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The Swing Phase of Human Walking Is Not a Passive Movement

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Many studies have assumed that the swing phase of human walking at preferred velocity is largely passive and thus highly analogous to the swing of an unforced pendulum. In other words, while swing-phase joint moments are generally nonzero during swing, it was assumed that they were either zero or at least negligibly small compared to gravity. While neglect of joint moments does not invalidate a study by default, it remains that the limitations of such an assumption have not been explored thoroughly. This paper makes five arguments that the swing phase cannot be passive, using both original data and the literature: (1) Computer simulations of the swing phase require muscular control to be accurate. (2) Swing-phase joint moments, while smaller than those during stance, are still greater than those due to gravity. (3) Gravity accounts for a minority of the total kinetics of a swing phase. (4) The kinetics due to gravity do not have the pattern needed to develop a normal swing phase. (5) There is no correlation between pendular swing times and human walking periods in overground walking. The conclusion of this paper is that the swing phase must be an actively controlled process, and should be assumed to be passive only when a study does not require a quantitative result. This conclusion has significant implications for many areas of gait research, including clinical study, control theory, and mechanical modeling.

Key Words: biomechanics, motor control, gait, modeling

Bernstein's degrees-of-freedom problem (1967) demonstrates that the production of coordinated movements using multiple joints and many motor units cannot be performed unless the body has self-organizing features. However, it is not yet clear exactly what these features are, even for relatively simple human movements. One possibility, discussed by Bernstein himself, is that the physical characteristics of the limbs may constrain human movements so that the control problem becomes drastically simplified. For instance, Greene (1982) suggested that in performing many reaching tasks with the arms, one need not employ a limb trajectory specification as would a robot. Rather, the arms might simply

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extend forward as relatively free-swinging pendulums, and thus the additional control needed from the nervous system would be minimal. A similar view has often been taken of the lower extremity, at least for the swing phase of normal human walking at velocities near those preferred by the individual.

There are many reasons to entertain the notion that the swing phase of human walking is a largely unforced, pendular movement. In other words, once the foot comes off the ground, the lower extremity segments swing forward with little control from the neuromuscular system. For example, net joint moments are smaller during swing compared to stance (DeVita, 1994; Winter, 1983; 1990, pp. 254-260). EMG recordings of lower-extremity muscles are relatively quiet during swing as well (Basmajian & DeLuca, 1985, p. 136; Crowninshield & Brand, 1981). Cavagna et al. (1976) suggested that 65% of the body's mechanical energy was exchanged between kinetic and gravitational potential energies while walking near preferred velocity. Pendular models have been used to predict many aspects of the swing phase, including timing (Holt et al., 1990; Westwell, 1994), kinematics (e.g., Bach et al., 1994a, 1994b; Mena et al., 1979, 1981; Menkveld et al., 1981), and the constraint of the toes needing to clear the ground as velocity increases for a given stride length (Mochon & McMahon, 1980a, 1980b).

In fairness to all researchers who have employed pendular models of human walking, the data cited above give ample reason to experiment with a passive approximation to the swing phase. Moreover, it should not be interpreted that the conclusions of the publications cited above were entirely invalid. To quote Mochon and McMahon (1980a), "We present our model, not with the intent of making the end-all statement on human walking, . . . but rather in the hope of saying something simple about it" (p. 56).

The intent of this paper is to demonstrate that assuming the swing phase to be passive limits one to qualitative descriptions at best. Passive models of the swing phase do not appear to offer quantitative accuracy. Moreover, models with active controls must adapt to test conditions correctly. Various models employed in our laboratory, virtually identical to those in published studies (Bach et al., 1994a, 1994b; Holt et al., 1990; Tsai & Mansour, 1986), did not yield quantitatively reasonable predictions of human walking under different conditions (Whittlesey, 1997).

Assuming the swing phase to be passive has many implications for our understanding of the control of walking. In a fundamental sense, it says we have evolved to employ a locomotion strategy that requires little neural control and energy consumption for 40% of its cycle. This view has many applications to different areas of research, including pathological gaits, rehabilitation, and motor learning. By way of example, if the human swing phase were passive, then unforced pendular models of lower-extremity prostheses should be able to predict inertial modifications that would render the amputee a more normal gait pattern. Similarly, a passive view of the swing phase might imply that human swing patterns from infancy into adulthood would largely reflect the changing inertial characteristics of the limbs, that gender differences in walking were largely due to limb inertial properties rather than strength differences, and that preferred walking velocity is determined by the pendular characteristics of the swing limb.

The purpose of this paper is not simply to suggest that muscular control can be a significant factor during the swing phase. Rather, this paper seeks to dismiss the possibility that the swing phase of normal human walking is a relatively passive process. Therefore, rather than presenting the results of a single study, this paper is divided into five sections, each of which makes the same conclusion using a different approach. Three of these sections use original data that is similar to those from many other published studies, one section presents novel data, and the final section uses published data exclusively.

Methods—Data Collection and Reduction

Four subjects ages 21 to 35, two men and two women, were recruited for this study. Subject body masses ranged from 54.5 kg to 92 kg, and heights ranged from 1.57 to 1.91 m. The subjects had no lower extremity dysfunction and all signed informed consent documents in accordance with university policy.

The subjects' lower extremities were marked at the greater trochanter, lateral femoral epicondyle, lateral malleolus, and fifth metatarsal head as approximations of joint centers. Sagittal kinematic data were collected at 200 Hz as the subjects walked at their preferred velocities. Piezoresistive sensors under the heel and toe were interfaced to an LED to record the instants of toe-off and heel contact in the camera images. Timing lights on each end of a 5-meter section of the walkway were used to measure walking velocity for each trial.

Since this study concerned only the swing phase, the leg and foot were modeled as a single segment (Cavanagh & Gregor, 1975). Anthropometric measures were calculated using proportions given by Winter (1990). Kinematic data were used to calculate the sagittal angles of the thigh and leg as well as the linear acceleration of the hip in the vertical and anteroposterior directions for the swing phase. Hip and knee joint moments were computed via inverse dynamics using the equations of motion for a two-segment open-chain system (e.g., Putnam, 1991).

1) Computer Simulations of the Swing Phase

Regardless of whether they were passive or actively controlled, swing phase simulations have been implemented based on a variety of mechanical models: two-segment open chain (Bach et al., 1994a, 1994b; Tsai & Mansour, 1986), three-segment open chain (Mena et al., 1979, 1981; Menkveld et al., 1981), and complete lower-extremity representation (e.g., Gilchrist & Winter, 1997; Mochon & McMahon, 1980a, 1980b). These simulations have replicated, at least in a qualitative sense, many aspects of human walking, including swing-phase kinematics and ground reaction forces during single support.

For the purposes of the present demonstration, a simulation was implemented using a two-segment open-chain system with linear accelerations of the hip and joint moments at the hip and knee. The equations of motion of this system may be written as Equations 1 and 2 below (Putnam, 1991; Whittlesey & Hamill, 1996):

$$I_L \alpha_L + m_L d_L (L_T \alpha_T \cos(\theta_L - \theta_T) + L_T \omega_T^2 \sin(\theta_L - \theta_T) + a_{Hx} \cos \theta_L + (a_{Hy} + g) \sin\theta_L) = M_K \tag{1}$$

$$(I_T + m_L L_T^2) \alpha_T + m_L L_T d_L (\alpha_L \cos(\theta_L - \theta_T) - \omega_L^2 \sin(\theta_L - \theta_T)) + (a_{Hx} \cos \theta_T + (a_{Hy} + g) \sin\theta_T) (m_T d_T + m_L L_T) + M_K = M_H \tag{2}$$

where	m	segment mass
	I	segment moment of inertia about its proximal end
	L	segment length
	d	distance from segment mass center to proximal end
	$\theta\ \omega\ \alpha$	segment angular position, velocity, and acceleration
	a_H	linear acceleration of the hip
	g	gravitational constant
	M	net joint moment

The subscripts T and L denote the thigh and leg, respectively, the subscripts H and K denote the hip and knee, respectively, and the subscripts x and y denote the horizontal and vertical coordinate system directions.

Equations 1 and 2 were used to simulate leg swing beginning at toe-off and ending at heel contact. To accomplish this, they were first solved simultaneously for the angular accelerations of the thigh and leg, α_T and α_L, thereby yielding segment angular accelerations as functions of angular positions and velocities. The resulting equations were numerically integrated using a fourth-order Runge-Kutta method with a time increment of 0.005 second. Simulations began with experimental values for segment angular positions and velocities at toe-off, and proceeded until the subject's heel touched the ground again. The computer simulations also required the input of data for the hip linear accelerations (a_H), knee moments (M_K), and hip moments (M_H). These were taken from the four subjects' data.

Three different sets of simulations were conducted that systematically removed kinetic input from the lower extremity, as shown in Figure 1. The first set used experimentally measured values for hip linear accelerations and hip and knee moments. This simulation was designed to replicate the experimentally measured movement. The second set of simulations omitted the hip and knee moments, and the third set additionally neglected the hip linear accelerations so that the lower extremity swung as a free double pendulum.

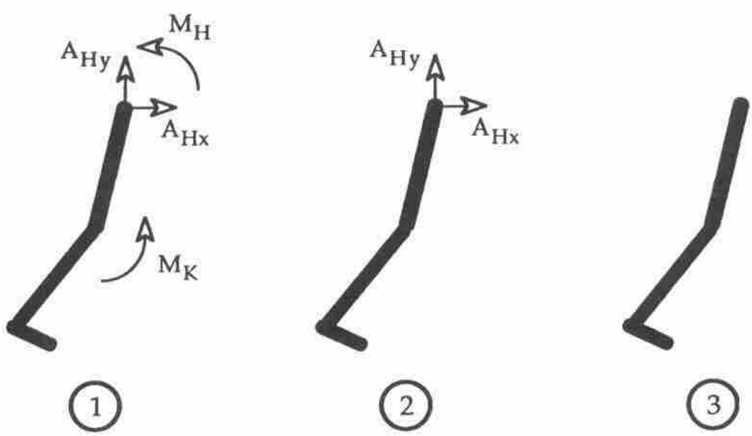

Figure 1 — Mechanical representations of the three swing phase simulations used in this study: (1) Complete representation of lower-extremity kinetics; (2) Knee and hip moments omitted; and (3) Completely passive double pendulum swing.

The parameters compared across the three simulations were the angles of the thigh and leg at the instant of heel contact of the experimental data. The angular differences provided measures by which the accuracy of the simulations could be tested.

2) Relative Size of Gravitational Moments

The relative sizes of swing- and stance-phase joint moments have often led to the assumption that the swing phase was passive. However, it seems that a better rationale for this assumption would be the relative sizes of swing-phase joint and gravitational moments. Thus, an estimate was made of the peak gravitational moments during a normal swing phase. These moments varied depending on subject anthropometry and the angular positions of the thigh and leg. As shown in Equations 3 and 4 (which are terms from Equations 1 and 2 above), the gravitational moment on the thigh had terms for the weights of both the thigh and leg, whereas the moment on the leg was a function of leg parameters only.

$$\text{Gravitational moment on thigh} = -(m_T\, d_T + m_L\, L_T)\, g \sin\theta_T \tag{3}$$

$$\text{Gravitational moment on leg} = -m_L\, d_L\, g \sin\theta_L \tag{4}$$

The following physical measures were used for calculating the moments due to gravity: thigh mass, $m_T = 9.2$ kg; leg and foot mass, $m_L = 5.6$ kg; thigh length, $L_T = 0.42$ m; distance from hip to thigh mass center, $d_T = 0.18$ m; distance from knee to leg/foot mass center, $d_L = 0.24$ m; thigh angle from the vertical, $\theta_T = -15°$; leg angle from the vertical, $\theta_L = -55°$.

The geometry of the lower extremity above is roughly representative of its position about 50 ms after toe-off. However, it should be noted that peak thigh and leg angles generally do not occur simultaneously. Overall, the goal here was to offer an estimate of the largest gravitational moments, as they would be the most conservative for the purposes of the present argument. Thus the anthropometric measures above were slightly greater than the largest of the four subjects in this study, and the angles of the thigh and leg were slightly greater than the largest angles measured during the swing phases of the subjects in this study. These angles were also greater than those reported by several studies (Putnam, 1991; Winter 1983; 1990, pp. 238-244). Peak swing-phase thigh and leg angles of all of these studies were 7.5° and 53.3° from the vertical, respectively.

3) Kinetic Inputs to the Lower Extremity

It was important within the context of this paper to provide an estimate of the percentage of the total kinetics of the swing phase that was due to gravity. Putnam (1991), among others, presented time histories of gravity moments, joint moments, and segment interactions during swing. Qualitatively, these figures suggested that gravity was smaller than any other kinetic factor during swing. Thus the kinematic data obtained from the four subjects of this paper were used to provide a basis for comparison between gravity and the other moments acting on the lower extremity. To accomplish this, the absolute angular impulses were calculated over the course of subject swing phases for the joint moments, the moments due to gravity, and the three segment interactions collectively (i.e., for each segment, these were the moments due to the accelerations of the hip and the other segment). The total impulses for each swing phase were then used to compute the proportions of the total due to the three kinetic factors.

The proportions calculated in this section seem not to have appeared before in the literature. However, for the purpose of comparing them to published data, they were calculated using data of Putnam (1991, Table 2). Putnam's table compiled average values of the various moments on each segment for four sections of the swing phase. Thus, neglecting the effects of intervals where a particular moment had both positive and negative values, these data allowed the calculation of the relative contribution of each kinetic parameter to the swing phase.

The assumption of this analysis was that joint moments represented the net actions of active muscle contractions. It has been suggested that joint moments in walking may be largely the result of passive muscle tensions (Jeng et al., 1996; Yoon & Mansour, 1982). This suggestion may be questioned because integrated EMGs have been shown to correspond with joint moments during swing (Cavanagh & Gregor, 1975; DeVita, 1994). In addition, the moments due to passive tensions have been measured to be small until the limits of the joints were approached (Mansour & Audu, 1986; McFaull & Lamontagne, 1993; Vrahas et al., 1990; Yoon & Mansour, 1982). To show that this was true for the present data, the kinematic data of our four subjects were used to estimate the moments due to the passive tensions acting about the joints using the passive moment studies cited above.

4) The Pattern of Kinetic Events

A first examination of the moments due to gravity might suggest they have the potential to develop a normal swing phase: they tend to accelerate the lower extremity at the start of swing and decelerate it at the end of swing. This fact explains why passive devices (McGeer, 1990a; McMahon, 1984) can locomote in a gravitational field. The present demonstration was designed to show that gravity alone yields simple kinetic patterns during swing since it is only a function of segment angles. By contrast, the total moments acting on the thigh and leg can be shown to have more complex patterns, quite aside from their exact magnitudes. This was accomplished by plotting time histories of the moments due to gravity and the sum of all moments acting on the thigh and leg. This section was designed to offer only qualitative statements on the complexity of the moments due to gravity, and thus no quantitative comparisons were made.

5) Limb Size and Stride Periods

Holt et al. (1990), among others, applied a prediction equation of Kugler and Turvey (1987) for walking cycle periods known as the force-driven harmonic oscillator (FDHO) model. The model equation was:

$$\text{Walking cycle period} = 2\pi \sqrt{\frac{L}{2g}} \tag{5}$$

where g is the gravitational constant and L is the length of a simple pendulum having the same natural period as the limb in question. Without the constant of 2 in the denominator, this formula gave the period of an unforced swinging limb. In other words, without this factor, the model predicted gait cycle periods that were about 1.4 times longer than those of experimental subjects. This constant was first suggested by Kugler and Turvey (1987) as taking values of 2, 7, and 10 for quadruped walking, trotting, and cantering, respectively. Holt et al. (1990) concluded that this correction factor also worked for human walking.

Holt et al. (1990, Table 2) compiled actual and predicted swing times for the 25 subjects of their study. However, only the mean periods of all subjects were presented graphically. Thus the FDHO's ability to predict intersubject variability was not discussed. This was seen here as a critical test: if this model of the lower extremity were accurate, then it should be able to estimate the differences between individuals with different body sizes. To that end, Holt et al.'s normal walking data were plotted against FHDO model predictions (Table 2 of Holt et al.), and these data points were then subjected to a linear regression.

Results and Discussion

Subjects' walking velocities ranged from 1.20 to 1.28 m/s, well within the range of preferred velocities determined by several studies (Andriacchi et al., 1977; Cavagna et al., 1976; Holt et al., 1990; 1991; Westwell, 1994).

Computer Simulations of the Swing Phase

The computer simulations using Equations 1 and 2 demonstrated that representations of muscle activity were needed to achieve accurate results. Using experimental joint moment and hip acceleration data, each computer simulation was able to reproduce the kinematics of the subjects' swing phase very closely (Figure 2A). The slight differences at the end of the simulations (averaging less than 0.4°) reflected the accumulated errors of the numerical integrations over approximately 85 iterations.

The simulations became considerably less accurate when the joint moments were removed (Figure 2B). The thigh and leg angles were about 22° less than measured. Finally, as shown in Figure 2C, a completely passive lower extremity finished its swing with a similar lag as the simulations in Figure 2B. These differences are summarized in Table 1.

Overall, these simulations demonstrated that three kinetic factors—the hip and knee moments and the hip acceleration moments—play important roles in the development of swing-phase kinematics. Leg angle discrepancies of 5 to 10° from normal have been associated with clinical abnormalities (e.g., Mueller et

Table 1 Average Thigh & Leg Angles at Heel Contact for the 3 Sets of Simulations

	Measured	No M_K, M_H	No M_H, M_K, a_H
Thigh	24.9° (±2.10)	–2.5° (±5.58)	2.6° (±5.70)
Leg	18.9° (±2.08)	–7.2° (±6.81)	–2.1° (±6.39)

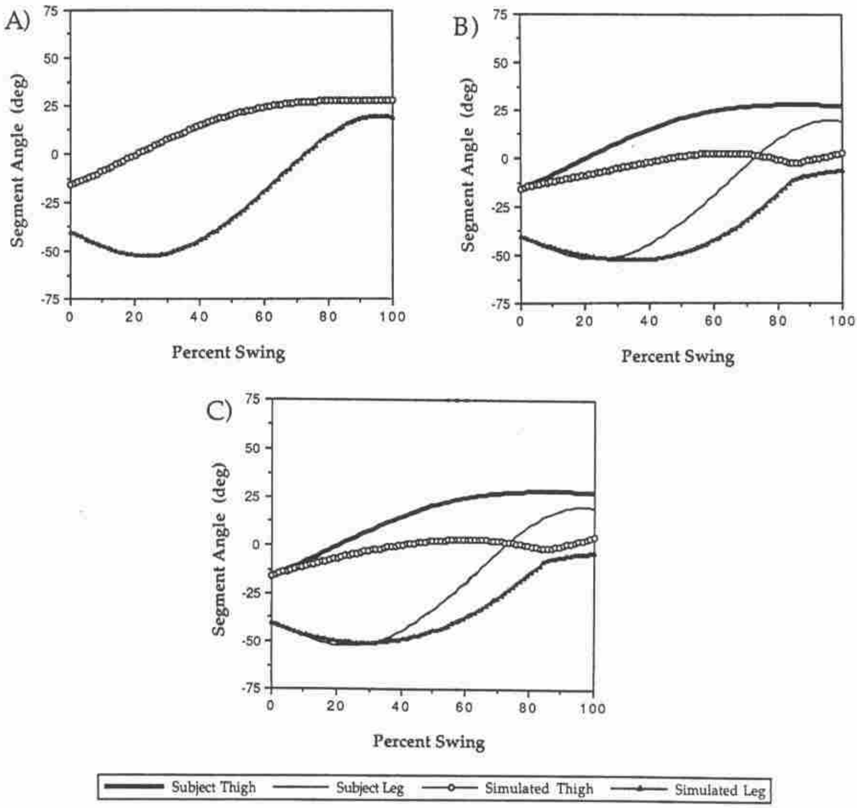

Figure 2 — Simulation of a swing phase using: A) measured hip accelerations and hip and knee moments; B) Measured hip accelerations (hip and knee moments omitted); C) A free- swinging double pendulum. Note in Figure A that the simulated and actual swing phase data overlap.

al., 1994). The completely passive case (Figure 2C), being some 22° less than normal, demonstrated that gravity alone could not perform the swing phase normally. This simulation also showed there can be qualitative resemblances between passive and active swings, at least in the sense that the lower extremity may be observed to swing forward from behind the hip. However, the motions are quantitatively distinct.

Relative Size of Gravitational Moments

Using Equations 3 and 4, the gravitational moment acting about the hip was computed to be 10.2 N•m, and about the knee, 10.8 N•m. Again, these values were extreme: segment angles θ_T and θ_L were greater than any that had been observed among the subjects in this paper and in the literature, and the anthropometric measures were greater than those of a subject who was 1.91 m tall and weighed

92 kg. An average subject would record even smaller values. Without resorting to further discussion of intersubject variability, these calculations show that swing-phase joint moments should not be ignored despite being small compared to stance-phase values. Even "small" joint moments on the order of 5 to 10 N•m are comparable to gravitational input. Thus the 20 to 40 N•m joint moments typically observed during swing (e.g., Figure 3; also Cavanagh & Gregor, 1975; Winter, 1990) are still much greater than those due to gravity. Based on these numbers, it is difficult to argue that the swing phase can be thought of as gravitationally controlled.

Kinetic Inputs to the Swinging Lower Extremity

To demonstrate the relative magnitudes of the kinetic inputs to the lower extremity, specific terms were extracted from Equations 1 and 2 above and plotted.

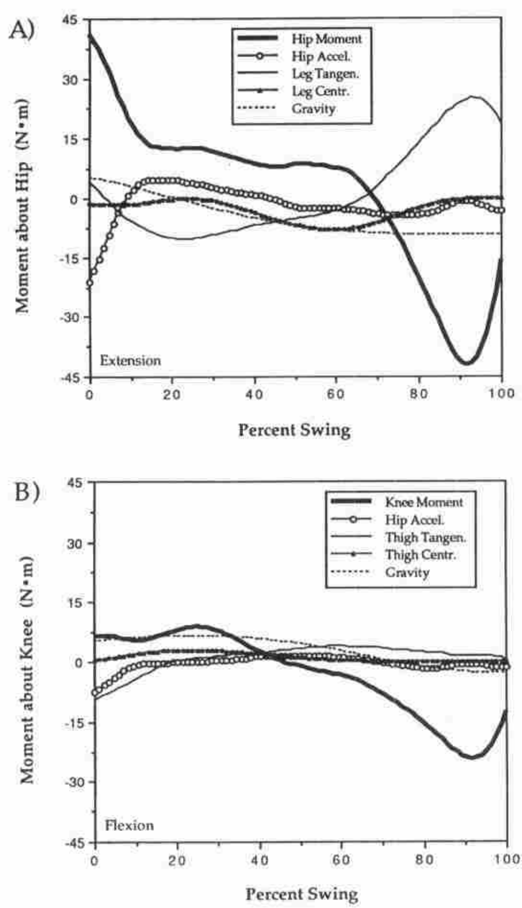

Figure 3 — Moments acting on the thigh (A) and leg (B) during the swing phase.

Table 2 Average Proportions of Total Absolute Angular Impulses to the Thigh & Leg During the Swing Phases

Kinetic source	Present Data		Putnam (1991)	
	Thigh	Leg	Thigh	Leg
Joint moment	36.6% (±5.2)	45.8% (±3.6)	29.8%	40.8%
Segment interaction	45.0% (±4.6)	31.7% (±3.5)	60.6%	36.7%
Gravity	18.4% (±1.1)	22.5% (±0.8)	9.6%	22.5%

Note: Data from Putnam were calculated slightly differently from the present data; see text for details.

Equation 2 partialed the kinetics acting on the thigh into moments due to hip and knee musculature, hip linear acceleration, leg angular acceleration, leg centripetal acceleration, and gravity. Similarly, Equation 1 modeled the leg as subject to moments due to knee musculature, hip linear acceleration, thigh angular acceleration, thigh centripetal acceleration, and gravity. These moments are plotted in Figure 3 for the thigh and leg for one swing phase of a subject walking at 1.2 m/s. Putnam (1991) offers a comprehensive discussion of these five moments. For the purposes of this paper, the exact patterns of these moments are not relevant; rather, it is important to note their relative magnitudes. Gravity and centripetal accelerations were never observed to exceed 10 N•m, whereas joint moments and segment acceleration moments peaked in the range of 25 to 50 N•m, depending on the subject's body size and locomotion style.

The proportions of the total absolute angular impulses over the swing phase are listed in Table 2. Gravity was the smallest of the three kinetic factors acting on either the leg or thigh, with an average of 22.5% and 18.4% of the total kinetic contribution, respectively. These results were roughly confirmed using the data of Putnam (1991). In fact, the same 22.5% of the leg kinetics were due to gravity. However, gravity was calculated to contribute only 9.6% of the thigh kinetics. While there may be explanations for these differences, such as sensitivity to the thigh angle at toe-off, Putnam's data at least confirm the minority role of gravity determined from the present data.

There appears to be some debate concerning the nature of the three kinetic factors in Table 2. Joint moments are generally interpreted as active controls because they represent muscle activity, and gravity moments are clearly passive. However, segment interactions have at times been discussed as passive (e.g., Eng et al., 1997; Patla & Prentice, 1995) since they result from joint reaction forces. Such interpretation appears to be inaccurate. While it may be simply a matter of arguing the nomenclature, it is important to clarify it in this paper. Segment interactions result from both muscular and gravitational actions on segments, and thus the relative contributions to a segment interaction should be considered. In other words, collectively labeling segment interactions as passive ignores the muscular actions on other segments. For example, the thigh acceleration moment results from thigh acceleration, which is due to the actions of both muscles and gravity on the thigh. In the case of toe-off, one can see in Figure 3A that the hip

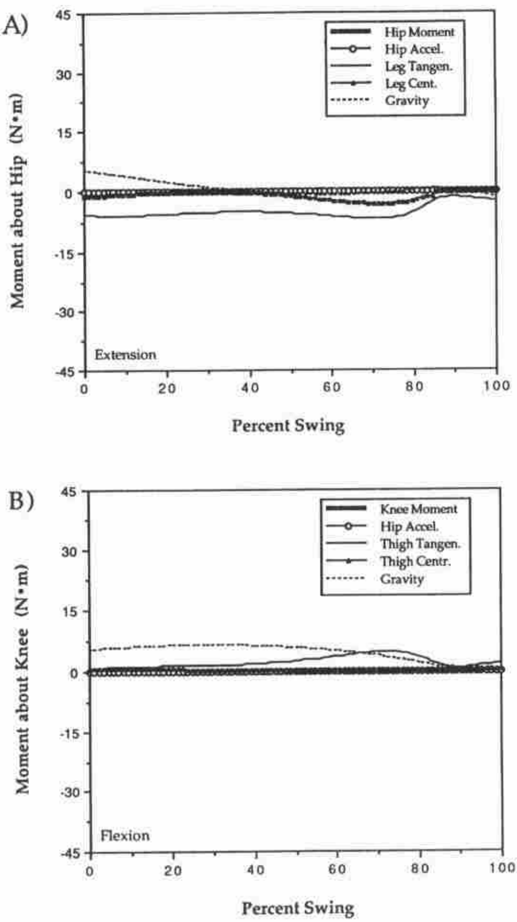

Figure 4 — The interaction moments acting on the thigh (A) and leg (B) from the swing-phase simulation of Figure 2C. When compared to Figure 3, these graphs show that absence of muscle action greatly reduces the segment interactions.

moment was the largest kinetic factor acting on the thigh at toe-off, and the result of this action was a thigh acceleration moment acting on the leg (Figure 3B). Thus, at this instant the thigh acceleration moment may be thought of as a primarily active effect. As another example, the moments due to hip linear acceleration may be thought of as external, active inputs because they largely reflect the effort of the stance limb.

To demonstrate this point in another manner, the thigh and leg kinetics in the passive simulation (Figure 2C) were decomposed into their five component torques—muscle, hip acceleration, tangential and centripetal accelerations, and gravity—and plotted in Figure 4. In this case, all the segment interactions are dramatically smaller as compared to Figure 3 because the muscle actions were

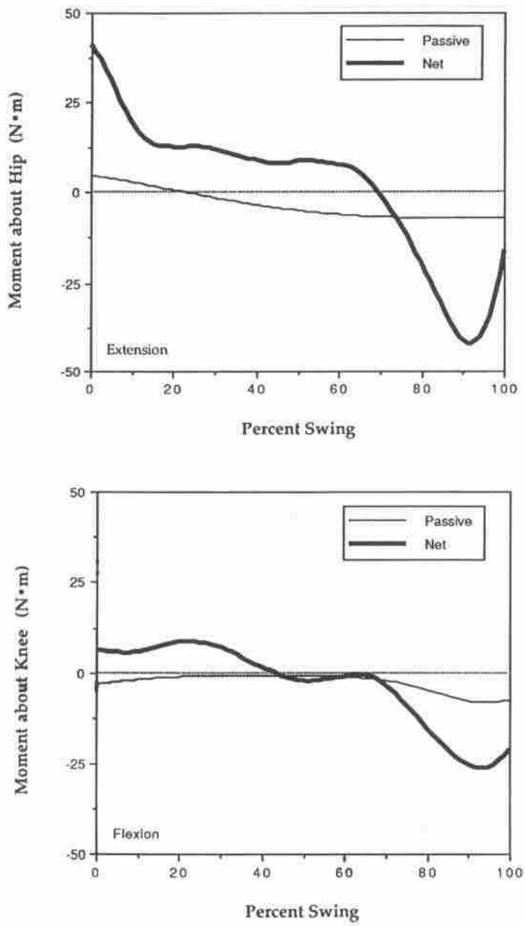

Figure 5 — A) Hip moment as measured for one subject for one swing phase, and the passive moment calculated using subject kinematics and the measurements of Yoon and Mansour (1982); B) Knee moment and the passive moment using measurements of McFaull and Lamontagne (1993).

removed from the simulations. Thus again, it should be clear that segment interactions combine active and passive effects, and that labeling them as passive is inaccurate. Therefore the segment interactions have not been labeled here as either passive or active. As a conservative approximation, one may use the relative proportion of the joint moment to gravity, or about 2:1. This leaves roughly two-thirds of the total kinetic input to either the thigh or leg as being active in nature.

The estimated moments due to passive structures are plotted for representative trials in Figure 5 using the data of Vrahas et al. (1990) for the hip and the data of McFaull and Lamontagne (1993) for the knee. The peaks of these moments are summarized in Table 3 for all subjects and trials.

Table 3 Average Peak Joint & Passive Moments About the Hip & Knee for the Swing Phase

Moment	Joint (N • m)	Passive (N • m)	Proportion
Hip, toe-off	56.9	4.2	0.07
Hip, heel contact	–50.7	–13.5	0.27
Knee, toe-off	11.3	–2.7	*
Knee, heel contact	–31.0	–8.6	0.28

Note: Rightmost column is the proportion of passive component to the joint moment. The value marked with an asterisk was not calculated since the passive moment in fact was a resistance overcome by the joint moment.

Both Vrahas et al. (1990) and Yoon and Mansour (1982) observed that passive moments about the hip did not exceed 10 N•m for hip angles between –5° (extension) and 40° (flexion). Given that hip moments in the present data peaked at about 50 N•m in both flexion and extension during swing, and that hip angles ranged between 5 and 30° flexion, consistent with published data (DeVita, 1994; Winter, 1983, 1990), it seems clear that the passive moments were a relatively minor proportion of the total muscular contribution. This conclusion is supported by Vrahas et al. (1990), who estimated about a 10% contribution of the passive structures about the hip to the hip joint moment during walking.

It is equally important to emphasize that calculations of joint moments do not include co-contractions or friction. Thus, while we have a precise measure of the moment due to gravity, we must also recall in this particular comparison that our estimates of muscular activity represent the minimal muscle action possible to achieve the observed task.

Overall, it is important to note the relative magnitudes of the parameters calculated in this section. Gravity was not close to having a majority of the kinetic input to the lower extremity: the contribution of gravity was estimated to be about one-third of the total, and then only by including some of the segment interactions. Similarly, passive joint tensions were a minority of the measured joint moments.

The Pattern of Kinetic Events

The direction of a gravitational moment is position-dependent. In other words, gravity acts only in the direction that minimizes the system's potential energy. Thus, rather than assisting in the development of swing-phase kinematics, gravity can at times be a hindrance that must be overcome. Figure 6 presents time histories of the moments due to gravity, along with the sum of all moments acting on the thigh and leg across a swing phase. Figure 6B shows that in the first 10% of swing, gravity resisted the upward acceleration of the leg off the ground. This can be understood simply from lower extremity kinematics: the foot will not leave the ground without muscular actions in one limb or the other. Figure 6A shows that gravity was also a resistance during the last 20% of swing. In this

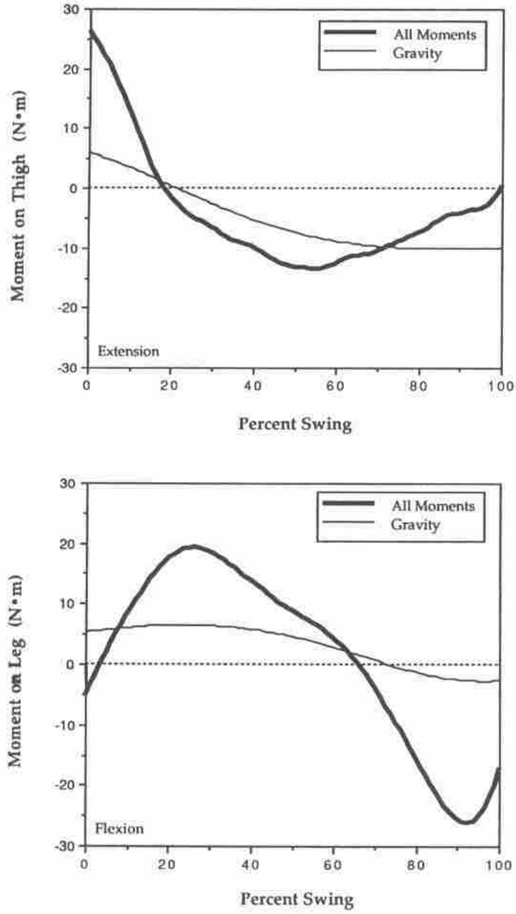

Figure 6 — A) Moments due to gravity and the sum of all moments acting on the thigh ($I_T\ \alpha_T$) over one swing phase; B) Same as above, only for the leg.

case, the thigh was held against gravity in an extended position while the leg rotated into its heel contact position.

Figure 6 also demonstrates that gravity's dependence on segment angle denies it the ability to change magnitude rapidly. Quite aside from the magnitudes of the total moments themselves, it may be argued that it is the rapid acceleration of the limbs at toe-off and their subsequent rapid deceleration before heel contact that allows the majority of the swing phase to be performed at a velocity suitable for locomotion. An entirely passive swing phase, while incurring no metabolic cost on its own part, would be slow and therefore would increase the metabolic cost to the stance limb. Given the relative size of the stance moments, it seems that the body opts to make relatively small muscle actions on the swing limb to decrease the period of single support, thereby reducing the total metabolic cost.

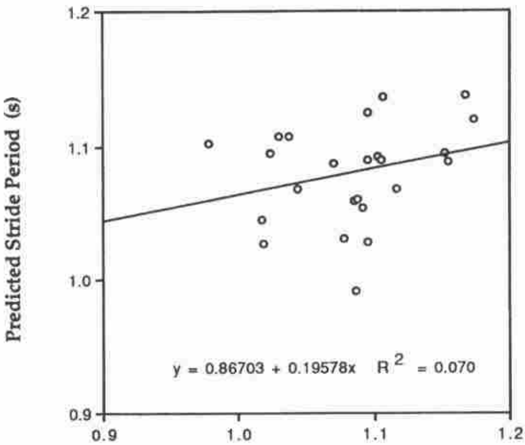

Figure 7 — Actual stride periods versus those predicted from a pendulum of identical characteristics to the swing limb, from data of Holt et al. (1990). $r^2 = 0.07$.

Limb Size and Stride Periods

Figure 7 shows that preferred stride periods and those predicted by the FDHO model were weakly correlated ($r^2 = 0.07$). In other words, the predicted swing times based on the pendular characteristics of the limbs had virtually no correlation to the periods at which the subjects chose to walk. This was not a result of a small range of observed body sizes: heights ranged from 1.57 to 1.91 m, and body masses ranged from 45.6 to 89.3 kg. Very low common variances (r^2 = 0.027 to 0.035) were also determined for data of Jeng et al. (1996), although for sample sizes of only 6. These correlations are well within the bounds suggested by the proportions of kinetic input in Table 2. If gravity delivers between 20 and 33% of the total kinetics to the lower extremity segments, then predictions of free-swinging pendulums should not be expected to generate predictions of human walking cadences with correlations higher than 0.33. The fact that the correlations herein were so much lower than this may owe to the fact that each lower extremity swings as a pendulum for only 40% of its cycle. This possibility may be supported by FDHO predictions for preferred swing periods of freely swinging lower extremities (Obusek et al., 1995, Table 2). The actual and predicted periods in this study had higher common variances ($r^2 = 0.487$), although the predictions were an average of 125 ms longer than those observed.

The FDHO model has been cited many times in the literature, but no explanation has been provided for the constant of 2 in its denominator. Kugler and Turvey (1987, p. 237) suggested that it was a "scalar multiple of the gravitational field intensity." It has also been interpreted as due to "elastic restoring torques" on the same order as gravity (Jeng et al., 1996; Obusek et al., 1995). Concerning this latter proposal, the "restoring torques" would have to be perfectly synchronized with gravity in order to be of the same magnitude. However, it is well understood from many studies that human muscle actions about the hip and knee do not have simple oscillatory patterns over the course of a gait cycle (DeVita, 1994; Putnam, 1991; Winter 1983, 1990).

Regardless of how the factor of 2 in the denominator is interpreted, Figure 7 shows that, even with this adjustment, the FDHO model fails to predict walking cadences based on the physical characteristics of the limb. However, it seems difficult to justify these "integer multiples of g" when they appear under a radical. In other words,

$$\frac{1}{\sqrt{2g}} = 0.707g,\ \frac{1}{\sqrt{3g}} = 0.577g,\ \frac{1}{\sqrt{4g}} = 0.500g,\ \frac{1}{\sqrt{5g}} = 0.477g, \text{ and so on.}$$

The differences between these factors become small for multiples of g greater than 2, thus casting into doubt the "integer multiples" for quadrupeds. This suggests that the factor of 2 is merely a constant of regression. In other words, it works for group averages, not individuals. This suggestion is supported by the observations that the FDHO model does not correctly predict differences between subjects, and that gravity accounts for less than one-third of the kinetic input to the lower extremity. Thus the implication here is that any physical significance drawn from this constant would be an overinterpretation.

Implications

The main conclusion of this paper is that the human swing phase cannot be a passive pendular movement. The corollary to this assertion is that the swing phase is a highly controlled, active process dominated by the actions of muscles both within and outside the swing limb. It has been shown in this study that: (1) Active control was essential for computer simulations of normal swing to be accurate. (2) Swing-phase joint moments were not small when compared to those due to gravity. (3) Gravity accounted for less than one-third of the total kinetic input to the lower extremity. (4) Moments due to gravity did not have the necessary pattern for normal human swing phases. (5) There was virtually no correlation between limb size and preferred stride period.

While the results of one study should not completely refute an approximation that has been used for years, the five demonstrations above, when woven together, make a collective statement. Their results concur strongly with each other, and there is data in the literature that replicates most of these five observations. Moreover, the results presented herein are extreme: a proportion less than one-third does not come close to a majority, and a correlation less than 0.1 does not approach a physically meaningful result.

Despite the extreme nature of the results presented here, it should never be interpreted that gravity is insignficant to the kinetics of the swing limb. On the contrary, one-third of the total kinetics is a significant proportion. Moreover, it has been shown through computer simulation (e.g., Garcia et al., 1998; McGeer, 1990b; Mena et al., 1981; Mochon & McMahon, 1980a) and physical models (e.g., McGeer, 1990a; McMahon, 1984) that a passive system can locomote under the influence of gravity alone. Simple models have served to promote our understanding of human movement, and the intent of this paper is not to discredit studies that used passive models. Rather, the present data demonstrate that such passive motions are similar to human walking only in a qualitative manner. Therefore, this paper suggests that future research should not expect passive models to explain the magnitude of a result. As a simple example, a pendular model might predict that walking velocity would be lower under a loading condition, but such a model would be inaccurate in predicting what the slower velocity would be.

It is important to address the literature in the context of the conclusions of this paper. There is much published data that supports the present paper. However, the point here is not to provide further support for the present work, but rather to suggest that the data offered herein may slightly alter the interpretation of certain data. Concerning the magnitudes of swing-phase joint moments and EMG compared to those during stance, it has been demonstrated here that the actions of gravity are even smaller. Thus, while muscles are significantly less active during swing, it appears that the lower extremity is nonetheless sensitive to them. This observation concurs with the musculoskeletal modeling of Piazza and Delp (1996), which found that swing-phase knee flexion angles were sensitive to muscular actions. Similarly, Caldwell and Forrester (1992) used the segmental power model of Aleshinsky (1986) and found the "power" due to gravity to be the smallest of any energy source during walking.

The dominance of active control offers many suggestions for clinical study. For example, in terms of amputee locomotion, simulation studies have suggested that the current practice of making lightweight prostheses is inappropriate (Bach et al., 1994b; Tsai & Mansour, 1986). The dominance of muscular control demonstrated here concurs with the prosthetic trade because it suggests that the inertia of a prosthesis should be roughly proportional to the relative strength of the amputee's stump. It would thus seem that future research should not be directed toward the development of better-swinging pendulums, but rather toward legs that are most sensitive to the hip and thigh movements that amputees employ for active control. As another example, it seems that preferred walking velocities should not be imposed on subjects based on pendular models. This should be particularly clear for pathological subjects because it has already been demonstrated for cerebral palsy patients (Jeng et al., 1996). This may also be inferred from the slower walking velocities of the elderly (e.g., Gibbs et al., 1996). In these and other clinical cases, the overall indication is that walking mechanics should be viewed from the standpoint of the individual's control/perceptual mechanisms, not as a pendular swing.

Even if a computer simulation does not assume a passive swing phase, the simulations presented in this paper demonstrated a potential problem with using experimental data as input. Specifically, such inputs constitute a majority of the kinetics of the simulations, and thus should adapt correctly to different test conditions. This task is far from straightforward, as it requires much knowledge of the mechanics of the test conditions before the simulations can be designed.

Whittlesey (1997) reported that simulations like those in the first section of this paper failed to predict changes in subjects' gait patterns when their leg inertial characteristics were altered. In that case, and also in Tsai and Mansour (1986), the joint moments used in the models were the dominant kinetic factors, but they were fixed across various loading conditions. Gilchrist and Winter (1997) reported dissatisfaction with simulated kinematics despite added constraints to keep the torso vertical and joints within their natural ranges of motion. In this case, the authors noted the need to model other constraints, particularly a reciprocal swing/stance pattern. Another problem was that while the magnitudes of the joint moments were adjusted, the time courses of these inputs did not change. In other words, conditions with different walking velocities would require input data with different time courses. To extend the comments of Gilchrist and Winter (1997), it would appear that a sound simulation approach would adjust both the magnitudes and time courses of model controls in order to simulate the effects of different test conditions.

Summary

Five arguments have been developed here to demonstrate that the swing phase of human walking is an actively controlled process. In addition, the data herein concerning the forces due to passive tensions suggest that joint moments during walking are largely active in nature. However, the variability of current data may leave room for debate, and thus a full resolution of this issue must be left to the continued development of our understanding of the magnitude and functional significance of tissue elasticity in human movement. At present, this paper seeks to demonstrate beyond debate that the swing phase is not a passive pendular swing, and that modeling it as such for most purposes would be ill-advised.

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