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Orthopedic Biomechanics of Pathologic Gaits

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ORTHOPEDIC BIOMECHANICS OF PATHOLOGIC GAITS

Peter Mitchel Quesada

ABSTRACT

Orthopaedic pathology is often accompanied by gait abnormalities. Investigations relating to gait associated with selected pathologic conditions are described here.

The proposed hypothesis that excessive ground reaction loading rates predispose to tibio-femoral osteoarthrosis was evaluated by measuring vertical ground reaction loading rates for osteoarthrotic subjects, who were asymptomatic following arthroscopic debridement. The hypothesis was not born out; and no relationships between loading rate and pre-heelstrike velocity or average walking velocity were detected.

External moments about the knee were measured for subjects with anterior cruciate ligament (ACL) reconstructions. Injured limb, external flexion moments at midstance were significantly lower than those for control subjects and those for the contralateral, uninjured limbs. It's likely that lower net quadriceps reactions are being exhibited to effect lower tensions in the ACL reconstructions, and to reduce anterior, tibial displacement. Net quadriceps reactions, however, were positive rather than negative, as previously demonstrated in a majority of ACL deficient patients. Tibial and vertical ground reaction loading rates were lower in the reconstruction subjects. Thus, no support was provided for the previously mentioned hypothesis for the etiology of osteoarthrosis despite

high a prevalence of osteoarthrosis among patients with ACL injuries.

Knee joint proprioception was examined for a group of above-knee (AK) amputees, using a specially designed testing apparatus. Subjects reproduced prosthetic limb knee angle deflections between approximately 5° and 25°, as well as they reproduced such sound limb motions. Onset of prosthetic limb motion, however, was not detected as well as was sound limb motion onset. The difference in the two tests' results indicates that they are evaluating different neural mechanisms. The association of greater hip angle deflections in reproduction testing suggests that AK amputees make use of hip motion cues when available to aid knee joint proprioception.

A proposed prosthetic foot/ankle system was analyzed to assess its energy return capacity. Motion and ground reaction data of a normal subject during stance phase were used as input. Energy storage and motion of the system were calculated. Peak energy storage was greater than previous single-axis and greissinger system measurements, but less than normal ankle energy generation.

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INTRODUCTION

A wide variety of measurements have been used to characterize human gait. Basic measurements of temporal parameters, ground reaction forces, and electromyograms (Emgs) have been collected by various investigators to identify ranges of these parameters for normal subjects (51,79). Studies involving subjects with orthopedic pathologies have been performed which have compared many of these parameters for such subjects with those for normal subjects (1,19,87). Ranges of normal values for many of these basic gait parameters are large; consequently, it is, generally, only for subjects with more dramatic pathologies that differences in such basic parameters can be detected. Such pathologies, however, do not require measurement of such gait parameters to suggest or confirm their presence or to evaluate patients' responses to treatments.

The use of video motion analysis techniques, often in conjunction with ground reaction force data, has made possible more sophisticated measurements, such as joint angles and moments (7,69). Further analysis of ground reaction force data as well, beyond determination of local maxima and minima of the vertical, fore-aft, and medial-lateral components, can yield such information as center of pressure and ground reaction loading rates (13,85). Deviations for some pathologic groups in measurements of these types from those on normal subjects can be significant without the "by eye" appearance of such gaits differing markedly from normal subjects' gaits.

In the next two chapters, investigations of the gaits of tibio-femoral osteoarthrotic subjects following arthroscopic debridement, and anterior cruciate ligament injured subjects following surgical reconstruction have been described. Vertical ground reaction loading rates, pre-heelstrike velocities, and free walking velocities were examined for the osteoarthrotic subjects and compared with values measured for an age-matched group of control subjects. For the anterior cruciate ligament reconstruction patients, these measurements, in addition to calculations of external knee joint moments during midstance and following heelstrike, as well as the tibially directed component of the ground reaction loading rate were obtained, and were also compared with the results from an age-matched group of controls.

In addition to altering gait, orthopedic pathologic conditions can also affect an individual's joint proprioception (i.e. his ability to sense joint position and motion without external cues such as visual or auditory inputs). Much of the study of joint proprioception has been devoted to differentiating the contributions of joint capsular and extracapsular receptors (50,66). In a number of these studies, proprioception measurements have been recorded for subjects with conditions involving loss of anatomic structures (e.g., total joint arthroplasty, anterior cruciate ligament injury) in order to assess the proprioceptive contributions of remaining structures (30,73,5). Typically, proprioception has been assessed by quantifying an individual's ability to reproduce a prescribed joint movement, or to detect very slow passive joint motion.

The study described in the fourth chapter focusses on lower limb joint proprioception in above-knee (AK) amputees. For the subjects' prosthetic limbs, measurements of passive motion reproduction and detection of slow passive motion were recorded and compared with those of their sound limbs. The results of these comparisons, as well as the relationships between the prosthetic limb measurements and other variables were analyzed to assess proprioceptive mechanisms for AK amputees.

Biomechanical measurements of normal subjects can also be used as a basis for analyzing designs of devices required for ambulation with certain pathologies. This is particularly appropriate for lower limb prosthetic devices, which many amputees rely upon to replace the function of the missing portion of their limb (86,62). Measurements of normal gait, in conjunction with design criteria, can be used to determine specifications for the components of prosthetic designs.

The last investigation presented here is an analysis of a proposed design for a prosthetic foot/ankle system for lower limb amputees intended to provide greater energy return during the push off portion of gait. The analysis uses the lower limb motion from heelstrike to maximum plantar flexion and the ground reaction forces throughout stance, measured for a normal subject, as the system input, and calculates the motion and energy storage of the system during stance phase.

PEAK GROUND REACTION LOADING RATES WITH TIBIO-FEMORAL OSTEOARTHROSIS

INTRODUCTION

The terms osteoarthrosis and osteoarthritis are often used synonymously by orthopedists. These terms, however, can be differentiated by adhering to more narrowly formulated definitions (57). Osteoarthrosis has been used to describe a mechanical breakdown of the joint tissues resulting from an imbalance between the mechanical stresses imposed on a joint and the ability of the joint tissues to withstand such stresses. Osteoarthritis, rather, denotes an inflammation of the joint (57).

Primary mechanical damage, osteoarthrosis, of a joint can lead to secondary inflammation, or osteoarthritis. Likewise, a primary osteoarthritis can change the distribution of load to the joint leading to a secondary osteoarthrosis. As a result, many patients will suffer from both conditions upon presenting to a physician, perhaps contributing to the lack of distinction between the two terms.

The hypothesis has been proposed that primary tibio-femoral osteoarthrosis may be due to excessive peak loading rates at the heelstrike portion of gait (85). Peak loading rate is meant to refer to the rate of increase of the vertical ground reaction force, rather than the magnitude of the force itself. A number of studies have demonstrated results which tend to support such a hypothesis.

In a study using the knees of sheep as a model, two groups of sheep were investigated (58). One group was walked daily on a concrete surface and housed in an area with a tarmac floor. The other group was walked daily on a surface covered with wood chips, which would tend to attenuate the rate of loading at heelstrike, and was pastured. Upon examination of the knees of the two groups two and a half years later, the group walking on concrete was found to have developed significant changes in articular cartilage and subchondral bone at the knee joint consistent with osteoarthrosis.

Another study used the knees of two groups of rabbits as a model (84). Each rabbit was put through a daily protocol in which one of their knees was immobilized in a splint and axially loaded cyclically. The immobilized knees of one group was subjected to a load amplitude of 58% of body weight, administered at a loading rate of 840 N/sec. The other group received a load amplitude of 83% of body weight applied at a loading rate of 700 N/sec. Thus, the group with the higher load amplitude had the lower loading rate. Upon subsequent examination of these knees, none of the rabbits from the group with the higher load amplitude and lower loading rate developed osteoarthrotic conditions. Conversely, almost half of the lower load amplitude, higher loading rate group sacrificed at 3-6 weeks showed evidence of osteoarthrotic changes, and 90% of these rabbits sacrificed at 9 weeks demonstrated signs of osteoarthrosis.

In a third study a group of human subjects with "presumed early stage tibio-femoral osteoarthrosis" was composed of persons with activity related knee pain, but who were without pain during testing, and had a normal appearing gait (85). Additionally, the subjects had negative radiographic findings, no history of lower extremity trauma or surgery, and no evidence of rheumatoid arthritis. These subjects were instructed to walk across a walkway at their natural velocity. The starting point on the walkway for each subject was adjusted so that the subject would strike a force plate located within the walkway. Recordings were made with both the affected and non-affected limbs striking the force plate. A control group of subjects with "normal" knees was put through the same protocol. The results demonstrated a significantly greater rate of vertical loading for the affected knees than for the control knees. This study relied in part on the assumption that the "pre-arthrotic" subjects would in fact develop tibio-femoral osteoarthrotic conditions at some future time.

In the investigation presented here the vertical ground reaction loading rate, "pre-heelstrike" vertical foot velocity, and the free walking velocity were measured in a group of tibio-femoral osteoarthrotic subjects with arthroscopically documented and debrided joint tissue damage in one of their knees, and in a group of age-matched control subjects with no history of lower extremity pathology. Osteoarthrotic subjects for this study were chosen on the basis of having a "good" or better clinical result of their arthroscopy, such that the individual can now ambulate asymptomatically. A post-arthroscopic debridement osteoarthrotic group was selected in order to obtain a group of subjects who, with the relief of pain during walking, could reasonably be expected to return to their gait patterns prior to the onset of osteoarthrotic conditions.

METHODS

Measurement Considerations

It has been previously demonstrated that the rate of increase of the vertical ground reaction load at heelstrike increases monotonically with the average walking velocity (75,17). Basic physics dictates, however, that as force is proportional to acceleration, i.e.

$$F = ma \tag{2.1}$$

where, F is force, m is mass, and a is acceleration, then the rate of change of force should depend upon the rate of change of acceleration or

$$\dot{F} = m\dot{a}. \tag{2.2}$$

Considering only the vertical components yields,

$$\dot{F}_z = m\dot{a}_z. \tag{2.3}$$

Difficulty in measuring the rate of change of acceleration hampers any effort to directly assess its relationship to ground reaction loading rate. Reflective tape markers placed directly on bony landmarks can be used to record the positions of landmarks at specific time intervals. Calculation of the rate of change of acceleration would then, however, require three numerical differentiations to calculate the rate of change of acceleration during heelstrike. Although the error, or noise, in position measurements can be less than 1 mm in some video motion analysis systems, the magnification of the noise after three numerical differentiations can still be very substantial.

The motion of accelerometers and the skin to which they are attached relative to the underlying tissues calls into question the validity of such measurements. Light, et al have previously measured lower limb accelerations at heelstrike by attaching accelerometers via kirschner wires (1.14 mm diameter) directly to tibial bone (40). For a given footwear condition (such as barefoot), duration and shape of the deceleration phase immediately following heelstrike was observed to be consistent; consequently, the greater the limb velocity at heelstrike, the larger the peak deceleration that would be required to bring the limb to rest within the deceleration phase duration. Larger peak deceleration rates would then be necessary to achieve the greater peak decelerations during the deceleration phase. This would seem to be consistent with the findings of Radin, et al that the heelstrike transient was related to the angular velocity of the shank at heelstrike (59). Some measure of lower limb velocity just prior to heelstrike, thus, seems to provide a measure of the rate of change of acceleration as well.

When calculating ground reaction loading rates from force plate data, the sampling rate at which ground reaction data is acquired should be considered. While larger amplitude frequency components of the ground reactions at heelstrike have been generally found below 10-20 Hz for normal subjects at comfortable walking velocities, some smaller amplitude frequency components (approximately 1% of the fundamental amplitude) have been reported in the range of 50-75 Hz (71,2). Consequently, to avoid aliasing of components with amplitudes of about 1% of the fundamental, sampling rates in the 100-150 Hz range would be needed. For reasonably normal gaits, such sampling rates should allow for one numerical differentiation of the ground reactions to obtain loading rates. For subjects with markedly abnormal gaits or at very different walking velocities, the frequency content may be noticeably increased; and similarly increased sampling rates may, thus, be required.

Subjects

Twelve tibio-femoral osteoarthrotic subjects (seven male and five female) and sixteen control subjects (nine male and seven female) were recruited for this study (see Tables 2.1 and 2.2). The osteoarthrotic group had a mean age of 62.9 ± 9.7 years, and ranged from 45 to 77 years. The average age of the control group was 65.6 ± 8.5 years, ranging from 48 to 80 years. The mean weight of the osteoarthrotic group was 89.3 ± 25.9 kg with a range of 49.9 to 131.0 kg. The control group had an average weight of 75.5 ± 12.9 kg, ranging from 55.3 to 95.2 kg. The osteoarthrotic subject group was drawn from patients from the United States Naval Hospital in Oakland, California, who had undergone arthroscopic debridement between October, 1982 and January, 1986. These subjects were recruited on the basis of having a "good" or better clinical result of their arthroscopy. After obtaining informed consent, a medical history of each subject was obtained to identify any disqualifying gait limitations such as rheumatoid arthritis, neurological disorders, cardiovascular or respiratory disease. The subjects reported no limitation in their daily activity; and each subject had the ability to walk at least one mile without discomfort. The control subjects were assembled to approximately age-match the osteoarthrotic group. Additional criteria for the control group included having no history of lower extremity trauma or surgery, and having no other condition affecting ambulation (e.g. neurologic, cardiovascular, or respiratory problems).

Table 2.1. Summary of demographic data on osteoarthrotic subjects.

Subject	Gender	Age (years)	Weight (kg)
1	F	77	63.9
2	M	56	93.9
3	F	75	85.7
4	F	71	68.5
5	F	62	55.3
6	M	65	95.7
7	M	66	123.8
8	M	59	131.0
9	M	45	110.2
10	M	52	91.1
11	M	71	102.0
12	F	56	49.9
Mean		62.9	89.3
Std. Dev.		9.7	25.9

Table 2.2. Summary of demographic data on control subjects.

Subject	Gender	Age (years)	Weight (kg)
1	F	73	64.6
2	F	62	58.5
3	F	63	64.4
4	F	66	63.9
5	M	80	55.3
6	F	65	66.7
7	M	69	88.6
8	F	65	71.2
9	F	54	76.6
10	M	65	95.2
11	M	77	73.2
12	M	48	89.3
13	M	69	90.7
14	M	53	88.0
15	M	74	72.1
16	M	65	88.6
Mean		65.6	75.5
Std. Dev.		8.5	6.1

Motion and Ground Reaction Data

Data for the subjects were collected in the gait laboratory at the Oakland Naval Hospital using a three camera VICON video motion analysis system (Oxford Metrics, Inc., Tampa, FL), an AMTI force plate (Newton, MA), and a Rancho Los Amigos stride analyzer. The VICON cameras were placed on one side of a raised platform, and the force plate was located in the middle of the platform.

After recording the age and weight of each subject, a circular patch of retroreflective tape was placed on each lateral malleolus of every subject. The lateral malleolus was chosen because it was a prominent bony landmark which is easily reproduced and has a relatively small amount of soft tissue between skin and bone. Each subject also wore a waist belt with the Rancho Los Amigos battery and signal pack, and an arm band with a light sensor to trigger the stride analyzer to measure the free walking velocity.

Each subject was then instructed to walk across the raised platform at his natural velocity while looking straight ahead and not watching the force plate. A few practice walks were taken by each subject to get comfortable walking with the experimental setup, and in order to determine the appropriate starting point on the platform such that the subject would strike the force plate with the leg wearing the marker that was visible to the VICON cameras. Such adjustments were made in order to prevent the subject from "targeting" the force plate in order to strike it. Such "targeting" could result in a shorter or longer stride prior to heelstrike which would not be representative of the subjects usual gait. For each subject six trials were recorded, three with each foot striking the force platform, and in view of the cameras.

The motion of the marker and the ground reaction forces were acquired synchronously at 200 Hz on a DEC micro-PDP-11 (Digital Equipment Corporation, Nashua, NH). From the subject group, only data from the leg with documented tibio-femoral osteoarthrosis was analyzed because the contralateral knee had not been evaluated by arthroscopy. All data from the control group was considered.

From the ground reaction force data, the maximum rate of change of the vertical reaction force was determined as the maximum change between consecutive samples divided by the sampling time (i.e. $\dot{F}_{z,max} = \Delta F_{z,max}/\Delta t_{samp}$, where $\dot{F}_{z,max}$ is the maximum rate of change of the vertical ground reaction force, $\Delta F_{z,max}$ is the maximum increase in the vertical ground reaction force between consecutive samples, and Δt_{samp} is the time between samples) (see Figure 2.1) (16). This quantity was also normalized by dividing by subject weight.

The marker trajectories were then tracked in order for the VICON system software to be able to calculate the marker positions and velocities. Subsequent to these VICON software calculations, all data was converted to ASCII format and then transmitted via a serial line to a PC compatible computer. All further data analysis was conducted on PC compatible machines. Since the software employed an algorithm which for a given frame used the marker positions at the previous and succeeding three frames to calculate the marker velocity, the velocity determined four frames (approximately 20 milliseconds) prior to the onset of force platform data was selected as the "pre-heelstrike" velocity.

Since laboratory space restrictions did not allow for the Rancho Los

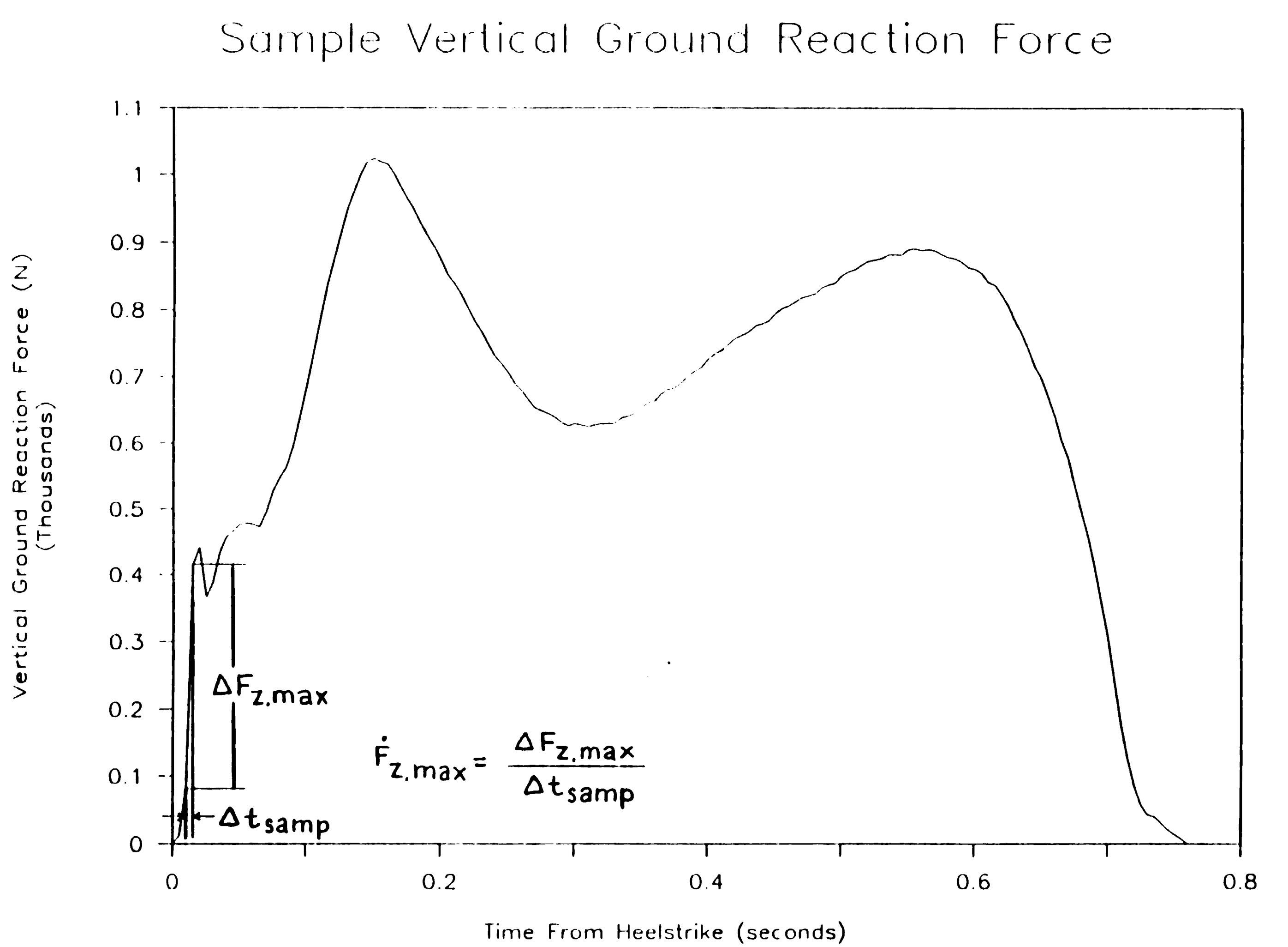


Fig. 2.1. **Sample plot of vertical ground reaction force versus time, with a graphical demonstration of the maximum loading rate.**

Amigos light triggers to be separated by the fully prescribed spacing, the free walking velocities obtained were multiplied by an appropriate correction factor in order to calculate the actual average free walking velocities. The calibration coefficients provided by the manufacturer were verified by using the force plate to measure the weight of an individual standing on it; and then checking the measurement with one obtained with a standard hospital scale.

Statistical Analysis

Statistical analysis was performed on a PC-compatible computer using the PC version of the Minitab statistical package (Minitab, Inc., State College, PA). Multivariate analysis was performed to compare the maximum vertical ground reaction loading rates, vertical pre-heelstrike velocities, and average free walking velocities of the osteoarthrotic group to those of the control subjects. Multiple regression was used to assess the relationship between loading rate, pre-heelstrike foot velocity, and free walking velocity.

RESULTS

The mean maximum rate of increase of the vertical ground reaction for the osteoarthrotic subjects was 49.7 ± 18.6 BW/sec (see Table 2.3). The maximum rate of vertical loading for the control subjects was 48.7 ± 23.0 BW/sec (see Table 2.4).

The osteoarthrotic subjects had a mean pre-heelstrike velocity of 21.1 ± 10.5 m/min (see Table 2.3). Pre-heelstrike velocity for the

Table 2.3. Summary of experimental results of osteoarthrotic subjects.

Subject	Maximum Vertical Loading Rate (BW/sec)	Pre-heelstrike Vertical Velocity (m/min)	Average Free Walking Velocity (m/min)
1	22.7	27.2	53.5
2	74.8	30.0	85.2
3	29.3	19.3	60.3
4	62.6	13.6	81.8
5	61.9	25.0	80.0
6	57.5	30.3	69.3
7	69.3	17.6	72.8
8	50.6	29.3	64.2
9	57.8	30.5	88.9
10	37.1	12.1	62.6
11	29.6	10.5	59.6
12	43.1	8.3	70.4
Mean	49.7	21.1	70.7
Std. Dev.	18.6	10.5	11.2

Table 2.4. Summary of experimental results of osteoarthrotic subjects.

Subject	Maximum Vertical Loading Rate (BW/sec)	Pre-heelstrike Vertical Velocity (m/min)	Average Free Walking Velocity (m/min)
1	47.5	13.7	60.6
2	30.2	10.2	54.7
3	49.9	20.4	68.0
4	40.8	13.3	71.5
5	38.5	11.6	57.4
6	44.2	23.6	73.3
7	42.1	21.4	54.1
8	16.7	16.9	47.5
9	43.6	21.3	74.7
10	58.1	26.0	77.9
11	33.3	15.2	66.7
12	65.7	24.5	73.1
13	30.4	15.8	66.2
14	89.5	20.7	83.0
15	62.2	19.0	74.4
16	87.1	20.2	79.6
Mean	48.7	18.4	67.7
Std. Dev.	23.0	8.2	10.2

control subjects averaged 18.4 ± 8.16 m/min (see Table 2.4). Free walking velocity of control subjects had an average of 70.7 ± 11.2 m/min (see Table 2.3). The control subjects' mean free walking velocity was 67.7 ± 10.2 m/min (see Table 2.4).

Analysis of variance performed using multiple linear regression with a series of dummy variables to account for repeated measures and unbalanced data indicated no significance for the differences in maximum rate of loading, free walking velocity, and pre-heelstrike velocity between the osteoarthrotic and control groups (see Table 2.5).

Multiple regression of maximum loading rate versus pre-heelstrike velocity and the previously mentioned dummy variables produced a P-value of 0.186 for the pre-heelstrike velocity regression coefficient. Similar regression with free walking velocity instead of pre-heelstrike velocity yielded a P-value of 0.451 for the free walking velocity regression coefficient. Multiple regression of pre-heelstrike velocity with free walking velocity and the dummy variables produced a P-value of 0.008 for the regression coefficient.

DISCUSSION

Although a substantial amount of recent evidence has been accumulated which links excessive loading rate with the initiation and progression of tibio-femoral osteoarthrosis (58,84,85), Much of this data involves animal models in very controlled experimental environments. Such animal studies have either speculated on the ground reaction loading rate based upon knowledge of the walking surface (58), or imposed prescribed loads at

Table 2.5. Summary of regression analysis.

Dependent variable	Independent variable	Regression coefficient	P-value
Maximum rate of loading	subject group	0.476 BW/sec	>0.05
Pre-heelstrike velocity	subject group	1.39 m/sec	>0.05
Free walking velocity	subject group	1.53 m/sec	>0.05
Maximum rate of loading	Pre-heelstrike velocity	0.219 (BW/sec)/(m/min)	0.19
Maximum rate of loading	Free walking velocity	-0.323 (BW/sec)/(m/min)	0.45
Pre-heelstrike velocity	Free walking velocity	0.666 (m/min)/(m/min)	0.008

specific rates upon statically braced limbs (84). Actual measurement of ground reaction loading rate of humans during walking has not been recorded for subjects with documented early tibio-femoral osteoarthrotic changes, but rather for subjects believed to be headed that way (85).

Measurement of gait parameters of subjects with tibio-femoral osteoarthrosis presents a problem, however. Such a condition is likely to be painful, and cause an altered gait pattern compared to the gait of the patient prior to the onset of osteoarthrotic changes. An attempt has been made here to circumvent this problem by testing patients who have undergone arthroscopic debridement and have achieved a good or better clinical result such that they can ambulate asymptomatically. The underlying assumption requires that the patients' return to their gait patterns prior to the onset of symptoms (including pain) that led to the diagnosis of and arthroscopic debridement for tibio-femoral osteoarthrosis.

The lack of any significant difference in the vertical ground reaction loading rates of the osteoarthrotic and control groups would seem to not support the hypothesis that tibio-femoral osteoarthrosis is associated with excessive ground reaction loading rate. Osteoarthrotic subjects would also not appear to differ from control subjects in their free walking velocity or their pre-heelstrike velocity. Such conclusions based upon comparison of osteoarthrotic and control subject data should be evaluated with consideration of the assumption of the patients' return to their pre-osteoarthrotic gait patterns. It seems likely that some, who might continue to support the hypothesis that excessive loading rate is associated with initiation of tibio-femoral osteoarthrosis, would contend that it is the assumption of the return to pre-osteoarthrotic gait patterns which is invalid.

A more complete assessment of the role of ground reaction loading rate may require cohort studies involving measurement of ground reactions of large numbers of subjects prior to any diagnosis of osteoarthrosis. At some point in the long term future, data from subjects who develop osteoarthrotic changes could be compared to data from subjects whose tibio-femoral joints remain healthy. Alternatively, post-arthroscopic debridement patients might be used as a cohort group for the measurement of ground reaction loading rate with the progression rather than the initiation of osteoarthrotic changes being of long-term interest.

FREE WALKING GAIT ANALYSIS FOLLOWING ANTERIOR CRUCIATE LIGAMENT RECONSTRUCTION

INTRODUCTION

The anterior cruciate ligament (ACL) has been a topic of intensive orthopedic research. The ACL is the most commonly injured major ligament of the knee (36). Long-term follow-up studies of patients with ACL deficient knees have indicated that the functional abilities of these subjects are diminished (44,45,53). A number of reports on long-term follow-ups of patients who had ACL reconstructions have been encouraging (37,77); although some complications including flexion contractures and patellofemoral pain have also been noted (65). With non-operative treatment it has been suggested, however, that a rigorous rehabilitation regime, notably one which stresses hamstring development, can improve the functional outcome of these patients (26). A high prevalence of long-term knee joint degeneration has also been reported, particularly with non-operative treatments (22,43,53).

Biomechanical analyses are useful in order to clarify the mechanisms underlying the results of clinical studies. Experimental studies with cadaveric limbs have indicated that the ACL provides significant restraint to anterior displacement of the tibia relative to the femur during quadriceps activity (9). It has also been reported that the ACL is significantly strained during quadriceps activity when the knee joint is between full extension and 45° of flexion (3). Hamstring activity alone has been demonstrated to decrease strain in the ACL relative to normal passive strains; however, co-contraction of the quadriceps and hamstrings still increases the ACL strains between 0° and 30° of flexion (60). The quadriceps force required to resist an external flexion moment has been shown to increase markedly during the last 15° of extension, indicating that there is a decrease in the effective moment arm for the quadriceps force during this motion (31).

Simple clinical procedures are often desired by physicians to assist in determining diagnoses and prognoses for knee injury patients. ACL deficient knees have quantitatively exhibited significantly greater antero-posterior displacement (as measured by a Medmetric KT-2000 knee arthrometer) than have the knees of normal subjects without history of knee injury, as well as compared with the uninjured contralateral knees (18). Knee joint stiffness has, conversely, been measured to be much less in ACL deficient knees, compared with both the knees of normal subjects and uninjured contralateral limbs (42). Clinical measures of knee stability have been reported, however, to have little relationship to the functional outcome of patients following ACL reconstruction (68,33); and it has been suggested that the development of more specific, dynamic tests may be required to obtain stronger correlations between test results and subjective, patient evaluations of knee status following ACL reconstruction.

While simulated motion studies with cadaveric limbs provide insight to the kinematic and kinetic mechanisms, and clinical measurements of displacement and knee stiffness may provide the clinician with useful diagnostic information, analyses of subjects performing useful motions are necessary to assess the dynamic functional capabilities of patients with injuries to the anterior cruciate ligament. Subjects with an ACL deficient knee have been reported to exhibit significantly lower external flexion moments about the knee at the midstance portion of gait at free walking velocities, to the point that many of the subjects tested actually maintained external knee extension moments throughout midstance (7). At free walking velocities, knee joint angles at midstance in patients with ACL deficient knees have also been demonstrated to be less (i.e., the knees are less flexed) than those in control subjects (69). Following loading at free walking velocities, the magnitude of quadriceps electromyograms (EMGs) in ACL deficient limbs have been reported to be lower than those of control subjects, while ACL deficient hamstring EMGs have been measured to be greater than control hamstring EMGs (41). Subjects with ACL reconstructions have previously been evaluated using a cutting index (intended to quantify difficulty of the cutting motion) which utilized video motion data to determine the angle of the cut and the time required to travel the last 100 inches prior to the cut, and which required ground reaction force data to measure the three components of the ground reaction force and the time spent on the force plate (77). The ACL reconstruction subjects, who had pre-operatively demonstrated a cross-cut index difference, were shown at about 2 years post-operatively to have no difference between their injured and uninjured limb cutting indices for straight and cross-cuts.

The investigation described here focussed on analyzing the free walking gait of patients following ACL reconstruction. Although a knee with an ACL reconstruction may withstand a greater load than an ACL deficient knee, the strength and stiffness of reconstructions have been found to be less than those of intact ACLs in an animal model (46). An ACL reconstruction, therefore, may not provide the same capacity for resisting anterior displacement of the tibia or for sustaining load imposed during quadriceps contractions. External moments about the knee joint at midstance and following heelstrike were calculated to provide a measure of the net muscular forces acting about the knee, and to determine whether subjects with ACL reconstructions exhibit lower external flexion moments about the knee than do control subjects. Such lower moments could indicate a mechanism employed, perhaps subconsciously, by patients to shield reconstructions from the level of loading imposed upon intact ACLs in restraint of undesired anterior tibial displacement. Since high prevalences of osteoarthrosis have been reported for ACL injured patients, ground reaction loading rates directed tibially and vertically were obtained to evaluate the hypothesis proposed by previous investigators that excessive loading rates predispose to osteoarthrosis (58,84,85). Additionally, pre-heelstrike foot velocities and free walking velocities were measured to assess their relationships with any of the other measurements (external, midstance and heelstrike knee moments, and tibial and vertical ground reaction loading rates) for which a difference between the ACL reconstruction and control subjects was detected. Previous investigations have examined the relationship between ground reaction loading rates and walking velocity, and reported loading rates to increase monotonically with walking velocity (75,17). Ultimately, this analysis was undertaken to provide some understanding of the differences between the free walking gait of ACL reconstructed patients and individuals without knee injuries, and the mechanisms behind these differences.

METHODS

Subjects

Ten male subjects with ACL reconstructions and ten male control subjects were recruited for this study. Age, weight, and height for each of the subjects are listed and summarized in Table 3.1. The ACL reconstruction group was assembled from a list of patients, between 20 and 30 years of age, from the United States Naval Hospital in Oakland, California, who had documented tears of their ACLs, had undergone surgical ACL reconstructions during the course of their treatment, had subjectively reported satisfaction with the results of their surgeries, and who agreed to participate in this study. ACL reconstruction subjects were tested between nine and fifteen months following surgery. These subjects were in the latter stages of the major portion of their outpatient rehabilitation, and were near their return from limited active duty to full active duty. At the time of evaluation, the mean subject reported percentage activity level was 75 ± 11 %; and the mean difference between injured and uninjured limb KT-1000 measured knee displacement was 2.83 ± 1.89 mm. Neither ACL reconstruction subjects nor control subjects displayed any additional disqualifying gait limitations such as respiratory, cardiovascular, or neurological disorders. Control subjects were selected, from unlimited active duty naval personnel who volunteered to enroll in this study, to approximately age-match the ACL reconstruction subjects. As military personnel, all control subjects and all ACL reconstruction subjects, prior their injuries, were subjected to periodic Physical Readiness Testing (PRT). Consequently, it was assumed that all subjects maintained some

Table 3.1. Summary of demographic data on ACL reconstruction and control subjects.

ACL/Control Number	Age (years)	Weight (kg)	Height (m)
ACL 1	30	79.4	1.74
ACL 2	21	76.2	1.70
ACL 3	22	88.5	1.88
ACL 4	28	111.1	1.92
ACL 5	29	81.6	1.74
ACL 6	26	114.8	1.87
ACL 7	23	127.5	1.93
ACL 8	23	78.5	1.78
ACL 9	23	85.3	1.75
ACL 10	26	81.4	1.78
Mean (SD)	25.1 (3.0)	92.4 (17.4)	1.81 (0.08)
Control 1	25	89.8	1.84
Control 2	20	71.2	1.71
Control 3	22	89.1	1.80
Control 4	25	82.1	1.80
Control 5	22	87.5	1.82
Control 6	21	88.9	1.80
Control 7	27	75.3	1.84
Control 8	26	75.8	1.69
Control 9	26	72.6	1.72
Control 10	26	66.5	1.75
Mean (SD)	24.0 (2.4)	79.9 (8.2)	1.78 (0.05)

level of activity consistent with being prepared for these examinations, and that activity levels were comparable between subjects. No significant differences were noted in age, weight, and height between the control and ACL reconstruction groups. Informed consent was obtained from all subjects at the time of their enrollment in this study.

Motion and Ground Reaction Data

Kinematic and kinetic gait data were acquired with a three camera VICON video motion analysis system (Oxford Metrics, Inc., Tampa, FL), and an AMTI force plate (Newton, MA), located in the gait laboratory at the Oakland Naval Hospital. The cameras were situated on one side of a raised platform, and the force plate was at the middle of the platform.

Height, weight, and age were recorded for each subject. Circular patches of retroreflective tape were placed on the greater trochanter, knee joint space, lateral malleolus, and fifth metatarsal head of each limb of every subject. Subjects were instructed to walk at comfortable free walking velocities across the raised platform directing his gaze straight ahead, away from the force plate. Practice walks were taken in order to acclimate each subject to walking under the experimental conditions, and in order for the experimenter to determine the proper starting point such that the limb with markers visible to the cameras would strike the force plate. These measures were taken to avoid "targeting" of the force plate by the subject, which could result in a short or long stride prior to heelstrike on the force plate. For each subject six trials were recorded, three with each limb striking the force plate, and in view of the cameras. All video and ground reaction data were acquired synchronously at 200 Hz, on a DEC micro-PDP-11 (Digital Equipment Corporation, Nashua, NH).

The external knee moment for each frame during which there was contact with force plate was determined as the cross product of the sagittal plane vector from the ground reaction center of pressure to the knee marker, and the sagittal plane ground reaction force vector, or

$$m_k = (c_z - k_z)F_x - (c_x - k_x)F_z \tag{3.1}$$

where, m_k is the external knee moment, c_x and c_z are the fore-aft and vertical coordinates of the center of pressure, k_x and k_z are the fore-aft and vertical coordinates of the knee marker position, and F_x and F_z are the fore-aft and vertical, ground reaction force components. The heelstrike and midstance knee moments were taken as the maximum knee extension moment following heelstrike and the maximum knee flexion moment during midstance, respectively. The moments were normalized by dividing by the product of body weight (after converting to Newtons) and height (in meters) (7).

The maximum vertical ground reaction loading rate was calculated as

$$\dot{F}_{z,max} = \frac{\Delta F_{z,max}}{\Delta t_{samp}} \tag{3.2}$$

where, $\dot{F}_{z,max}$ is the maximum vertical ground reaction loading rate, $\Delta F_{z,max}$ is the maximum increase in the vertical ground reaction force between consecutive samples, and Δt_{samp} is the sampling time (16). To provide a measure of the ground reaction force directed at the knee joint, the tibially directed ground reaction force was calculated in the sagittal plane as the scalar dot product of the ground reaction force vector and a unit vector directed from the ankle to the knee, or

$$F_{tib} = F_x u_{ak,x} + F_z u_{ak,z} \tag{3.3}$$

where F_{tib} is the tibially directed ground reaction force, and $u_{ak,x}$ and $u_{ak,z}$ are the fore-aft and vertical components of the unit vector between the ankle and knee. The unit vector components were determined as

$$u_{ak,x} = \frac{a_x - k_x}{((a_x - k_x)^2 + (a_z - k_z)^2)^{1/2}} \text{, and} \tag{3.4}$$

$$u_{ak,z} = \frac{a_z - k_z}{((a_x - k_x)^2 + (a_z - k_z)^2)^{1/2}} \tag{3.5}$$

where, a_x and a_z are the fore-aft and vertical coordinates of the ankle marker position. Similar to the maximum vertical loading rate, the maximum tibially directed ground reaction loading rate was determined as

$$\dot{F}_{tib,max} = \frac{\Delta F_{tib,max}}{\Delta t_{samp}} \tag{3.6}$$

where $\dot{F}_{tib,max}$ is the maximum tibially directed ground reaction loading rate, and $\Delta F_{tib,max}$ is the maximum increase calculated for consecutive samples in the tibially directed ground reaction force. These maximum loading rates were normalized by dividing by subject body weight.

The VICON system software was employed to obtain the marker positions and velocities from the raw video data. The vertical component of the ankle marker velocity (calculated by the VICON software) which occurred four frames (approximately 20 milliseconds) prior to heelstrike on the force plate was taken as the pre-heelstrike velocity, since the software used an algorithm which involved the previous and succeeding three frames to determine the velocity of a marker in a given frame. To determine the average free walking velocity, a motion data sampling frame during the swing phase prior to heelstrike was chosen arbitrarily. The angle of the shank was obtained as,

$$\theta_s = \tan^{-1} \frac{(k_{z,i} - a_{z,i})}{(k_{x,i} - a_{x,i})} \tag{3.7}$$

where, θ_s is the shank angle relative to vertical, $k_{x,i}$ and $k_{z,i}$ are the fore-aft and vertical coordinates, respectively, of the knee marker at the selected initial frame, and $a_{x,i}$ and $a_{z,i}$ are the fore-aft and vertical coordinates of the ankle marker at the selected initial frame. The end of the stride was the frame, during the swing phase following toe off from the force plate, at which the difference in shank angle relative to the angle for the initial frame was a minimum. The average walking velocity was then obtained as

$$v_{avg} = \frac{h_{x,f} - h_{x,i}}{\Delta t_{samp}(f - i)} \tag{3.8}$$

where, v_{avg} is the average free walking velocity, $h_{x,i}$ and $h_{x,f}$ are the fore-aft coordinates of the greater trochanter marker at the initial and final sampling frames of the stride, respectively, and i and f are numbers of the initial and final sampling frames, respectively.

Statistical Analysis

Statistical analysis was performed with the PC version of the Minitab statistical software package (Minitab, Inc., State College, PA). Multivariate analysis was used to compare the maximum loading rates, pre-heelstrike and free walking velocities, and external heelstrike and midstance knee moments of the ACL reconstruction subjects to those of the control subjects. Multiple regression analysis was used to assess the relationships between the loading rates, velocities, and moments.

RESULTS

The mean external, midstance knee flexion moment and mean external, heelstrike knee extension moment for the ACL reconstruction group's injured limbs were 2.02 ± 1.44 %BW-H and 2.69 ± 1.85 %BW-H, respectively (see Table 3.2). For the uninjured limb of the ACL reconstruction group the average midstance knee flexion moment and average heelstrike knee extension moment were 3.10 ± 1.35 %BW-H and 2.70 ± 1.25 %BW-H, respectively (see Table 3.3). The mean midstance and heelstrike moments

Table 3.2. Summary of experimental results of ACL reconstruction subjects' injured limbs.

Subject	Maximum Midstance Knee Flexion Moment (%BW-H)	Maximum Heelstrike Knee Extension Moment (%BW-H)	Maximum Tibially Directed Loading Rate (BW/sec)	Maximum Vertical Loading Rate (BW/sec)	Pre-heelstrike Vertical Velocity (m/min)	Free Walking Velocity (m/min)
1	2.90	1.12	65.4	64.9	19.6	87.0
2	1.05	2.02	26.9	28.5	18.4	57.0
3	0.60	4.03	59.9	62.2	19.5	65.3
4	3.66	1.28	25.8	26.7	20.2	62.7
5	0.62	3.03	32.1	33.4	23.8	62.0
6	3.15	7.34	114.4	120.1	35.1	89.7
7	2.48	1.80	48.5	48.7	16.7	68.0
8	1.43	1.82	34.6	35.4	19.8	64.1
9	1.24	1.96	43.6	44.6	11.7	68.2
10	3.02	2.45	51.9	54.6	25.3	76.6
Mean	2.02	2.69	50.3	51.9	21.0	70.1
Std. Dev.	1.14	1.85	26.3	27.5	6.2	10.6

Table 3.3. Summary of experimental results of ACL reconstruction subjects' uninjured limbs.

Subject	Maximum Midstance Knee Flexion Moment (%BW-H)	Maximum Heelstrike Knee Extension Moment (%BW-H)	Maximum Tibially Directed Loading Rate (BW/sec)	Maximum Vertical Loading Rate (BW/sec)	Pre-heelstrike Vertical Velocity (m/min)	Free Walking Velocity (m/min)
1	4.16	0.24	31.0	30.8	21.2	80.4
2	1.82	2.33	31.0	32.3	20.9	54.4
3	1.82	3.90	64.9	68.2	18.8	67.1
4	2.76	1.75	27.0	27.6	14.3	60.8
5	3.07	2.88	36.2	38.0	22.1	62.2
6	6.30	3.59	84.1	88.6	37.5	87.0
7	2.54	2.24	51.4	52.7	23.2	73.3
8	2.47	1.94	34.6	34.8	20.4	63.8
9	2.33	3.88	73.7	77.3	17.8	68.2
10	3.72	4.27	69.6	74.0	26.9	77.2
Mean	3.10	2.70	50.3	52.4	22.3	69.5
Std. Dev.	1.35	1.25	21.2	22.7	6.3	10.0

for the control subjects were 3.74 ± 0.82 %BW-H and 2.83 ± 1.54 %BW-H, respectively (see Table 3.4).

The ACL reconstruction group had a mean maximum tibially directed ground reaction loading rate for the injured limbs of 50.3 ± 26.3 BW/sec and a mean maximum vertical ground reaction loading rate of 51.9 ± 27.5 BW/sec (see Table 3.2). The ACL reconstruction subjects' average maximum tibially directed loading rate and average maximum vertical loading rate for the uninjured limb were 50.3 ± 21.2 BW/sec and 52.4 ± 22.7 BW/sec, respectively (see Table 3.3). The mean maximum tibially directed and vertical loading rates for the control group were 64.1 ± 17.4 BW/sec and 66.6 ± 18.6 BW/sec, respectively (see Table 3.4).

The average pre-heelstrike vertical velocity and average free walking velocity for the ACL reconstruction group's injured limb trials were 21.0 ± 6.2 m/min and 70.1 ± 10.9 m/min, respectively (see Table 3.2). The mean pre-heelstrike and free walking velocities for the trials with the ACL group's uninjured limbs were 22.3 ± 6.3 m/min and 69.5 ± 10.0 m/min, respectively (see Table 3.3). The control subjects had mean pre-heelstrike and free walking velocities of 24.7 ± 4.9 m/min and 73.6 ± 8.6 m/min, respectively (see Table 3.4).

Examination of the measurements associated with heelstrike (i.e. heelstrike knee moments and tibially directed and vertical loading rates) indicated that ACL reconstruction subject number 6 had unusually high standard deviations of these measurements for the six data collection trials, compared with the other subjects (see Table 3.5), possibly indicating that measures taken to avoid "targeting" were not completely effective for this subject. Additionally, this subject reported a

Table 3.4. Summary of experimental results of control subjects.

Subject	Maximum Midstance Knee Flexion Moment (%BW-H)	Maximum Heelstrike Knee Extension Moment (%BW-H)	Maximum Tibially Directed Loading Rate (BW/sec)	Maximum Vertical Loading Rate (BW/sec)	Pre-heelstrike Vertical Velocity (m/min)	Free Walking Velocity (m/min)
1	4.28	2.87	68.9	71.2	25.4	74.4
2	3.42	3.18	59.9	62.2	22.7	67.7
3	4.59	3.25	74.3	76.7	24.3	71.7
4	3.39	2.62	38.1	39.9	24.7	66.0
5	4.35	4.17	84.3	87.5	26.9	71.6
6	2.79	5.69	97.0	103.3	30.4	78.8
7	5.11	1.86	59.1	60.2	28.6	93.9
8	2.61	2.30	51.2	52.8	19.2	63.8
9	3.15	3.98	56.3	58.8	30.0	70.2
10	3.76	1.37	51.9	52.9	14.7	78.1
Mean	3.74	2.83	64.1	66.6	24.7	73.6
Std. Dev.	0.82	1.54	17.4	18.6	4.9	8.6

Table 3.5. Summary description of within subject standard deviations for measurements associated with heelstrike for the ACL reconstruction group.

	Maximum Heelstrike Knee Extension Moment (%BW-H)	Maximum Tibially Directed Loading Rate (BW/sec)	Maximum Vertical Loading Rate (BW/sec)
Mean of Within Subject Stand. Deviations	0.85	12.0	12.6
Stand. Dev. Within Subject Number 6	2.27	28.4	30.1
Mean (SD) for Injured Limb Without Subject Number 6	2.17 (0.90)	43.2 (14.3)	44.3 (14.3)
Mean (SD) for uninjured Limb Without Subject Number 6	2.60 (1.28)	46.5 (18.6)	48.4 (20.0)

significant weight gain during his period of inactivity following ACL injury, which may have affected the early portion of stance phase. Data for variables associated with heelstrike for this subject were, subsequently, removed from further statistical analysis. The mean heelstrike knee moment, mean tibially directed loading rate, and mean vertical loading rate for the ACL reconstruction group with the values for subject number 6 removed, were 2.17 ± 0.90 %BW-H, 43.2 ± 14.3 BW/sec, and 44.3 ± 14.3 BW/sec, respectively, for the injured limbs and 2.60 ± 1.28 %BW-H, 46.5 ± 18.6 BW/sec, and 48.2 ± 20.0 BW/sec, respectively, for the uninjured limbs.

Comparison of measurements for the injured limbs of the ACL reconstruction group and the control group using a multiple regression analysis of variance technique, which employed a series of subject dummy variables to account for repeated measures, indicated significantly lower midstance knee flexion moments and significantly lower tibially directed and vertical loading rates ($P < 0.05$) (see Table 3.6). Heelstrike knee extension moments, pre-heelstrike velocities, and free walking velocities for the injured ACL trials were not found to be significantly different from those of the control group ($P > 0.05$) (see Table 3.6). Within the ACL reconstruction group, paired t-tests of the mean measurements for the injured and uninjured limbs revealed significantly lower midstance knee moments for the ACL injured limbs (uninjured - injured = 1.08 ± 1.13 %BW-H, $P = 0.014$) (see Table 3.7). No significant differences were found between any other injured and uninjured limb measurements within the ACL reconstruction group (see Table 3.7).

Multiple regressions of the measured values with each other,

Table 3.6. Summary of analysis of variance comparing the ACL reconstruction subjects' injured limbs with the control group.

Measurement	F statistic	P-value
Maximum Midstance Knee Flexion Moment	15.10	<0.01
Maximum Heelstrike Knee Extension Moment	2.80	>0.05
Maximum Tibially Directed Loading Rate	6.54	<0.05
Maximum Vertical Loading Rate	6.63	<0.05
Pre-heelstrike Vertical Velocity	2.08	>0.05
Free Walking Velocity	0.63	>0.05

Table 3.7. Summary of paired t-test comparisons between the injured and uninjured limbs of the ACL reconstruction group.

Measurement	Uninjured Minus injured value Mean (Stand. Dev.)	P-value
Maximum Midstance Knee Flexion Moment (%BW-H)	1.08 (1.13)	0.014
Maximum Heelstrike Knee Extension Moment (%BW-H)	0.44 (0.91)	0.19
Maximum Tibially Directed Loading Rate (BW/sec)	3.34 (17.2)	0.58
Maximum Vertical Loading Rate (BW/sec)	4.08 (17.9)	0.51
Pre-heelstrike Vertical Velocity (m/min)	1.28 (3.63)	0.30
Free Walking Velocity (m/min)	-0.60 (3.15)	0.57

employing the subject dummy variables, previously mentioned, to account for repeated measures, indicated significant correlations between: midstance knee moment and tibially directed loading rate, midstance knee moment and vertical loading rate, midstance knee moment and free walking velocity, tibially directed loading rate and free walking velocity, tibially directed loading rate and vertical loading rate, and vertical loading rate and free walking velocity (see Table 3.8). Other correlations were not found to be significant.

DISCUSSION

A substantial research effort has been devoted to the area of ACL injury. Clinical studies have provided physicians with simple quantitative measurements of knee joint laxity which are useful for diagnosing ACL tears, as well as with some indication of the prognoses to be expected for patients with ACL injuries (18,22,43,53,44,45,26). Biomechanical studies, particularly more recent investigations involving the ambulation of human subjects, however, have begun to address the mechanisms involved with the gait of ACL injury patients (7,69,41). These gait studies have generally focussed on subjects with ACL deficiencies and have not addressed the gait of patients with ACL reconstructions.

The finding of lower than normal midstance knee flexion moments for the injured limbs of the ACL reconstruction subjects indicates that the subjects with ACL reconstructions are not functioning in entirely the same manner as those with an intact ACL. ACL reconstruction subjects appear to be shielding their reconstructions from the level of loading imposed upon

Table 3.8. Summary of regression analysis.

Dependent Variable	Independent Variable	Regression Coefficient	P-value
Maximum Midstance Knee Flexion Moment	Maximum Tibially Directed Loading Rate	0.053 %BW-H / BW/sec	<0.01
Maximum Midstance Knee Flexion Moment	Maximum Vertical Loading Rate	0.051 %BW-H / BW/sec	<0.01
Maximum Midstance Knee Flexion Moment	Pre-heelstrike Vertical Velocity	0.065 %BW-H / m/min	>0.05
Maximum Midstance Knee Flexion Moment	Free Walking Velocity	0.20 %BW-H / m/min	<0.01
Maximum Tibially Directed Loading Rate	Pre-heelstrike Vertical Velocity	1.03 %BW-H / m/min	>0.05
Maximum Tibially Directed Loading Rate	Free Walking Velocity	2.33 %BW-H / m/min	<0.05
Maximum Tibially Directed Loading Rate	Maximum Vertical Loading Rate	0.95	<0.01
Maximum Vertical Loading Rate	Pre-heelstrike Vertical Velocity	1.12 %BW-H / m/min	>0.05
Maximum Vertical Loading Rate	Free Walking Velocity	2.45 %BW-H / m/min	<0.05

intact ACLs in restraint of anterior tibial displacement. The lower midstance knee moments of the injured limbs compared with those of the uninjured contralateral limbs supports this contention as well. While it is also possible that lower midstance knee moments might be due to subjects' unwillingness or inability to sustain load due to weak quadriceps, such an explanation is unlikely here, since the majority (60%) of reconstruction subjects exhibited no clinically discernable difference between uninjured and injured limb thigh size. Additionally, the difference between uninjured and injured limb midstance moments among those with thigh size discrepancies was similar to uninjured and injured limb midstance moment differences for those without thigh size discrepancies. It is also improbable that patello-femoral pain is associated with lower midstance knee moments since the incidence of patello-femoral pain at the Oakland Naval Hospital has been reported at only 10% (10).

The most encouraging result, however, appears to be that the external knee flexion moments at midstance were exhibited by all subjects, which suggests that patients with ACL reconstructions, by having some loading imposed on their reconstructions to resist anterior tibial displacement, are making more of a return to a normal gait pattern than are ACL deficient patients, who have been reported to maintain external extension moments about the knee throughout midstance as a mechanism for preventing anterior tibial displacement (7). These results, like those of Tibone and Antich (77) involving improved cross-cutting ability, indicate that improved dynamic function can be achieved following ACL reconstruction.

The lower ground reaction loading rates of the ACL reconstruction subjects compared with those of control subjects is similar to previous reports (16). Although, high prevalences of knee joint degenerative changes have been reported following ACL injuries (22,43,53), the mechanism hypothesized involving higher than normal ground reaction loading rates (58,84,85) has not been born out by this study.

The high correlations between midstance knee moments and loading rates, coupled with the significant differences for these measurements between the two groups, suggests that lower rates of loading may contribute to lower midstance knee moments. Although the correlation between midstance knee moments and free walking velocities was significant, the free walking velocities of the two groups were not different. Thus, within subject variations in midstnace knee moments appear to be affected by within subject free walking velocity variations. Variations in midstance knee moments between the subject groups, however, can not be related to between subject group variations in free walking velocity. The correlations between loading rates and free walking velocities indicate similar relationships. Within subject variations in loading rates appear to be associated with within subject free walking velocity variations, while variations between subject groups are unrelated.

The significant correlation between the tibially directed and vertical loading rates indicates that use of the vertical component of the ground reaction force alone should provide a good measure of the ground reaction force directed at the knee joint. Such a result can be useful since measurement of the vertical loading rate requires only a force plate. Conversely, measurement of the tibially directed loading rate requires video motion analysis capabilities in addition. Future studies addressing the ground reaction loading rate directed towards the knee should be able to use the vertical component as a reasonable approximation of the tibially directed loading rate.

Evaluation of the results of this investigation should include some consideration of the composition of the injured subject group. Being drawn from a pool of ACL injury patients from a Naval Hospital, these subjects were relatively young, male, active duty military personnel on temporary limited duty during their rehabilitation period. These subjects generally described their lifestyles, previous to their injuries, as being fairly active; and most expressed a desire to be fully active again and were somewhat active already. A high level of motivation to achieve a successful result may, subsequently, have some effect on whether and to what degree these subjects return to normal gait patterns.

With regard to comparisons of the results of this study and those of Berchuck, et al. (7), it bears noting that all ACL deficient patients tested, subsequently, underwent ACL reconstructions. The choice of surgery over non-operative treatment indicates that these patients were not managing well without an ACL, and, thus, may not be representative of all ACL deficient individuals, but may be a valid comparison group for the present study.

Future studies could analyze patients both pre-operatively and post-operatively. Longer term follow-up measurements of patients undergoing both surgical and non-operative treatment could be beneficial as well. Such investigations may help clarify issues pertaining to the dynamic function of a broader range of patients with injuries to their ACL, such as the effectiveness of hamstring development in restraint of anterior tibial displacement, and the development of osteoarthrosis.

LOWER LIMB PROPRIOCEPTION IN ABOVE-KNEE AMPUTEES

INTRODUCTION

Joint position sensation is a complicated process believed to involve a wide variety of specialized mechanoreceptors. These receptors have been demonstrated in a wide variety of capsular and extracapsular tissues, including the joint capsule (8,14,15) joint space ligaments, menisci, and muscles and tendons crossing the joint (29,39,54,55,56,63,66,67). When stimulated, they provide feedback regarding both the static joint position and rate of joint movement, depending on the specific type of receptors involved.

Much work has been devoted in recent years towards determining the type and location of the primary receptors involved in joint proprioception. Traditionally, the joint capsule has been thought to be the site of the predominant receptors involved in signaling joint position (30,50,49,76,80). Recently, however, several studies have challenged this concept, suggesting instead that extracapsular receptors located in muscle spindles and joint space ligaments may assume a substantial responsibility for conscious appreciation of joint position and motion (25,29,66,67,74).

Regardless of whether capsular or extracapsular receptors provide primary proprioceptive feedback, the importance of their aggregate contribution can be inferred from analysis of gait patterns in above-knee (AK) amputees. Because AK amputees are necessarily without input from either type of joint position receptor, one can hypothesize that they would have significantly diminished joint proprioception, and that this may partially contribute to the gait abnormalities which have been documented in AK amputees (72).

For instance, Murray et al. measured lower limb motion in AK amputees using an interrupted light method (52). Prior to heel strike, when knee flexion should assist in deceleration, they documented persistence of knee extension. Continued extension of the prosthetic limb was observed during early stance, in contrast to the slight knee flexion occurring in the normal (52). It is possible that the amputee's inability to sense when the position of the prosthetic limb reaches full extension may lead to excessive delay during late swing phase and contribute to the already uncoordinated gait pattern.

It is believed that the flexion of the knee at heel strike in the normal individual (involving active lengthening of the quadriceps) provides "shock absorption" during heel strike impact and can thereby protect the joints from microdamage (38). Thus, proprioception sensation in the AK amputee knee may have wider implications. In addition, controlled muscle activity at the knee in normal gait is important for energy storage and transfer between limb segments (72). The absence of adequate knee proprioception may thus be responsible for the jarring impulse created by the repetitive planting of the prosthetic limb, which is believed to register pain at the amputee's stump. This, in turn, is thought to be a rate-limiting factor in the gait of AK amputees.

In light of this, a protocol has been designed to test the hypothesis that AK amputees have compensatory proprioceptive sensations, ostensibly arising from remaining receptors in the skin and soft tissues of the stump and from proprioceptors in and around the hip, which are used to sense the limb's position in space during gait. They may therefore adjust the impact with which the prosthesis strikes the ground, and this feedback may correlate with a faster average ambulatory velocity. In addition, proprioceptive measurements in AK amputees, which have heretofore been ignored (72), could provide fundamental information regarding the stability of these individuals with their current prostheses and aid in the development of better gait training and prosthetic design.

METHODS

Subjects

Subjects were recruited from the STAMP program and the prosthetic service at the Veterans Administration Medical Center (VAMC). Only male traumatic and tumor amputees were included in this study group and all subjects signed consent forms for the University of California, San Francisco, Committee on Human Research and the VAMC Human Studies Committee. All amputees had to have a stable, painfree residual limb with no problem areas. Further, all subjects were required to have a well-fitting satisfactory prosthesis, and to have had the present prosthesis for at least four months. Demographic data are shown in Table 4.1.

Measurements of Prosthetic and Sound Limbs

Anthropomorphic data was obtained on each subject. The stump length (greater trochanter to end of stump) measurements were obtained on each residual limb. Additional measurements of the prosthesis and sound limb were made to standardize the tests from subject to subject. These

Table 4.1. Summary of data on the amputees.

Subject	Cause For Amput.	Age (years)	Time Since Amputation (years)	Time With Current Prosth. (years)	Stump Length (cm)	Free Walking Velocity (m/min)
1	tumor	30	2	0.33	24	70.3
2	trauma	64	45	1	30	66.5
3	trauma	69	45	5	31.5	83.2
4	trauma	55	5	2	28.5	47.7
5	trauma	43	22	1.5	30.5	89.1
6	trauma	58	37	5	25	76.2
7	trauma	44	21	5	40	68.5
8	trauma	64	39	1.2	31	63.5
9	trauma	60	38	1	32	56.4
10	trauma	71	45	5	31	62.4
Mean		55.8	29.9	2.7	30.4	68.4
Std. Dev.		13.0	16.4	2.0	4.4	12.2

measurements were subsequently used to determine the change in linear distance necessary to represent five standard angular deflections. This allowed for the testing of the same angular displacement in all subjects regardless of individual variation in sound limb, stump, or prosthesis length.

Measurement of Position Sense

Testing of joint position and kinesthesia was done using a specially-designed truck as depicted in Figure 4.1. The test apparatus permitted consistent positioning of subjects and eliminated all external cues to limb motion except those emanating from the hip, stump (including skin and soft tissue structures), and prosthesis. Subjects were asked to stand on their normal leg with the prosthetic limb supported by a freely-rotating pedal located on the truck. The subjects were tested in this position to approximate the lower limb motion in the late swing phase of gait. A velcro sock slipped onto the prosthetic foot with sponge rubber and velcro applied to the surface of the pedal prevented the foot from slipping on the pedal and aided in preventing pedal motion from serving as a cue to the subject. Handrails were provided to assist the subject with stability. In addition, a support brace was positioned across the handrails to offer further lower back and pelvic support and to provide the subject with a reference point in order to minimize sway. The prosthetic lower limb was positioned such that it was vertical with the ground. This provided for a resting knee angle of approximately 120°. The truck supporting the prosthesis was attached by a cord to a shaft driven by a variable speed motor. The motor and shaft were designed such

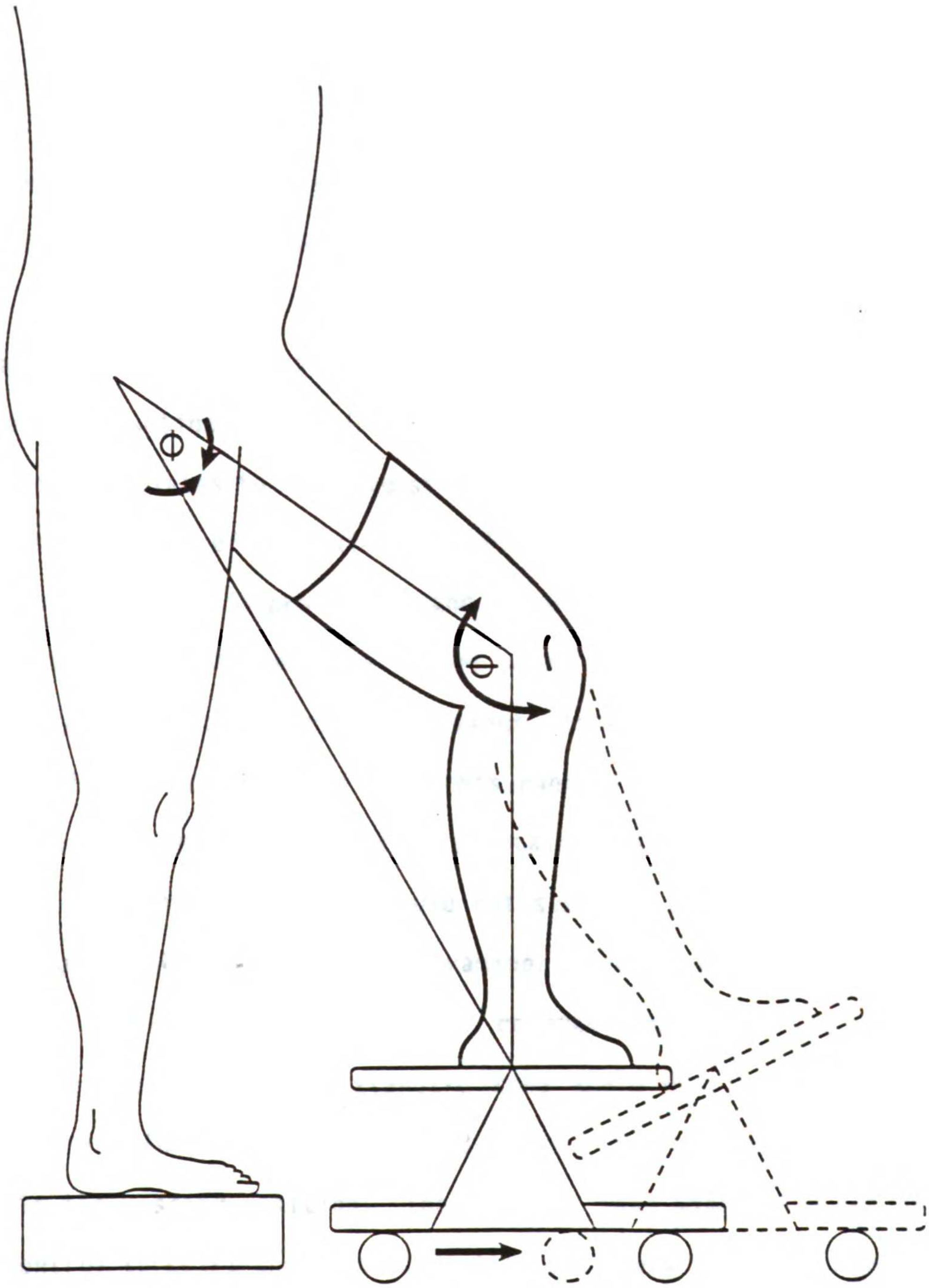


Fig. 4.1. Schematic diagram of the experimental setup showing the angles measured in this study.

that the time between the start of the motor and the beginning of shaft rotation could be varied (i.e. by varying the time before the gear became engaged). All subjects were blindfolded to remove visual input. Tests of both reproduction of passive positioning and the threshold detection of passive motion were performed to measure joint position sense.

Reproduction of Passive Positioning

The reproduction of passive positioning was begun by obtaining the five linear correlates to the five standardized angular displacements from the resting position of approximately 120° (full extension = 180°), given the individual subject's prosthetic and sound limb measurements. The five angular deflections chosen to be tested were: +5°, +10°, +15°, +20°, and +25°. They were chosen because previous studies had shown that joint proprioception is most sensitive in this range (i.e., approximately 120° to 145°) (5). Once these values were obtained, they were randomized so that their order of application would not serve as a cue to the subject. The testing was initiated by the examiner, who started the motor and allowed it to run until the truck covered the requisite distance representing one of the five standard knee angle displacements. After holding the leg in this position for 8-10 seconds, the examiner returned the leg to the resting position. The examiner then handed the subject an on/off control to start the motor and allow the truck to proceed until the leg had returned to the position at which it had been previously held.

The subject was asked to use only cues emanating from his leg to reproduce the angle. To guard against the use of auditory cues in repositioning the leg, the amount of time between when the motor was started and when the shaft actually began to rotate was varied. In addition, the motor speed was varied between trials; this prevented the subject from sensing the beginning of leg motion and then counting until the motor was switched off.

The position selected by the subject was recorded, and the difference between the original position and that selected by the subject was calculated. The subject was given two practice runs to familiarize himself with the procedure. The subject was then tested with the sound limb on the truck and the prosthetic limb providing support. Before positioning the sound limb on the pedal, the ankle was braced with a velcro air cast to minimize ankle dorsiflexion/plantar flexion and internal rotation/external rotation. A velcro sock was then placed on the sound foot to minimize slipping on the pedal. In this manner, the subject's sound limb provided the control for the study.

Threshold Detection of Passive Motion

The threshold detection of passive motion was measured with the prosthesis starting from the same resting position (approximately 120° extension). The handrails and other supports to improve stability and minimize sway were as described above for reproduction of angular deflection. The subject was given the control box with the on/off switch and was told that, following a variable amount of time, his leg (supported by the truck) would begin to change position. He was instructed to switch the machine off when this change occurred. With visual and auditory cues eliminated as described above, the leg was slowly moved into extension 5 to 30 seconds after starting the motor at a speed of approximately

0.5°/sec. The variation in starting the motor prior to shaft engagement served as a control of the validity of the response. The linear movement of the truck was measured in millimeters and converted to angular deflection as previously described. Five repetitions were made on the prosthetic limb, followed by five repetitions on the sound limb. For each measurement obtained, an angular correlate was calculated.

Gait Velocity

The final test involved the determination of average gait velocity. Each subject underwent four measurements timed by a hand-held stopwatch while ambulating at a free-walking rate on a 20-foot walkway wearing his normal shoes. After the four runs, the subject's average ambulatory velocity was obtained.

Statistical Analysis

Statistical analysis was performed using the Minitab (Minitab, Inc., State College, PA) statistical package. For comparison of proprioception measurements of sound limb to prosthetic limb, the paired t test was used. Linear regression was used to test for correlation between proprioceptive measures and stump length, gait velocity, and age.

RESULTS

Data are shown in Tables 4.1, 4.2, and 4.3. Free walking velocities for the amputees averaged 68.4 ± 12.2 m/min (range 47.7 to 89.1 m/min). Stump length (greater trochanter to end of stump) was 30.4 ± 4.4 cm

Table 4.2. Average Knee Angle Reproduction Accuracy Measurements and Change in the Hip Angle Associated with These Knee Results.

	Prosthetic Limb		Sound Limb	
Subject	Knee Angle (°)	Hip Angle (°)	Knee Angle (°)	Hip Angle (°)
1	5.76	0.65	3.50	0.46
2	5.42	0.63	6.36	0.70
3	2.11	0.24	1.86	0.28
4	2.92	0.39	4.03	0.48
5	4.06	0.49	1.04	0.12
6	1.93	0.25	1.61	0.27
7	3.09	0.30	4.02	0.41
8	1.46	0.17	3.23	0.39
9	2.19	0.29	1.80	0.21
10	1.85	0.22	3.81	0.43
Mean	3.08	0.36	3.13	0.37
Std. Dev.	1.52	0.17	1.59	0.16

Table 4.3. Average Knee Angle Threshold/Detection Measurements and Change in the Hip Angle Associated with These Knee Results

	Prosthetic Limb		Sound Limb	
Subject	Knee Angle (°)	Hip Angle (°)	Knee Angle (°)	Hip Angle (°)
1	3.26	0.08	1.65	.01
2	3.22	0.08	0.76	.005
3	4.76	0.18	0.66	.004
4	2.23	0.02	1.75	.02
5	17.27	1.44	2.74	.06
6	2.00	0.05	0.55	.003
7	2.26	0.06	0.35	0.002
8	3.43	0.11	1.23	.01
9	2.60	0.07	0.94	.01
10	5.82	0.21	2.52	.04
Mean	4.69	0.21	1.32	0.02
Std. Dev.	4.58	0.44	0.83	0.02

(range, 24 to 40 cm). Time from amputation and age averaged 29.8 ± 16.4 years and 55.8 ± 13.0 years, respectively.

Comparison of the sound limb knee angle reproduction to the prosthetic side using a paired t test revealed no difference ($P = 0.43$). The comparison of the groups' prosthetic and sound limb knee angle measurements for threshold detection revealed a significant difference (prosthetic limb minus sound limb = 3.37 ± 4.05°; $P = 0.027$). Subject 5, however, demonstrated a threshold detection for the prosthetic limb that was almost 3 standard deviations worse than the other amputees (Table 4.3). Thus, this subject was discarded from further analysis of threshold detection resulting in a highly significant threshold detection difference (prosthetic limb minus sound limb = 2.13 ± 1.07; $P = 0.0003$). These comparisons are demonstrated in Figure 4.2. Comparison of reproduction measurement data for the hip angle changes using a paired t test revealed no difference between the prosthetic and sound limb ($P = 0.88$). Prosthetic and sound limb hip angle threshold detection measurements were found to be significantly different after removal of subject 5 (prosthetic limb minus sound limb = 0.082 ± 0.056; $P = 0.0023$).

Figure 4.3 demonstrates the relationship between prosthetic knee angle reproduction versus age. This regression has a correlation coefficient of 0.64 ($P = 0.048$). Figure 4.4 shows a plot of prosthetic knee angle reproduction versus time since amputation; however, the regression coefficient, 0.48, was not significant ($P = 0.17$). Regressions of threshold detection for the prosthetic side versus age and versus time since amputation, with subject 5 excluded as mentioned previously, yielded correlation coefficients of 0.5 ($P = .166$), and 0.04 ($P = 0.91$).

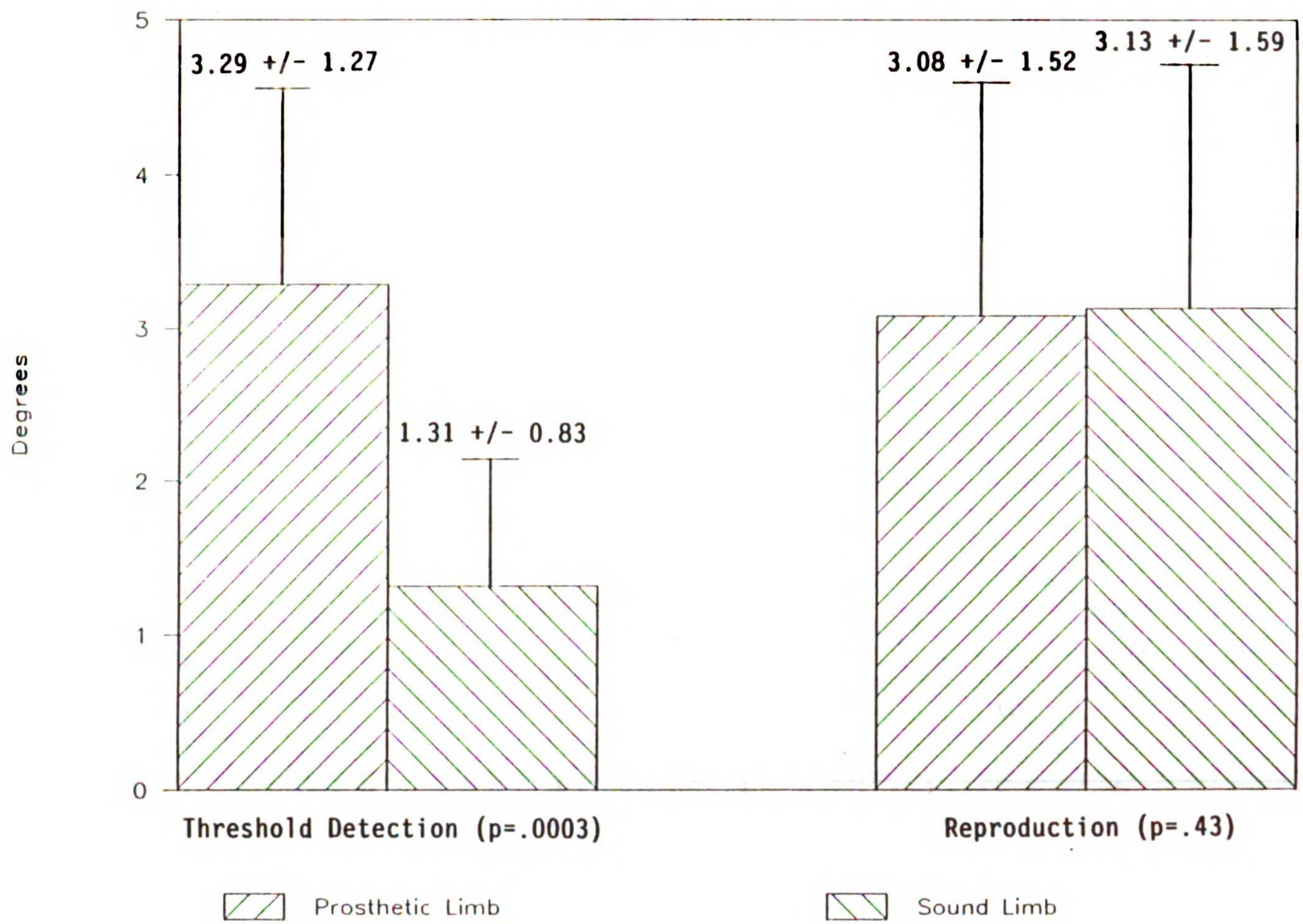


Fig. 4.2. Comparison of sound and prosthetic limb knee angle measurements of passive motion threshold detection and passive motion reproduction.

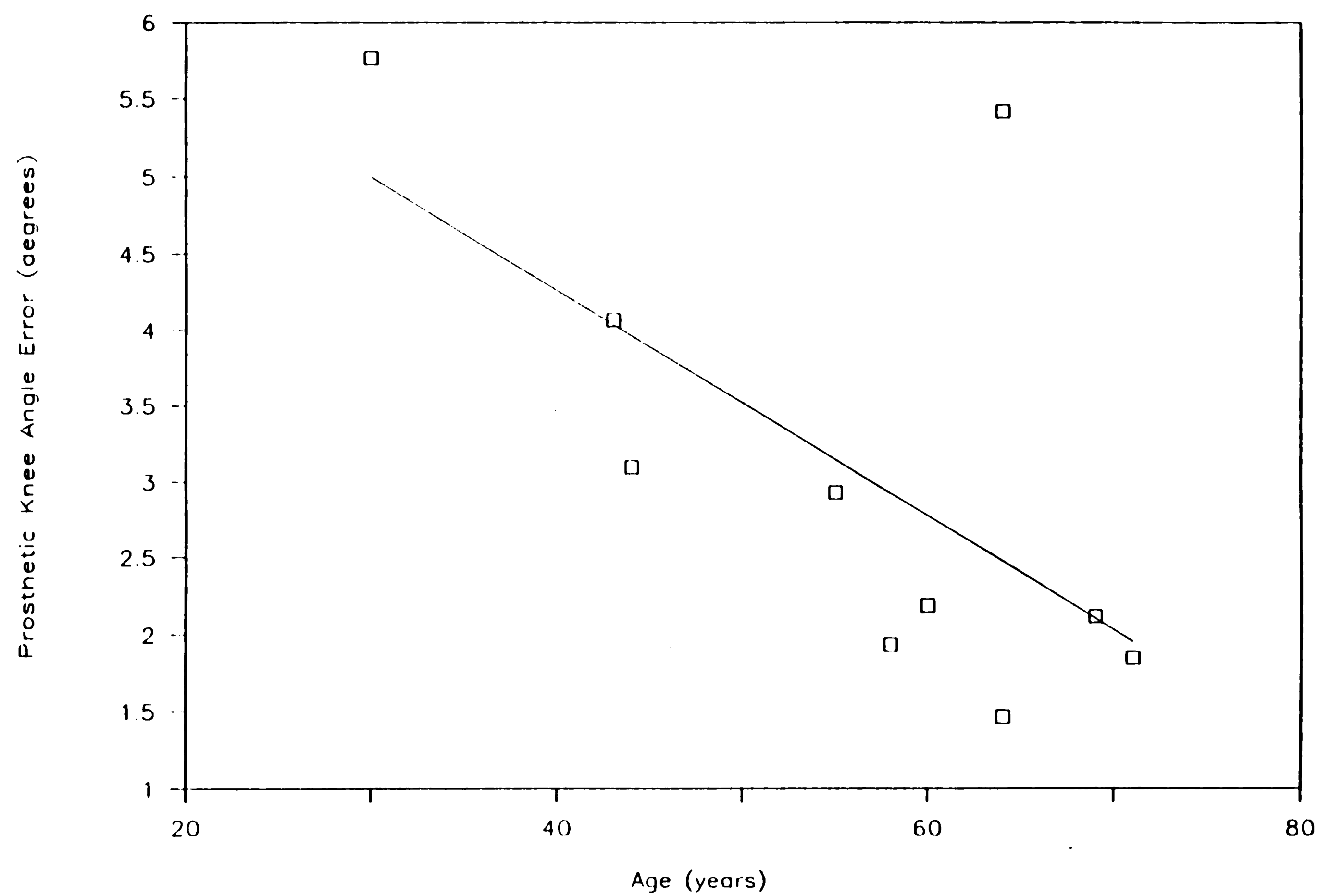


Fig. 4.3. Correlation of prosthetic limb knee angle reproduction measurements with age.

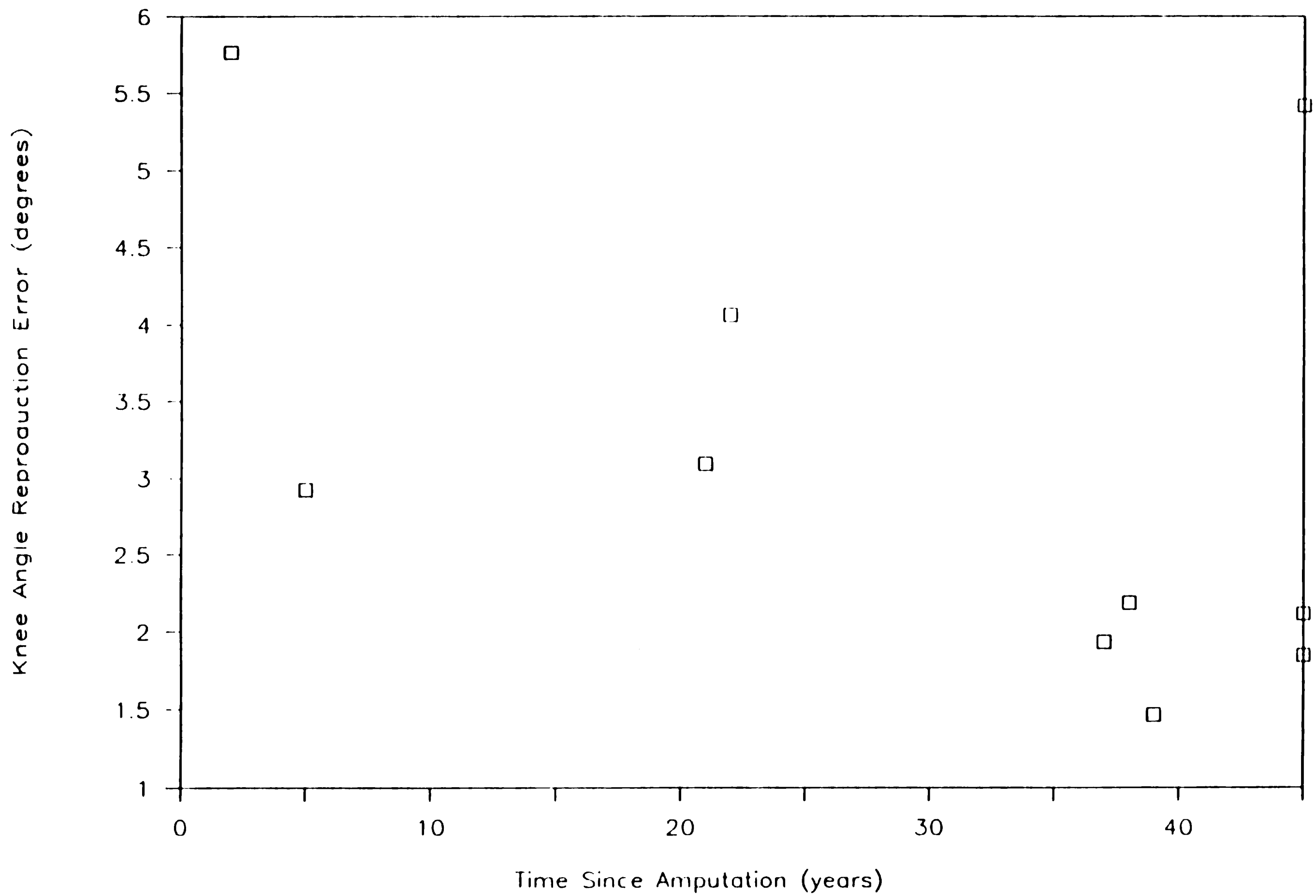


Fig. 4.4. Plot of prosthetic limb knee angle reproduction measurements with time since amputation.

Regression of prosthetic limb knee angle reproduction with average free walking velocity and with stump length did not yield significant correlations ($P = 0.65$ and $P = 0.47$, respectively). Regressions of prosthetic limb threshold detection versus velocity of gait and stump length resulted in P-values for the correlations of 0.53 and 0.94, respectively. Regression of knee angle reproduction versus threshold detection for the prosthetic limb yielded a P-value for the correlation of 0.66 ($n = 9$). Average free walking velocity versus stump length did not correlate significantly ($P = 0.40$).

DISCUSSION

Of the various tests used to assess "proprioception," two of the more commonly accepted methods, passive motion threshold detection and passive motion reproduction, have been used here.

It has been hypothesized that AK amputees compensate for the absence of these structures by relying on alternative mechanisms to provide an indication of prosthesis position. These may include the hip joint, differential pressure of the prosthesis on the deep soft tissue structures of the limb, and on the skin-prosthesis interface position.

As has been mentioned, the prosthetic knee joint of the AK amputee is obviously without any of the structures known to contain proprioceptive elements. Even the quadriceps and hamstrings do not have normal function as a cue to proprioception. Thus, it is not surprising that these subjects had measurably diminished proprioception on their prosthetic side as measured by threshold detection of passive motion. That no difference was seen between prosthetic and sound limbs on angle reproduction tests seems to indicate that these two methods of proprioceptive assessment involve different neural mechanisms, as has been indicated in previous studies (4,6) A difference in the mechanisms being examined by the two tests is further suggested by the contrast between the lack of a relationship between age and prosthetic limb threshold detection and the demonstrated improvement in prosthetic limb reproduction with age as well as the lack of any relationship between such threshold detection and reproduction.

The amputees' better performance on prosthetic limb knee angle reproduction as compared with threshold detection with respect to the results of the sound limb may indicate the importance of hip motion cues to the amputees. Hip angle threshold detection in normal limbs has been measured at about 0.6° (30). Since the corresponding hip angle changes for prosthetic limb knee angle threshold detection were about 0.2°, it seems unlikely that hip motion afforded much of a cue to the amputees during this test. Conversely, for prosthetic limb knee angle reproduction, corresponding hip angle changes ranged up to about 3° for the largest deflections. Consequently, the use of hip motion as a cue to knee motion during reproduction testing seems likely as long as the two motions are associated.

The lack of correlation between walking velocity and stump length in previous studies of below-knee amputees (28) supports the findings of this investigation in above-knee amputees. Despite the lack of correlation, the use of sensory cues from the surface or deep tissues of the stump to signal position would still seem to be a possibility, particularly during threshold detection testing when hip motion is minimal. Subjective comments of some amputees subsequent to testing included remarks concerning sensation of separation between subject and stump, socket/stump interface pressure changes, and binding of the skin along the socket brim that would seem to support this hypothesis.

ANALYSIS OF A PROPOSED PROSTHETIC FOOT/ANKLE SYSTEM

INTRODUCTION

Background

Loss of active ankle plantar flexion during the push off portion of the stance phase of gait is a deficit resulting from lower extremity amputation. At natural velocities in normal subjects, ankle plantar flexion energy generation at push off has been measured to account for approximately 80% (26 J) of lower extremity stance phase energy generation (82). A similar investigation of amputee gait indicated a tendency of amputees to increase early stance phase hip energy generation to compensate for the loss of ankle plantar flexion at push off (83).

Traditional SACH feet have demonstrated very little energy storing capacity (83). Single axis and Greissinger foot/ankle systems have been shown, at push off, to return approximately 20-30% (3-4 J) of the energy stored during stance phase dorsiflexion (83). Other comparisons of single axis and SACH feet have demonstrated that single axis feet tend to provide plantar and dorsiflexion more closely resembling the normal ankle (20,27). One investigation, utilizing single axis feet with adjustable plantar and dorsiflexion bumper stiffnesses as well as adjustable toe joint stiffness, demonstrated a preference by several amputees for more compliant bumpers allowing greater rotation about the ankle axis and for stiffer toe joints (25). Such characteristics were noted to be of interest in that they were contrary to such energy storing feet as the Seattle Foot which stores energy in a more compliant toe joint and permits relatively less ankle rotation (25). Some other commonly used energy storing prosthetic feet have been described with advantages and disadvantages based upon clinical subjective experience (21,81). Quantitative or objective evaluations of most foot/ankle systems are lacking.

Previously, Catranis Inc. of Syracuse, N.Y. attempted to develop a foot/ankle assembly for AK amputees which would store energy at heelstrike plantar flexion as well as at midstance dorsiflexion for return at push off plantar flexion (12). Development of this device was eventually discontinued at some point subsequent to the termination of their government contract.

More recently, Rigas proposed a preliminary design for a total AK prosthesis with foot and ankle which was intended to provide improved energy generation at ankle plantar flexion (62). Such a system, however, could not be used by BK amputees since the prosthetic knee joint of the AK prosthesis was utilized as part of the energy storing mechanism.

Although a number of other currently available prosthetic foot/ankle systems have been reported to possess energy storage and return capabilities (11,21,34,47,48,61,81), the designs of these systems dictate that only energy stored during positive stance phase dorsiflexion, which will be defined here as dorsiflexion beyond the neutral position of the prosthesis, can be returned at push off. Energy associated with plantar flexion following heelstrike is not available at push off because it is either dissipated or returned prior to positive dorsiflexion. Some of these energy storing prosthetic feet have been described with advantages and disadvantages based upon clinical subjective experience (21,81). Quantitative or objective evaluations, however, of most foot/ankle systems are lacking.

The motivation for this project was the potential for a prosthetic foot/ankle system to return increased energy at push off if energy could be stored prior to positive dorsiflexion, in addition to during positive dorsiflexion. A design for a prosthetic foot/ankle system that would store energy prior to as well as during positive dorsiflexion is proposed. The proposed design has been modeled and an analysis of the system has been performed which has produced estimates of the parameters of the components necessary to achieve the intended function.

Proposed System

In the proposed system (see Figure 5.1), rotation of the foot is about a single axis ankle joint (A). A series spring-damper (DB) connection is attached between the shank and foot, posterior to the ankle joint. Another spring (DE) is attached between the shank and foot, anterior to the ankle joint, and is much more compliant than the posterior spring. The posterior spring is the primary energy storage component providing subsequent return at push off. The damper essentially alters the neutral position of the system by adjusting its resistance during gait between three levels. The mechanism of this adjustment is not discussed but could be accomplished mechanically or electrically. The anterior spring mainly functions during swing phase to restore the system to its original neutral position. Although energy stored in the anterior spring will negate a portion of the energy stored in the posterior spring, this amount will be minimal if the difference in compliance of the two springs is substantial.

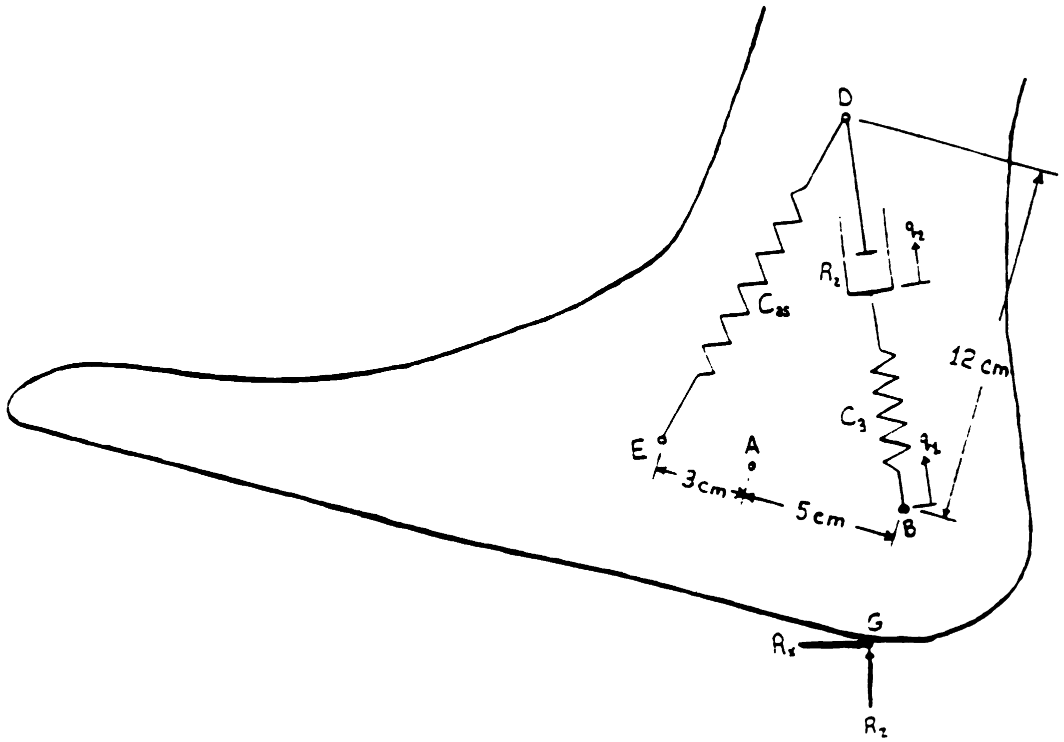


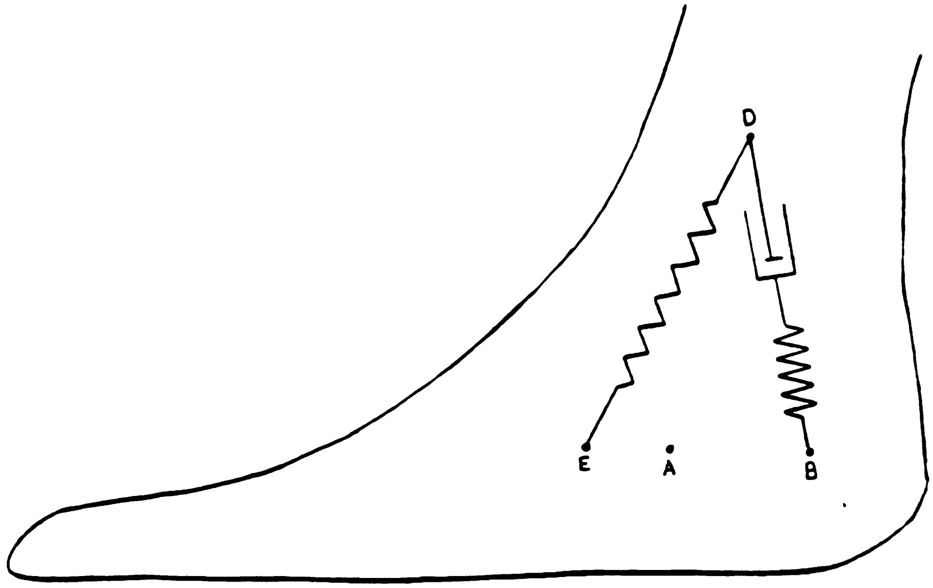
Fig. 5.1. Schematic diagram of proposed system at heelstrike.

The function of the proposed foot/ankle system will be explained in the context of one gait cycle beginning just prior to prosthetic heelstrike. Upon heelstrike, the damper will adjust to a middle level resistance and be capable of sustaining ground reaction loading conditions as the spring-damper assembly displaces during the period from prosthetic heelstrike (see Figure 5.1) to maximum plantar flexion (see Figure 5.2a). After maximum plantar flexion, the degree of ankle plantar flexion will decrease. Before the ankle reaches its original neutral position (see Figure 5.2b), however, the posterior spring will return to its free length. At this point the resistance of the damper adjusts to approach infinite resistance (i.e. the position of the damper locks). As the ankle then dorsiflexes, more energy can be stored in the spring than if its free length coincided with the original neutral position at zero degrees of plantar flexion. After sound limb heelstrike the load on the prosthetic foot subsides, and the stored energy is returned as a plantar flexion push off. After prosthetic foot toe off (see Figure 5.3), the damper adjusts such that the resistance becomes very small. In this condition little force can be maintained in the spring-damper assembly, and the more compliant, anterior spring, whose compliance was too high to affect stance, now contains sufficient stored energy to restore the system during swing to its original neutral position.

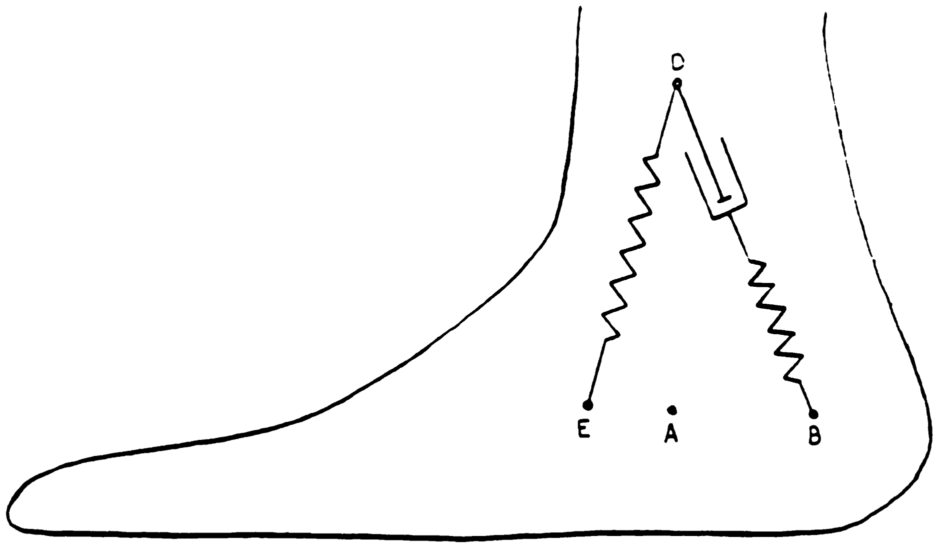
METHODS

Motion and Ground Reaction Data

Kinematic and kinetic gait data were obtained as a normal subject



(a)



(b)

Fig. 5.2. Schematic diagram of proposed system: (a) at maximum plantar flexion, (b) as it passes the neutral position.

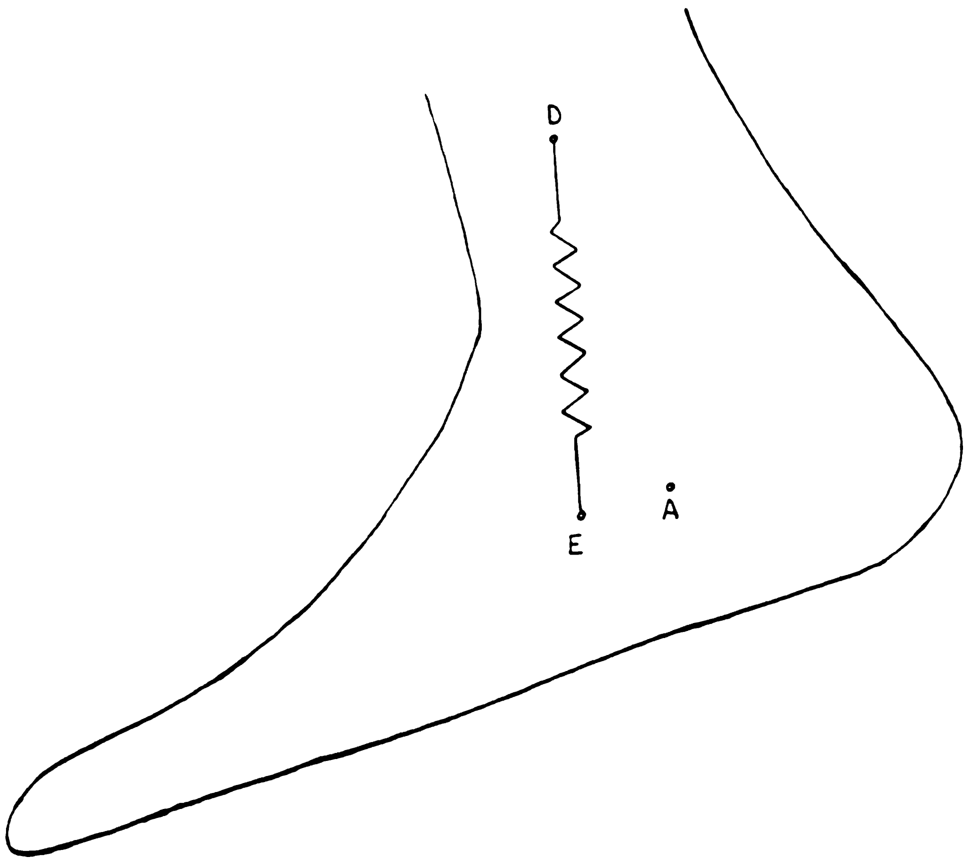


Fig. 5.3. Schematic diagram of proposed system following toe off.

walked across a raised walkway at a self-selected comfortable velocity. Retroreflective markers were placed on the right knee, ankle, and fifth metatarsal head (see Figure 5.4), and motion data were then acquired with a three camera VICON video motion analysis system (Oxford Metrics, Inc., Tampa, FL) sampling at 200 Hz. Ground reaction force data were acquired with a six channel AMTI force plate (Newton, MA), sampling synchronously with the cameras at 200 Hz. The coordinate system of the data acquisition apparatus was a left-handed one in which the x-axis was fore-aft and the z-axis was vertical and is the system used for all measurements.

System Modeling

In order to obtain estimates for the parameters of the springs and damper based on the video and force plate data of the normal subject's gait, two models were developed: a sagittal plane, schematic model of the overall foot/ankle system (see Figure 5.1), and a bond graph model (64) of the posterior spring-damper assembly (see Figure 5.5).

The positions of points A, B, D, and E in the schematic model are determined from heelstrike to toe off using video data of the normal subject's knee, ankle, and fifth metatarsal motions (see Figure 5.6). The ankle angle for the sampling frame prior to heelstrike was chosen to correspond to the neutral position and was calculated as

$$\theta_{ankle,np} = \tan^{-1}\frac{(m_{z,ph} - a_{z,ph})}{(m_{x,ph} - a_{x,ph})} + \pi - \tan^{-1}\frac{(k_{z,ph} - a_{z,ph})}{(k_{x,ph} - a_{x,ph})} \tag{5.1}$$

where, $\theta_{ankle,np}$ is the ankle angle for the neutral position, $k_{x,ph}$ and $_{kz,ph}$ are

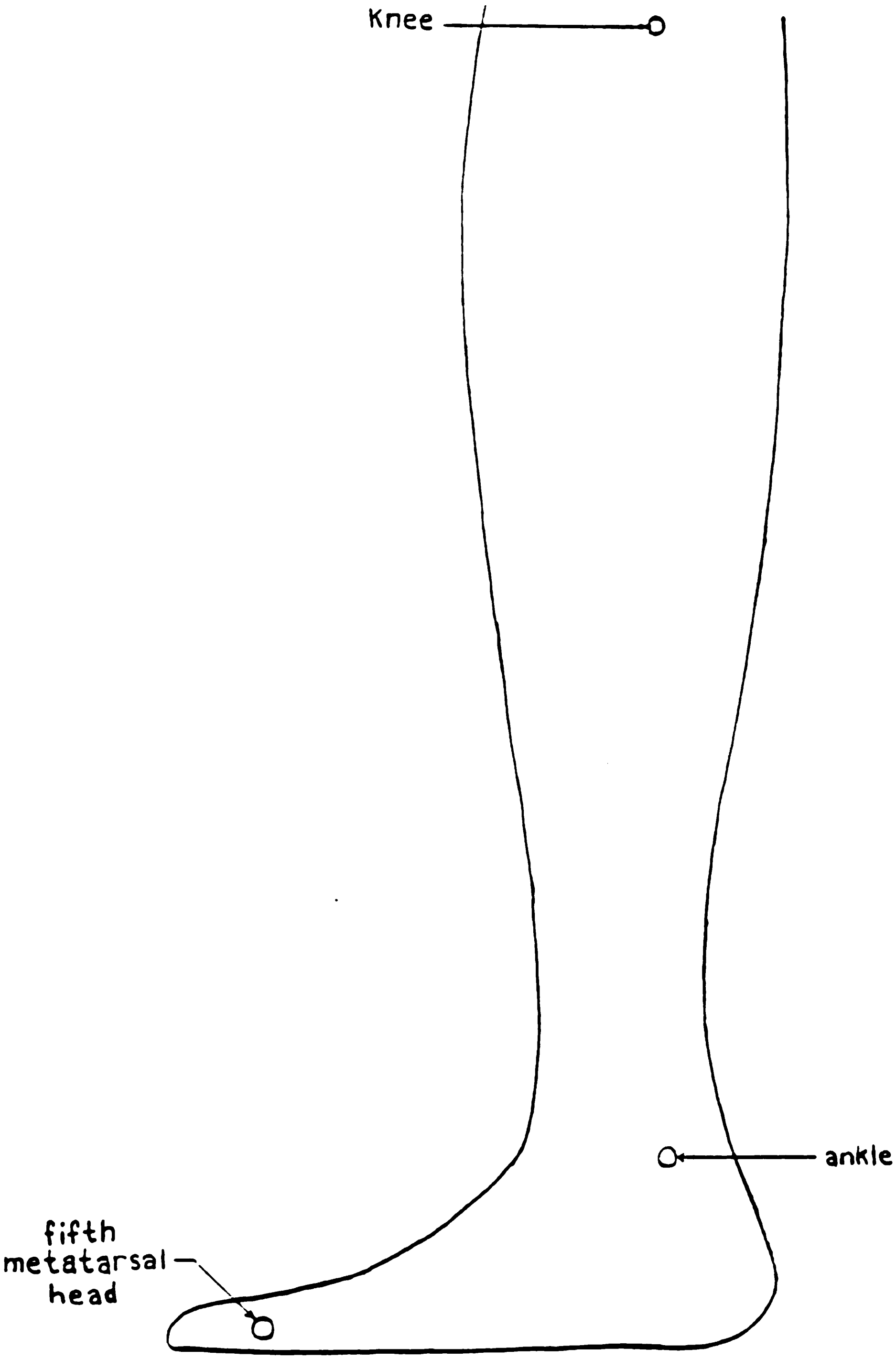


Fig. 5.4. Diagram of lower leg showing locations of retroreflective markers.

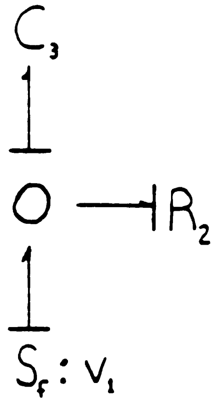


Fig. 5.5. Bond graph model of posterior spring-damper assembly.

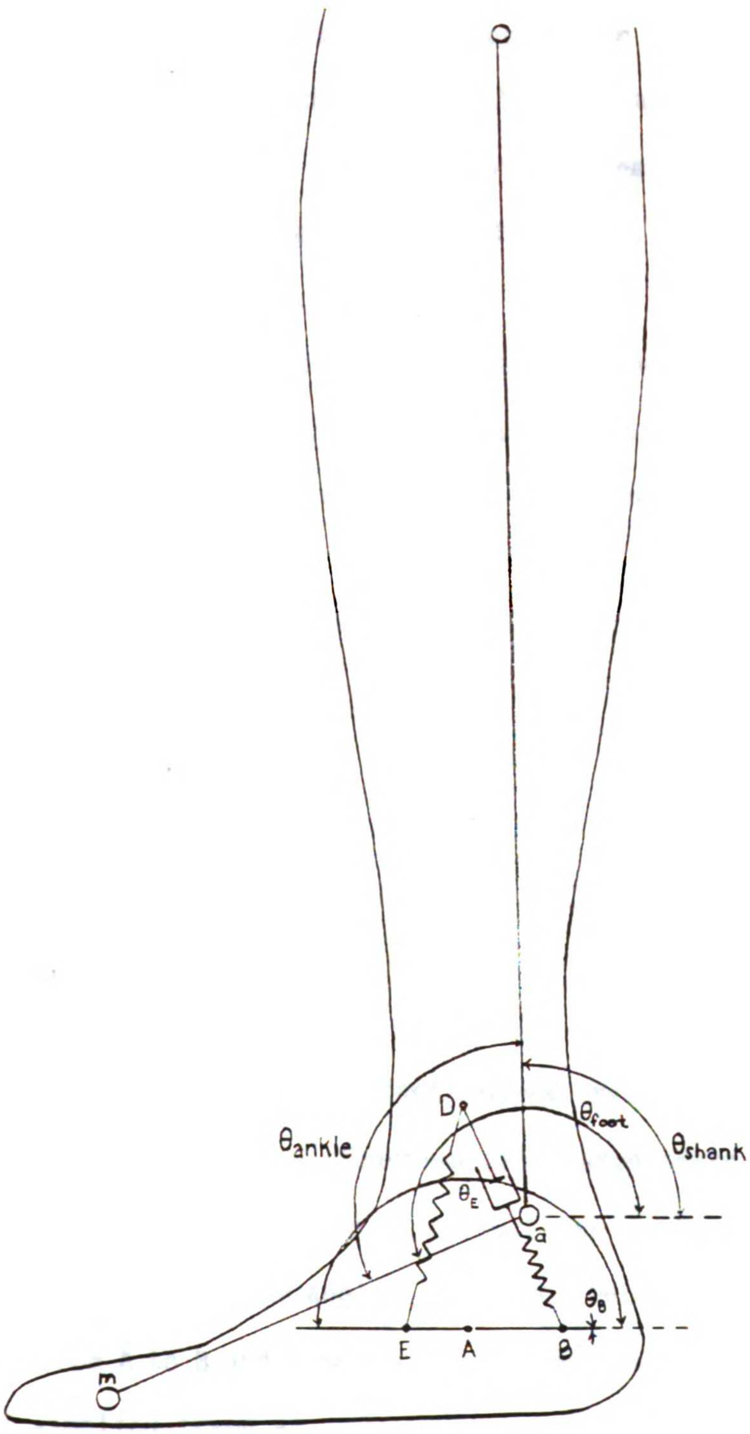


Fig. 5.6. Superposition of proposed system schematic diagram onto lower leg marker location diagram.

the fore-aft and vertical coordinates of the knee marker for the sampling frame prior to heelstrike, $a_{x,ph}$ and $a_{z,ph}$ are the fore-aft and vertical coordinates, respectively, of the ankle marker for the sampling frame prior to heelstrike, $m_{x,ph}$ and $m_{z,ph}$ are the fore-aft and vertical coordinates of the fifth metatarsal head marker for the sampling frame prior to heelstrike, and π is added as a correction because the $\tan^{-1}$ function calculates an angle between π and $3\pi/2$ as angle between 0 and $\pi/2$. The shank and foot angles were then determined for each sampling frame, i, during stance as

$$\theta_{shank,i} = \tan^{-1} \frac{(k_{z,i} - a_{z,i})}{(k_{x,i} - a_{x,i})} \text{ ,and} \tag{5.2}$$

$$\theta_{foot,i} = \tan^{-1} \frac{(m_{z,i} - a_{z,i})}{(m_{x,i} - a_{x,i})} \tag{5.3}$$

where $\theta_{shank,i}$ and $\theta_{foot,i}$ are the shank and foot angles, respectively, for a given sampling frame. When the $\tan^{-1}$ function calculated these angles to be in the wrong quadrants, π was added to correct the angles. With the coordinates still undetermined, the angles, θ_B and θ_E, between the fore-aft axis and the lines from A to B and from A to E, respectively, obtained for each stance phase sampling frame as

$$\theta_{B,i} = \theta_{foot,i} - \theta_{ankle,np} - \pi/2 \text{ , and} \tag{5.4}$$

$$\theta_{E,i} = \theta_{B,i} + \pi. \tag{5.5}$$

The components of the prosthetic ankle joint position were then obtained as a function of time as

$$A_x(t) = a_{x,i} + x_{offset} \ ; \ t = (\Delta t_{samp})(i), \text{ and} \tag{5.6}$$

$$A_z(t) = a_{z,i} + z_{offset} \tag{5.7}$$

where, $A_x(t)$ and $A_z(t)$ are the time functions of the fore-aft and vertical components, respectively, of the prosthetic ankle joint position, x_{offset} and z_{offset} are fore-aft and vertical corrections, respectively, to account for the difference in location of the anatomic and prosthetic ankle joints, and were approximated from direct measurement of the investigator's ankle and that of a previously used prosthetic single-axis foot, Δt_{samp} is the sampling time between frames, and i is the sampling frame, taken to be zero at heelstrike. The components of the positions of B, D, and E as functions of time were then determined as

$$B_x(t) = A_x(t) + |AB|\cos(\theta_{B,i}) \ , \tag{5.8}$$

$$B_z(t) = A_z(t) + |AB|\sin(\theta_{B,i}) \ , \tag{5.9}$$

$$D_x(t) = A_x(t) + |AD|\cos(\theta_{shank,i}) \ , \tag{5.10}$$

$$D_z(t) = A_z(t) + |AD|\sin(\theta_{shank,i}) \ , \tag{5.11}$$

$$E_x(t) = A_x(t) + |AE|\cos(\theta_{E,i}) \ , \tag{5.12}$$

$$E_z(t) = A_z(t) + |AE|\sin(\theta_{E,i}) \ , \tag{5.13}$$

where, $|AB|$, $|AD|$, and $|AE|$ are the lengths between A and B, A and D, and A and E, respectively, which were taken to be 0.05 m, 0.12 m, and 0.03 m, respectively, for purposes of this analysis.

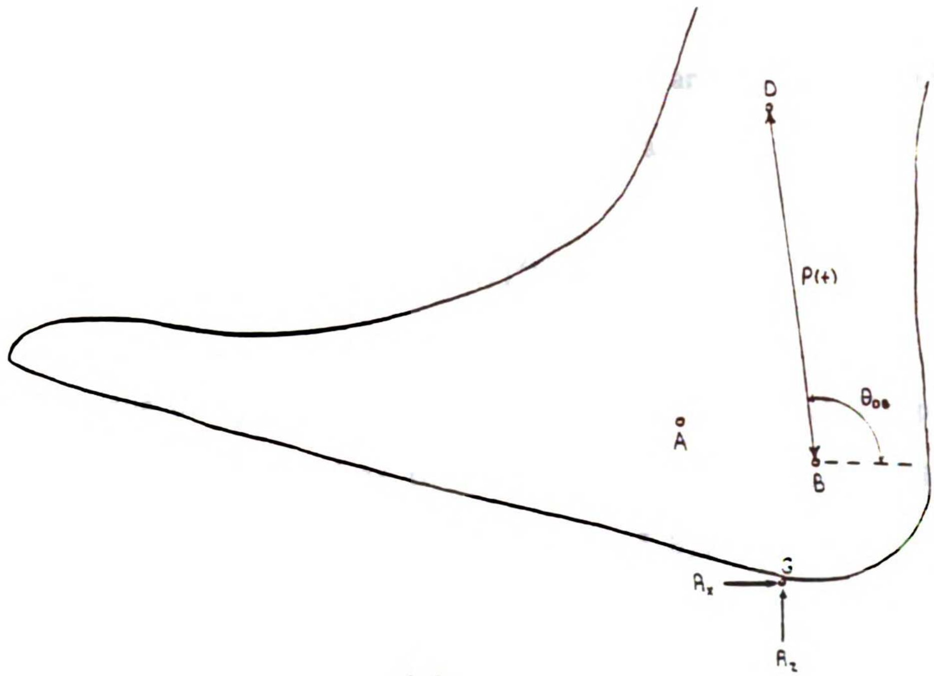
The load on the posterior spring-damper assembly, P(t) (see Figure 5.7a), is then determined from heelstrike to toe off as

$$P(t)= \frac{(G_x(t)-A_x(t))R_z(t)-(G_z(t)-A_z(t))R_x(t)}{(B_z(t)-A_z(t))\cos(\theta_{DB}(t))-(B_x(t)-A_x(t))\sin(\theta_{DB}(t))} \tag{5.14}$$

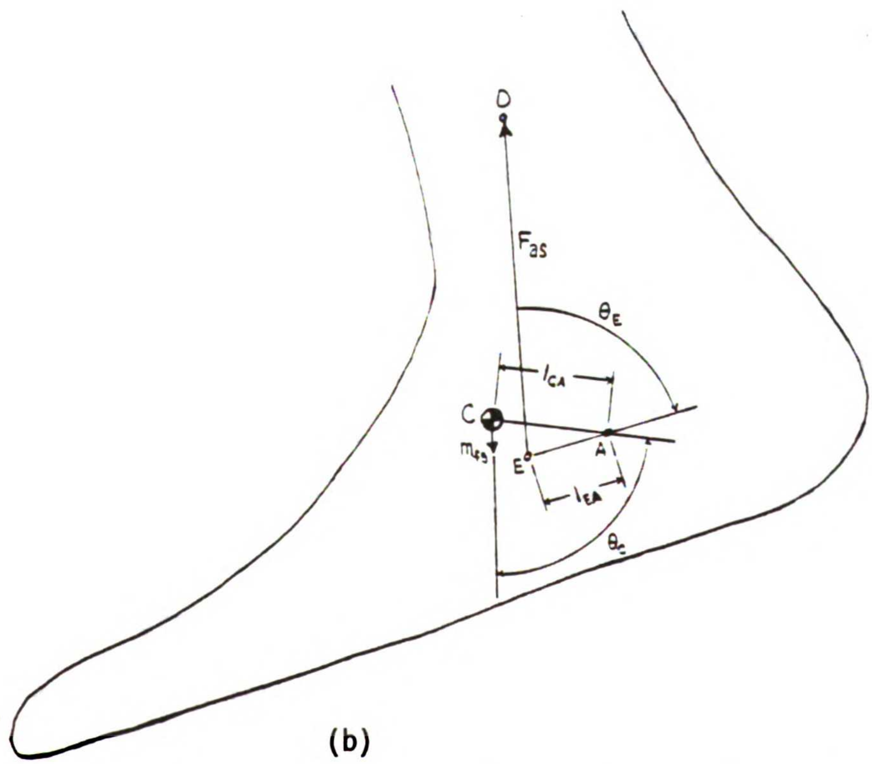
where $R_x(t)$ and $R_z(t)$ are, respectively, the fore-aft and vertical components of the ground reaction force, $G_x(t)$, and $G_z(t)$ are, respectively, the x and z components of the point of ground reaction force application, and $\theta_{DB}(t)$ is the angle between the horizontal axis and the posterior spring-damper assembly, and t is time.

The displacement of the spring-damper assembly, $q_1(t)$, is then determined from heelstrike to maximum heelstrike plantar flexion as

$$q_1(t) = DB(t_0)-DB(t) \tag{5.15}$$



(a)



(b)

Fig. 5.7. Schematic model of proposed system: (a) showing the load on the posterior assembly and with the anterior spring removed, (b) with the posterior spring-damper assembly removed.

where $DB(t)$ is the distance from $D(t)$ to $B(t)$, and t_0 is the time at heelstrike. The distance, $DB(t)$, is calculated as

$$DB(t) = ((D_x(t)-B_x(t))^2+(D_z(t)-B_z(t))^2)^{1/2} \ . \tag{5.16}$$

The rate of change of $q_1(t)$ from heelstrike to maximum heelstrike plantar flexion, v_1, is then estimated as the slope of the least-squares fit line through a plot of $q_1(t)$ versus time (see Figure 5.8).

With the estimated v_1, the bond graph model of the posterior spring-damper assembly (see Figure 5.5) is then used to calculate the displacement of the spring, $q_3(t)$, and the rate of change of displacement of the damper, $v_2(t)$, based upon initial estimates for the posterior spring compliance, C_3, and damper resistance, R_2. This bond graph model utilizes a zero-junction, a flow source, S_f, a C-element, and an R-element (64). For a zero-junction, the efforts (or in linear mechanical terms, the forces) associated with all bonds are equal, and the sum of the flows (i.e. velocities) associated with all bonds is zero. The flow source, S_f, represents a known flow (i.e. rate of change of posterior assembly length for this model) to the system. The C-element is an energy storage element (i.e. spring for this model) for which displacement is proportional to effort. The R-element is an energy dissipation element (i.e. damper for this model) for which effort is proportional to flow. For $q_3(t)$ and $v_2(t)$ the bond graph model, with initial estimates for C_3 and R_2, predicts

$$q_3(t) = R_2C_3v_1 - R_2C_3v_1\exp(-t/(R_2C_3)) \ , \text{ and} \tag{5.17}$$

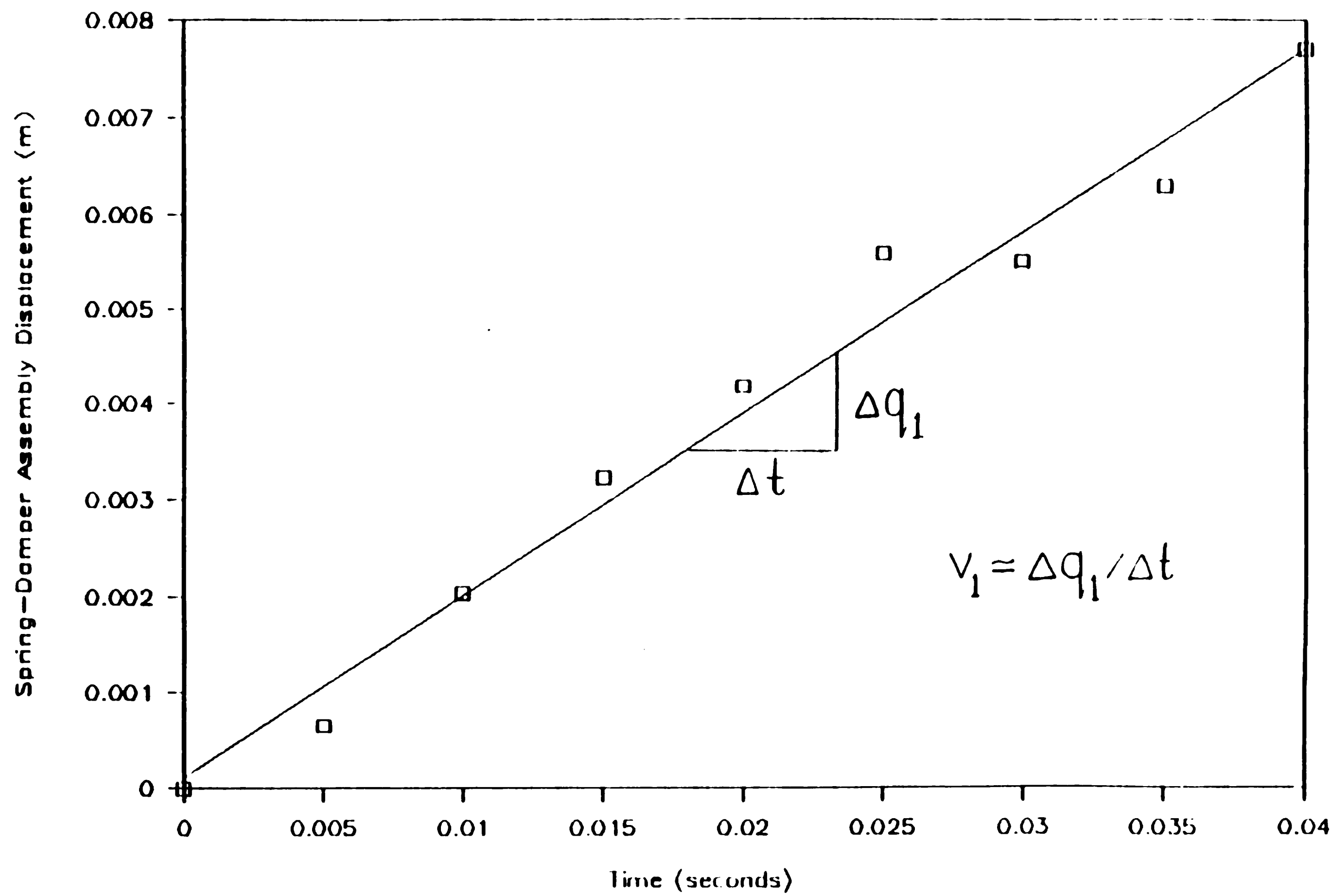


Fig. 5.8. Regression of spring-damper assembly displacement versus time from, heelstrike to maximum plantar flexion, to determine input rate of spring-damper assembly displacement.

$$v_2(t) = v_1 - v_1\exp(-t/(R_2C_3)) \,. \tag{5.18}$$

where initial estimates for C_3 and R_2 were taken to be 15 x 10^{-6} m/N and 1.5 x 10^3 N-sec/m, respectively.

The spring compliance is then estimated as the time average of spring displacement divided by spring-damper assembly load

$$C_{3,est} = \frac{\int (q_3(t)/P(t))dt}{\int dt} \,. \tag{5.19}$$

This estimate is then averaged with the preliminary estimate for C_3 to obtain a new estimate for C_3

$$C_{3,est,new} = \frac{C_{3,est} + C_{3,est,old}}{2} \,. \tag{5.20}$$

Similarly the damper resistance is estimated as the time average of spring-damper assembly load divided by the rate of change of damper displacement

$$R_{2,est} = \frac{\int (P(t)/v_2(t))dt}{\int dt} \,. \tag{5.21}$$

The estimated resistance is then averaged with the preliminary estimate for R_2 to obtain a new estimate for R_2

$$R_{2,est,new} = \frac{R_{2,est} + R_{2,est,old}}{2} \cdot \tag{5.22}$$

This iterative process is continued until the differences between successive values for C_3 and R_2 are below a specified tolerance of 1%.

The final estimate for C_3 and P(t), from maximum heelstrike plantar flexion to toe off, are then used to determine the displacement and energy storage of the spring-damper assembly for the remainder of stance as

$$q_1(t) = C_3(P_{mpf}-P(t)) + q_{mpf}, \text{ and} \tag{5.23}$$

$$e(t) = \tfrac{1}{2}C_3P^2(t) \tag{5.24}$$

where, P_{mpf}, and q_{mpf} are the load on and displacement of the spring-damper assembly at maximum plantar flexion following heelstrike, and e(t) is the energy stored in the posterior spring.

To estimate the compliance for the anterior spring the overall schematic model was considered with the posterior assembly removed (see Figure 5.7b). This approximation was made because during swing phase the damper resistance will be very low. Consequently, very little force can be sustained in either the posterior spring or damper since they are in series.

To return the system to its original neutral position the anterior spring must act against the weight of the foot applied at the foot center of gravity. Equilibrium at the original neutral position can be obtained when the sum of moments about the prosthetic ankle equals zero, or

$$M_A = 0 = l_{EA}F_{as}\sin\theta_E - l_{CA}m_f g\sin\theta_C \tag{5.25}$$

where, M_A is the summation of moments about the prosthetic ankle joint, A, F_{as} is the force exerted by the anterior spring, m_f is the foot mass, g is the acceleration of gravity, l_{EA} is the distance from E, the anterior spring connection at the foot, to A, l_{CA} is the distance from C, the foot center of gravity, to A, θ_E is the angle between F_{as} and l_{EA}, and θ_C is the angle between $m_f g$ and l_{CA}. The force exerted by the anterior spring can be obtained as

$$F_{as} = \frac{\Delta l_{as}}{C_{as}} \tag{5.26}$$

where, C_{as} is the anterior spring compliance, and Δl_{as} is the displacement of the anterior spring from its free length. Substituting for F_{as} and rearranging yields the estimate for anterior spring compliance as

$$C_{as} = \frac{l_{EA}\Delta l_{as}\sin\theta_E}{l_{CA}m_f g\sin\theta_C}\ . \tag{5.27}$$

Estimation of the anterior spring compliance was based upon a displacement from free length at equilibrium of about 1 mm, a foot weight of about 4.5 N, a ratio of l_{EA}/l_{CA} of about 1.1, and letting the ratio of $\sin\theta_E/\sin\theta_C$ be 1.

RESULTS

Using the equations derived from the bond graph model in the iterative manner described yielded an estimate for the damper resistance of approximately 2.3 x 10^3 N-sec/m (see Figure 5.9). Compliance of the posterior spring was estimated from this procedure as 10 x 10^{-6} m/N (see Figure 5.10). Figures 5.11 and 5.12 show the calculated posterior spring-damper assembly displacement, and posterior spring energy storage during stance. Peak energy storage during late stance was approximately 14 J. Use of the schematic model with the posterior assembly removed produced an estimate for anterior spring compliance of about 0.2 x 10^{-3} m/N.

DISCUSSION

System analyses of proposed prosthetic systems can be useful in evaluating potential function and feasibility of such designs. In the present analysis motion and ground reaction data from a subject at a self-selected velocity have been imposed on the system and energy storage during the stance phase of gait has been estimated. In light of energy storage measurements of 3-4 J for uniaxial and Greissinger foot/ankle systems (83), the peak energy storage estimate of 14 J in the posterior spring would seem encouraging. Increased energy return at push off could

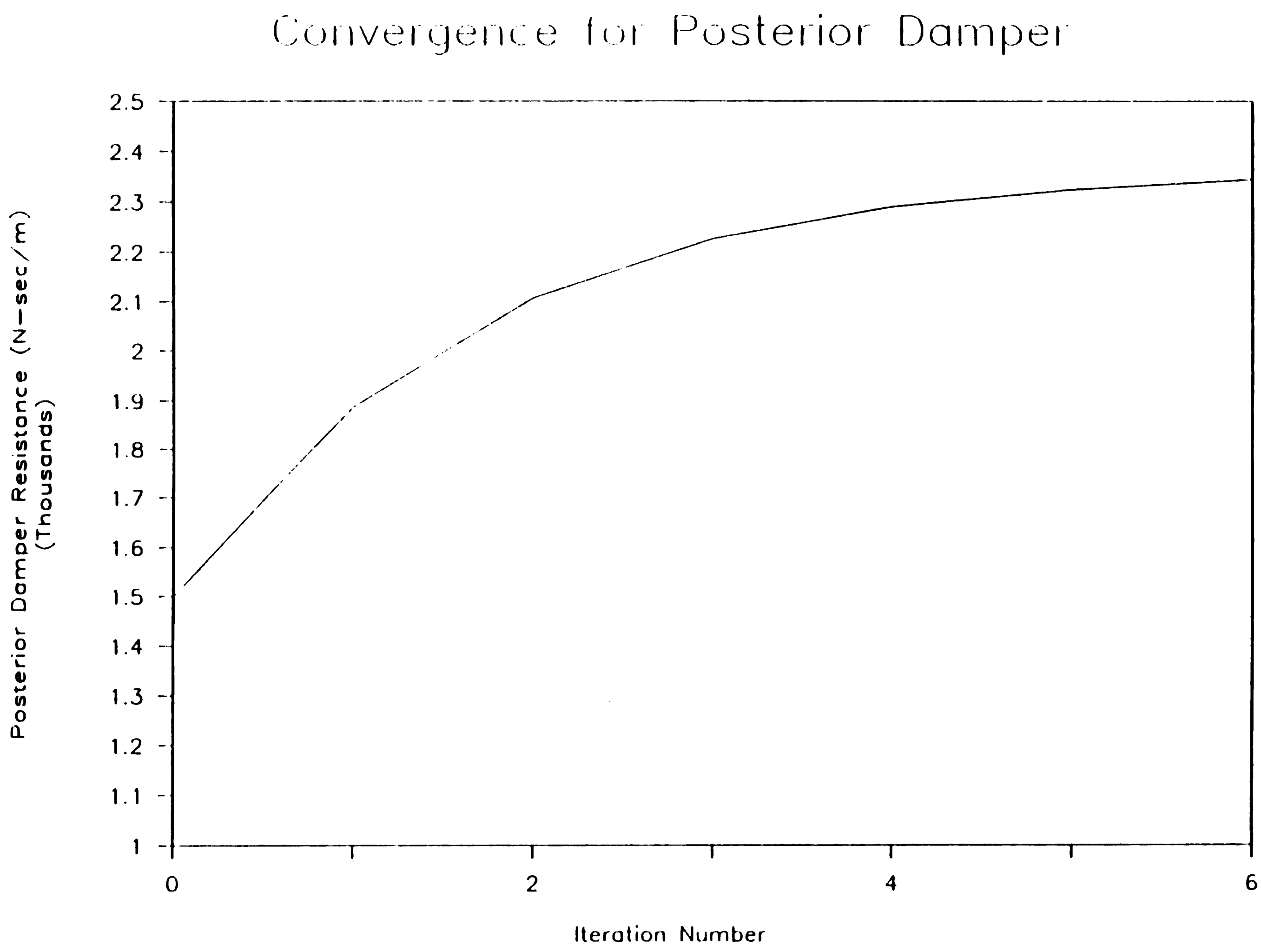


Fig. 5.9. Posterior damper resistance estimate at each iteration until convergence.

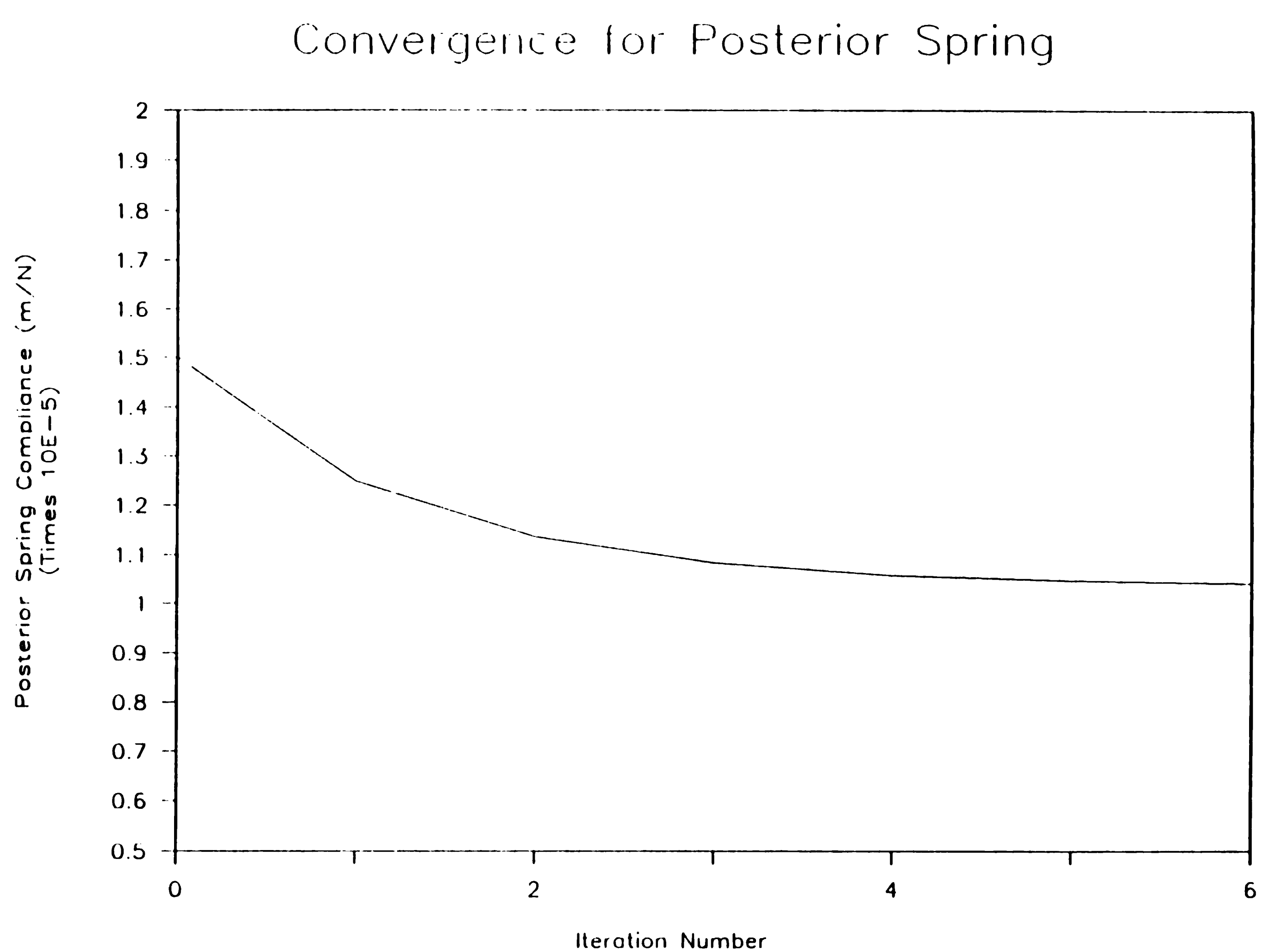


Fig. 5.10. Posterior spring compliance estimate at each iteration until convergence.

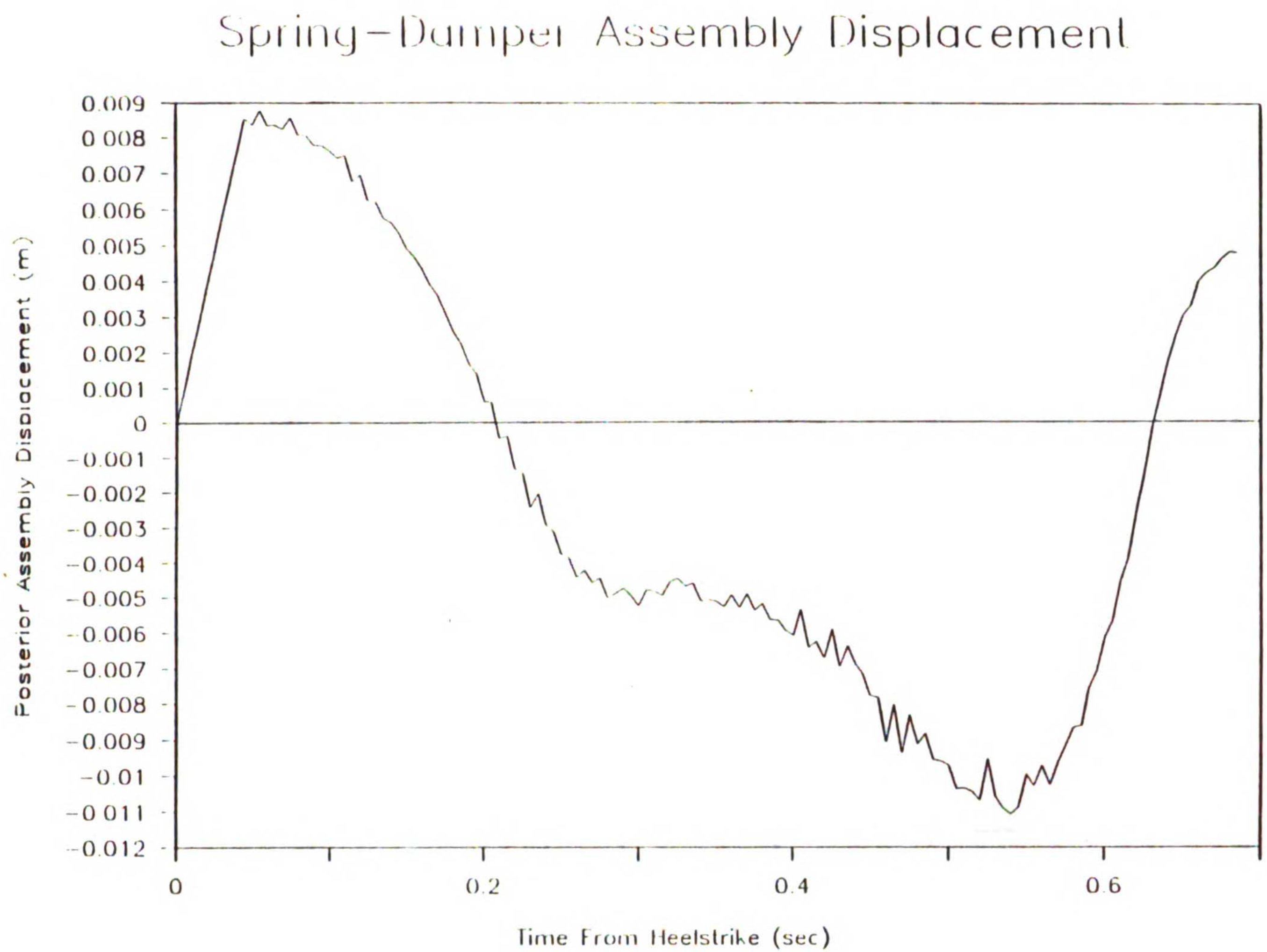


Fig. 5.11. Displacement of the posterior spring damper assembly during stance.

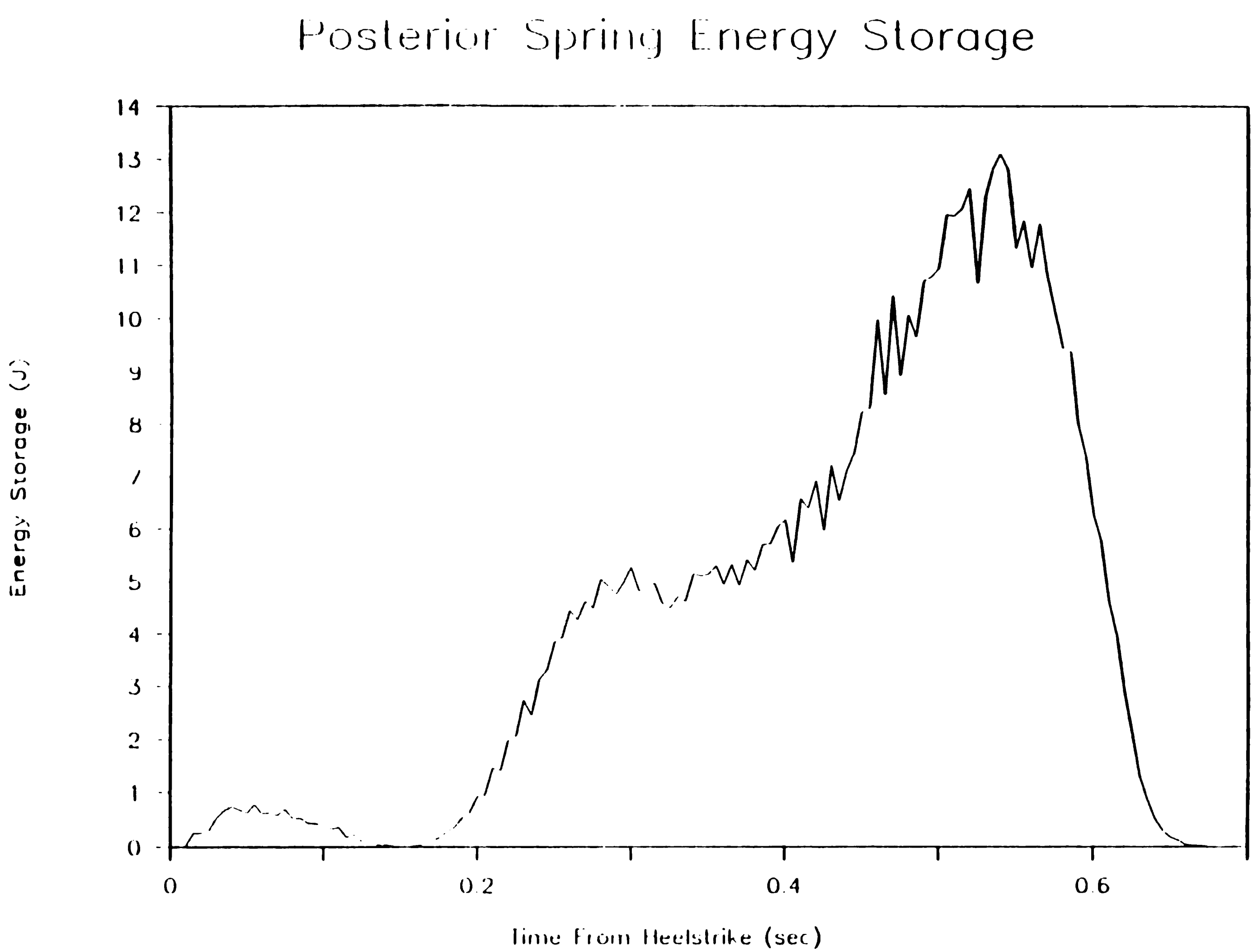


Fig. 5.12. Energy storage of the posterior spring during stance.

improve ambulation of lower limb amputees by diminishing compensatory mechanisms such as increased hip extension energy (83). Decreased hip energy generation may result in reduced oxygen consumption during gait.

Specification of the compliance and resistance of the spring and damper permits determination of the physical characteristics (i.e. materials and dimensions) of those components. For example, these properties are related to the compliance of a helical spring (70), and resistance of a fluid filled damper (24) by

$$C = \frac{8D^3N}{(d^4G)} \quad \text{,and} \tag{5.28}$$

$$R = \frac{\rho A v D_1^{\,4}}{(2K^2 D_t^{\,4})} \tag{5.29}$$

where C is the spring compliance, d is the spring wire diameter, D is the mean spring diameter, G is the spring material shear modulus, N is the number of spring coils, R is the damper resistance, ρ is the damper fluid density, A is the damper cross-sectional area, v is the rate of change of damper length, D_1 is the diameter of the damper, D_t is the diameter of the damper orifice, and K is an empirical constant based on the ratio (D_t/D_1), and the Reynolds number ($Re = \rho v D_1/\mu$, where μ is the fluid viscosity).

Posterior spring compliance has been estimated to be approximately 10 x 10^{-6} m/N; and damper resistance immediately following heelstrike has been estimated at about 2.3 x 10^3 N-sec/m. From these relationships it has been estimated that such the compliance for the posterior spring could be obtained with a titanium helical spring with 5 coils, 30 mm mean spring diameter, 7.1 mm wire diameter, and approximately 70 mm free length. Such a spring would weigh about 3.1 oz. An oil filled piston type damper with the indicated resistance has been estimated to be about 32.5 mm in diameter, 3.1 mm in orifice diameter and 50 mm in length, and to weigh approximately 3.6 oz. For the anterior spring, a titanium helical type with 12 coils, 19 mm mean spring diameter, 3 mm wire diameter, and 70 mm free length would have approximately the indicated compliance. In view of the recent work by Foerster et al (23), indicating that minimizing mass should be a foremost consideration in the specification of components for the foot, as well as the dimensional constraints involved with lower limb prosthetics, these specifications appear reasonable.

Realizing the limitations resulting from the use of gait data from a single subject, further analyses should use gait data from several subjects walking at a range of velocities in order to assess the effects of size, weight, and velocity on the system component parameter estimates. While theoretical relationships for spring compliance can be expected to estimate spring specifications rather well, the effects of such variables as coil spacing, and end condition (i.e. plain, ground, squared, ground and squared) may not be completely described. Spring compliance measurements and modifications may be necessary to fully determine spring specifications. Such measurement and modification of damper specifications would likely be needed to account for the effects of multiple orifices with non-circular geometries.

CONCLUSIONS

Vertical ground reaction loading rates were measured for subjects with tibio-femoral osteoarthrosis who were asymptomatic following arthroscopic debridement, and for age-matched control subjects. No difference was found between the loading rates of the two age groups; consequently, these results do not support the hypothesis that has been proposed for the etiology of osteoarthrosis involving excessive ground reaction loading rates. Additionally, no relationship was found between vertical ground reaction loading rates and vertical pre-heelstrike velocities or average free walking velocities.

Lower external flexion moments about the knee at midstance were found for the injured limbs of subjects, who had received ACL reconstructions, and compared with midstance knee moments of age-matched control subjects or those for the ACL reconstruction subjects' contralateral limbs. Lower midstance knee flexion moments indicate lower net quadriceps reactions which suggest lower tensions in the reconstructed ACLs. It appears that these subjects, whether consciously or not, are somewhat, but not completely, shielding their ACL reconstructions from experiencing loads to which the control subject and contralateral ACLs were subjected. The lack of complete "quadriceps avoidance gait" (as identified by previous investigators in a majority of ACL deficient patients) indicates that the ACL reconstructions have benefitted the patients since they are able and willing to sustain loading requiring a net quadriceps reaction. High prevalences of osteoarthrosis have been reported for patients with ACL injuries, however, the previously mentioned mechanism proposed for the etiology of osteoarthrosis was not born out by the measurement of lower ground reaction loading rates for the ACL reconstruction subjects compared with those for control subjects.

Above knee amputees demonstrated no difference between their ability to reproduce prosthetic limb knee angle deflections, in the range of approximately 5° to 25°, and their ability to reproduce sound limb knee angle deflections. However, the threshold for detection of very slow passive motion was significantly greater for the prosthetic limbs than for the sound limbs. Consequently, different neural mechanisms appeared to be involved in these two measurements. The generally larger hip motions produced during reproduction testing, which are typically greater than previously reported hip motion threshold detection values, indicate that proprioception cues arising from hip motion aided prosthetic limb reproduction. Smaller hip motions during knee motion threshold detection testing were generally below the reported hip motion detection thresholds, suggesting that for their prosthetic limbs, subjects had to rely on cutaneous sensations to detect the onset of motion. Such cues were not sufficient for the subjects to detect prosthetic limb motion as well as sound limb motion.

Energy storage was calculated during simulated stance phase for a proposed prosthetic foot/ankle system. Peak energy storage was considerably greater than previous measurements of energy return for single-axis and greissinger systems, although it was about half the ankle energy generation previously measured for normal subjects. Parameters for the components of the proposed system also were calculated from the analysis. Approximation for component specifications were determined which appeared reasonable for prosthetic foot/ankle systems. The calculated improvement in energy return over previously measured systems seems to encourage a design of this type. Analyses such as this may be useful in evaluating the potential of other proposed prosthetic systems.

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