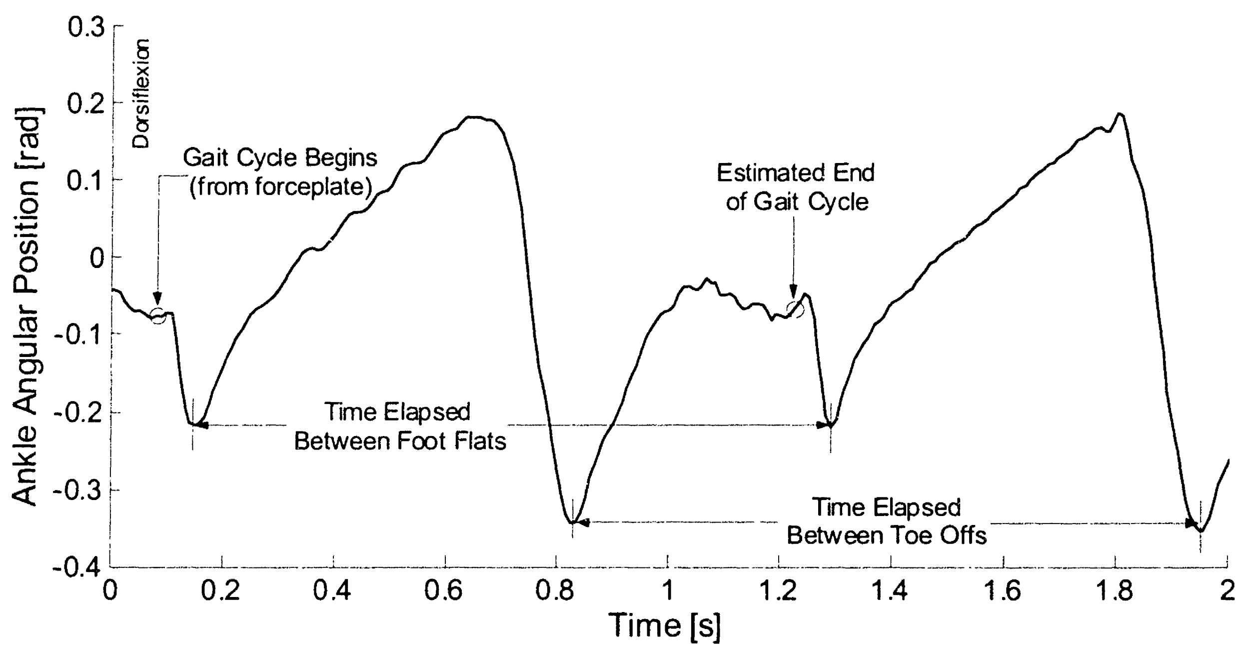

Figure 2.1 Example of using the ankle angular position to determine when the gait cycle ends for a single trial. The time when the gait cycle ends is estimated by adding the time elapsed between foot flat and the subsequent foot flat to the time when the gait cycle begins. The results are similar if the time elapsed between toe off and the subsequent toe off is used.

Characterization of the ankle was done in three parts, each part considering a different period of the stance phase of the gait cycle (Figure 2.2). The first period of stance was controlled plantarflexion (CP). The beginning of CP was defined as the first instant after initial HS when the ankle began to plantarflex. Using this definition for when CP began, the beginning of CP coincided with HS in 40% of all the trials analyzed, but for the rest of the trials the ankle dorsiflexed a small

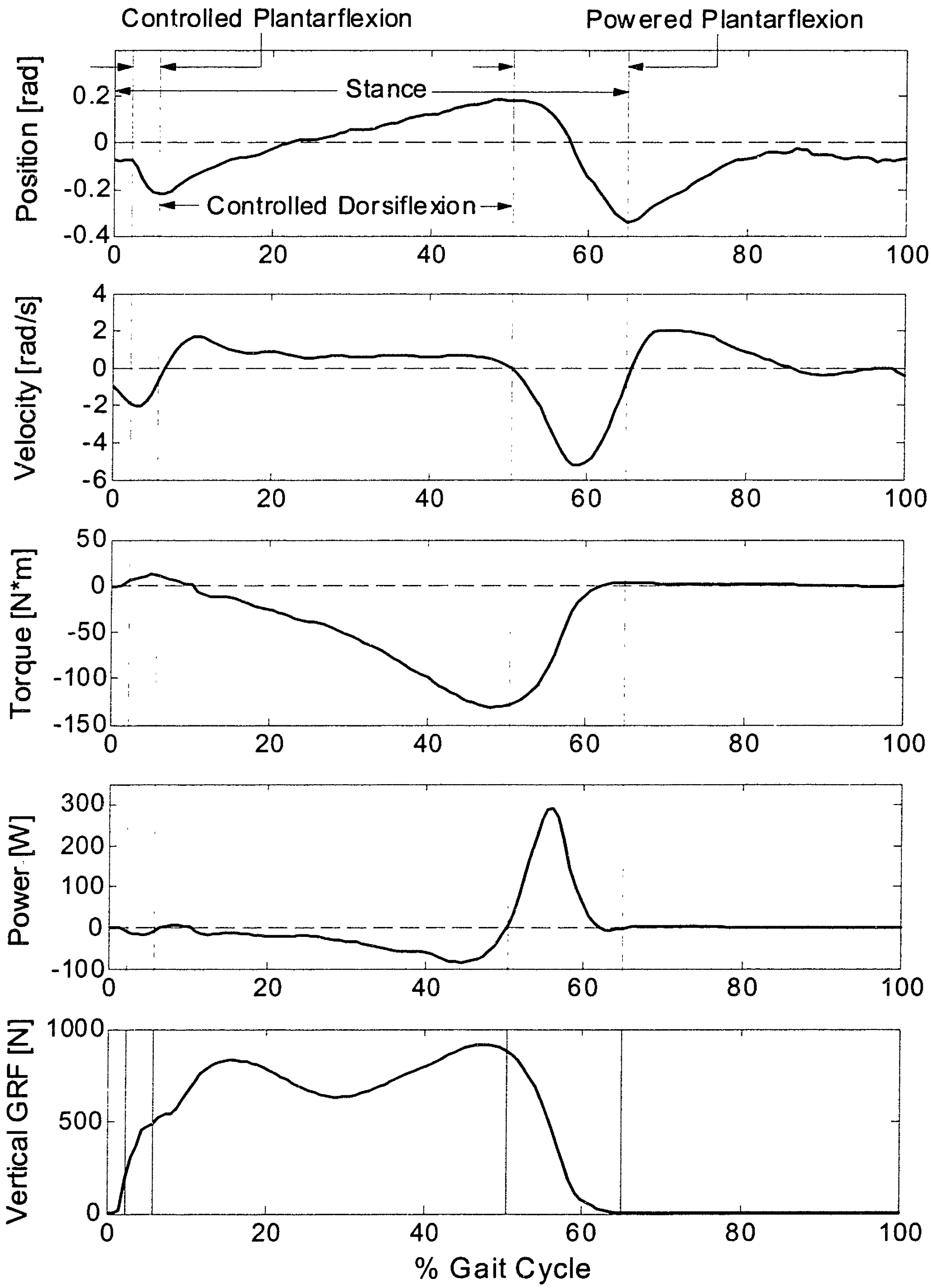

Figure 2.2 The three periods of the stance phase of a walking gait for which ankle function was characterized. The beginning and ending of each phase was determined from the kinematics and kinetics of the ankle along with the vertical ground reaction force (GRF). Data shown are from a single trial of subject FJI walking at the self-selected normal gait speed.

amount (always less than 0.025 rad) after HS before beginning to plantarflex resulting in a time delay between HS and when CP began. The time delay between HS and the beginning of CP was never greater than 33 ms and occurred more often for slow trials than for fast trials.[2] The end of CP was defined as the instant of FF. FF was identified as the instant when a local minimum was reached in the ankle angular position versus time curve.

Because of a tendency for the noise-to-signal ratio of the ankle torque data to be large immediately after HS, a criterion was established for consistently rejecting data when the noise-to-signal ratio was unacceptable. The large noise-to-signal ratio immediately after HS was likely caused by "ringing" of the deck in which the forceplates were mounted occurring at a time when the calculated ankle torque was near zero. This problem was overcome by analyzing the data during CP only after the ankle power became negative and remained negative until FF. Rejecting data this way gave rise to a time delay between the point when CP actually began and the first data point used for the analysis in 64% of all the trials. This time delay was rarely (less than 3% of all trials) greater than 17 ms and was never greater than 42 ms.

The second period of stance considered was controlled dorsiflexion (CD), and powered plantarflexion (PP) was the third and final period of stance for which the ankle was characterized. CD was defined as beginning when CP ended and ending at the instant when the ankle power became positive and remained positive until TO. The beginning of PP was defined as when CD ended, and

[2] The trend in the time delay between HS and the beginning of CP suggests two possible explanations for the cause of this time delay. HS occurs at the end of the swing phase of gait during which the ankle is typically dorsiflexing. It is likely that the ankle continued to dorsiflex after HS while the heel pad was being compressed, and CP began only when the heel pad bottomed out. The dorsiflexion after HS might also have been the result of the subject anticipating the instant of HS. The dorsiflexors are active during swing and CP, but compared to the swing phase, it's likely that the dorsiflexors are required to generate greater torque about the ankle during CP as the heel is loaded. The ankle would tend to dorsiflex after HS if the dorsiflexors began to generate that greater amount of torque before the heel is completely loaded.

the end of PP was defined as the instant of TO. The instant of TO was identified in a fashion similar to how the instant of initial HS on the forceplate was identified.

2.4 CHARACTERIZING ANKLE FUNCTION

The method of gait analysis used for this research employs a black box to model the ankle kinetically during the stance phase of gait. Kinematic data were collected by a motion capture system while kinetic data were collected via forceplates. These kinematic and kinetic data were then fed into a model of the lower limbs (Ramakrishnan et al., 1987; Kadaba et al., 1990; Davis et al., 1991), and through the wizardry of inverse dynamics, ankle joint position and ankle joint torque were calculated. In this model of the lower limbs, the ankle is represented kinematically as a spherical joint with three degrees of freedom. Ankle kinetics are simply represented by the ankle joint torque that satisfies the equations of motion that govern the model. The model offers no explanation of the mechanisms that produce the ankle torque.

The aim of this research was to characterize ankle function by revealing the contents of the black box that represented the ankle during the gait analysis. Ankle angular position and angular velocity were taken as the inputs to the black box, and ankle torque was considered the output. The problem was then to derive the simplest combination of mechanical elements that could produce the observed position/torque and velocity/torque relationships. It is emphasized that the goal was not to identify the roles of individual muscles, tendons, bone structures, etc. in the dynamics of the ankle joint but to characterize the function of the overall system. Thus the contents of the black box represent the sum effect of the lengthening and shortening of muscles, tendons, and ligaments as well as deformation of the foot and any other mechanisms of generating torque about the ankle joint.

The first step in identifying the contents of the black box model of the ankle was to define the set of mechanical elements to be considered and to establish some rules of thumb for selecting the right element at the right time. The set of mechanical elements consisted of torsional springs, torsional dampers, and torque actuators. This set can be subdivided by classifying springs and dampers as passive elements, leaving torque actuators as the only active elements. Since it would be inefficient to do negative work with an active element, only passive elements were considered to characterize ankle function while the power at the ankle was negative. Active elements were considered when the power at the ankle was positive and the amount of positive work done was greater than any energy that might have been stored in the system representing the ankle.

When considering passive elements to characterize ankle function, it was desirable to distinguish between springlike behavior and damper-like behavior. This was possible during periods of stance when ankle angular position and ankle angular velocity did not co-vary (periods when position was increasing and velocity was either decreasing or constant and vice versa). During these periods, springlike behavior was distinguished from damper-like behavior by considering the amount of time between the occurrence of the maximum in the magnitude of ankle position and the occurrence of the maximum in the magnitude of ankle torque. This amount of time was called the "phase difference" between the maximum position and maximum torque. The phase difference was taken to be positive if the maximum torque occurred before the maximum position. A small phase difference would be evidence that ankle function was dominated by springlike behavior during that period of stance.

It should be noted that the phase difference was defined as the amount of time between the occurrence of the maximum in the magnitude of ankle position and the occurrence of the maximum in the magnitude of ankle torque *during the period of stance being considered.* Often the magnitude versus time curves for either position or torque would reach a local maximum *after* the period of stance being considered had come to an end. In these cases, the maximum was taken to be the endpoint of the magnitude curve for that period, instead of the point when the magnitude reached the local maximum, since the local maximum occurred after the period had concluded.

Ignoring local maximums in the magnitude curves for either position or torque that occurred after the period of stance had concluded did not bias the phase difference in a way that would lead to improper conclusions in distinguishing springlike behavior from damper-like behavior. During the periods of stance for which passive elements were considered, the magnitude of the ankle position was always approaching a local maximum. This means that at the same time, the magnitude of the ankle velocity was approaching zero, or in other words, that any local maximum in the ankle velocity magnitude had to occur before the end of the period. Thus if the ankle were characterized by velocity-dependent, damper-like behavior the effect would be that the maximum in the magnitude of ankle torque would occur sometime before the period came to an end. Consequently, ignoring local maximums in the ankle torque magnitude that occurred after the period ended did not ignore the effects of damper-like behavior.[3]

The category of passive elements was subdivided further by differentiating between linear and nonlinear elements. For example, during periods when the ankle was characterized as behaving like a spring, the expected relationship between the ankle torque and ankle angular position for a linear spring would be

$$\tau = k\theta + \tau_0 \tag{2.3}$$

where τ is ankle torque and θ is ankle position. Assuming this model, the spring could be characterized by the two parameters k and τ_0, the spring stiffness and the torque when the position is

[3] Ignoring local maximums in the ankle torque magnitude that occurred after the period ended and soon after the ankle position magnitude reached a local maximum most likely ignored the effects of active elements.

zero. Values for these parameters were determined by simple linear regression,[4] and the adequacy of the simple linear model was assessed. If the simple linear model given by Equation 2.3 was determined to be inadequate in describing the measured position/torque relationship, the spring was considered to be nonlinear. Testing for a linear damper would be similar for periods when the ankle was characterized as behaving like a damper whereas multiple linear regression would be required for periods when ankle behavior was both springlike and damper-like.

Assessment of the adequacy of a simple linear model was based on significance of the model, the coefficient of determination, and scatter plots of the residual versus the independent variable. A simple linear model was considered significant when $p < 0.05$. If the model was significant, the coefficient of determination was used as the metric to assess the model's goodness of fit. Finally, a scatter plot of the residual versus the independent variable was generated in order to assess whether or not the assumptions of a simple linear model held. The assumptions made by a simple linear model that involve the residual are that the residuals are mutually independent as well as normally distributed with mean zero and constant variance (Hogg and Ledolter, 1992). Consequently, in a scatter plot of the residual versus the independent variable, one would expect to see the residuals distributed in a band of constant width centered about zero. Any other patterns in the residual would be an indication that the assumptions of a simple linear model were violated and that the simple linear model is not adequate.

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[4] One of the assumptions of simple linear regression is that the independent variable, ankle position, is measured without error. Performing simple linear regression of the ankle torque on the ankle position is certainly a violation of this assumption since measurement error is ever present in the ankle position data. However, since the measurement error is uncorrelated with the ankle position, the simple linear regression model may still be applied (Sokal and Rohlf, 1995).

CHAPTER 3
RESULTS

3.1 GAIT PARAMETERS

Time and distance gait parameters as well as kinematic and kinetic gait parameters for the ankle were calculated for each subject at the self-selected slow, normal, and fast speeds (Table 3.1 and Table 3.2).

3.2 CONTROLLED PLANTARFLEXION

The power at the ankle during controlled plantarflexion (CP) was always negative (Figure 2.2). Thus only passive elements were considered to characterize ankle function. Both damper-like and springlike behavior were considered by examining relationships between ankle torque and ankle angular velocity as well as relationships between ankle torque and ankle angular position.

Scatter plots of the ankle torque versus ankle angular velocity (Figure 3.1) did not suggest a cause and effect relationship between torque and velocity. Although ankle torque increased with increasing velocity during the first half of CP, during the latter half of CP the torque continues to increase at the same rate while velocity decreases.

Scatter plots of the ankle torque versus ankle angular position (Figure 3.2) did suggest a cause and effect relationship between torque and position. Simple linear regression of the ankle torque on the ankle angular position during CP was significant ($p < 0.05$) for 87.6% of the trials for all subjects and all self-selected speeds (Table 3.3 and Figure 3.2). In addition, the mean coefficient of determination (r^2 in Table 3.3) for each self-selected speed ranged from 0.85 to 0.98. Scatter plots of the residual versus the ankle position (Figure 3.2) showed that the residual appeared to be distributed in a band of constant width centered about zero. Because presenting scatter plots of the residual versus ankle position for every trial would be impractical, the mean number of times the residual changed signs is given (Table 3.3) as an indicator of to the extent to which the residual was distributed about zero.

Scatter plots of the residual versus ankle angular velocity (Figure 3.3) were generated as an additional check for relationships between ankle torque and ankle angular velocity. In these scatter plots, the magnitude of the residuals did not appear to be correlated with the magnitude of the ankle velocity. Furthermore, the residuals were both negative and positive at points where the magnitude of the ankle velocity was approximately equal.

The mean phase difference between the maximum magnitude of ankle torque and the maximum magnitude of ankle position during CP (Phase in Table 3.3) as a percentage of the duration of CP (T_{CP} in Table 3.3) ranged between 0% and 21%. The mean phase difference was greater than 15% for 5 out of the 30 self-selected speed groups for all subjects. CP was defined as ending at the instant of foot flat (FF), and FF was identified by a local minimum in the ankle position. Thus the maximum in the ankle position's magnitude during CP always occurred at FF. Therefore, the phase difference could only be positive or zero since the maximum in the ankle torque's magnitude could only occur before or simultaneously with FF.

Table 3.1 The subjects' mean (SD) time and distance gait parameters.

Subject Label	Self-selected Speed	Number of Trials	Gait Speed [m/s]	Stride Length [m]	Duration of Gait Cycle [ms]	Duration of Stance [% Gait Cycle]
	Slow	9	0.99 (0.060)	1.312 (0.049)	1334 (64)	63 (1.1)
DEF	Normal	9	1.49 (0.052)	1.494 (0.032)	1018 (18)	63 (0.4)
	Fast	8	1.91 (0.027)	1.690 (0.071)	895 (31)	62 (1.4)
	Slow	7	0.77 (0.019)	1.090 (0.029)	1412 (23)	65 (1.4)
DMG	Normal	8	1.30 (0.060)	1.437 (0.045)	1107 (20)	62 (0.7)
	Fast	7	1.69 (0.038)	1.689 (0.034)	999 (17)	61 (1.0)
	Slow	7	0.85 (0.063)	1.186 (0.029)	1426 (117)	67 (2.4)
EAS	Normal	8	1.33 (0.034)	1.380 (0.016)	1044 (26)	64 (1.4)
	Fast	5	1.89 (0.048)	1.387 (0.306)	810 (107)	63 (9.0)
	Slow	6	0.85 (0.042)	1.100 (0.018)	1283 (47)	68 (1.0)
EEB	Normal	3	1.15 (0.061)	1.241 (0.030)	1067 (22)	65 (1.2)
	Fast	4	1.79 (0.080)	1.511 (0.022)	852 (29)	61 (1.8)
	Slow	6	0.87 (0.022)	1.163 (0.044)	1351 (50)	67 (1.4)
FJI	Normal	9	1.21 (0.016)	1.354 (0.014)	1119 (20)	65 (1.4)
	Fast	5	1.69 (0.063)	1.607 (0.035)	950 (19)	62 (0.6)
	Slow	7	0.94 (0.060)	1.244 (0.035)	1321 (61)	63 (1.4)
JLM	Normal	8	1.46 (0.058)	1.393 (0.019)	959 (32)	61 (0.8)
	Fast	6	1.88 (0.007)	1.509 (0.020)	804 (19)	59 (1.3)
	Slow	6	0.84 (0.049)	1.159 (0.048)	1371 (43)	67 (2.0)
MJK	Normal	9	1.23 (0.024)	1.312 (0.027)	1076 (25)	64 (1.3)
	Fast	7	1.85 (0.044)	1.550 (0.031)	840 (14)	62 (1.7)
	Slow	6	0.88 (0.052)	1.144 (0.024)	1328 (45)	66 (1.1)
MKK	Normal	6	1.38 (0.052)	1.409 (0.040)	1014 (14)	62 (0.7)
	Fast	6	1.84 (0.040)	1.577 (0.016)	856 (13)	61 (1.5)
	Slow	6	0.89 (0.027)	1.077 (0.026)	1218 (29)	66 (1.9)
RAH	Normal	5	1.28 (0.025)	1.260 (0.009)	988 (40)	62 (1.9)
	Fast	5	1.70 (0.061)	1.448 (0.037)	863 (22)	60 (1.3)
	Slow	8	1.06 (0.043)	1.266 (0.025)	1195 (28)	66 (0.9)
RWW	Normal	8	1.47 (0.047)	1.468 (0.073)	1021 (13)	64 (1.8)
	Fast	7	1.90 (0.071)	1.635 (0.285)	892 (86)	62 (5.7)

Table 3.2 The subjects' mean (SD) kinematic and kinetic gait parameters for the ankle. Zero ankle position was arbitrarily assigned to be the position at which the segment representing the foot was perpendicular to the segment representing the shank.

Subject Label	Self-selected Speed	Number of Trials	Gait Speed [m/s]	Max. Ankle Dorsiflexion [rad]	Max. Ankle Plantarflexion [rad]	Max. Ankle Dorsi. Torque [N*m]	Max. Ankle Plant. Torque [N*m]
	Slow	9	0.99 (0.060)	0.14 (0.029)	-0.24 (0.060)	10 (1.7)	-104 (4.8)
DEF	Normal	9	1.49 (0.052)	0.11 (0.013)	-0.42 (0.080)	11 (2.2)	-115 (3.0)
	Fast	8	1.91 (0.027)	0.14 (0.020)	-0.39 (0.105)	15 (3.5)	-124 (3.8)
	Slow	7	0.77 (0.019)	0.18 (0.039)	-0.30 (0.068)	6 (1.1)	-101 (4.9)
DMG	Normal	8	1.30 (0.060)	0.12 (0.040)	-0.43 (0.081)	9 (2.4)	-111 (6.5)
	Fast	7	1.69 (0.038)	0.10 (0.046)	-0.45 (0.046)	11 (2.1)	-127 (4.3)
	Slow	7	0.85 (0.063)	0.16 (0.024)	-0.45 (0.048)	8 (1.6)	-144 (5.3)
EAS	Normal	8	1.33 (0.034)	0.18 (0.018)	-0.50 (0.056)	15 (1.6)	-169 (9.6)
	Fast	5	1.89 (0.048)	0.14 (0.018)	-0.55 (0.058)	20 (1.3)	-172 (5.5)
	Slow	6	0.85 (0.042)	0.17 (0.014)	-0.55 (0.054)	5 (1.0)	-93 (5.2)
EEB	Normal	3	1.15 (0.061)	0.16 (0.008)	-0.60 (0.036)	8 (1.5)	-97 (3.6)
	Fast	4	1.79 (0.080)	0.08 (0.022)	-0.66 (0.032)	11 (1.2)	-97 (14.4)
	Slow	6	0.87 (0.022)	0.19 (0.010)	-0.33 (0.037)	11 (2.3)	-120 (8.8)
FJI	Normal	9	1.21 (0.016)	0.18 (0.008)	-0.37 (0.032)	16 (2.3)	-133 (3.1)
	Fast	5	1.69 (0.063)	0.15 (0.025)	-0.37 (0.021)	27 (3.0)	-144 (9.0)
	Slow	7	0.94 (0.060)	0.17 (0.044)	-0.38 (0.058)	5 (0.3)	-83 (3.5)
JLM	Normal	8	1.46 (0.058)	0.13 (0.041)	-0.40 (0.039)	8 (1.4)	-90 (3.2)
	Fast	6	1.88 (0.007)	0.07 (0.025)	-0.46 (0.035)	10 (1.8)	-93 (3.7)
	Slow	6	0.84 (0.049)	0.23 (0.016)	-0.44 (0.109)	5 (1.5)	-77 (2.8)
MJK	Normal	9	1.23 (0.024)	0.21 (0.013)	-0.44 (0.043)	6 (1.3)	-89 (4.0)
	Fast	7	1.85 (0.044)	0.15 (0.019)	-0.43 (0.027)	11 (2.3)	-90 (5.2)
	Slow	6	0.88 (0.052)	0.16 (0.029)	-0.37 (0.053)	5 (1.4)	-75 (6.0)
MKK	Normal	6	1.38 (0.052)	0.18 (0.029)	-0.44 (0.046)	9 (2.3)	-89 (4.7)
	Fast	6	1.84 (0.040)	0.20 (0.085)	-0.43 (0.020)	8 (2.8)	-103 (13.1)
	Slow	6	0.89 (0.027)	0.08 (0.023)	-0.37 (0.034)	7 (1.5)	-67 (5.4)
RAH	Normal	5	1.28 (0.025)	0.06 (0.016)	-0.38 (0.045)	9 (1.5)	-72 (2.9)
	Fast	5	1.70 (0.061)	0.14 (0.077)	-0.43 (0.023)	14 (2.7)	-77 (8.1)
	Slow	8	1.06 (0.043)	0.12 (0.019)	-0.36 (0.056)	9 (1.9)	-130 (4.2)
RWW	Normal	8	1.47 (0.047)	0.11 (0.029)	-0.38 (0.044)	13 (2.8)	-148 (9.4)
	Fast	7	1.90 (0.071)	0.11 (0.022)	-0.39 (0.019)	19 (2.6)	-160 (9.8)

Table 3.3 Results of simple linear regression of ankle torque on ankle angular position during controlled plantarflexion ($p < 0.05$ for all trials except where noted). T_{CP} is the duration of controlled plantarflexion. Phase is the time difference between the maximum ankle position and the maximum ankle torque. Values given are the mean (SD) for each speed group.

Subject Label	Self-selected Speed	Number of Trials	Gait Speed [m/s]	# of Points Used for Regression	r^2	# of Sign Changes in Residual	T_{CP} [ms]	Phase [% T_{CP}]
	Slow	8	0.99 (0.060)	7 (1.4)	0.89 (0.062)[a]	3.1 (0.64)	57 (19.1)	13 (5.1)
DEF	Normal	7	1.49 (0.058)	5 (0.8)	0.87 (0.065)	2.4 (0.53)	50 (6.8)	5 (8.1)
	Fast	4	1.90 (0.025)	4 (0.5)	0.89 (0.037)	2.3 (0.50)	50 (0.0)	4 (8.3)
	Slow	6	0.76 (0.021)	6 (1.4)	0.92 (0.106)	2.5 (0.55)	53 (4.3)	11 (12.9)
DMG	Normal	3	1.31 (0.036)	5 (1.0)	0.96 (0.019)	2.7 (0.58)	44 (4.8)	13 (10.8)
	Fast	6	1.69 (0.039)	5 (0.5)	0.94 (0.045)	2.8 (0.98)	47 (6.8)	12 (9.1)
	Slow	7	0.85 (0.063)	9 (1.1)	0.89 (0.115)[a]	3.1 (0.90)	63 (9.4)	17 (10.0)
EAS	Normal	8	1.33 (0.034)	6 (0.5)	0.92 (0.037)[a]	2.5 (0.93)	51 (11.3)	16 (0.0)
	Fast	2	1.91 (0.028)	5 (0.0)	0.84 (0.041)	2.0 (0.00)	42 (0.0)	20 (0.0)
	Slow	5	0.85 (0.048)	6 (1.0)	0.96 (0.016)[a]	2.4 (0.55)	62 (12.6)	5 (7.4)
EEB	Normal	3	1.15 (0.061)	5 (1.2)	0.95 (0.049)	2.3 (0.58)	50 (8.3)	17 (0.0)
	Fast	3	1.81 (0.073)	5 (0.6)	0.90 (0.054)	3.0 (1.00)	39 (4.8)	0 (0.0)
	Slow	6	0.87 (0.022)	8 (0.9)	0.92 (0.038)[b]	3.7 (0.52)	61 (11.4)	11 (10.3)
FJI	Normal	8	1.21 (0.016)	6 (0.5)	0.87 (0.072)	3.1 (0.64)	49 (8.3)	13 (7.9)
	Fast	5	1.69 (0.063)	5 (0.4)	0.89 (0.040)	2.6 (0.55)	42 (8.3)	4 (8.9)
	Slow	7	0.94 (0.060)	7 (1.3)	0.90 (0.070)	2.0 (0.00)	51 (8.9)	14 (14.6)
JLM	Normal	7	1.46 (0.064)	5 (0.8)	0.98 (0.022)[b]	2.7 (0.49)	51 (5.8)	2 (6.2)
	Fast	6	1.88 (0.007)	5 (0.5)	0.93 (0.072)	2.2 (0.41)	44 (4.3)	3 (7.7)
	Slow	6	0.84 (0.049)	8 (1.0)	0.87 (0.126)	3.5 (0.55)	74 (9.7)	9 (8.5)
MJK	Normal	9	1.23 (0.024)	6 (0.4)	0.92 (0.071)	2.6 (0.73)	58 (5.9)	10 (7.1)
	Fast	3	1.82 (0.026)	4 (0.6)	0.90 (0.096)	2.3 (0.58)	47 (4.8)	12 (10.2)
	Slow	6	0.88 (0.052)	10 (1.8)	0.85 (0.208)	2.2 (0.41)	85 (16.2)	21 (19.1)
MKK	Normal	6	1.38 (0.052)	6 (0.8)	0.91 (0.014)	2.0 (0.00)	58 (7.5)	14 (9.0)
	Fast	6	1.84 (0.040)	6 (0.5)	0.93 (0.051)	2.5 (0.55)	57 (3.4)	7 (8.0)
	Slow	6	0.89 (0.027)	7 (0.6)	0.96 (0.022)[b]	3.0 (0.89)	57 (9.7)	10 (7.6)
RAH	Normal	5	1.28 (0.025)	6 (0.4)	0.96 (0.054)[a]	2.8 (0.84)	52 (9.1)	3 (7.2)
	Fast	5	1.70 (0.061)	5 (0.0)	0.97 (0.028)[a]	2.4 (0.55)	42 (8.3)	0 (0.0)
	Slow	8	1.06 (0.043)	7 (0.9)	0.95 (0.067)[a]	2.4 (0.74)	57 (13.7)	11 (6.7)
RWW	Normal	8	1.47 (0.047)	6 (0.8)	0.94 (0.041)[a]	2.5 (0.76)	48 (7.4)	15 (6.1)
	Fast	7	1.90 (0.071)	5 (0.7)	0.92 (0.070)	2.3 (0.49)	43 (5.8)	11 (10.4)

[a] $p < 0.01$ for all trials.

[b] $p < 0.001$ for all trials.

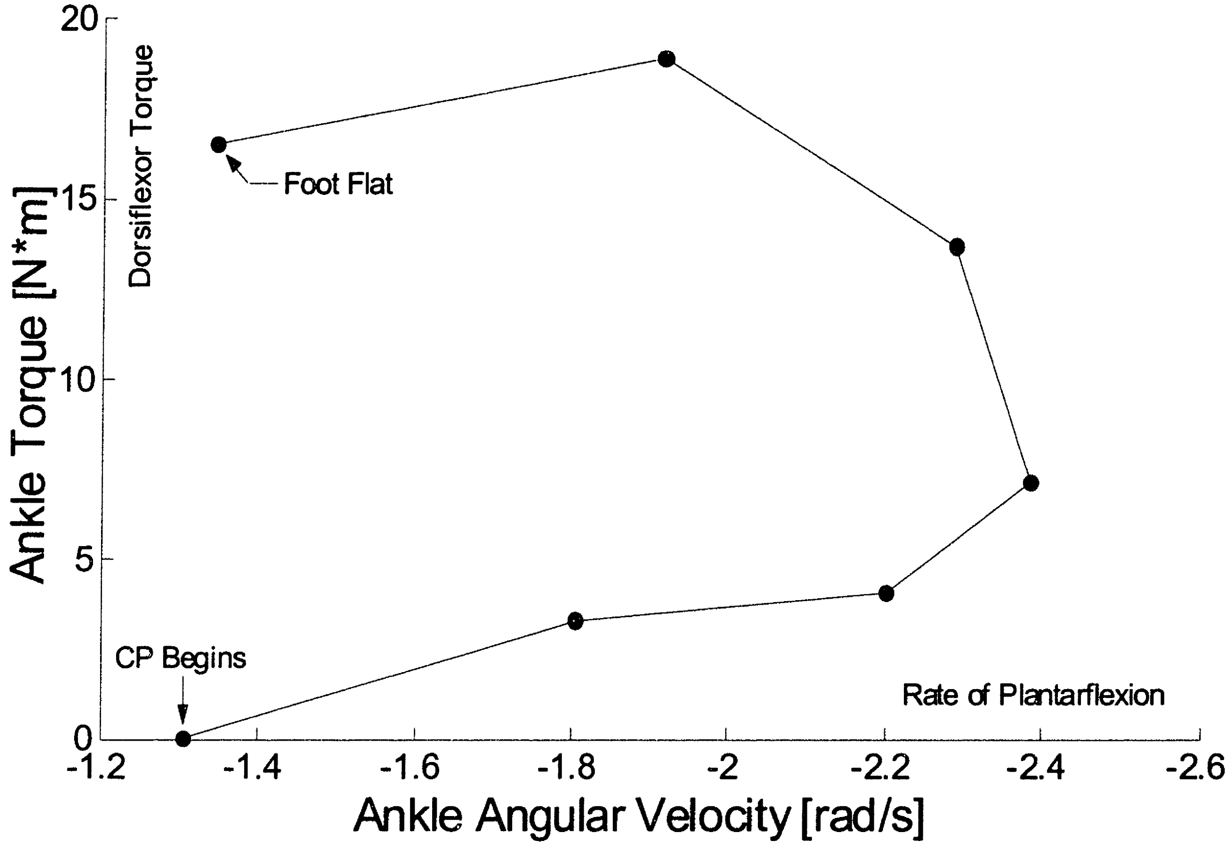

Figure 3.1 Scatter plot of ankle torque versus ankle angular velocity during controlled plantarflexion (CP). Data shown are from a single trial of subject FJI walking at the self-selected normal gait speed and are representative of the subject's trials. Data points are equally spaced in time.

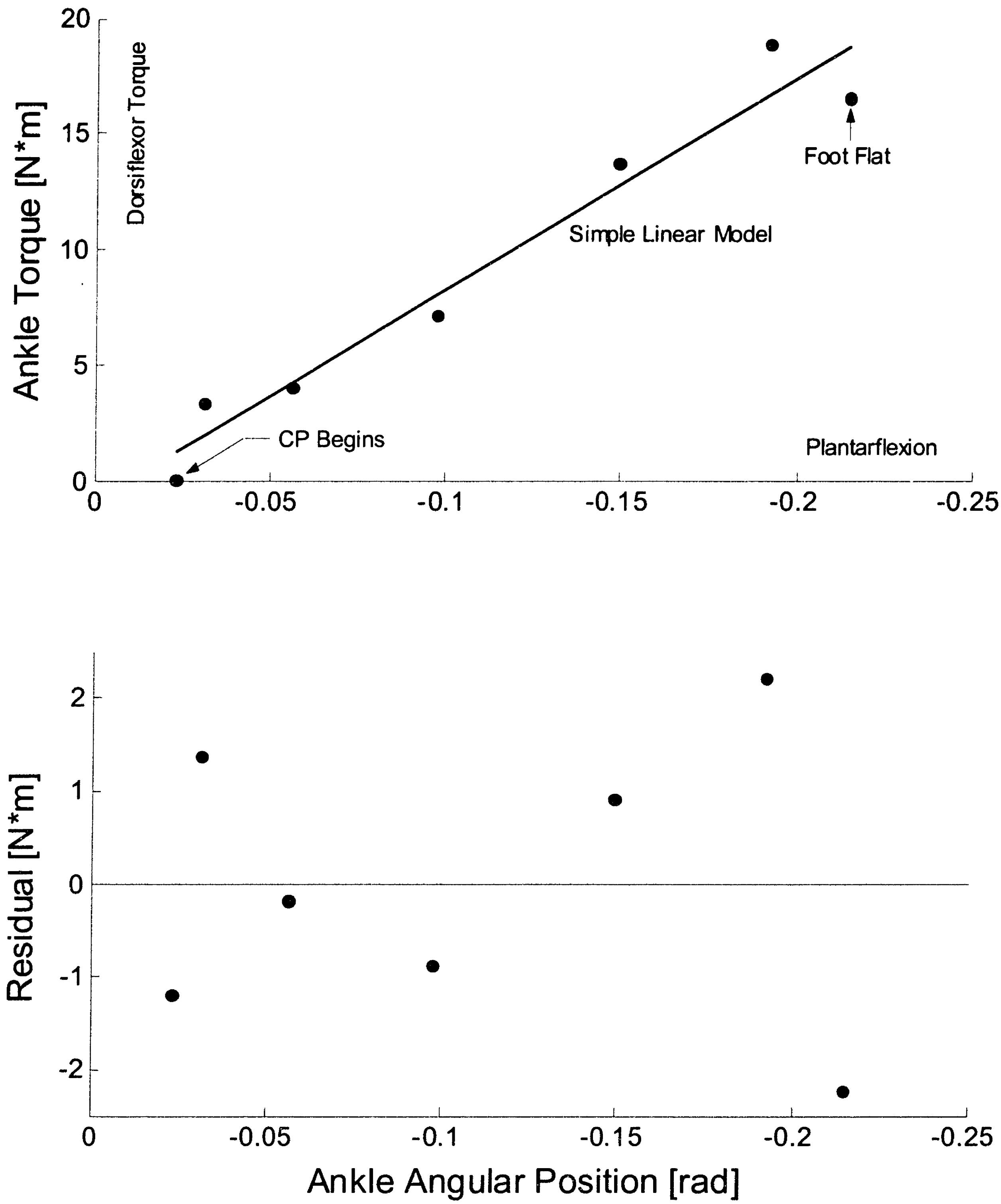

Figure 3.2 Results of simple linear regression of ankle torque on ankle angular position during controlled plantarflexion (CP) for a single trial ($p < 0.001$, $r^2 = 0.95$) including a scatter plot of the residuals. Data shown are from the same trial as Figure 3.1. Data points are equally spaced in time.

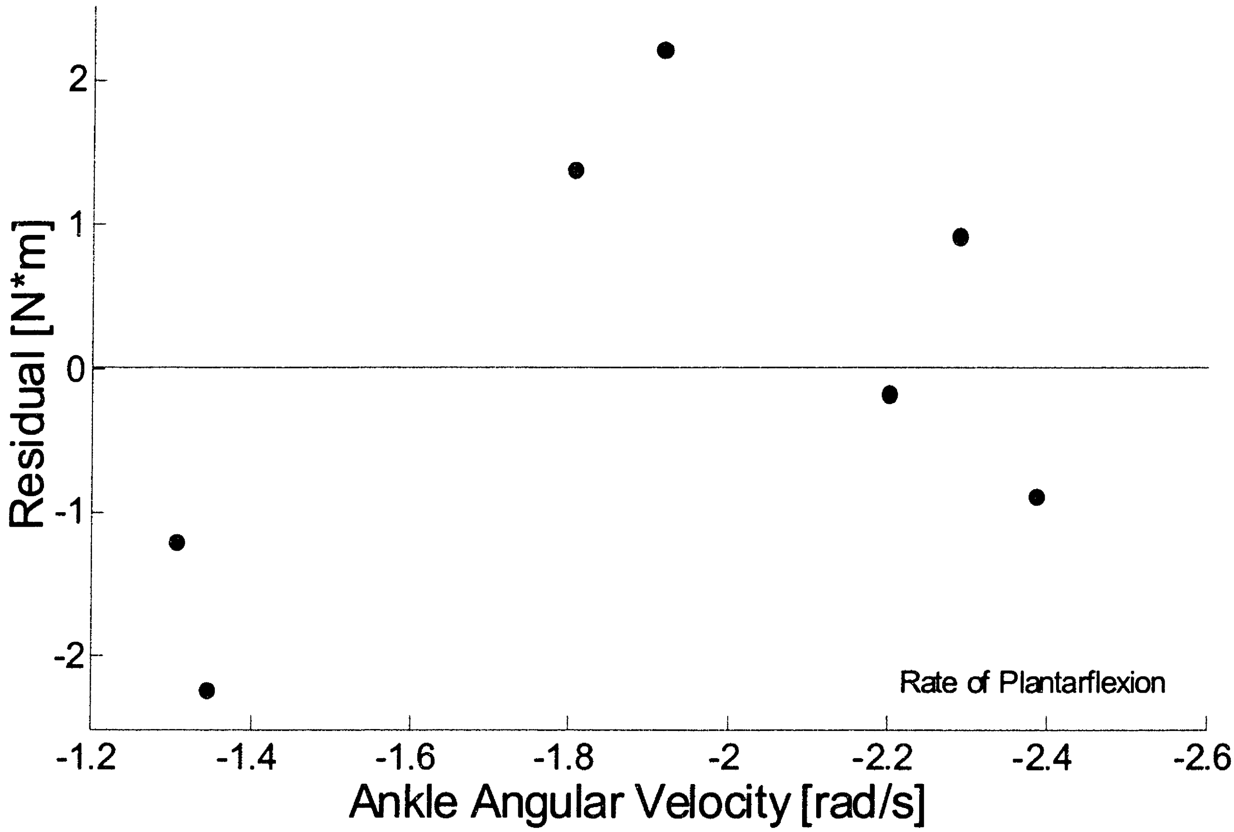

Figure 3.3 Scatter plot of the residual (from simple linear regression of ankle torque on ankle angular position during controlled plantarflexion) versus ankle angular velocity. Data shown are from the same trial as Figure 3.1 and Figure 3.2. Data points are equally spaced in time.

3.3 Controlled Dorsiflexion

The power at the ankle during controlled dorsiflexion (CD) was predominately negative (Figure 2.2). Typically a burst of positive power would occur near the beginning of CD and would last for about 5% of the duration of CD. Since the power was negative for the majority of CD, only passive elements were considered to characterize ankle function. Both damper-like and springlike behavior were considered by examining relationships between ankle torque and ankle angular velocity as well as relationships between ankle torque and ankle angular position.

Scatter plots of the ankle torque versus ankle angular velocity (Figure 3.4) did not suggest a cause and effect relationship between torque and velocity. As CD begins, ankle velocity starts near zero and quickly rises to a positive peak. At the same time ankle torque is increasing, but the torque starts negative and becomes positive, meaning the direction in which it is acting changed while the direction of the velocity did not. However, this fact alone was not deemed sufficient to rule out a causal relationship between torque and velocity. As CD continues, the ankle velocity decreases to about one-third of the peak velocity and then remains fairly constant until just before the end of CD. During this time, torque increases monotonically.

Scatter plots of the ankle torque versus ankle angular position (Figure 3.5) did suggest a cause and effect relationship between torque and position. Simple linear regression of the ankle torque on the ankle angular position during CD was significant ($p < 0.001$) for every trial of all subjects at all self-selected speeds with mean values of the coefficient of determination ranging from 0.50 to 0.98 (Table 3.4 and Figure 3.5). However, scatter plots of the residual versus ankle position (Figure 3.5) did not appear to be distributed in a band of constant width centered about zero. Typically these scatter plots could be divided into regions where the residual was either purely negative or purely positive. Because it would be impractical to present a residual scatter plot for each trial, the mean number of times the residual changed signs is given (Table 3.4) as an indicator of to what extent the scatter plots were apportioned into regions where all the residuals had the same sign.

Scatter plots of the residual versus ankle angular velocity (Figure 3.6) were generated as an additional check for relationships between ankle torque and ankle angular velocity. As was the case for controlled dorsiflexion, the magnitude of the residuals did not appear to be correlated with the magnitude of the ankle velocity. Furthermore, the residuals were both negative and positive at points where the magnitude of the ankle velocity was approximately equal.

The mean phase difference between the maximum magnitude of ankle torque and the maximum magnitude of ankle position during CD (Phase in Table 3.4) as a percentage of the duration of CD (T_{CD} in Table 3.4) ranged between -64% and 8%. Relatively large negative values of the phase difference (at least 2.25 times greater than the next smaller negative value) were observed at the normal and fast self-selected speeds for the subject labeled DEF as well as at the fast speed for both subjects EEB and RWW.

These large negative values of the phase difference are due to the fact that, for these subjects at these speeds, the ankle angular position versus time curve often did not increase monotonically during CD. Instead of the typical single peak in the position curve at about 45% of the gait cycle, there would also occur an earlier peak in position at around 20% of the gait cycle. In these cases, the magnitude of the ankle torque did not increase monotonically either. The magnitude of the ankle torque had two peaks, but the second peak was always greater than the first. Since the phase difference was defined to be negative when the maximum position occurred before the maximum torque, large negative values of the phase difference occurred when the first peak in position was greater than the second one. Ignoring negative values of the phase difference that are at least 2.25 times greater than the next smaller negative value, the mean phase difference during CD ranged between -12% and 8% and was positive for 8 out of the remaining 26 speed groups for all subjects.

Table 3.4 Results of simple linear regression of ankle torque on ankle angular position during controlled dorsiflexion ($p < 0.001$ for all trials). T_{CD} is the duration of controlled dorsiflexion. Phase is the time difference between the max. ankle position and the max. ankle torque. Values given are the mean (SD) for each speed group.

Subject Label	Self-selected Speed	Number of Trials	Gait Speed [m/s]	# of Points Used for Regression	r^2	# of Sign Changes in Residual	T_{CD} [ms]	Phase [% T_{CD}]
	Slow	9	0.99 (0.060)	67 (8.0)	0.92 (0.029)	4 (1.7)	553 (66.7)	8 (12.7)
DEF	Normal	9	1.49 (0.052)	53 (1.8)	0.66 (0.102)	1 (0.5)	430 (15.1)	-64 (3.9)
	Fast	8	1.91 (0.027)	41 (8.5)	0.60 (0.224)	3 (1.7)	330 (71.0)	-54 (22.5)
	Slow	7	0.77 (0.019)	79 (3.4)	0.96 (0.018)	4 (2.9)	646 (28.0)	4 (5.2)
DMG	Normal	8	1.30 (0.060)	58 (1.8)	0.92 (0.027)	3 (1.2)	472 (14.7)	0 (3.1)
	Fast	7	1.69 (0.038)	50 (1.9)	0.75 (0.101)	2 (0.0)	410 (15.5)	-2 (2.9)
	Slow	7	0.85 (0.063)	78 (8.7)	0.97 (0.032)	6 (2.2)	645 (72.6)	2 (3.5)
EAS	Normal	8	1.33 (0.034)	56 (3.1)	0.94 (0.029)	6 (2.1)	454 (25.6)	0 (3.8)
	Fast	5	1.89 (0.048)	38 (2.0)	0.79 (0.037)	2 (0.0)	308 (16.7)	-1 (1.5)
	Slow	6	0.85 (0.042)	75 (3.8)	0.94 (0.043)	4 (1.4)	614 (31.5)	5 (2.4)
EEB	Normal	3	1.15 (0.061)	60 (1.7)	0.95 (0.007)	5 (2.1)	492 (14.4)	3 (2.9)
	Fast	4	1.79 (0.080)	39 (8.1)	0.59 (0.259)	2 (0.0)	313 (67.2)	-35 (21.9)
	Slow	6	0.87 (0.022)	77 (4.5)	0.92 (0.041)	2 (0.8)	636 (37.5)	3 (2.3)
FJI	Normal	9	1.21 (0.016)	62 (1.3)	0.92 (0.038)	2 (0.0)	505 (11.1)	0 (2.8)
	Fast	5	1.69 (0.063)	50 (1.3)	0.84 (0.048)	2 (0.9)	407 (10.9)	0 (1.4)
	Slow	7	0.94 (0.060)	59 (4.3)	0.91 (0.050)	3 (1.5)	486 (36.2)	-3 (1.9)
JLM	Normal	8	1.46 (0.058)	42 (4.2)	0.79 (0.077)	2 (0.0)	343 (34.9)	-12 (15.2)
	Fast	6	1.88 (0.007)	24 (4.8)	0.78 (0.153)	2 (0.0)	194 (40.0)	-6 (7.5)
	Slow	6	0.84 (0.049)	76 (5.2)	0.97 (0.015)	3 (1.6)	621 (43.7)	-1 (1.6)
MJK	Normal	9	1.23 (0.024)	57 (1.6)	0.92 (0.025)	2 (0.7)	467 (13.2)	-3 (3.1)
	Fast	7	1.85 (0.044)	37 (8.1)	0.73 (0.132)	3 (1.1)	301 (67.8)	-6 (9.6)
	Slow	6	0.88 (0.052)	68 (2.0)	0.97 (0.020)	5 (3.3)	556 (16.4)	-1 (2.8)
MKK	Normal	6	1.38 (0.052)	49 (3.2)	0.95 (0.029)	4 (2.0)	403 (26.7)	0 (4.0)
	Fast	6	1.84 (0.040)	40 (1.6)	0.81 (0.063)	3 (1.0)	326 (13.4)	-3 (2.8)
	Slow	6	0.89 (0.027)	64 (3.6)	0.92 (0.045)	4 (2.5)	521 (30.2)	1 (1.7)
RAH	Normal	5	1.28 (0.025)	46 (3.0)	0.94 (0.043)	5 (1.3)	377 (25.3)	-6 (8.0)
	Fast	5	1.70 (0.061)	37 (7.4)	0.79 (0.142)	4 (2.3)	300 (61.8)	-6 (6.5)
	Slow	8	1.06 (0.043)	64 (3.0)	0.98 (0.017)	7 (2.6)	522 (25.2)	1 (3.2)
RWW	Normal	8	1.47 (0.047)	53 (1.3)	0.87 (0.054)	3 (1.1)	435 (10.7)	-1 (1.0)
	Fast	7	1.90 (0.071)	43 (2.2)	0.50 (0.142)	2 (1.0)	354 (18.5)	-27 (28.8)

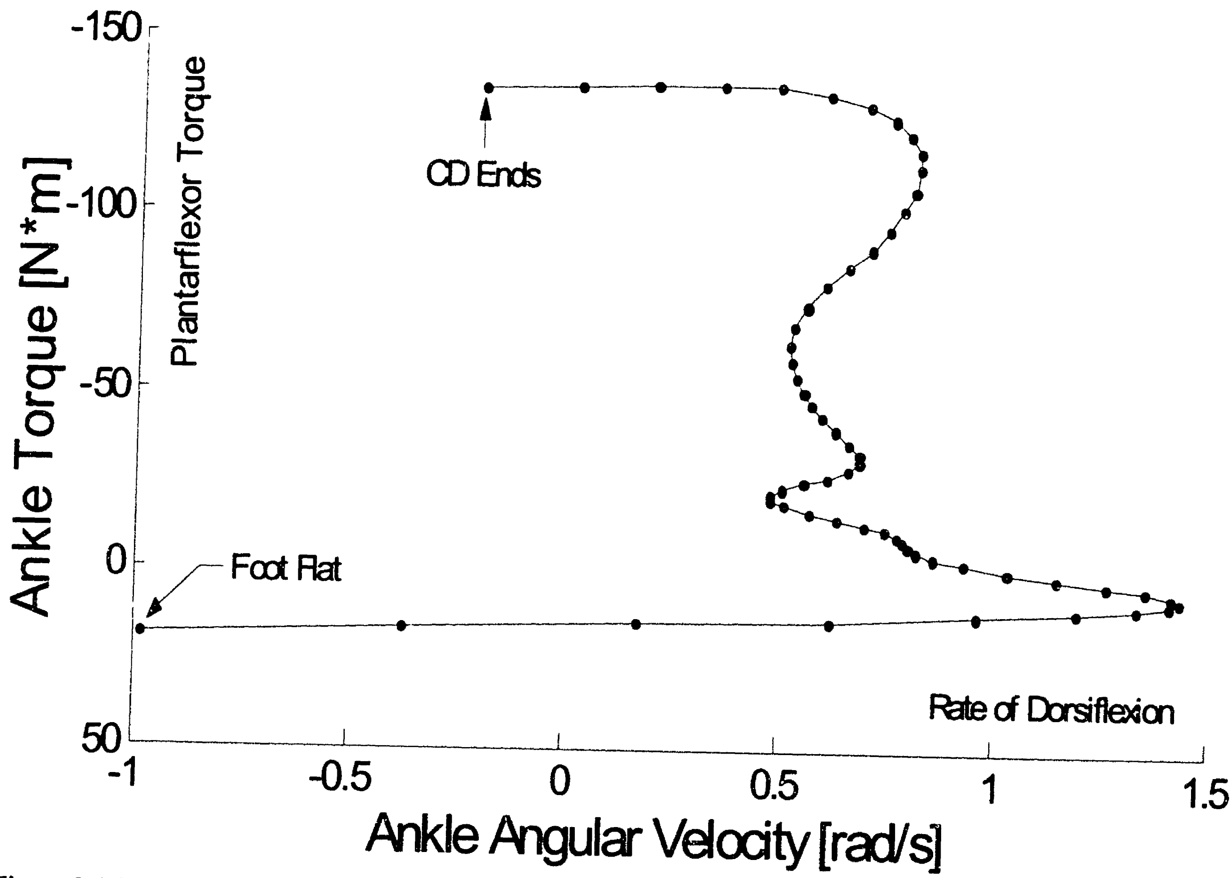

Figure 3.4 Scatter plot of ankle torque versus ankle angular velocity during controlled dorsiflexion (CD). Data shown are from a single trial of subject FJI walking at the self-selected normal gait speed and are representative of the subject's trials. Data points are equally spaced in time.

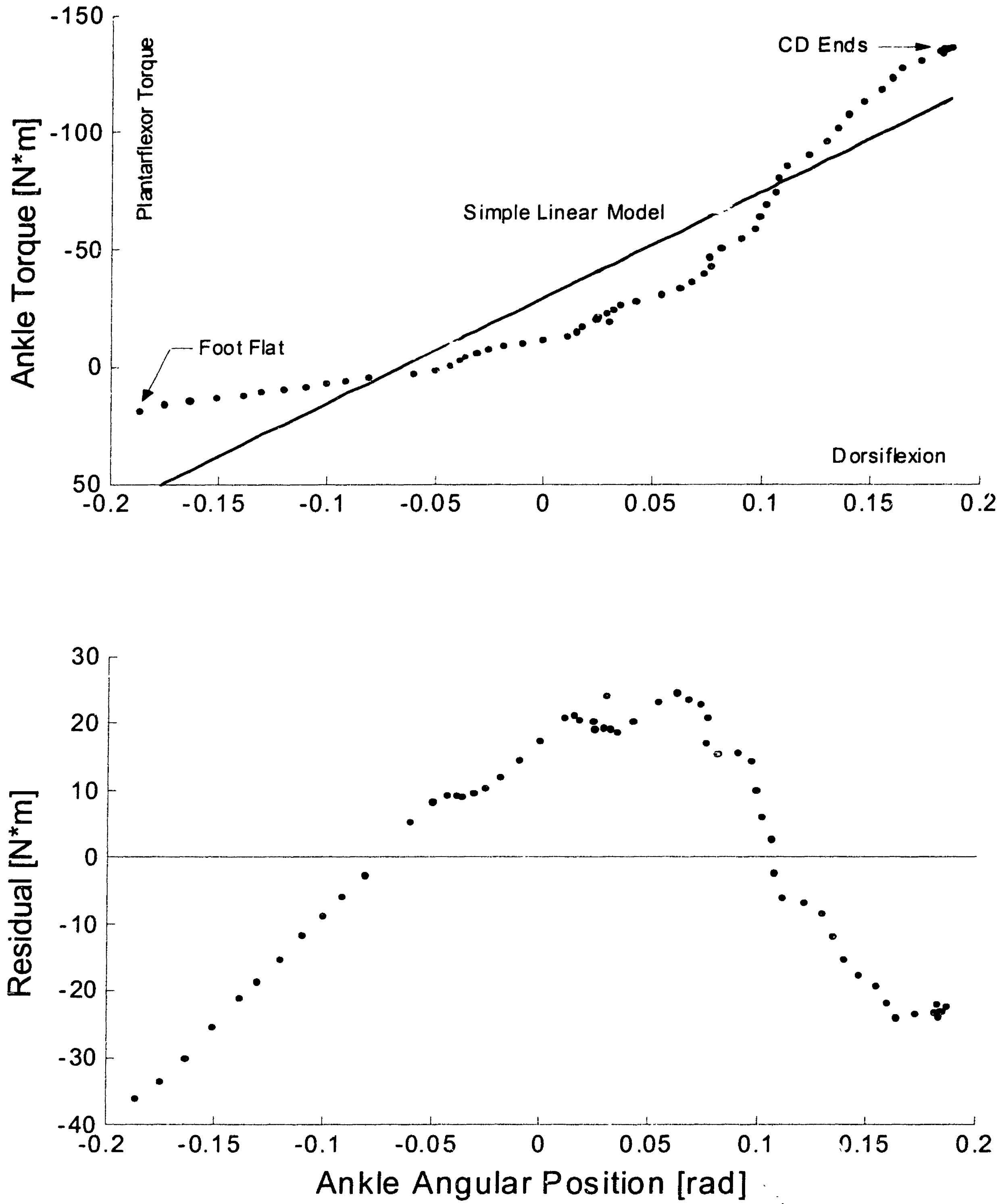

Figure 3.5 Results of simple linear regression of ankle torque on ankle angular position during controlled dorsiflexion (CD) for a single trial ($p < 0.001$, $r^2 = 0.87$) including a scatter plot of the residuals. Data shown are from the same trial as Figure 3.4. Data points are equally spaced in time.

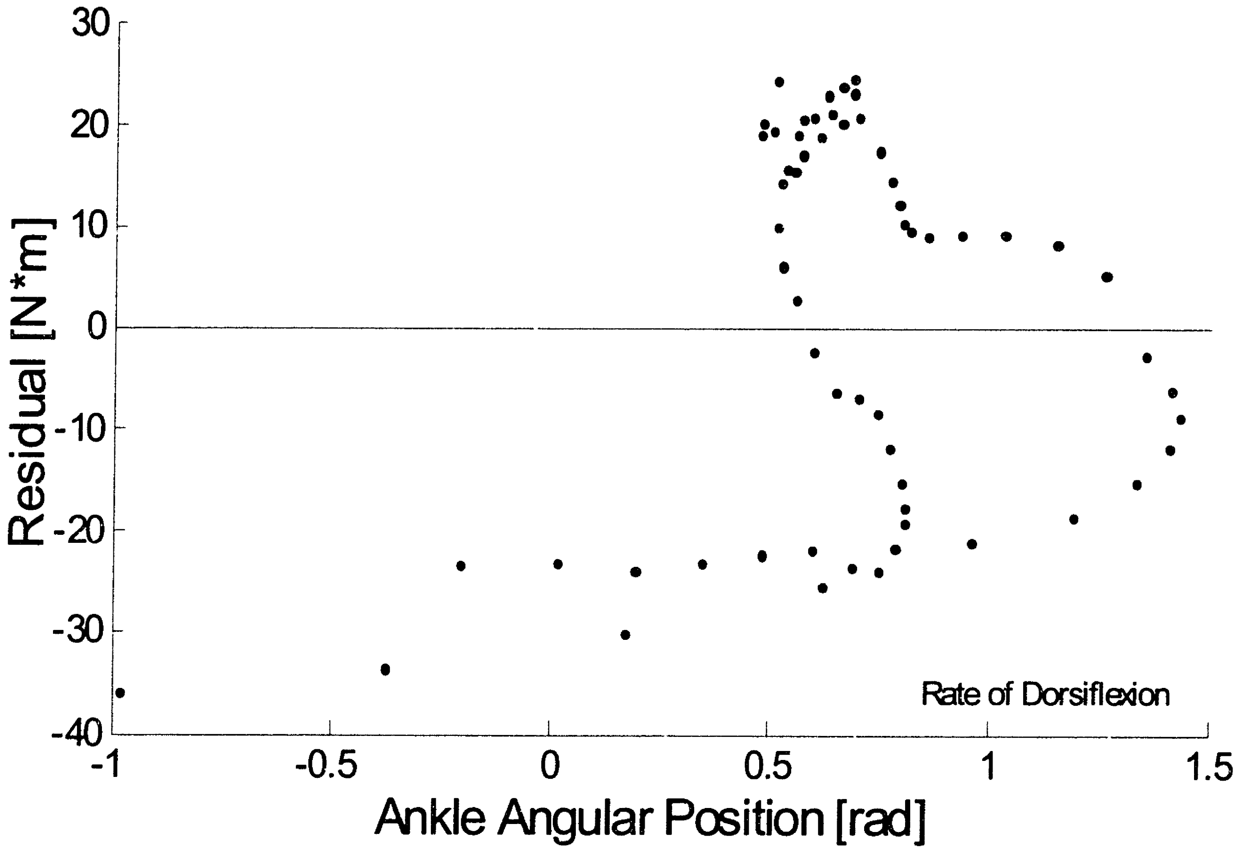

Figure 3.6 Scatter plot of the residual (from simple linear regression of ankle torque on ankle angular position during controlled dorsiflexion) versus ankle angular velocity. Data shown are from the same trial as Figure 3.4 and Figure 3.5. Data points are spaced equally in time.

3.4 POWERED PLANTARFLEXION

The power at the ankle was always positive during powered plantarflexion (PP) (Figure 2.2) so both active and passive elements were considered to characterize ankle function. For all 10 subjects, the mean work done at the ankle during PP monotonically increased from the self-selected slow speed to the fast speed (Table 3.5). The increase from slow to fast was significant ($p < 0.05$) for all 10 subjects as was the increase from slow to normal. The net increase from slow to fast ranged between 47% and 109%. Furthermore, the work done at the ankle previous to PP (i.e., the sum of the work done at the ankle during both controlled plantarflexion and controlled dorsiflexion) was always negative (Table 3.5). The magnitude of the mean positive work done at the ankle during PP was greater than the magnitude of the mean negative work done at the ankle previous to PP for 26 out of the 30 self-selected speed groups for all subjects.

Table 3.5 Work done at the ankle during powered plantarflexion (W_{PP}) and the sum of the work done at the ankle during controlled plantarflexion and controlled dorsiflexion (W_{CP+CD}). Values given are the mean (SD) for each speed group.

Subject Label	Self-selected Speed	Number of Trials	Gait Speed [m/s]	W_{PP} [J]	W_{CP+CD} [J]	W_{PP} - $\|W_{CP+CD}\|$ [J]
	Slow	9	0.99 (0.060)	12.4 (2.52)[a,b]	-11.0 (1.97)[a,b]	1.5 (3.55)[a,b]
DEF	Normal	9	1.49 (0.052)	16.8 (2.83)[b]	-4.1 (1.50)	12.7 (3.00)[b]
	Fast	8	1.91 (0.027)	22.5 (3.92)	-5.6 (2.74)	16.9 (3.27)
	Slow	7	0.77 (0.019)	14.0 (3.79)[a,b]	-14.7 (1.62)[b]	-0.7 (2.70)[a,b]
DMG	Normal	8	1.30 (0.060)	19.7 (4.63)[b]	-14.2 (2.89)[b]	5.5 (3.31)[b]
	Fast	7	1.69 (0.038)	26.7 (5.15)	-11.1 (2.31)	15.6 (3.45)
	Slow	7	0.85 (0.063)	24.3 (4.34)[a,b]	-23.3 (2.02)[b]	1.0 (5.36)[a,b]
EAS	Normal	8	1.33 (0.034)	33.4 (5.29)[b]	-22.1 (2.67)[b]	11.3 (3.64)[b]
	Fast	5	1.89 (0.048)	44.5 (3.06)	-15.2 (1.20)	29.4 (3.30)
	Slow	6	0.85 (0.042)	10.7 (2.15)[a,b]	-14.2 (2.03)[a,b]	-3.4 (1.07)[b]
EEB	Normal	3	1.15 (0.061)	14.5 (1.95)[b]	-17.5 (1.60)[b]	-3.0 (3.30)[b]
	Fast	4	1.79 (0.080)	21.0 (2.23)	-5.8 (2.34)	15.2 (3.02)
	Slow	6	0.87 (0.022)	17.7 (3.13)[a,b]	-14.5 (1.77)	3.2 (2.94)[a,b]
FJI	Normal	9	1.21 (0.016)	23.5 (2.50)[b]	-16.2 (1.88)[b]	7.3 (3.78)[b]
	Fast	5	1.69 (0.063)	29.0 (6.77)	-12.8 (3.31)	16.2 (9.23)
	Slow	7	0.94 (0.060)	16.0 (1.76)[a,b]	-9.2 (1.67)[a,b]	6.7 (1.33)[a,b]
JLM	Normal	8	1.46 (0.058)	19.8 (2.00)[b]	-5.8 (1.28)[b]	14.0 (2.46)[b]
	Fast	6	1.88 (0.007)	23.5 (1.43)	-2.9 (0.88)	20.6 (2.20)
	Slow	6	0.84 (0.049)	14.7 (2.62)[a,b]	-12.6 (1.36)[b]	2.1 (2.13)[a,b]
MJK	Normal	9	1.23 (0.024)	19.5 (1.56)[b]	-12.4 (1.38)[b]	7.1 (1.87)[b]
	Fast	7	1.85 (0.044)	22.8 (2.14)	-6.3 (0.95)	16.5 (2.09)
	Slow	6	0.88 (0.052)	12.2 (1.99)[a,b]	-12.6 (1.11)	-0.4 (2.75)[a,b]
MKK	Normal	6	1.38 (0.052)	18.6 (2.44)[b]	-11.4 (0.97)	7.2 (3.03)[b]
	Fast	6	1.84 (0.040)	25.5 (3.97)	-10.5 (5.41)	15.0 (1.68)
	Slow	6	0.89 (0.027)	8.5 (1.41)[a,b]	-5.9 (0.93)[a]	2.6 (1.76)[a,b]
RAH	Normal	5	1.28 (0.025)	13.3 (2.65)	-4.5 (0.47)	8.8 (2.92)
	Fast	5	1.70 (0.061)	17.2 (2.96)	-4.5 (3.95)	12.7 (6.43)
	Slow	8	1.06 (0.043)	18.2 (1.62)[a,b]	-16.8 (1.11)[b]	1.3 (1.70)[a,b]
RWW	Normal	8	1.47 (0.047)	24.6 (3.75)[b]	-14.9 (2.40)[b]	9.7 (4.83)[b]
	Fast	7	1.90 (0.071)	32.7 (3.32)	-10.1 (3.77)	22.6 (2.62)

[a] Significantly different from the mean of the same parameter at the self-selected normal speed ($p < 0.05$).

[b] Significantly different from the mean of the same parameter at the self-selected fast speed ($p < 0.05$).

CHAPTER 4
DISCUSSION

4.1 NORMALITY OF THE SUBJECTS

The classification of the subjects who provided the data for this study as "normal" was based on comparison of the subjects' gait parameters with those that have been reported previously. The temporal, distance, and ankle kinematic and kinetic gait parameters for the subjects' self-selected normal gait speed are in agreement with those reported in the literature (Murray et al., 1966; Murray et al., 1970; Kadaba et al., 1989; Kadaba et al., 1990; Borghese et al., 1996). The trends across gait speed of the temporal and distance parameters agree with those observed by Murray et al. (1966) and Borghese et al. (1996). These comparisons suggest that the subjects did indeed represent a normal cross section of the population.

4.2 CHARACTERIZATION OF ANKLE FUNCTION DURING CONTROLLED PLANTARFLEXION

Ankle function during controlled plantarflexion (CP) was characterized by a linear, torsional spring for all self-selected gait speeds. The power was always negative during CP (Figure 2.2) so only passive elements were considered in modeling the ankle. The scatter plots of ankle angular velocity versus ankle torque (Figure 3.1) suggested that velocity-dependent effects on ankle torque caused by damping elements were negligible. This conclusion was supported by plotting the residual from simple linear regression of ankle torque on ankle position against ankle velocity (Figure 3.3) and by the fact that the mean phase difference between the maximum ankle torque and the maximum ankle angular position was rarely greater than 15% of the mean duration of CP.

The possibility that error due to quantization introduced a bias into the estimates of the mean phase difference was considered. The phase difference was calculated by taking the difference between two measurements of time. The gait data were processed at 120 Hz (i.e., 8.3 ms elapsed between each data sample). Thus any measurement of time was subject to a quantization error that was uniformly distributed between -4.2 and 4.2 ms with a mean of zero (Oppenheim and Schafer, 1989). In the worst-case scenario, the magnitude of the quantization error in a single measurement of time could be as great as 11% of the duration of CP. Because the magnitude of the quantization error could be large relative to the duration of CP, the effect of the quantization error on the estimation of the mean phase difference from a given sample was considered. By considering each measurement of time as the sum of the actual time plus a quantization error it can be shown that, while the quantization error did bias the estimate of the variance, it did not bias the estimate of the mean phase difference (Ang and Tang, 1975).

Once it had been concluded that ankle function was dominated by springlike behavior during CP, simple linear regression was used to show that the ankle could be modeled as a linear spring. A simple linear model was determined to be adequate based on the mean coefficient of determination always being greater than 0.85 and on the pattern observed in the resulting scatter plots of the residual versus the ankle position. In these scatter plots, the residuals did appear to be normally distributed in a band of constant width centered about zero (Figure 3.2) indicating that a simple linear model was appropriate. The ratio of the mean number of times the residual changed sign to the number of data points used for regression ranged between 0.22 and 0.64 suggesting that this pattern in the residual scatter plots held for all the subjects' trials.

4.3 Ankle Spring Stiffness Variability and Ankle Adaptation to Gait Speed

After establishing that ankle function can be characterized by a linear, torsional spring during controlled plantarflexion (CP), the slope of the regression line fit to the ankle torque versus ankle angular position curves for CP took on the meaning of an ankle spring stiffness. There was a surprising amount of stride-to-stride variability in this calculated ankle spring stiffness. Comparing values of the ankle spring stiffness only to values calculated for other trials in the same self-selected speed group for the same subject, it was seen that the largest ankle spring stiffness was at least 2 times greater than the smallest ankle spring stiffness in 18 out of the 30 self-selected speed groups. In addition, no systematic trend in the variation of ankle spring stiffness with gait speed was observed (Table 4.1) because there was often a considerable amount of overlap between the range of ankle spring stiffness at a certain self-selected speed and the range of ankle spring stiffness at the other speeds for the same subject (Figure 4.1 and Figure 4.2).

As was mentioned in Chapter 1, determining when and to what extent the ankle model's parameters varied could lead to insight about the control system governing ankle function. But was the unexpectedly large amount of stride-to-stride variation in ankle spring stiffness evidence that the ankle spring stiffness was being modulated purposefully by an intelligent control system, or was it simply the result of large random error in estimating this parameter? In what follows, it is argued that comparison of the calculated values of ankle spring stiffness to those measured by other researchers, the dynamics of the foot-ankle system during CP, and statistical considerations all indicated that stride-to-stride adjustments made by an intelligent control system was the more likely of these two explanations.

Table 4.1 Parameters that affect the work done by the ankle spring (W_{spring}) during controlled plantarflexion. θ_{NP} is the ankle spring neutral position. θ_{HS} and θ_{FF} are the ankle position at heel strike and at foot flat, respectively. Values given are the mean (SD) for each speed group.

Subject Label	Self-selected Speed	Ankle Spring Stiffness (Slope) [N*m/rad]	θ_{NP} [rad]	θ_{HS} [rad]	θ_{FF} [rad]	W_{spring} [J]
DEF	Slow	59 (14.9)	0.02 (0.035)	0.01 (0.032)	-0.15 (0.033)	-0.8 (0.15)^{a,b}
	Normal	80 (31.1)	0.00 (0.033)	0.00 (0.019)	-0.13 (0.021)^{b}	-0.6 (0.15)^{b}
	Fast	70 (22.8)	0.01 (0.036)	0.00 (0.007)	-0.18 (0.020)	-1.1 (0.24)
DMG	Slow	75 (28.8)	-0.12 (0.036)^{b}	-0.12 (0.024)	-0.21 (0.017)^{b}	-0.3 (0.12)^{a,b}
	Normal	66 (5.6)	-0.12 (0.020)^{b}	-0.12 (0.006)	-0.24 (0.021)	-0.5 (0.06)^{b}
	Fast	68 (18.4)	-0.06 (0.021)	-0.09 (0.020)	-0.25 (0.017)	-1.1 (0.18)
EAS	Slow	42 (9.8)^{a}	-0.06 (0.040)^{a,b}	-0.09 (0.047)^{b}	-0.24 (0.033)	-0.6 (0.10)^{a,b}
	Normal	80 (9.9)^{b}	-0.02 (0.022)^{b}	-0.05 (0.024)^{b}	-0.22 (0.012)	-1.4 (0.22)^{b}
	Fast	28 (3.4)	0.42 (0.095)	0.05 (0.009)	-0.21 (0.016)	-3.7 (0.45)
EEB	Slow	35 (2.3)^{a}	-0.10 (0.023)^{b}	-0.09 (0.028)^{b}	-0.24 (0.046)	-0.3 (0.13)^{a,b}
	Normal	49 (4.1)	-0.11 (0.024)^{b}	-0.10 (0.016)^{b}	-0.29 (0.041)	-0.8 (0.30)^{b}
	Fast	34 (15.2)	0.09 (0.096)	-0.03 (0.009)	-0.30 (0.031)	-2.0 (0.19)
FJI	Slow	72 (15.0)^{b}	-0.02 (0.020)^{a,b}	-0.04 (0.014)^{b}	-0.17 (0.015)^{b}	-0.8 (0.19)^{a,b}
	Normal	70 (15.9)^{b}	0.02 (0.029)	-0.04 (0.023)^{b}	-0.20 (0.016)	-1.6 (0.29)^{b}
	Fast	112 (32.6)	0.05 (0.049)	0.00 (0.022)	-0.19 (0.034)	-2.8 (0.14)
JLM	Slow	43 (6.5)	-0.07 (0.054)^{b}	-0.09 (0.057)	-0.21 (0.043)	-0.4 (0.08)^{a,b}
	Normal	47 (8.0)	-0.03 (0.071)	-0.05 (0.056)	-0.20 (0.031)	-0.7 (0.29)^{b}
	Fast	49 (17.4)	0.00 (0.049)	-0.04 (0.051)	-0.22 (0.036)	-1.1 (0.16)
MJK	Slow	25 (15.8)^{a}	0.00 (0.098)	-0.05 (0.067)	-0.22 (0.028)	-0.4 (0.08)^{b}
	Normal	28 (7.1)	-0.02 (0.079)^{b}	-0.05 (0.024)	-0.23 (0.020)	-0.6 (0.29)^{b}
	Fast	32 (6.0)	0.10 (0.046)	-0.02 (0.013)	-0.21 (0.024)	-1.3 (0.16)
MKK	Slow	25 (8.9)	-0.06 (0.035)^{a,b}	-0.09 (0.012)^{a,b}	-0.28 (0.014)^{a,b}	-0.6 (0.23)^{b}
	Normal	43 (12.3)	0.00 (0.050)	-0.02 (0.021)	-0.22 (0.022)	-1.0 (0.38)
	Fast	37 (9.8)	0.03 (0.050)	0.00 (0.016)	-0.20 (0.020)	-0.9 (0.58)
RAH	Slow	40 (10.3)^{a,b}	-0.03 (0.025)^{a,b}	-0.05 (0.028)^{a,b}	-0.22 (0.026)^{a,b}	-0.7 (0.18)^{b}
	Normal	52 (4.0)	0.01 (0.016)	-0.01 (0.018)	-0.17 (0.012)	-0.8 (0.21)
	Fast	80 (26.8)	0.01 (0.038)	-0.01 (0.029)	-0.17 (0.024)	-1.2 (0.39)
RWW	Slow	55 (16.2)^{b}	-0.03 (0.016)^{a,b}	-0.03 (0.015)^{a,b}	-0.19 (0.024)	-0.7 (0.18)^{a,b}
	Normal	63 (20.2)	0.02 (0.035)	-0.01 (0.018)	-0.21 (0.020)	-1.5 (0.30)^{b}
	Fast	84 (24.8)	0.03 (0.024)	-0.01 (0.018)	-0.20 (0.019)	-2.1 (0.37)

[a] Significantly different from the mean of the same parameter at the self-selected normal speed ($p < 0.05$).
[b] Significantly different from the mean of the same parameter at the self-selected fast speed ($p < 0.05$).

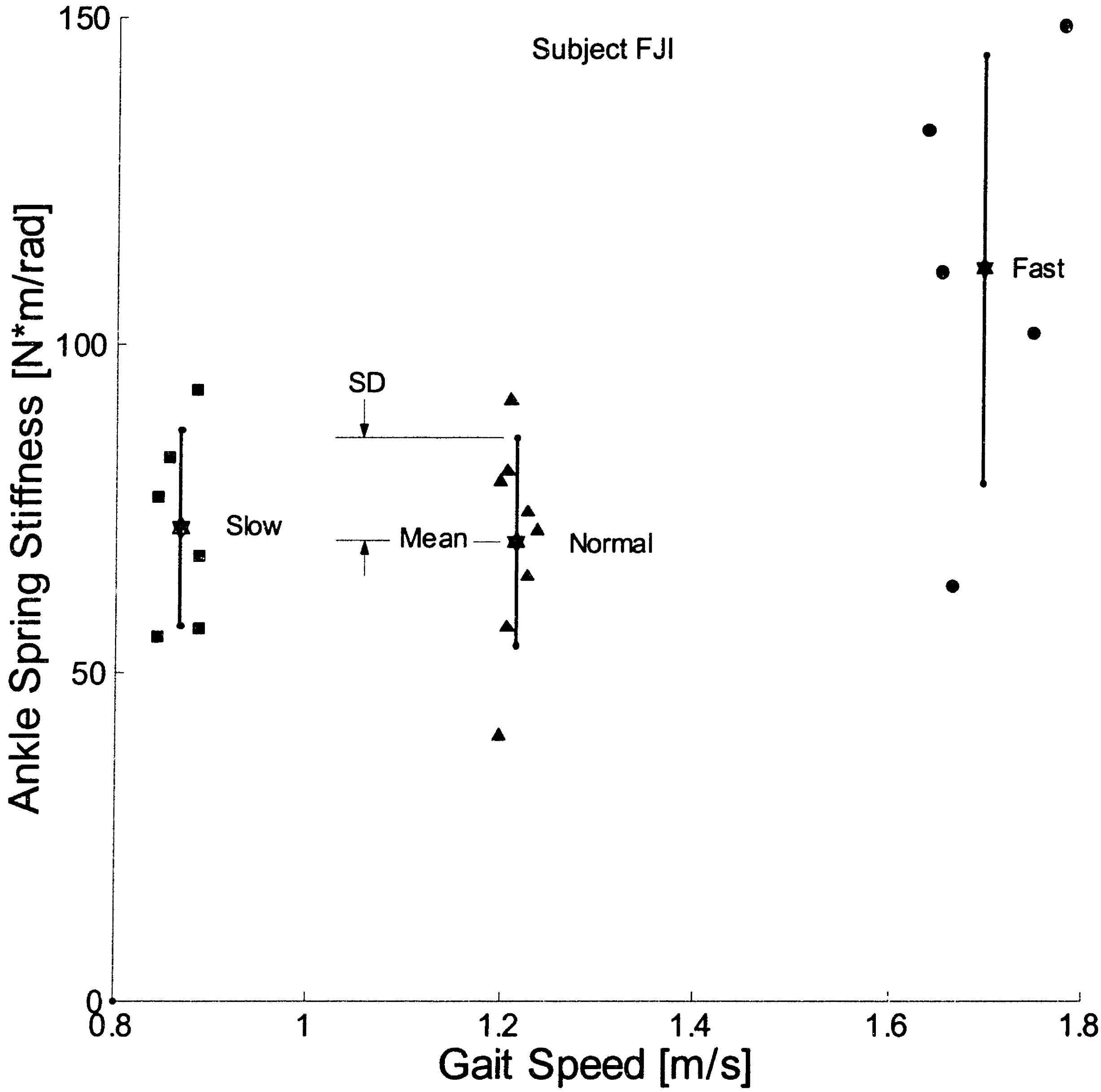

Figure 4.1 Ankle spring stiffness during controlled plantarflexion grouped by the subject's self-selected slow, normal, and fast gait speeds. Each point represents the ankle spring stiffness (slope) obtained for a single trial by simple linear regression of the ankle torque on the ankle angular position. The data for this subject are presented because of the relatively large number of trials for each self-selected speed.

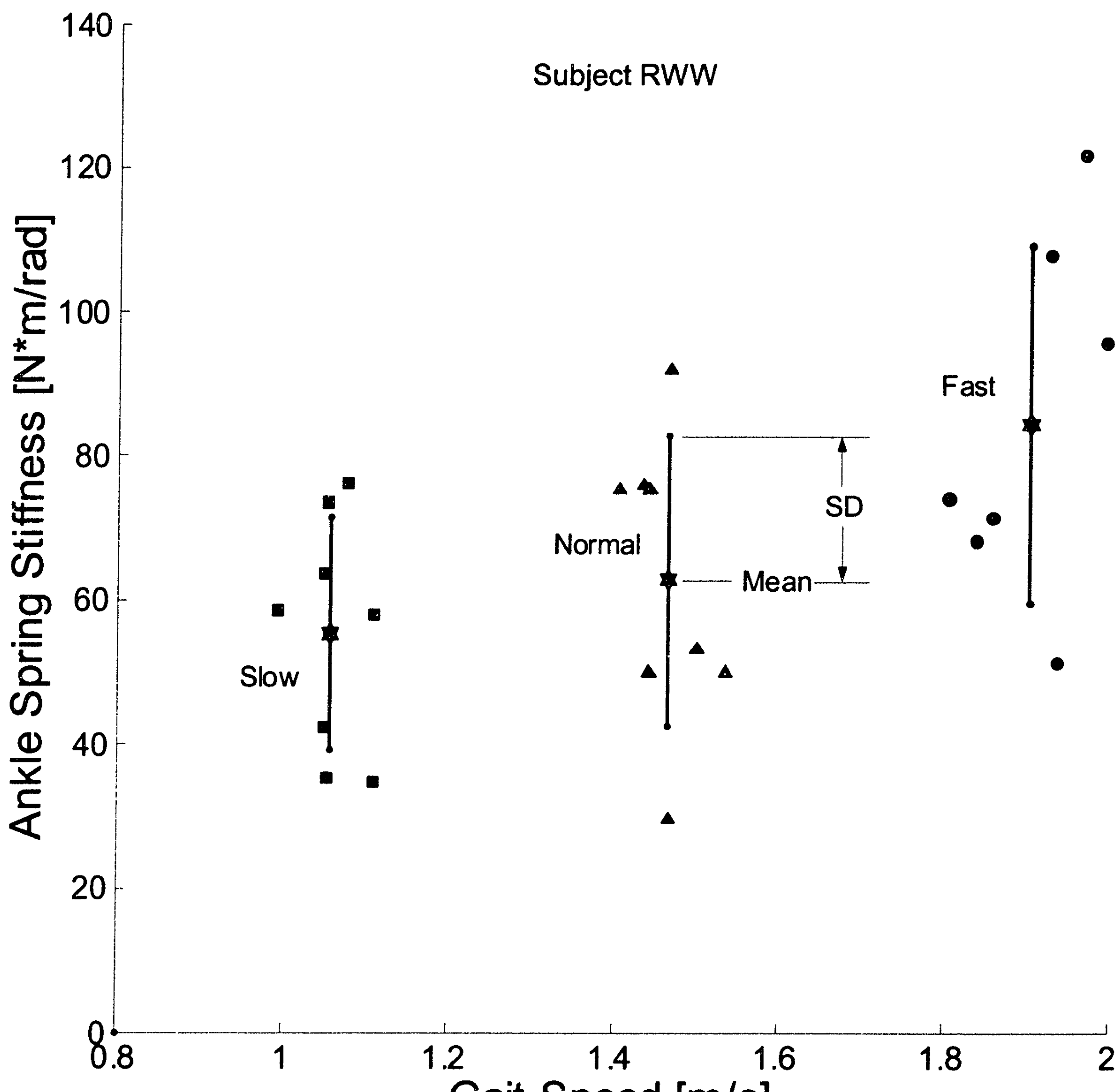

Figure 4.2 Same as Figure 4.1 but for a different subject.

The reliability of the estimated values for the ankle spring stiffness was certainly in doubt. After all, at the self-selected fast speed the estimates were sometimes based on no more than 4 points. The ankle spring stiffness for individual trials of all subjects derived using simple linear regression ranged between 8 and 149 N*m/rad. In stationary conditions with constant mean dorsiflexor torque, Weiss et. al (1988) measured a maximum ankle spring stiffness for 4 subjects ranging between 200 and 400 N*m/rad. Hunt et al. (2000) reported that a maximum ankle spring stiffness of about 286 N*m/rad could be achieved through functional electrical stimulation (FES) of a fit, healthy, male subject's dorsiflexors. Clearly the values of ankle spring stiffness derived in this study were biologically possible. Furthermore, the range of ankle spring stiffness observed was a likely operating range for the ankle during walking because the maximum ankle spring stiffness was never greater than 75% of estimates of the maximum possible ankle spring stiffness. Encouraged by these facts, hypotheses were developed that could explain the stride-to-stride variation in ankle spring stiffness in terms of a plausible control strategy.

The term "controlled plantarflexion" comes from the concept that, during this period of stance, the ankle plantarflexes as the foot descends to the ground under control of the dorsiflexors. One quickly learns the advantages of *controlled* plantarflexion after taking a few steps while forcing the dorsiflexors to relax after heel strike (HS), or in other words by simulating *uncontrolled* plantarflexion. A controlled descent of the foot after HS reduces the impulse to the foot as it collides with the ground, thereby making the process smooth as well as energy efficient since little kinetic energy is lost in an inelastic collision with the ground.

With this in mind, it was hypothesized that the primary control objective during CP was to keep the impulse to the foot at the instant of ground-foot collision below an acceptable limit that does not change with gait speed. At the same time, it was considered that energy expenditure is one of the driving factors that has been used to explain patterns in legged locomotion (Inman et al., 1981; Alexander, 1989). Thus it was also hypothesized that a reasonable secondary control objective would be to minimize the amount of energy expended in maintaining the impulse to the foot below the acceptable limit. As was discussed previously, it was concluded that the function of the dorsiflexors, tendons, etc. that control the foot's descent during CP could be characterized by a linear, torsional spring. Speaking in terms of this model of ankle function, the hypothesized control objectives could be achieved by tuning the ankle spring's parameters to get the desired response. Thus adjustment of the model's parameters according to a control strategy could explain the stride-to-stride variation in the ankle spring stiffness.

Considering the sagittal plane dynamics of the foot and ankle during CP gave insight into what behavior one might expect to see from a system being governed by the hypothesized control strategy. During CP, the foot experiences pure rotation about a point located at the heel. Therefore the primary control objective of keeping the impulse to the foot at foot flat (FF) below an acceptable limit is equivalent to keeping the angular velocity of the foot at FF, relative to an inertial reference frame, below a specified limit. Accordingly, it would be expected that the angular velocity of the foot at FF would be limited by an upper bound regardless of gait speed. At the same time, the secondary control objective requires that the energy expended to achieve the primary objective be minimized. The work done at the ankle during CP would be minimized for each stride if the foot velocity at FF was always allowed to reach the upper bound. Thus it was expected that the angular velocity of the foot at FF would be invariant across gait speeds.

Additional insight was gained by considering the dynamics of the foot and ankle in terms of work and energy. In terms of the primary control objective, the purpose of the negative work done by the ankle spring during CP is to keep the kinetic energy of the foot at FF below an acceptable limit.

As weight is transferred to the foot after HS,[1] an internal force acting through the ankle joint generates a torque about the heel that tends to accelerate the rotation of the foot towards the ground. The ankle spring applies a torque to the foot that opposes that angular acceleration of the foot. Thus, neglecting the work done by gravity, the kinetic energy of the foot at FF can be expressed as

$$KE_{FF} = KE_{HS} + W_{force} + W_{spring} \tag{4.1}$$

where KE_{FF} is the kinetic energy of the foot at FF, KE_{HS} is the kinetic energy of the foot at HS, W_{force} is the work done on the foot by the internal force acting at the ankle, and W_{spring} is the work done on the foot by the ankle spring. W_{force} is always positive and increases if the magnitude of the force acting at the ankle increases. Therefore, if the primary control objective is to keep KE_{FF} bounded, the negative work done by the ankle spring on the foot would have to increase if the magnitude of the force acting through the ankle increased enough to cause the limit on KE_{FF} to be exceeded.

Neglecting inertial and gravitational forces acting on the foot, the magnitude of the internal force acting through the ankle is equal to the magnitude of the ground reaction force (GRF) during CP. In order to easily compare the magnitude of the GRF during CP across the three self-selected speed groups for each subject, the time average of the magnitude of the sagittal plane GRF during CP was first calculated for each trial in a group and then these time averages were ensemble averaged to obtain a mean for each self-selected speed group. This metric was called the "mean magnitude of the sagittal plane GRF".

As would be expected, the mean magnitude of the sagittal plane GRF during CP was seen to monotonically increase with increasing gait speed (Table 4.2). The net increase from the self-selected slow speed to the fast speed ranged between 27% and 211% and was significant ($p < 0.05$) for all 10

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[1] For the purposes of this discussion, the instant of HS is considered to be the beginning of CP even though, as was discussed in Chapter 2, this was not always the case in the data analysis.

Table 4.2 The mean magnitude of the sagittal plane ground reaction force (GRF) during controlled plantarflexion and the angular velocity of the foot (ω_{foot}) at foot flat (FF). Values given are the mean (SD) for each speed group.

Subject Label	Self-selected Speed	Mean |GRF| in the Sagittal Plane [N]	ω_{foot} at FF [rad/s]
	Slow	347 (47.8)[b]	-2.9 (0.84)[a,b]
DEF	Normal	385 (79.7)	-4.0 (0.34)[b]
	Fast	440 (36.9)	-4.5 (0.27)
	Slow	211 (27.0)[a,b]	-1.5 (0.33)[a,b]
DMG	Normal	343 (47.6)	-2.4 (0.24)[b]
	Fast	373 (42.5)	-3.6 (0.22)
	Slow	254 (30.2)[a,b]	-2.1 (0.22)[a,b]
EAS	Normal	453 (67.0)[b]	-3.2 (0.24)[b]
	Fast	792 (11.9)	-4.6 (0.01)
	Slow	152 (26.2)[a,b]	-2.8 (0.29)[b]
EEB	Normal	236 (29.5)[b]	-3.1 (0.10)[b]
	Fast	409 (35.8)	-5.0 (0.28)
	Slow	243 (20.1)[a,b]	-2.4 (0.20)[a,b]
FJI	Normal	353 (40.5)[b]	-3.4 (0.28)[b]
	Fast	481 (75.5)	-4.8 (0.15)
	Slow	205 (35.1)[b]	-2.3 (0.27)[a,b]
JLM	Normal	225 (35.2)[b]	-2.6 (0.25)
	Fast	316 (54.7)	-2.9 (0.41)
	Slow	113 (20.2)[a,b]	-2.5 (0.23)[a,b]
MJK	Normal	145 (22.3)[b]	-3.6 (0.32)[b]
	Fast	252 (26.9)	-5.0 (0.09)
	Slow	142 (12.3)[a,b]	-2.3 (0.63)[a,b]
MKK	Normal	207 (24.5)[b]	-3.9 (0.31)[b]
	Fast	270 (23.0)	-4.6 (0.32)
	Slow	173 (20.7)[a,b]	-3.2 (0.22)[a,b]
RAH	Normal	226 (27.1)[b]	-4.0 (0.17)[b]
	Fast	326 (74.3)	-4.9 (0.22)
	Slow	276 (31.9)[a,b]	-3.1 (0.30)[a,b]
RWW	Normal	436 (60.5)[b]	-3.9 (0.36)[b]
	Fast	569 (32.4)	-4.6 (0.45)

[a] Significantly different from the mean of the same parameter at the self-selected normal speed ($p < 0.05$).

[b] Significantly different from the mean of the same parameter at the self-selected fast speed ($p < 0.05$).

subjects. Consequently, it was expected that the magnitude of the negative work done by the ankle spring on the foot during CP would have to increase with gait speed in order to achieve the primary control objective.

This expected increase in the magnitude of the mean work done by the ankle spring that would accompany an increase in gait speed also seemed in accordance with the trend in work done at the knee during CP reported by Winter (1983). Winter did not report the work done at the ankle during CP separately from the work done during controlled dorsiflexion so it was not possible to determine the trend in work done at the ankle during CP from the data that was reported. However, the work done at the knee during a negative power phase beginning at HS and ending at about 12% of the gait cycle was reported. The work done at the knee during that phase was seen to increase with increasing gait speed.

Assuming the ankle spring to be ideal, the work done by the ankle spring on the foot during CP is the difference between the energy stored in the ankle spring at HS and the energy stored in the ankle spring at FF given by

$$W_{spring} = \frac{1}{2}k\left(\delta_{HS}^2 - \delta_{FF}^2\right) \tag{4.2}$$

where W_{spring} is the work done by the ankle spring on the foot and k is the ankle spring stiffness. δ_i is the ankle spring's displacement from its neutral position at position i given by

$$\delta_i = \theta_i - \theta_{NP} \tag{4.3}$$

where θ_i is the ankle angle at position i, and θ_{NP} is the neutral position of the ankle spring.[2] Thus the

[2] The neutral position of the ankle spring was taken to be $\theta_{NP} = -\tau_0/k$ where τ_0 is the intercept given by simple linear regression of ankle torque on ankle angular position and k is the slope or ankle spring stiffness. θ_{NP} was nearly always an extrapolated point since it was nearly always outside the range of the ankle position data used for regression.

work done by the ankle spring during CP is dependent on four parameters: the ankle spring stiffness, the ankle spring's neutral position, the position of the ankle at HS, and the position of the ankle at FF.

Biologically any of these four parameters could be actively controlled by the central nervous system to modulate the amount of work done at the ankle during CP, but a change in the ankle position at either HS or FF does not represent a change in the ankle spring. In developing a model of ankle function, it was assumed that ankle angular position was one of the inputs to the black box representing the ankle during the gait analysis. Accordingly, any trends in the variation of ankle position must then be interpreted as changes in the input to the ankle system while trends in the variation of the ankle spring's stiffness and neutral position are interpreted as changes to the system itself. In other words, it is one thing to achieve the control objectives by changing the input to the ankle system from stride to stride, and quite another to achieve the control objectives by tuning the ankle system's parameters to get the desired dynamic response to the input.

When considered individually, no trend across gait speed was observed in the variation of any of the four parameters that affect the work done by the ankle spring (Table 4.1), but when these four parameters were combined to calculate the work according to Equation 4.2 and Equation 4.3, a systematic trend across gait speed was seen (Table 4.1, Figure 4.3, and Figure 4.4). The mean work done by the ankle spring during CP significantly ($p < 0.05$) increased from the self-selected slow speed to the fast speed for all 10 subjects. The net increase from slow to fast ranged between 37% and 531% and was monotonic for nine of the subjects. The subject labeled DEF had the smallest percent increase from slow to fast and subject EAS had the greatest increase. Subjects DEF and EAS also had the smallest and greatest percent increase, respectively, in the mean magnitude of the sagittal plane GRF from slow to fast.

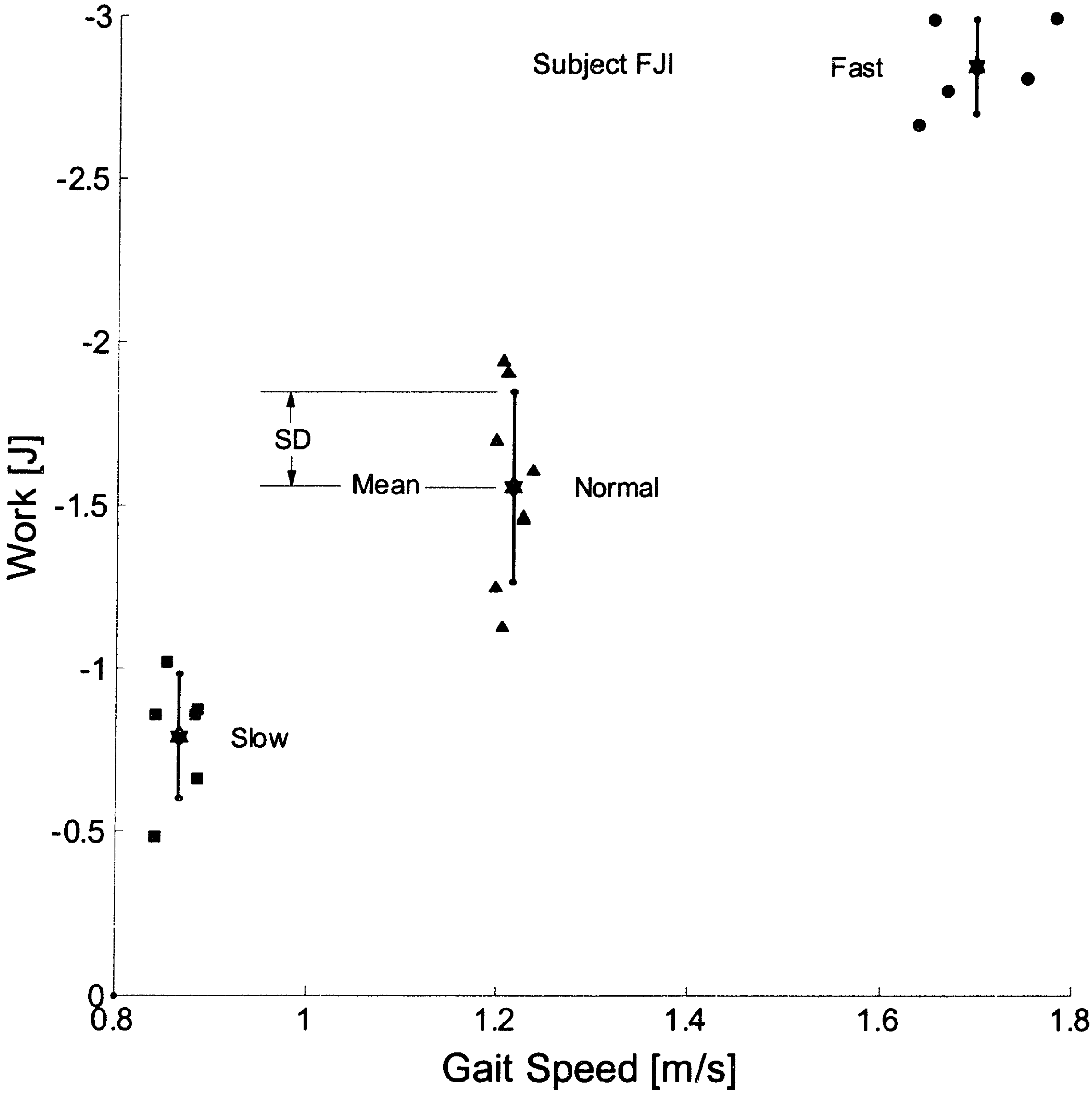

Figure 4.3 The work done by the ankle spring during controlled plantarflexion (CP) grouped by the subject's self-selected slow, normal, and fast gait speeds. Each point represents the work done by the ankle spring during CP for a single trial calculated by taking the product of the ankle spring stiffness and the change, from heel strike to foot flat, in the square of the ankle spring's displacement from its neutral position. The data for this subject are presented because of the relatively large number of trials for each self-selected speed.

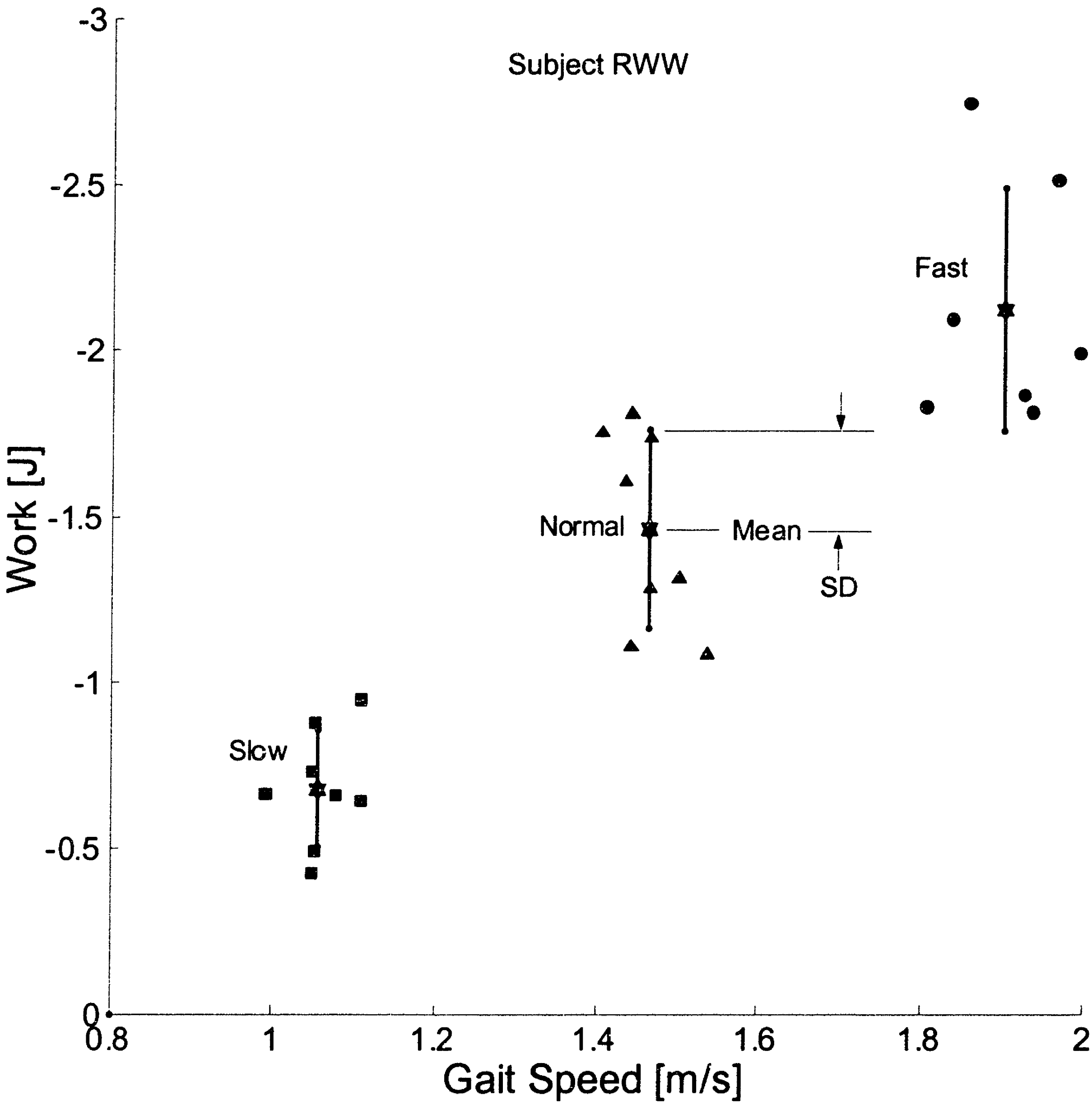

Figure 4.4 Same as Figure 3.3 but for a different

Thus four parameters that appear to vary independently of gait speed when taken one at a time were combined into a single parameter that does vary systematically with gait speed. This result was taken as evidence that stride-to-stride changes to these parameters made by an intelligent control system attempting to achieve a control objective was the more likely explanation for stride-to-stride variation of these parameters as opposed to random error in measuring or estimating them. Moreover, the trend across gait speed in the work done by the ankle spring during CP was the expected trend from a control strategy based on the hypothesized primary control objective.

Additional evidence in argument for a control system that intelligently modulates ankle spring stiffness from stride to stride can be gained through statistical considerations. Equation 4.2 can be rewritten as

$$W_{spring} = \frac{1}{2} K\Theta \tag{4.4}$$

where W_{spring} is now expressed as the product of two random variables, K and Θ. For any self-selected gait speed group, K takes on the values of the ankle spring stiffness calculated for each trial in that group, and Θ takes on the values of the change in the square of the ankle spring's displacement from HS to FF. If the large stride-to-stride variation in ankle spring stiffness was due to large random error in estimating this parameter, K would be independent of Θ as long as the estimation error and Θ are independent. If K and Θ are independent, the variance of W_{spring} (σ^2_{indep}) would be given by

$$\sigma^2_{indep} = \frac{1}{4}\left(E(K^2)E(\Theta^2) - \left(m_K m_\Theta\right)^2\right) \tag{4.5}$$

where $E(X^2)$ is the second initial moment and m_X is the mean of a random variable X (Ang and Tang, 1975).

A measure of the independence of K and Θ was made by evaluating, for each of the self-selected speed groups for all subjects, the ratio of the variance of W_{spring} given by Equation 4.5 and the variance of W_{spring} resulting from using Equation 4.2 and Equation 4.3 to calculate values for W_{spring}. This ratio can be seen as the ratio of the expected variance resulting from a large, independent estimation error to the "sample" variance. This ratio was greater than unity for 25 out of the 30 self-selected speed groups for all subjects (Table 4.3) indicating that, in the vast majority of cases, the variance of W_{spring} was less than that which would be expected if the predominant cause of stride-to-stride variability in the ankle spring stiffness was large estimation error.

The variance of W_{spring} was less than would be anticipated if K and Θ were independent because of a negative correlation between K and Θ. An analytical expression for the variance of W_{spring} given by Equation 4.4 is not easily derived, but the effect of a negative correlation between K and Θ on the variance of W_{spring} can be shown by using the linear terms of a Taylor series expansion about the means of K and Θ as a first-order approximation of W_{spring}. The variance of this first-order approximation of W_{spring} (σ_W^2) would be

$$\sigma_W^2 = \frac{1}{4}\left(m_\Theta^2 \sigma_K^2 + m_K^2 \sigma_\Theta^2 + \rho m_\Theta m_K \sigma_\Theta \sigma_K\right) \tag{4.6}$$

where σ_X^2 is the variance and m_X is the mean of a random variable X, and ρ is the correlation coefficient between K and Θ (Ang and Tang, 1975). According to Equation 4.6, if K and Θ are negatively correlated ($\rho < 0$), the effect is to reduce the variance of their product, W_{spring}. In every case where the variance of W_{spring} for a self-selected speed group was less than would be anticipated if K and Θ were independent, ρ was indeed negative (Table 4.3).

Table 4.3 Statistical considerations of the stride-to-stride variation in ankle spring stiffness during controlled plantarflexion. $\sigma^2_{indep} / \sigma^2_{sample}$ is the ratio of the expected variance of the work done by the ankle spring resulting from an independent estimation error to the sample variance. ρ is the correlation coefficient between the ankle spring stiffness and the change in the square of the ankle spring displacement.

Subject Label	Self-selected Speed	Number of Trials	$\sigma^2_{indep} / \sigma^2_{sample}$	ρ
	Slow	8	5.2	-0.86
DEF	Normal	7	5.4	-0.85
	Fast	4	4.1	-0.68
	Slow	6	3.6	-0.98
DMG	Normal	3	1.7	-0.85
	Fast	6	4.2	-0.82
	Slow	7	4.8	-0.77
EAS	Normal	8	2.3	-0.73
	Fast	2	2.6	-1.00
	Slow	5	0.7	0.44
EEB	Normal	3	0.6	0.18
	Fast	3	54.0	-0.93
	Slow	6	1.4	-0.25
FJI	Normal	8	2.0	-0.63
	Fast	5	81.7	-0.96
	Slow	7	2.2	-0.60
JLM	Normal	7	1.3	-0.66
	Fast	6	10.9	-0.83
	Slow	6	73.5	-0.71
MJK	Normal	9	3.8	-0.74
	Fast	3	7.6	-1.00
	Slow	6	1.3	-0.42
MKK	Normal	6	0.9	-0.14
	Fast	6	0.6	0.07
	Slow	6	2.4	-0.51
RAH	Normal	5	0.5	0.75
	Fast	5	3.0	-0.64
	Slow	8	2.2	-0.57
RWW	Normal	8	4.9	-0.73
	Fast	7	6.0	-0.83

This observation that the variance of W_{spring} was most often less than the variance that would result from large estimation error in calculating ankle spring stiffness was taken as more proof that stride-to-stride adjustments made by an intelligent control system were the more likely cause of the surprisingly large variation in ankle spring stiffness. Furthermore, the negative correlation between K and Θ gave insight into the control objective at a given self-selected speed. The negative correlation indicates that as the change in the square of ankle spring displacement from HS to FF increased, ankle spring stiffness decreased and vice versa. This suggests that, at a given self-selected speed, the ankle spring was tuned by the control system in order to achieve the secondary control objective of regulating the amount of work done at the ankle as the ankle position at HS and FF varied from stride to stride.

More about the hypothesized control objectives and how they interact was learned from the trend across gait speed in the angular velocity of the foot at FF. The observed trend in the angular velocity of the foot at FF was a monotonic increase from the self-selected slow speed to the fast speed for all of the subjects (Table 4.2, Figure 4.5, and Figure 4.6). The net increase from slow to fast ranged from 26% to 137% and was significant ($p < 0.05$) for all 10 subjects. This increase in the angular velocity of the foot at FF with increasing gait speed was not the trend that would be expected from a control system with the objectives of minimizing energy expenditure while keeping the foot velocity at FF bounded.

This unexpected trend across gait speed in the angular velocity of the foot at FF was taken as an indication that the original hypothesis about the primary control objective was not entirely correct and should be modified. Considering this trend, it would appear that a more accurate statement of the primary control objective would be that, during each stride, the control system attempts to *minimize* the angular velocity of the foot at FF instead of just keeping the angular velocity bounded. This change

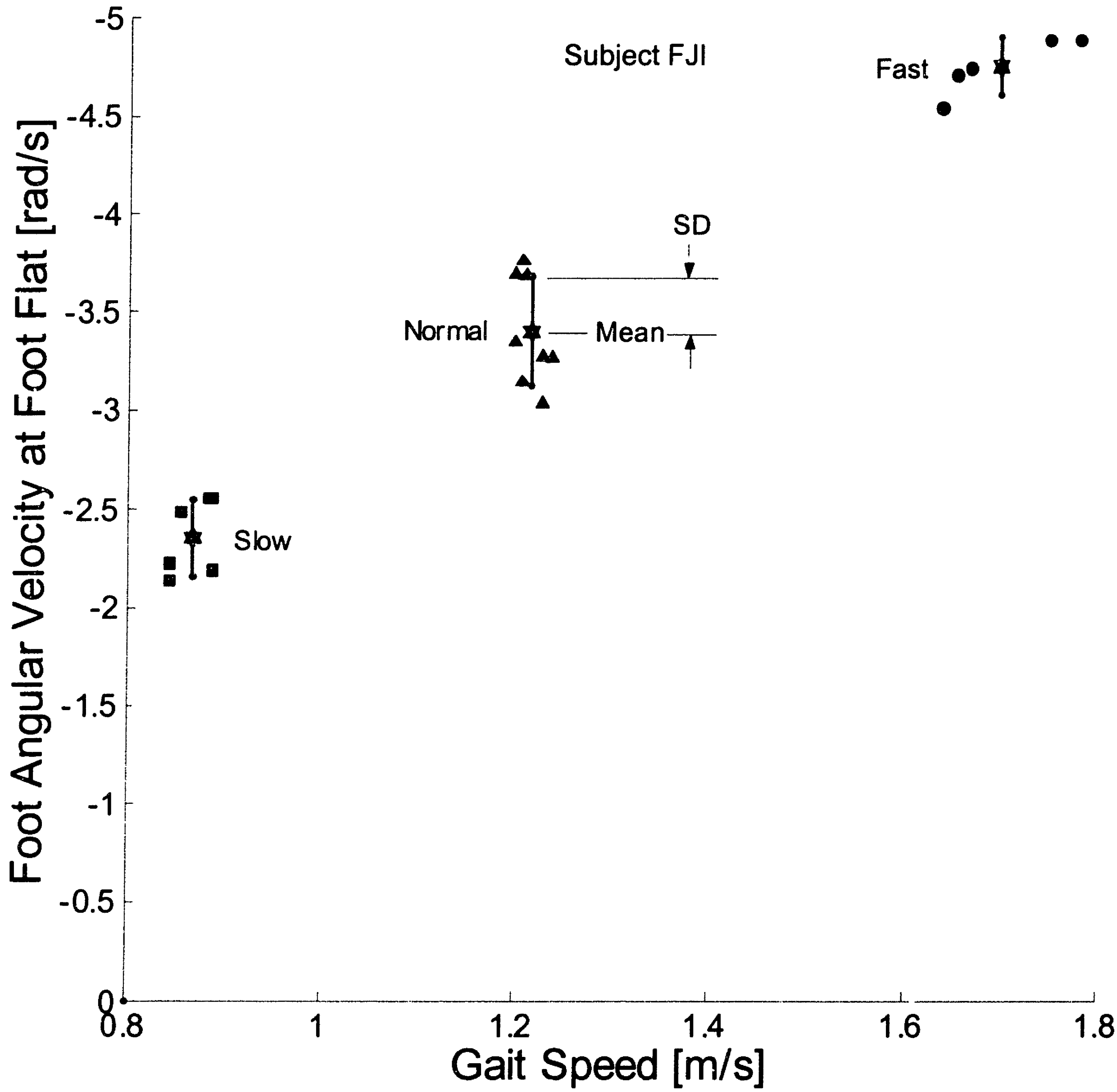

Figure 4.5 The angular velocity of the foot at foot flat (FF) grouped by the subject's self-selected slow, normal, and fast gait speeds. Each point represents the angular velocity of the foot at FF for a single trial. The data for this subject are presented because of the relatively large number of trials for each self-selected speed.

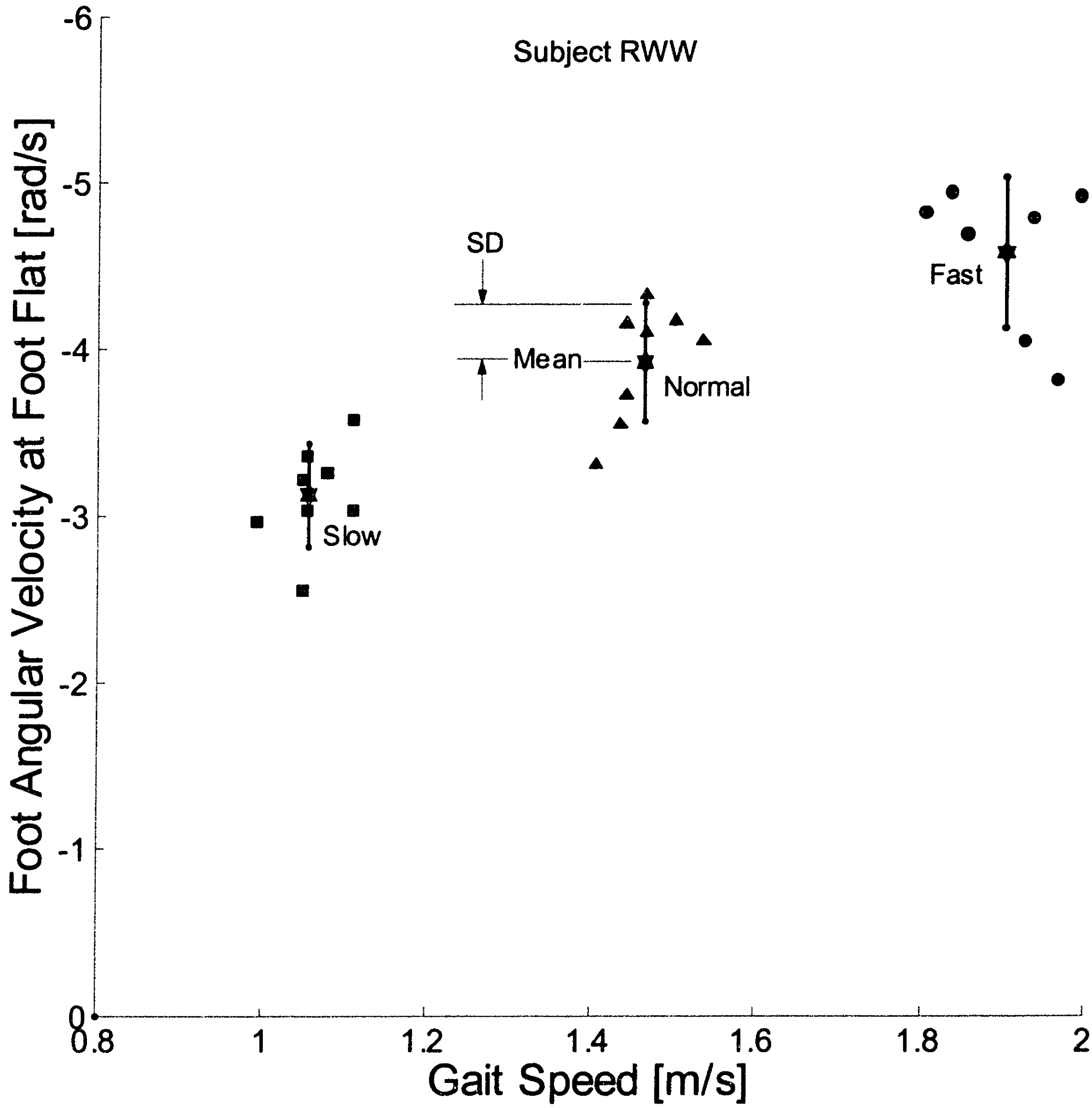

Figure 4.6 Same as Figure 4.5 but for a different subject.

to the primary control objective does not change the expected trend across gait speed in the work done by the ankle spring during CP. It only changes the expected trend in the angular velocity of the foot at FF to match what was observed.

At a given self-selected gait speed, the newly hypothesized primary control objective is fundamentally opposed to the second control objective. The primary objective of minimizing the angular velocity of the foot at FF requires as much work as possible to be done at the ankle during CP. The secondary objective of minimizing the work done at the ankle requires that the foot velocity at FF be as great as possible. The interplay between the objectives can be understood by proposing a penalty function

$$P = w_1 V + w_2 E \tag{4.7}$$

where P is the total penalty, V is the penalty associated with the foot velocity, and E is the penalty associated with energy expenditure. The weights of each term, w_1 and w_2, are positive constants and are not necessarily equal. V increases as the magnitude of the foot velocity at FF increases, and E increases as the magnitude of the work done at the ankle increases. The control system finds the optimal solution to the two opposing control objectives by minimizing P.

An understanding of how a compromise between the two hypothesized control objectives is reached through minimizing P and an explanation of the observed trends in the work done by the ankle spring and the foot velocity at FF can both be gained by considering what happens to V and E with an increase in gait speed. For reasons already explained, the work done at the ankle required to minimize the foot velocity at FF increases as gait speed increases. Thus minimizing V at a greater gait speed demands that more work be done at the ankle, but this demand for more work done at the ankle is tempered by E, the term of the penalty function that penalizes increased energy consumption. P is minimized at the greater gait speed when any decrease in V achieved by increasing the work done at the ankle would be outweighed by the increase in E. The net result is that work done at the ankle does indeed increase with an increase in gait speed, but it does not increase enough to keep the foot velocity at FF constant. Consequently, the angular velocity of the foot at FF also increases with an increase in gait speed.

In summary, a surprising amount of stride-to-stride variability in the ankle spring stiffness was observed and no consistent trend across gait speed was seen in the mean ankle spring stiffness. Trends across gait speed that held for every subject were observed in the mean work done by the ankle spring during CP and the mean angular velocity of the foot at FF. Based on inferences made from these trends, it was concluded that modulation of the ankle spring stiffness by an intelligent control system was the more likely cause of the stride-to-stride variability in the ankle spring stiffness as opposed to error in estimating this parameter. Additionally, the intelligent control strategy was defined in terms of two plausible control objectives that provide a possible explanation of how the ankle system is adapted to changes in gait speed. Finally, it was also observed that adaptation to gait speed was achieved both by changing the input to the ankle system during CP (ankle angular position) and by tuning the ankle system's parameters (ankle spring stiffness and neutral position) to get the desired system response to the input.

4.4 Characterization of Ankle Function During Controlled Dorsiflexion

Ankle function during controlled dorsiflexion (CD) was characterized by a nonlinear, hardening, torsional spring for all self-selected speed groups. The power was predominately negative during CD (Figure 2.2) so only passive elements were considered in modeling the ankle. The scatter plots of ankle angular velocity versus ankle torque (Figure 3.4) suggested that velocity-dependent effects on ankle torque caused by any damping elements were negligible. This conclusion was supported by plotting the residual from simple linear regression of ankle torque on ankle position against ankle velocity (Figure 3.6) and by the fact that the mean phase difference between the maximum ankle torque and the maximum ankle angular position was rarely positive and never greater than 8% of the mean duration of CD.

After determining that ankle function was dominated by a springlike element, it was concluded that a simple linear model was not adequate to characterize the springlike element. Simple linear regression of ankle torque on ankle angular position resulted in values of the coefficient of determination as low as 0.50. But even when the coefficient of determination was near 0.90, the pattern of the scatter plot of the residual versus the ankle position indicated that a simple linear model was not adequate. The ratio of the mean number of times the residual changed sign to the number of data points used for regression ranged between 0.03 and 0.12. Values of this ratio closer to 0.50 would be expected if the residual were a normally distributed in a band of constant width centered at zero. Hence the simple linear model was deemed inadequate and the springlike element during CD was classified as nonlinear.

The characterization of the ankle spring during CD can be taken beyond simply stating that it is nonlinear. The slope of the ankle torque versus ankle angular position curve for each self-selected gait speed group increased with increasing position (Figure 4.7), or in other words the spring became more stiff as it was displaced.

An understanding of why the ankle spring would tend to harden during CD can be arrived at by considering the time course of the center of pressure as well as the magnitude of the ground reaction force during CD. During the latter part of CD, the heel rises off the ground, the center of pressure moves more distally along the foot, and the magnitude of the ground reaction force increases (Czerniecki, 1988). The result is that the torque about the ankle generated by the ground reaction force

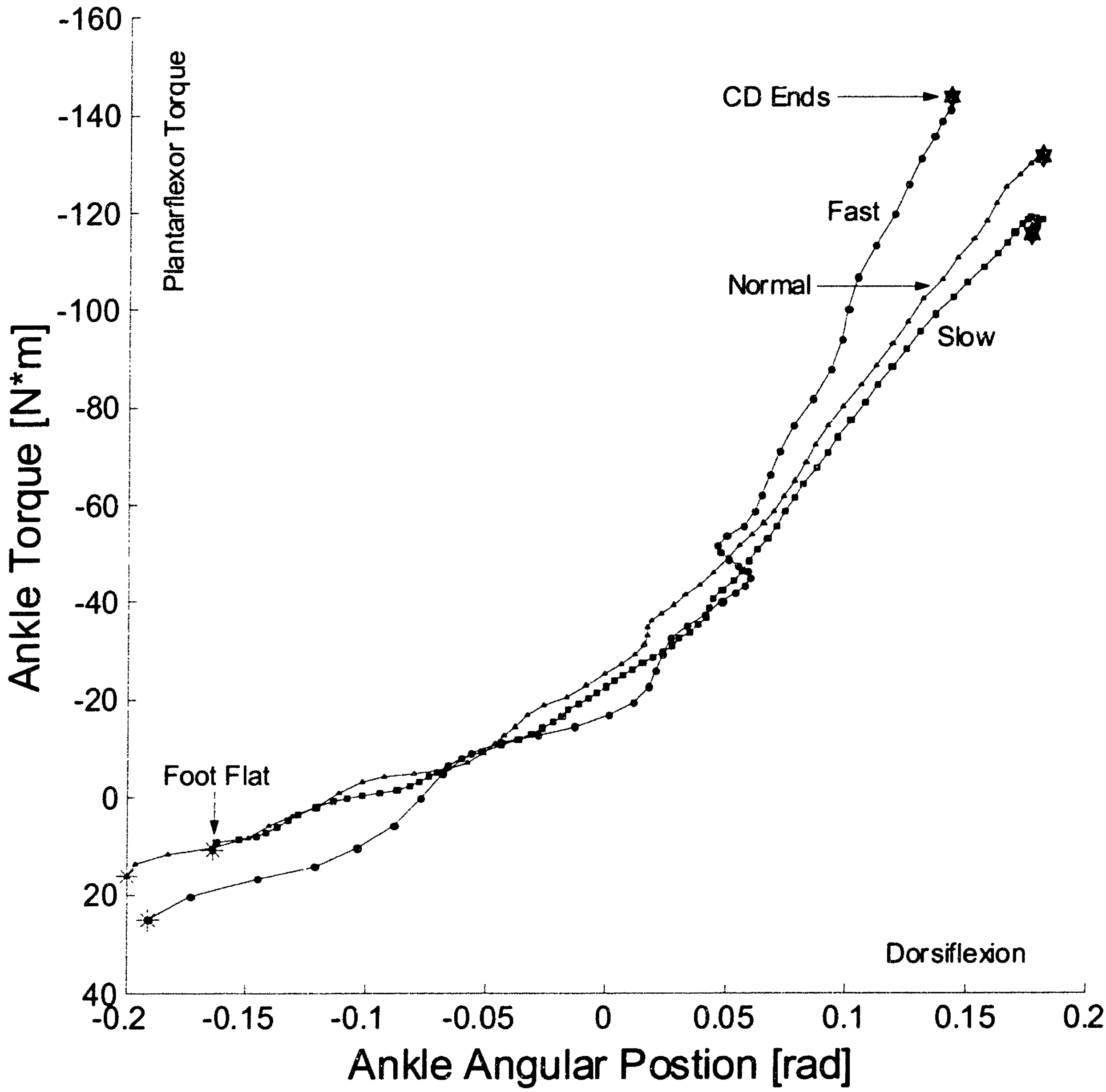

Figure 4.7 Ankle torque versus ankle angular position during controlled dorsiflexion (CD) for the subject's self-selected slow, normal, and fast gait speeds. The curve for each self-selected speed was obtained by first normalizing stride time to one and then ensemble averaging the data from all the subject's trials at that self-selected speed. Data shown are for the subject FJI and are characteristic of the torque/position curves that would be expected from a hardening spring.

increases not only because the force is increasing but also because the lever arm is increasing. The ankle spring becomes more stiff as it is displaced in order to match this increase in the torque due to the ground reaction force.

The large negative values of the phase difference between the maximum ankle torque and the maximum ankle position during CD are an indication that the assumption of a completely passive system was not appropriate in these cases. As was discussed in the Chapter 3, large negative values of the phase difference occurred when the ankle position did not increase monotonically. Instead the position would reach a local maximum at around 20% of the gait cycle, decrease, and then increase to another local maximum near 45% of the gait cycle. During this time the ankle velocity necessarily changed signs, but the ankle torque did not change signs. This means that for a period of time the ankle power was positive contrary to the assumption that the power was always negative during CD. However, passive elements can do positive work so this alone was not an indication that the assumption of a completely passive system was violated.

Showing that an active element would be needed to characterize ankle function on occasions when ankle position did not increase monotonically also requires consideration of how the magnitude of the ankle torque increased. In these cases, the magnitude of the ankle torque versus time curve would also have two peaks corresponding to the peaks in the position curve, but the second peak would always be the greater than the first. The large negative values of the phase difference occurred when the first peak of the position curve was greater than the second. If the passive mechanics of the ankle are characterized as being springlike, the only explanation for how the ankle torque could increase when ankle position actually decreased is that there was an active element generating torque about the ankle.

4.5 Characterization of Ankle Function During Powered Plantarflexion

Ankle function during powered plantarflexion (PP) must be characterized, at least in part, by an active torque actuator for all self-selected speeds. This conclusion was based on the fact that, in most cases, the positive work done at the ankle during PP was found to be greater than the amount of positive work that even a 100% efficient passive system could do using energy that was absorbed and stored before PP began.

The total amount of energy that a passive ankle system could store and use to do positive work during PP is the amount of energy stored in the ankle system at heel strike (HS) plus any changes to that amount of stored energy given by the work done at the ankle before PP begins. It has already been proposed that the ankle system be modeled as a passive spring during both controlled plantarflexion (CP) and controlled dorsiflexion (CD). Thus the amount of energy stored in the ankle system at HS was estimated by calculating the amount of energy stored in the spring used to characterize ankle function during CP. The mean work done at the ankle before PP began (i.e., the sum of the work done at the ankle during both CP and CD) was always negative indicating that the total amount of energy that could be stored in the ankle system had increased since HS. The estimates of the energy stored in the ankle system at HS were less than or equal to 0.2 J for 97% of all the trials for which an ankle spring stiffness and neutral position were calculated using linear regression. Since 0.2 J was always less than 7% of the mean work done at the ankle before PP began, the energy stored in the ankle system at HS was neglected. Therefore, assuming that the energy absorbed by the ankle spring during CP could, in some ingenious way, be stored to do positive work during PP and assuming that the ankle spring that absorbed energy during CD returned all that energy during PP (i.e., assuming there is no energy stored in the ankle system when PP ends), the work done at the ankle before PP began represents the total amount of energy that a passive ankle system could store and use to do positive work during PP.

Even if the passive ankle system model is considered to be 100% efficient, the amount of positive work that was done during PP was greater than what could be done by the passive system alone. The magnitude of the mean negative work done at the ankle before PP was less than the magnitude of the mean positive work done at the ankle during PP in most cases. The energy shortfall must have been made up by a torque actuator.

The inadequacy of a purely passive system in characterizing ankle function during PP was even more apparent as gait speed increased. The mean work done at the ankle during PP was observed to monotonically increase with gait speed. This trend agrees with that reported by Winter (1983). The mean amount of negative work done at the ankle before PP began decreased from the self-selected slow to the fast speed for all 10 subjects although this decrease was not always monotonic. The decrease was significant ($p < 0.05$) for 7 of the subjects. The result was that the shortfall between the maximum amount of energy that a passive ankle system could deliver and the total amount of work done at the ankle during PP increased with gait speed. Consequently, the torque actuator was required to do more work as gait speed increased. The amount of work that would have to be done by a torque actuator during PP at the self-selected fast speed, even if it was assisted by a 100% efficient passive system, was not negligible (ranging from 12.7 to 29.4 J).

4.6 Concluding Remarks on Applying the Results

The many people who choose to use prosthetic foot-ankle systems made from carbon-fiber springs will not be surprised to learn that human ankle function can be characterized by a spring throughout the greater part of the stance phase of walking. But some of the utility of what has been learned from this research can be demonstrated in applying it to measure the performance of commercially available prosthetic feet against that of a biological ankle.

The first differences between the performance of commercially available prosthetic feet and a biological ankle were seen in the results from controlled dorsiflexion. It was shown that human ankle function is dominated by springlike behavior during controlled plantarflexion (CP). Thus prosthetic feet that are characterized by viscoelastic behavior during CP would not perform the same as a biological ankle. It was also seen that, even when gait speed was constant, the stiffness of the spring characterizing ankle function was actively adjusted from stride to stride. It was hypothesized that the motivation for these adjustments in ankle spring stiffness was to make the process of CP smooth and energy efficient for every step. According to this hypothesis, even a prosthetic foot that acts as a spring during CP lacks the ability to ensure that CP is smooth and energy efficient for every step if the spring stiffness cannot be actively adjusted.

One of the major performance issues that arise from replacing a biological ankle with a prosthesis that acts as a passive spring in dorsiflexion can be seen in the results from powered plantarflexion. It was demonstrated that even the most efficient of passive springs could not put out as much power during powered plantarflexion as a biological ankle. Additionally, this shortfall would be apparent at self-selected normal gait speeds.

Thus the results of this study point to discrepancies between the performance of a biological ankle and the performance of commercially available prosthetic feet. These discrepancies are either compensated for by other members of the prosthetic user's lower limbs or are simply tolerated. The compensation required to overcome a given discrepancy could be minimal or that discrepancy may be easily tolerated. Consequently, it is left to the prosthesis designer to decide whether or not these specific performance issues present problems that should be addressed. It suffices to have shown that the simple model presented here for characterizing human ankle function during the stance phase of walking has led to an understanding of ankle function that could motivate future artificial ankle designs.

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