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Relative contributions of the lower extremity joint moments to forward progression and support during gait

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Abstract

A method, which was found to be accurate within 0.54 m/s^2, was developed to estimate the relative contributions of the net joint moments to forward progression and support in the gait of five normal subjects. Forward progression was produced primarily by the ankle plantar flexors with a significant assist from the knee extensors. Support was produced largely by the plantar flexors during single limb support and by a combination of ankle plantar flexors, knee extensors and hip extensors during double limb support. © 1997 Elsevier Science B.V.

Keywords: Plantar flexors; Joint moments; Support moments; Lower extremity; Forward and vertical acceleration

Introduction

The motor tasks used to transport the body in human gait can be divided into five major functions [1]: (1) generation or maintenance of forward velocity, (2) support of the upper body (prevention of lower limb collapse during stance), (3) balance of the body, (4) control of the foot trajectory during swing and (5) shock absorption. An understanding of how these functions are accomplished in healthy individuals can provide useful insight to clinicians examining a person with gait deficits. This study evaluated the relative contributions of the lower extremity joint moments to the first two functions: generation of forward velocity (forward acceleration) and support of the upper body (vertical acceleration against gravity).

Perry [2] stated that the generation of forward velocity in gait is characterized by a roll-off rather than a push-off, with the body undergoing a controlled fall as it moves over the foot. Perry's conclusions were based in part on a study by Simon et al. [3] that compared the gait of normal subjects to those with no significant calf muscle function. These authors found that subjects with normal calf muscle function extended the body's center of gravity more anterior to the center of pressure than subjects without calf muscle function. These results indicated that the generation of forward velocity in gait may arise from active control (plantar flexion) of a passive mechanism (falling). Sutherland et al. [4] presented additional evidence that a controlled fall may contribute to the generation of forward velocity in gait. The authors reasoned that if the plantar flexors had direct propulsive power, then there should be a drop in forward velocity when these muscles were paralyzed. The authors found that subjects who were given a tibial nerve block increased their forward velocity during late stance. Despite this finding, Sutherland et al. [4] were not convinced that forward acceleration was simply a function of body alignment. The authors stated that the action of the ankle plantar flexors on the center of mass required further study and interpretation.

In an attempt to clarify the role of the ankle joint muscles in forward progression, Winter [5] examined the power output at the ankle and knee joints during normal gait. Winter noted that the primary source of positive work came from the plantar flexors (11 J to 65 J). He also noted that the timing of the ankle plantar flexor power burst coincided with the second peak of

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the vertical ground reaction force. These results were interpreted to indicate that generation of forward velocity was characterized by a plantar flexor push-off rather than a passive roll-off.

Joints other than the ankle also have been associated with the generation of forward progression in gait. Inman [6] and Simon [3] stated that the deceleration of the swing leg may contribute to forward progression. Although the energetics of the swing leg have been examined in detail [7–9], the relative contribution of the swing leg to the generation of forward velocity has not been established.

A second motor function during gait is the support of the upper body (prevention of lower limb collapse during stance). Support of the upper body can be accomplished through passive transmission of forces through the joints or facilitated by muscles that act across the joints to produce rotation. Ideally, gait analysis would be able to determine the relative contribution of the individual muscles as well as the passive elements to the support of the body. However, the determination of individual muscle forces is complicated by the redundancy of the musculoskeletal system, and thus examination of support is usually limited to the net joint moments.

The support moment has been used to determine the relative contribution of the lower extremity joint moments in preventing lower limb collapse [10]. Winter [10] defined the support moment as the sum of all joint moments in a lower extremity. Positive values were assigned to extensor moments because they are believed to prevent collapse and negative values were assigned to flexor moments because they are believed to facilitate collapse. The general agreement between the shape of the ground reaction force and the support moment indicates that it is a useful measure for explaining how the lower extremity supports the upper body. Winter [11] also noted that variability of the ankle joint moment was small when compared with the knee and hip moments. Despite the high variability of the moments at the knee and hip, the support moment variability remained relatively low. This led Winter to conclude that neural control of walking involves a total lower limb pattern.

The support moment, however, can be limited when applied to some types of pathological gait. First, the support moment is based on the assumption that all flexor moments contribute to collapse; an assumption that may be incorrect for subjects with knee hyper-extension. A second limitation of the support moment is that it does not discriminate which portion of a given joint moment is being used to support the body and which portion is being used to generate forward progression. A final limitation of the support moment is that the contribution of each joint to support is assumed to be directly proportional to the magnitude of its moment. This assumption is only an approximation due to the multi-link characteristics of the human body. Zajac and Gordon [12] demonstrated that the torque at a single joint will produce accelerations at all joints and that the magnitude of these accelerations will be dependent upon the configuration of the body segments as well as on the magnitude of the torque.

Meglan [13] applied these multi-link principles to look at how the moments at each joint accelerate the pelvis segment along the direction of progression. Although Meglan's work represents an important attempt to estimate the relative contributions of the joint moments to the forward acceleration of the upper body, it did not examine the contribution of the joint moments to the vertical acceleration. A second limitation of Meglan's model was that it did not account for how the joint moments work in a closed kinetic chain; a situation that must be accounted for during the stance phase of gait.

The purpose of this study is to examine the relative contribution of the lower extremity joint moments to support and forward progression during gait. This study accomplishes this goal by using a model based on the principles outlined by Zajac and Gordon [12], including closed kinetic chain effects, to directly estimate the relative contribution of each joint to the forward and vertical acceleration of the upper body.

2. Theoretical model

The model used in this study estimates the forward and vertical acceleration of the upper body produced by the net moments estimated at each joint. The mathematical basis for the model was outlined by Zajac and Gordon [12], who demonstrated that the moments produced by muscle forces around a joint will generate accelerations at all joints of the body. This principle can be understood by stating the equations of motion in the following form:

$$q = M^{-1}T + M^{-1}C + M^{-1}G + M^{-1}F \qquad (1)$$

In this expression, q is the matrix containing the joint accelerations, M^{-1} is the inverse of the inertia matrix, T is the matrix containing the joint moments, C is the matrix containing the Coriolis terms, G is the matrix of gravitational terms and F is the matrix containing external forces.

The accelerations produced solely by the joint moments can be obtained by setting the matrices C, G and F to zero. This reduces the equation of motion to:

$$q = M^{-1}T \qquad (2)$$

The accelerations produced by an individual joint moment can be determined by setting all other joint moments to zero in the matrix T and calculating the elements of q. Thus, it is assumed that the contribution a joint moment makes to the acceleration of the other segments can be determined by applying that moment while considering all other joints to be frictionless joints with no torques or stiffness.

Eq. 2 confirms that a joint moment will act to accelerate all the joints of the body even if the other joints are considered to be frictionless joints with no stiffness. The magnitude of the accelerations produced by that moment will be a function of both the magnitude of the moment, as specified in T, and the configuration of the body segments, as specified in M. (The elements of the inertia matrix M are dependent on both the inertial properties of the segments and the position of the segments). Once the matrix of joint accelerations has been obtained, simple geometric principles can be used to determine the forward and vertical accelerations that a joint moment produces on the individual segments. This approach illustrates why the configuration of the body should be accounted for when determining the relative contribution of the joint moments to support or progression.

The physical model used in this study consisted of seven segments: two feet, two legs, two thighs and a single head-arms-trunk (HAT) segment. The inertial properties of the segments were customized to each subject based on geometric measurements [14] and scaled using anthropometric data compiled by Winter [15]. The ankles and hips were treated as spherical joints and the knees were treated as pin joints. A 2.5-degree-of-freedom constraint was added to each foot when the foot was in contact with the floor. This constraint, applied at the center of pressure, prevented the foot from translating down into the floor and from translating along the floor. Both feet were constrained during double limb support, and the constraint was removed from a foot during swing phase. No rotational constraints were applied to the feet at any time.

3. Methods

Five subjects, two men and three women, were examined while walking at a self selected pace with their arms crossed against their chest. The subjects ranged from 25 to 40 years in age and were free of any pathologies that may have affected their gait. Informed consent was given by all subjects prior to testing. Five data trials were collected on each subject with one representative trial used as input to the model. The forward acceleration data were evaluated only when the combined ground reaction force obtained from the two force platforms was anteriorly directed (forward acceleration interval). This interval began at right mid-stance and ended during double limb support. The support data were evaluated for one-half of a gait cycle starting from mid-stance on the right leg and ending with mid-stance on the left leg (half-cycle interval). (In this study, mid-stance was defined as the time in single limb support when the ground reaction force changes from posterior to anterior).

A six camera Vicon data collection system (Oxford Metrics, Inc., Oxford, England) measured the gait of the subjects over a $0.6 \times 1.2 \times 1.2$-m volume. The motions of the leg and thigh segments were tracked with clusters of four non-colinear reflective targets, while the motions of the feet were tracked with clusters of three targets. The motion of the HAT segment was tracked by placing four targets on the pelvis. (It was assumed that negligible errors resulted from using the pelvis to track the combined center of gravity of head, arms and trunk). Two strain gauge force platforms (Advanced Medical Technologies, Inc., Newton, MA) supplied the ground reaction force data required to estimate the net joint moments. The video data were sampled at 50 Hz and the force plate data were sampled at 200 Hz. A fourth order Butterworth filter was used to remove noise from both video and force plate data. Filter cut-off frequencies for the video data were dependent on the target locations (foot = 6 Hz, leg = 5 Hz, thigh = 4 Hz and pelvis = 3 Hz). Force platform data were filtered using a 25-Hz cut-off frequency.

The contribution of each joint moment to the acceleration of the HAT center of gravity was calculated using MOVE3D (National Institutes of Health, Bethesda, MD) and ADAMS (Mechanical Dynamics, Inc., Ann Arbor, MI) software. For each video sample, MOVE3D computed the net joint moments, the joint orientations, the locations of the center of pressure relative to the feet and the orientation of the right foot relative to the laboratory. The Android module of the ADAMS software configured the physical model based on the MOVE3D output and the anthropometric data. Translational constraints were added to the model during stance phase to prevent the foot from sliding along the surface or from going down into the floor. These constraints were applied to the foot at the center of pressure.

After configuring the model, for each data frame the gravitational constant and all but one of the joint moments were set to zero. The Solver module of the ADAMS software was then used to compute the forward and vertical acceleration of the HAT center of gravity produced by that joint moment (solve Eq. 2). Because ADAMS is a forward dynamics program that does not directly solve Eq. 2, a 0.001 s simulation was used to determine the accelerations. This short simulation interval guaranteed that the joint positions remained virtually unchanged, and that the joint velocities remained near zero. This resulted in accelerations that were virtually equivalent to the accelerations that would be obtained by direct application of Eq. 2.

This process was repeated for each joint moment, and the resulting vertical and horizontal HAT center of gravity accelerations were used to quantify that moment's contribution to support and generation of forward velocity at that data frame.

The accuracy of this approach was assessed by computing a final analysis that included all joint moments and gravity (matrices T and G in Eq. 1). Since the Coriolis forces were assumed to be small and there were no external forces other than those supplied by the constraints, the center of gravity acceleration determined from the model should be approximately equal to the subject's center of gravity acceleration as computed by dividing the ground reaction force vector (obtained from the force plates) by the subject's mass. The Solver module of the ADAMS software was used to calculate the acceleration of the model's center of gravity. Mean errors, determined from the absolute difference between the model and force plate derived accelerations, were computed over the half cycle interval for each subject. These errors were reported for both the anterior/posterior and vertical acceleration components.

4. Results

The accuracy of the model was assessed for each of the five subjects by computing the absolute differences between the force platform derived and model generated center of gravity accelerations (Table 1). These differences were averaged over the interval from right mid-stance to left mid-stance and were assumed to arise from errors in the model. The mean anterior/posterior errors ranged from 0.04 to 0.14 m/s^2, and the mean vertical errors ranged from 0.14 to 0.54 m/s^2. For all subjects the mean errors were < 5% of the total range of accelerations.

The low error values were the result of the excellent performance of the model during both single and double limb support. This point can be illustrated by presenting the data over the entire half-cycle interval (Fig. 1) for a single representative subject (subject 3). The sharp rise in the vertical acceleration that occurs at 32% of the half-cycle interval is produced as the left heel contacts the force platform. This rise marks the beginning of double limb support which continues until 52% of the half-cycle interval. The consistent agreement between the model and force platform center of gravity accelerations indicates that the model can be used throughout the gait cycle to estimate the accelerations generated by a set of joint moments.

Table 1
Mean acceleration errors for the five subjects over the interval from right mid-stance to left mid-stance (half cycle interval)

Subject	Anterior/posterior acceleration error (m/s^2)	Vertical acceleration error (m/s^2)
	0.08	0.14
2	0.04	0.25
3	0.14	0.28
4	0.09	0.54
5	0.12	0.22

These errors were computed from the absolute difference between the body center of gravity accelerations, measured by dividing the ground reaction force vector by the subject's mass, and the body center of gravity accelerations generated by the model.

After establishing the level of accuracy, the model was then applied to estimate the acceleration produced on the HAT segment by each joint moment. The forward accelerations of the HAT center of gravity were used to measure forward velocity generation, and the vertical accelerations of the HAT center of gravity were used to measure support.

The right ankle joint moments were the largest contributor to forward acceleration for all subjects (Fig. 2). The production of forward acceleration by the right ankle joint moments was not limited to late single limb support (~ 50–80% of the forward acceleration interval) when the ankle joint moments acted concentrically. The data reveal that the right ankle joint moments also produced notable forward acceleration of the HAT center of gravity when the ankle plantar flexors were acting eccentrically to control forward progression of the leg over the foot (~ 0–40% of the forward acceleration interval). This unexpected propulsion resulted from the accelerations at the other joints and the rotational acceleration of the foot relative to the floor.

The right knee joint moments also contributed to forward progression in all subjects (Fig. 2). The right knee musculature was found to generate forward acceleration of the HAT center of gravity when the moments were extensor and was found to generate negative acceleration when the moments were flexor. The variability in the contributions of the right knee moments to forward acceleration were due to the differences in the subjects' knee joint moment patterns.

The right hip joint moment produced a negative acceleration of the HAT center of gravity for all five subjects. These data indicate that the hip joint moments must be acting in some role other than generation of forward progression.

The combined left knee and left hip moments generated only a small amount of forward acceleration of the HAT center of gravity during the forward acceleration interval (Fig. 3). This lack of propulsion was due to the approximately equal and opposite contributions of the knee and hip joint moments during swing. The anterior/posterior accelerations produced by the left ankle joint also remained small throughout the forward acceleration interval.

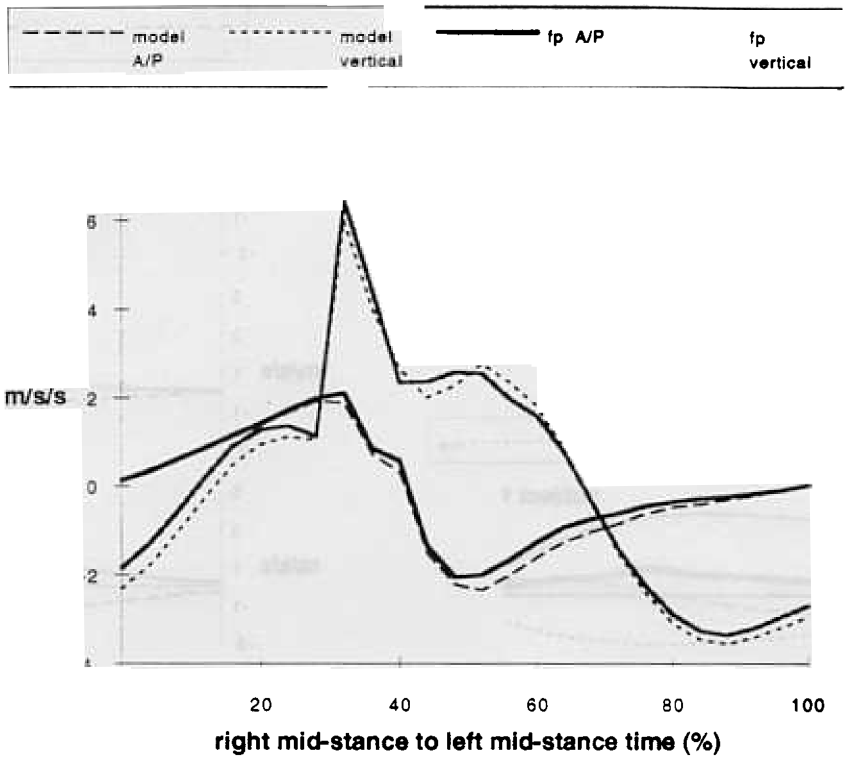

Fig. 1. Comparison of the body center of gravity acceleration as computed by the model and by the force platform for subject 3. The absolute difference between the two techniques is assumed to arise from errors in the model. The mean anterior/posterior (A/P) error was 0.14 m/s^2 and the mean vertical error was 0.28 m/s^2.

Support was examined by computing the relative contributions of the lower extremity net joint moments to the vertical acceleration of the HAT center of gravity (Fig. 4). Because the contributions of the right knee and right hip moments remained small throughout the half-cycle interval, the data for these two joints are not shown. The vertical accelerations produced by the hip abductors also remained small throughout the half-cycle interval (generally < 1.0 m/s^2).

The largest contributors to vertical acceleration of the HAT center of gravity were the left and right ankle joint moments (Fig. 4, Table 2). During the later period of right single limb support (~ 0–30% of the half cycle interval) the right ankle joint moments supplied > 90% of the total support. During double limb support (~ 30–50% of the half-cycle interval) the role of the right plantar flexors diminished. During this period the subjects used a variety of support strategies. All subjects generated some combination of support using their left hip and knee moments; however, the relative contribution of these two joint moments varied greatly from subject to subject. In addition, all subjects, except subject 2, received a small contribution to support from the left ankle dorsiflexor moments as they eccentrically lowered the foot to the floor. (Subject 2 made initial contact with a nearly flat foot and had virtually no dorsiflexor moment). As the subjects returned to left single limb support (~ 50–100% of the half cycle interval), the left ankle plantar flexors became the primary contributor to support.

5. Discussion

Two of the major motor functions that must be accomplished during human walking are the generation of forward velocity and support of the upper body [1]. Prior efforts to determine the contribution of the lower extremity joint moments to these tasks have been limited by the assumption that the relative contribution of each joint moment is proportional to the magnitude of the moment and is thus independent of the position of the body segments. This study extends existing gait analysis methodology by developing a mechanical linkage model that can provide direct estimates of the vertical and anterior accelerations produced by a given joint moment. The accuracy of the model was indirectly established by comparing the center of gravity accelerations computed by the model to the accelerations computed from the combined force platform data. The results (Table 1) indicate that the model can estimate the accelerations produced by the lower extremity joint moments with an average anterior/posterior error of < 0.14 m/s^2 and an average vertical error of < 0.54 m/s^2. The results also demonstrate that the predictive ability of the model remains excellent during both single limb and double limb support (Fig. 1).

Examination of the accelerations generated by the right ankle plantar flexors illustrates the advantage of this approach (Fig. 2). During the interval immediately following right mid-stance (0 to ~50% of the forward acceleration interval), the ankle plantar flexors acted eccentrically to control the forward progression of the leg over the foot. Eccentric activity is characterized by energy absorption that might be assumed to produce a negative acceleration of the HAT segment; however, the model indicates that this eccentric plantar flexor activity generated a forward acceleration of the HAT

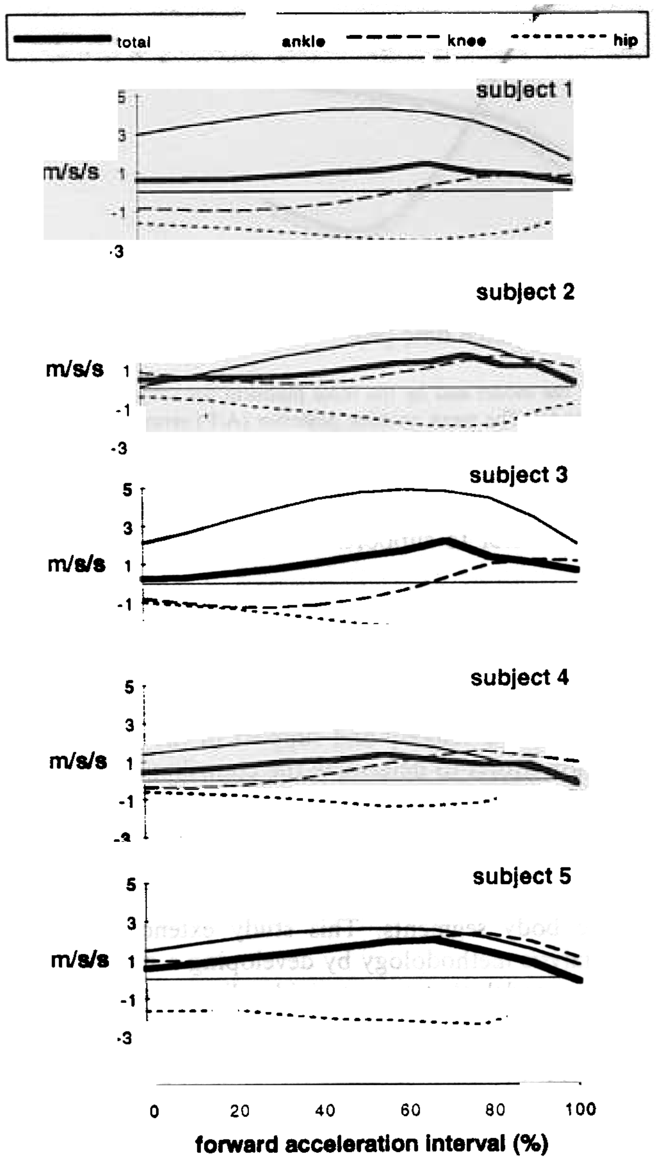

Fig. 2. Forward acceleration of the HAT center of gravity generated by the joint moments during the interval when the ground reaction force was anteriorly directed starting at right mid-stance (forward acceleration interval). The total acceleration produced by the joint moments is shown along with the accelerations produced by the moments at the right ankle, right knee and right hip. The right ankle and knee joint moments were the primary contributors to forward acceleration.

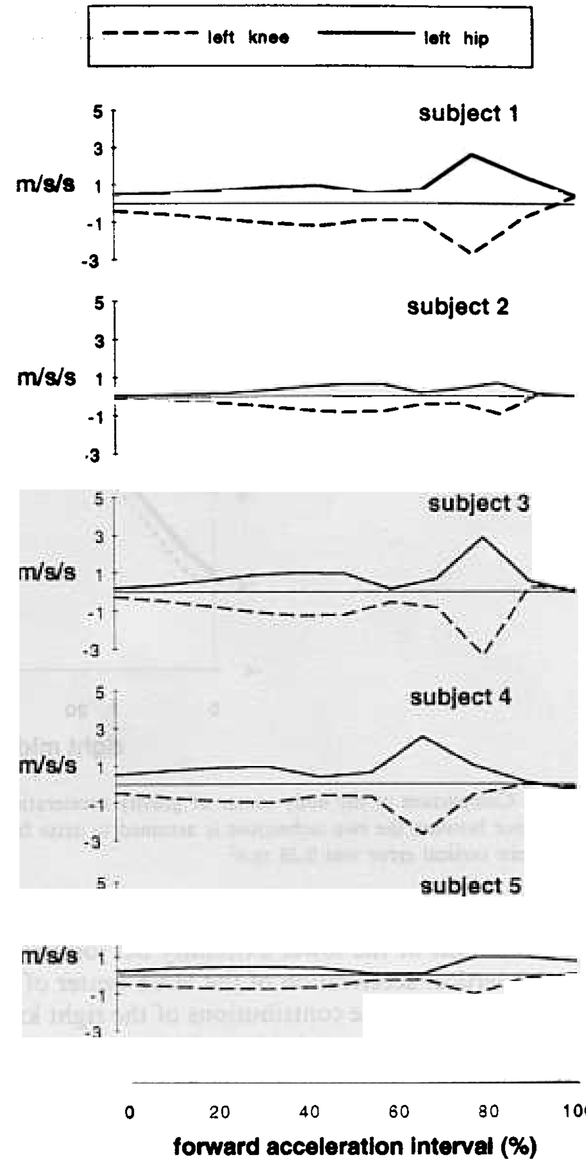

Fig. 3. Forward acceleration of the HAT center of gravity generated by the left knee and left hip joint moments during the forward acceleration interval. The left leg is in swing until ~80% of the interval. (The sharp peak in the accelerations occurs at left heel contact). The net acceleration produced by the two joint moments remains near zero in all subjects.

center of gravity.

The production of forward acceleration from eccentric plantar flexor activity arises because a moment acting across a joint will produce a reaction force at that joint. This reaction force will be transmitted throughout the linkage and will accelerate all of the segments. Eqs. 1 and 2 indicate that the accelerations produced by a joint moment are independent of velocity and thus are independent of the type of contraction. (The velocities only contribute to acceleration through the Coriolis terms). From these principles it should not be surprising that the combined effect of the observed eccentric plantar flexor activity was to produce a forward acceleration of the HAT center of gravity.

As the right ankle plantar flexors change to concentric action in late single limb support (~ 50–80% of the forward acceleration interval), they continue to be the primary source of forward acceleration (Fig. 2). This result supports Winter's [5] contention that generation of forward velocity was characterized by a plantar flexor push-off. However, the plantar flexors were not the only source of forward acceleration. Three of the five subjects (subjects 2, 4 and 5) received substantial contributions to propulsion from their right knee extensors. Thus, generation of forward velocity in normal gait may be more complicated than a simple plantar flexion push-off. These results also demonstrate that the generation of forward velocity did not arise through a controlled fall [2–4]. If gravity had been the primary source of forward acceleration, then the forward acceleration of the HAT produced by the lower extremity moments (Fig. 2, bold line, subject 3) would have been small when compared with the total acceleration (Fig. 1, A/P curves).

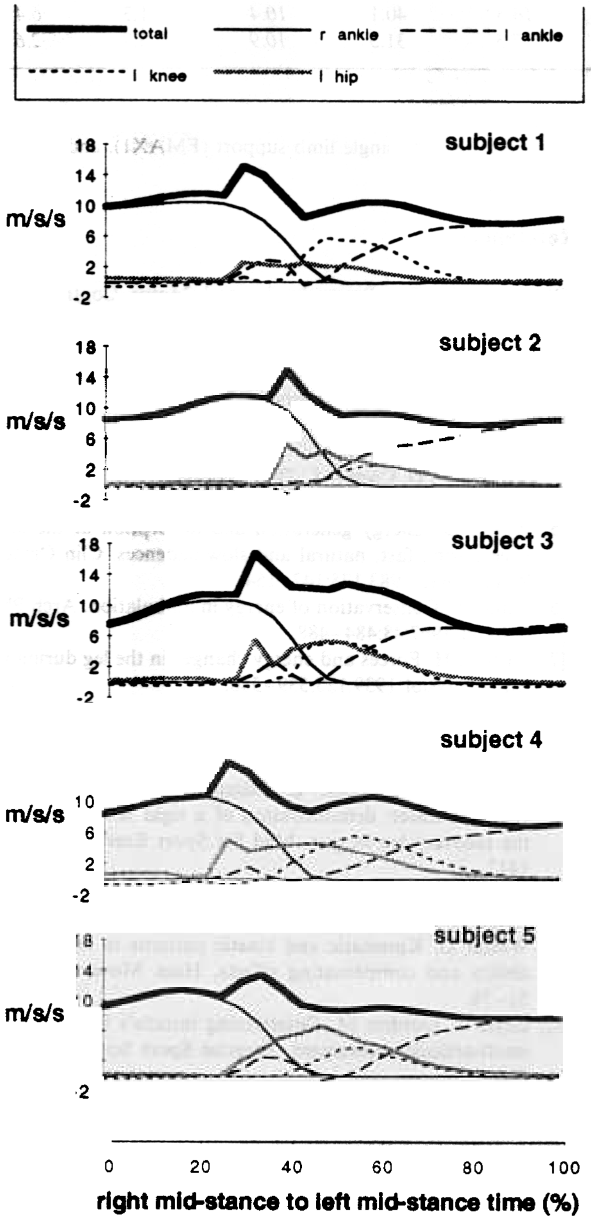

Fig. 4. Vertical accelerations of the HAT center of gravity generated by the joint moments over the interval from right mid-stance to left mid-stance. The total acceleration produced by the joint moments is shown along with the accelerations produced by the right ankle, left ankle, left knee and left hip. During 0 to ~ 30% of the half-cycle interval (late right single limb support) the right ankle moments were the primary contributors to the vertical acceleration. During double limb support (~ 30–50% of the half-cycle interval), the left knee and left hip moments produced significant vertical acceleration. As the subjects returned to left single limb support (~ 50–100% of the half-cycle interval), the left ankle moment became the primary contributor to support.

The deceleration of the swing limb has been proposed as an additional source of forward velocity generation [3,6]. The hip extensors, acting to decelerate the limb in late swing (~ 0–80% of the forward acceleration interval), produced a forward acceleration of the HAT center of gravity (Fig. 3). However, this acceleration was negated by the activity of knee flexors during the same period. The net result was that the swing limb moments did not make a significant contribution to the forward acceleration of the trunk during normal walking. The inverse relationship between the accelerations generated by the knee and hip joint moments may also indicate that the control of the swing limb strives to minimize the accelerations produced on the trunk. This finding agrees with Prince et al.'s [16] theory that one of the goals of balance during gait is the attenuation of the head's forward acceleration.

The relative contribution of the lower extremity joint moments to support was also examined in this study (Fig. 4, Table 2). During the single limb support phase of gait (~ 0–30% and 50–100% of the half-cycle interval), the ankle moments were found to generate the greatest contribution to support. In fact, during the second half of right single limb support (~ 0–30% of the half-cycle interval), the ankle joint moments produced nearly all the support generated by the five subjects. During double limb support (~ 30–50% of the half-cycle interval), the left knee and hip moments increased their contributions to support; however, the relative contribution of these two moments varied among subjects. This variation in the contribution to support can be related to the differences in the moment patterns observed at the knee and the hip. This verifies Winter's [11] concept of a flexible trade-off between the knee and hip muscles only for double limb support. During single limb support, the redundancy of moment patterns available to prevent the body from collapse diminishes as the ankle plantar flexors generate most of the vertical acceleration. Thus, this could be an additional reason [17] why the lowest variation in moment patterns have been found at the ankle joint [11].

Table 2
Percent contribution of the joint moments to the vertical acceleration averaged over all five subjects

Joint	MS		FMAX2		DLS		FMAX1		FMIN	
R ankle	102.4	*4.4*	96.2	*4.1*	34.8	*7.1*	−1.3	*0.2*	−0.6	*0.4*
R knee	1.9	*4.1*	2.0	*1.7*	6.4	*1.9*	0.6	*2.1*	0.0	*1.0*
R hip	0.4	*3.5*	3.0	*5.6*	−0.7	*2.0*	−1.7	*1.0*	−1.0	*0.7*
L ankle	−1.8	*0.5*	−0.4	*0.2*	5.1	*10.0*	30.8	*11.3*	94.6	*8.0*
L knee	−6.3	*1.9*	−3.3	*0.9*	14.8	*16.5*	40.1	*10.4*	1.3	*6.4*
L hip	3.3	*2.8*	2.6	*1.2*	39.6	*7.5*	31.5	*10.9*	5.7	*2.6*

The standard deviation is in italic.
The data are given at the following gait events: mid-stance (MS), the peak vertical ground reaction force during late single limb support (FMAX2), the middle of double limb support (DLS), the peak vertical ground reaction force during early single limb support (FMAX1), and the minimum vertical ground reaction force (FMIN).

Examination of the data also indicated that the substantial hip abductor moments generated during single limb support did not make a significant contribution to the vertical acceleration of the HAT center of gravity. In all instances, the hip abduction moments that acted to keep the HAT segment level also generated knee flexion accelerations that tended to collapse the lower limb. These knee flexion accelerations arose because the hip abduction moments generated a downward vertical reaction force on the proximal end of the thigh. This downward reaction force, which acted in the plane of the thigh and shank, tended to collapse the lower limb. Thus, the net effect of the hip abductors was to keep the HAT segment level without producing a net vertical acceleration. This result presents another example of the importance of using a mechanical linkage approach to determine the contribution of the net joint moments to support.

It was found that the generation of forward progression in gait was an active process produced primarily by the ankle plantar flexors often with a significant assist from the knee extensors. Support of the upper body was produced largely by the plantar flexors during single limb support and by a combination of ankle plantar flexors, knee extensors and hip extensors during double limb support. The process of obtaining these findings led to the development of a method for measuring the relative contributions of the net joint moments to forward progression and support.

Acknowledgments

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References

[1] Winter D. The Biomechanics and Motor Control of Human Gait: Normal, Elderly and Pathological. Waterloo: University of Waterloo Press, 1991.
[2] Perry J. Gait Analysis: Normal and Pathological Function. Thorofare: Slack Incorporated, 1992.
[3] Simon S, Mann R, Hagy J, Larsen L. Role of the posterior calf muscles in normal gait. J Bone Joint Surg 1978;60:465–472.
[4] Sutherland D, Cooper L, Daniel D. The role of the ankle plantar flexors in normal walking. J Bone Joint Surg 1980;62:354–363.
[5] Winter D. Energy generation and absorption at the ankle and knee during fast, natural and slow cadences. Clin Orthopaedics Related Res 1983;175:147–154
[6] Inman V. Conservation of energy in ambulation. Arch Phys Med Rehabil 1967;48:484–488.
[7] Elftman H. Forces and energy changes in the leg during walking. Am J Physiol 1939;125:339–356.
[8] Robertson D, Winter D. Mechanical energy generation, absorption, and transfer amongst segments during walking. J Biomech 1980;13:845–854.
[9] Caldwell G, Forrester L. Estimates of mechanical work and energy transfer: demonstration of a rigid body power model of the recovery leg in gait. Med Sci Sport Exercise 1992;24:1396–1412.
[10] Winter D. Overall principle of lower limb support during stance phase of gait. J Biomech 1980;13:923–927.
[11] Winter D. Kinematic and kinetic patterns in human gait: variability and compensating effects. Hum Movement Sci 1984;3:51–76.
[12] Zajac F, Gordon M. Determining muscle's force and action in multi-articular movement. Exercise Sport Sci Rev 1989;17:187–230.
[13] Meglan D. Enhanced Analysis of Human Locomotion. Unpublished doctoral dissertation, Ohio State University, 1991.
[14] Hanavan E. A Mathematical Model for the Human Body. Technical Report, Wright-Patterson Air Force Base 1964.
[15] Winter D. Biomechanics and Motor Control of Human Movement. New York: John Wiley and Sons, 1990.
[16] Prince F, Winter D, Stergiou P, Walt S. Anticipatory control of upper body balance during human locomotion. Gait and Posture 1994;2:19–25.
[17] Wells R. The projection of the ground reaction force as a predictor of internal joint moments. Bull Prosthetics Res 1981;18:15–19.