

Pendular model of paraplegic swing-through crutch ambulation

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Abstract—Kinematics of swing-through crutch ambulation for an individual with complete T11–T12 spinal cord injury was examined and quantitative aspects of the body-swing phase used to formulate and evaluate a 3-link pendular model. Model simulation parallels measured kinematics when shoulder motion is forced to follow the measured motion while hips and crutch tips are free pivots. Shoulder control contributes to increased ground clearance, influences timing and stride length, and gives flowing gait. Results indicate that mechanical work requirements during the body-swing phase are low. Metabolic energy demands exceed mechanical work requirements, due particularly to support of the body by the arms and shoulders. Exploiting low mechanical work requirements of the body-swing phase might be achieved through alternative mechanisms to assist ground clearance and to stabilize the wrists, arms, and shoulders while weight bearing.

Key words: *crutch, gait, paraplegic, kinematic, model.*

INTRODUCTION

Crutches have been used as an ambulation aid for the past 5,000 years (4). They remain a useful aid for disabled persons with a variety of disabilities. The energy cost and physical difficulty of using crutches is high for all individuals, although some-

what variable depending on the nature of the disability and the particular ambulation style adopted. Persons with paraplegia are sometimes effective crutch users. The purpose of this study is to examine some characteristics of paraplegic swing-through crutch ambulation.

Persons with paraplegia are encouraged to use crutches, when possible. Nonetheless, many choose wheelchairs as their exclusive or preferred method of locomotion. The rejection of crutches occurs more frequently in individuals with higher level lesions and with more complete lesions (17). The dislike of the orthoses which stabilize the joints of the lower limbs during stance is thought to be one reason for choosing wheelchairs over crutches. The knee-ankle-foot orthoses (KAFOs), needed by persons with upper lumbar and thoracic level lesions, are particularly cumbersome. Another deterrent to crutch use often cited is the physical difficulty of crutch ambulation. A greater understanding of crutch ambulation may stimulate ideas for reducing the difficulty of this ambulation style which may make crutch use an attractive alternative for some wheelchair users. The problems associated with long-term wheelchair use are well-documented. Crutch-assisted gait has the advantage of positioning users upright, at the level of their peers. It permits a greater freedom from architectural barriers and the upright stance may be beneficial to general health. On the other hand, crutch ambulation is energy-demanding, may be wearing on the joint cartilage and joint capsules, and the users may be endangered by falls.

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A 1974 survey article (16) illustrated that relatively few scientific analyses of crutch ambulation have been made. The largest group of scientific investigations have been metabolic energy studies. In these studies, crutch ambulation has been equated with moderate to heavy work (8), and with jogging or running (7,14). All investigators agree that crutch ambulation is more energy-consumptive than normal walking (5,10,11), although for some crutch users, this difference becomes insignificant at high walking speeds (6). What is not clear, is what necessitates this excess metabolic input; whether it is primarily related to the mechanical work requirements of the activity, or perhaps to the poor physiologic efficiency of the task. To answer these questions, one must analyze the mechanics of crutch-assisted gait, and the mechanics of the associated musculature.

Shoup *et al.* (16) performed a kinematic analysis of crutch gait by measuring the angles and elevations of body segments throughout the gait cycle. He concluded that one area of excessive energy use resulted from vertical height fluctuations of the body mass center. This conclusion assumed that there is no interchange between kinetic and potential energies, which is an unfounded assumption. Other sources of energy loss identified were excessive lateral crutch excursions, and the shock of crutches-strike. Wells (18) performed a more extensive study using similar techniques. Body segment angles and kinetic and potential energies were examined for able-bodied subjects. The participants were able-bodied subjects in which disabling conditions were also simulated by immobilization at either the knee joint or both the hip and knee joints. Looking at mechanical energy inputs (i.e., tally of kinetic and potential energies, as opposed to the metabolic energy cost), he concluded that the internal mechanical energy for crutch ambulation was similar to normal walking. However, this may not be the case for the spinal cord injured, where ground clearance becomes a difficult and important consideration. He also concluded that the use of the arms to support and propel the body, and the discontinuous nature of crutch ambulation, accounted for the increased difficulty and energy demand relative to normal biped walking.

A more recent study by McGill and Dainty (12) used the methods of Wells (18), in conjunction with force plate data, to analyze crutch ambulation in

children. This study was directed toward the ambulation styles common to persons using crutches for short periods during rehabilitation, rather than the ambulation styles of more permanent crutch users, such as the spinal cord injured.

In this paper, we describe results from a kinematic analysis of crutch ambulation performed by a paraplegic subject using the swing-through ambulation style. In addition, we present a mathematical model that simulates these kinematics during the period when the body swings. The 3-link pendular model was a simplified 2-dimensional representation of the crutches, body, and legs. The model was solved as both a conservative system with freely pivoting joints at the crutch tips, shoulders and hips, and also as a forced system where the time history of the shoulder kinematics was specified. The purpose of this study was not to provide a universal model of crutch ambulation. Rather, we have used a model to reveal some of the mechanical aspects of paraplegic swing-through crutch ambulation. This study used quantitative data taken from only 1 subject. We have observed this ambulation mode in other individuals. However, the prevalence of this form of ambulation among other paraplegic ambulators would be the subject of another study. We hope this work may form the basis for future studies, and that it may suggest design alternatives for the assistive aids currently used by persons who walk using crutches.

METHODS

The subject was a 40-year-old male diagnosed as having a complete spinal cord lesion between thoracic vertebral levels 11 and 12. He used forearm crutches in a swing-through gait style and had relied primarily on that method of locomotion for almost 20 years. Knee-ankle-foot orthoses immobilized his ankles, knees, and feet, which permitted weight bearing during body stance. Flexion and extension at the hip joints were passive phenomena, since he had no voluntary control of his hip flexor and extensor muscles. The range of motion at the hips was not obstructed and hyperextension of up to 20 degrees was observed.

Knowledge of the masses and moments of inertia for the various body segments was needed for mathematical modeling of the body-swing phase of

gait. Published data is available which allows estimation of these parameters based on body height and weight. This anthropomorphic data was adjusted for our subject, since it is not representative of active paraplegics whose weight distribution may be non-standard (e.g., more developed in the upper body and less developed in the lower limbs). Total body height and weight were measured. The weight of the KAFOs were also measured. The weight of the subject's lower limbs was then measured, utilizing a method proposed by Williams and Lissner (19). A sling attached to a spring scale was looped under the heels, and the legs were suspended horizontally. Assuming a normal weight distribution in the legs, the scale reading was a given proportion of the total weight based on a simple mechanical force and moment balance. The weight of the orthoses was included in this measurement. The weight of the rest of the body was total weight minus leg weight. Using the data of Dempster (2) and Drillis and Contini (3), 2 adjusted body weights were then computed. The first of these adjusted body weights reflected the total weight of an average individual whose legs would have had the same weight as those of our subject, while the second body weight similarly reflected an individual whose upper body weight would match that of our subject. These adjusted body weights were used to estimate separately the segment masses and moments of inertia for the legs and for the rest of the body (**Table 1**).

In this study, body length refers to total body height, and the other lengths are functional segment lengths. Leg length is the distance from the sole of

the foot to the hip joint with the knee in extension. Trunk length is the distance from the hip joint to the gleno-humeral joint, and crutch length is the distance from the gleno-humeral joint to the crutch tip with the arm extended. The angular velocities follow a sign convention where angles are measured with respect to the vertical and defined positive in the counterclockwise direction.

White circular targets 4 cm in diameter were used to mark the rotational centers of the ankle, knee, hip, shoulder, elbow, wrist, and crutch tip. The hip marker was fixed to the subject's shorts, the ankle marker to the boot, and all other markers were applied directly to the skin. The ear was used as an additional landmark. Videographic gait analysis techniques were used. A JVC HR2200U video cassette recorder and RCA CC0011 color television camera with zoom lens were used to record the walking trials. The camera was positioned 10 meters from the walkway and orthogonal to it. The focal length of the lens was adjusted to give an image width of at least 2 stride lengths. With this camera setup, errors in body segment angles due to visual perspective were estimated to be no greater than 1.5 degrees. A grid with lines every 10 cm was used as a backdrop and was positioned 1 meter behind the walkway. The video cassette recorder had the capacity for playback with variable speeds, including freeze frame display (**Figure 1**) with single frame advance. The recording rate was 30 images per second.

Ten ambulation trials were recorded as the subject traversed the walkway at his normal, comfortable speed. Seven of the recordings had the subject crossing the walkway, entering the field of view after the second stride. Two recordings had the subject starting from a standstill at the center of the field of view. The final recording had the subject approaching the camera, perpendicular to the walkway. A 2-dimensional sagittal plane analysis was performed.

The data were transcribed from videotape on a frame-by-frame basis. A sheet of transparent mylar plastic was placed on the face of the television monitor and a small dot was marked on the plastic over each body segment landmark. The videotape was then advanced 1 frame, and the new positions of the landmarks were marked on the same plastic sheet. The resulting transparency contained a locus of points for each of the landmarks, from heels-

Table 1.

Body segment parameters used in the solution of the pendular models.

	Total Body*	Legs*	Trunk	Crutches
Mass (kg)	94.4	32.5	51.3	13.8
Length (m)	1.83	0.84	0.51	1.51
Moment of inertia about mass center (kg m ²)		2.4	3.6	2.0
Distance from proximal pivot to mass center (m)		0.42	0.17	0.47
Initial angular velocity at toes-off (deg/sec)		-4.1	-13.2	-35.0

*Includes 5.5 kg mass of KAFO.

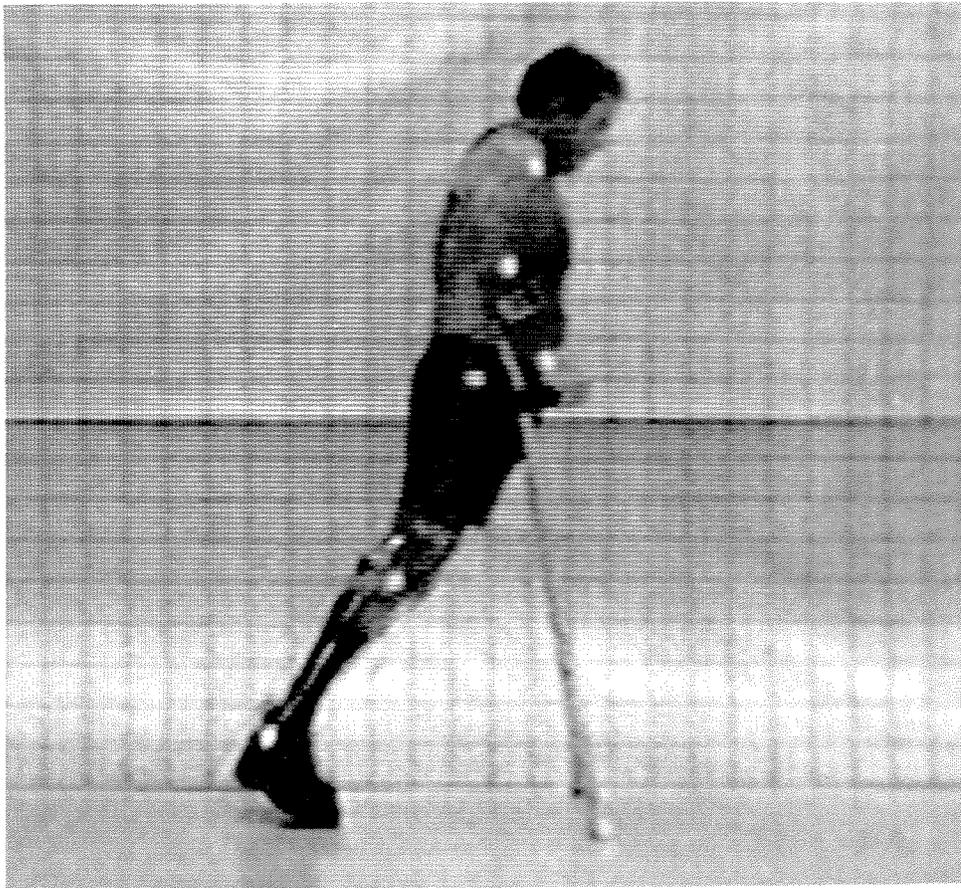


Figure 1.
Subject at toes-off: Photograph of television monitor during still-frame videotape playback.

strike of 1 stride to heels-strike of the following stride (approximately 50 points). To prevent parallax errors while transcribing the videotape, a simple jig was used to hold the head in a stable position, and an eye patch covered 1 eye. From the resulting assemblage of points (**Figure 2**), the angular orientation of each body segment was determined as it progressed through the gait cycle. The angular transcription and measurement was estimated to be accurate to within 0.5 degrees based on the scatter of the angle data after curve smoothing. The combined transcription and projection error, therefore, was within 2 degrees. Comparing the data from different experimental runs showed a consistency in the subject's gait pattern. In each successive stride, the pattern of motion was repeated both in the magnitudes of angular motion as well as the associated time histories of that motion.

MOTION CHARACTERISTICS

A visual examination of the subject's gait revealed the following characteristics: the gait cycle is initiated from a stance posture by bringing the crutches forward, shifting balance, and falling forward. The crutches are planted, and there is a short double-support phase before the body-swing phase begins. At crutches-strike, the elbows are slightly flexed. To begin body-swing, the toes are lifted off the ground by extension of the elbows and depression of the shoulder girdle. Therefore, body-swing phase does not flow directly from the preceding phase; rather, it must be initiated by an active lifting of the toes (through the trunk and legs) off the walking surface.

During the first half of the body-swing phase, the trunk appears to maintain a nearly constant orientation relative to the vertical, while the crutches pivot forward about the crutch tips, and the legs swing

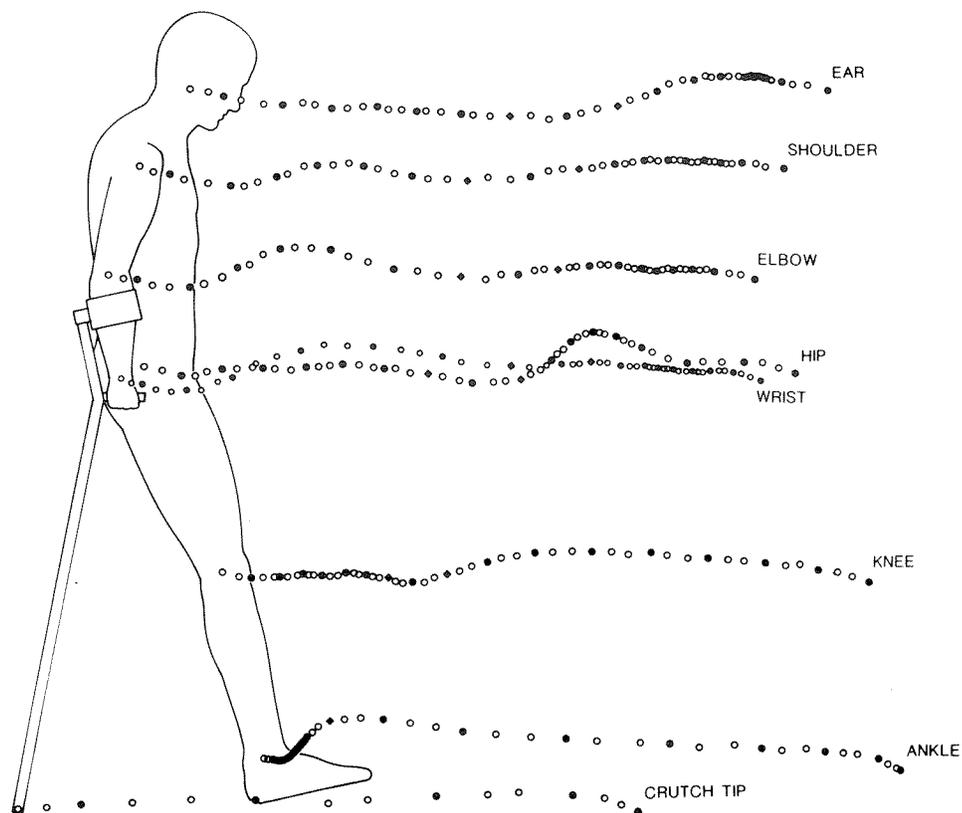


Figure 2.

Locus map used to determine body segment orientations. The map shows the time-varying position of each body landmark for 1 complete gait cycle, beginning at heels-strike. Separation between points is 1/30 second. Every third point is darkened. The events of crutches-strike and toes-off are shown with diamonds rather than circles.

from the hips, past the crutch tips, in pendular fashion. The trunk swings through in the last half of the body-swing phase; the crutches continue to pivot and the body straightens, preparing for heels-strike. When the heels strike the ground, there is a second short double-support phase before the body-stance phase begins. During the body-stance phase, the crutches are brought forward in preparation for the next cycle. The orientation of the crutches is nearly vertical as the weight is transferred to the crutches in preparation for the body-swing phase. The composite image of **Figure 3** shows these body positions in relation to the trajectories of the joint markers.

Averaging over all the experimental runs, the subject ambulated at a speed of 0.9 meters per second, with a stride length of 1.5 meters. This velocity is faster than those reported by Huang (8) in his metabolic energy studies using paraplegics (mean 0.23 m/s, range 0.04–0.75 m/s), and slower than the

mean velocities reported by Wells (18) in his kinematic studies using non-paraplegics (mean 1.02 m/s, range 0.27–1.42 m/s). On average, the relative proportion of time spent in each of the 4 phases of swing-through gait were: body-stance phase, 30 percent; first double-support phase, 12 percent; body-swing phase, 48 percent; second double-support phase, 10 percent. The body-swing phase followed equally well from a stationary standing posture as it did from flowing gait (i.e., steady state seemed to be reached almost immediately after beginning ambulation).

MATHEMATICAL MODELS

Swing-through crutch gait has 2 major phases separated by 2 transition phases. They are: 1) the body-swing phase, where the weight is supported on

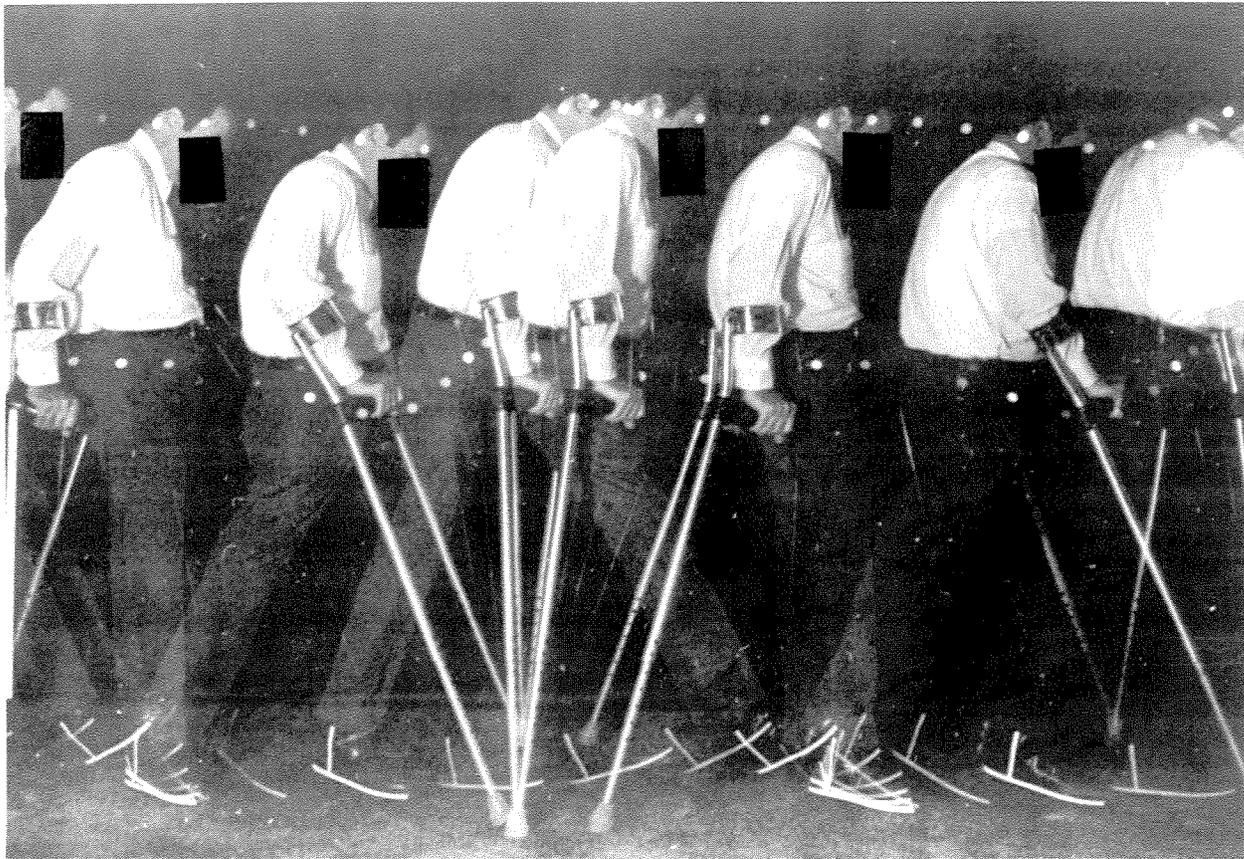


Figure 3. Stroboscopic photograph of the subject in swing-through crutch gait. Reflective markers are fixed along the shoe at the hip and on the ear lobe. The strobe frequency is 10 Hz with an intensified image every fourth strobe (2.5 Hz) to highlight the body position. This photograph was taken for illustration purposes only, and it should be noted that the stroboscopic techniques were not used for data collection.

the crutches and the feet are off the ground; and, 2) the body-stance phase, where the body weight is supported entirely on the feet and the crutches are off the ground. In the transition phases, both the feet and the crutches are on the ground, and support is being transferred from one to the other. For the crutch ambulator with paraplegia, the body-stance phase was thought to be relatively unimportant from an energy and modeling standpoint. The hips remain extended throughout this phase and, with the exception of bringing the crutches forward, the body acts more or less as a rigid column pivoting over the feet. So long as there is sufficient kinetic energy to carry the body past the vertical position and on to crutches-strike, the body-stance phase will progress to the next phase. In contrast to the body-stance phase, the body-swing phase has more complex kinematics and greater metabolic energy requirements. During the body-swing phase, there is signif-

icant angular motion at the hips and shoulders, in addition to pivoting at the crutch tips. There is also the possibility of active involvement of the shoulder musculature, restraining or driving the motion of the shoulder. The body-swing phase is presumably a phase of large metabolic activity, since the elbows must be kept extended during weight bearing to prevent flexion under the force of body weight. Similar muscular activity is needed to maintain the neutral position of the wrist and the stability of the scapula and the gleno-humeral joint.

From a mechanical point of view, it has been suggested that fluctuations in the potential energy are responsible, in part, for the high metabolic energy cost (16). Mathematically modeling the body-swing phase could facilitate the analysis and understanding of this apparently complicated motion. In paraplegic crutch gait, there are many subtleties used for control of the swinging body: examples are,

thrusting of the head, depression of the shoulders, and extension of the elbows (1). As a first approach to the modeling effort, these subtleties were not considered, and a simplified representation was adopted. A 3-link pendular model of the body-swing phase was chosen because, in the sagittal plane, there are 3 main points of rotation connected by more or less rigid links. The points of rotation are projections of 3 parallel axes which connect the approximated centers of rotation of the 2 hips, the 2 shoulders, and the 2 crutch tips. The 3 links were assumed to be rigid for the purpose of the model. The 3 pendular links were defined as: link 1) a rigid body composed of the arms and crutches with a single pivot at the crutch tips and a single pivot at the estimated center of rotation of the glenohumeral joints; link 2) a second rigid body composed of the head, neck, and trunk, sharing a common pivot with link 1 at the shoulders and having a second pivot at the estimated center of rotation of the hips; link 3) a third rigid body composed of the legs, feet, and orthoses, sharing a common pivot with link 2 at the hips. The pendular model is depicted schematically in **Figure 4**.

The differential equations which govern the conservative motion of these 3 links were derived using the methods of Lagrange, and are as follows:

$$\begin{aligned}
 -(M_L * L) * g * \sin(\theta) &= (I_L + M_L * L^2) * \ddot{\theta} \\
 &- (M_L * C * L) * \ddot{\sigma} * \cos(\sigma - \theta) \\
 &+ (M_L * C * L) * \dot{\sigma}^2 * \sin(\sigma - \theta) \\
 &+ (M_L * B * L) * \ddot{\phi} * \cos(\phi - \theta) \\
 &- (M_L * B * L) * \dot{\phi}^2 * \sin(\phi - \theta) \\
 -(M_T * T + M_L * B) * g * \sin(\phi) &= (I_T + M_T * T^2 + M_L * B^2) * \ddot{\phi} \\
 &+ (M_L * B * L) * \ddot{\theta} * \cos(\phi - \theta) \\
 &+ (M_L * B * L) * \dot{\theta}^2 * \sin(\phi - \theta) \\
 &- (M_T * C * T + M_L * C * B) * \ddot{\sigma} * \cos(\sigma - \phi) \\
 &+ (M_T * C * T + M_L * C * B) * \dot{\sigma}^2 * \sin(\sigma - \phi) \\
 -(M_C * A + M_T * C + M_L * C) * g * \sin(\sigma) &= (I_C + M_C * A^2 + M_T * C^2 + M_L * C^2) * \ddot{\sigma} \\
 &- (M_T * C * T + M_L * C * B) * \ddot{\theta} * \cos(\sigma - \phi) \\
 &- (M_T * C * T + M_L * C * B) * \dot{\theta}^2 * \sin(\sigma - \phi) \\
 &- (M_L * C * L) * \ddot{\theta} * \cos(\sigma - \theta) \\
 &- (M_L * C * L) * \dot{\theta}^2 * \sin(\sigma - \theta)
 \end{aligned}$$

Symbol meanings are given in the **Appendix** and segment angles are measured counterclockwise from the vertical. The equations are similar in form to those developed by Mochon and McMahon (13) for a pendular model of biped walking.

The solution of these equations was carried out with a digital computer utilizing Gear's method for solving ordinary differential equations. For our numerical solution, a step size of 1/3000 second was used. Initial angular velocities and accelerations were determined by differentiation of angular position versus time curves after fitting our kinematic data with least squares cubic splines in the vicinity of toes-off. Solution began at toes-off and proceeded until heels-strike was expected (approximately 0.8 seconds).

This first phase of modeling assumed a frictionless system operating entirely under the influence of gravity with no torques applied at the joint pivots, and all links assumed rigid. The solution of this freely swinging model is shown in **Figure 5**, along with the corresponding measured data. The model response in the form of stick figures is shown in **Figure 9a** for the first half of the body-swing phase. The model response does not correlate well with the measured data; trunk motion deviates markedly in phase, crutch motion deviates in magnitude, and leg

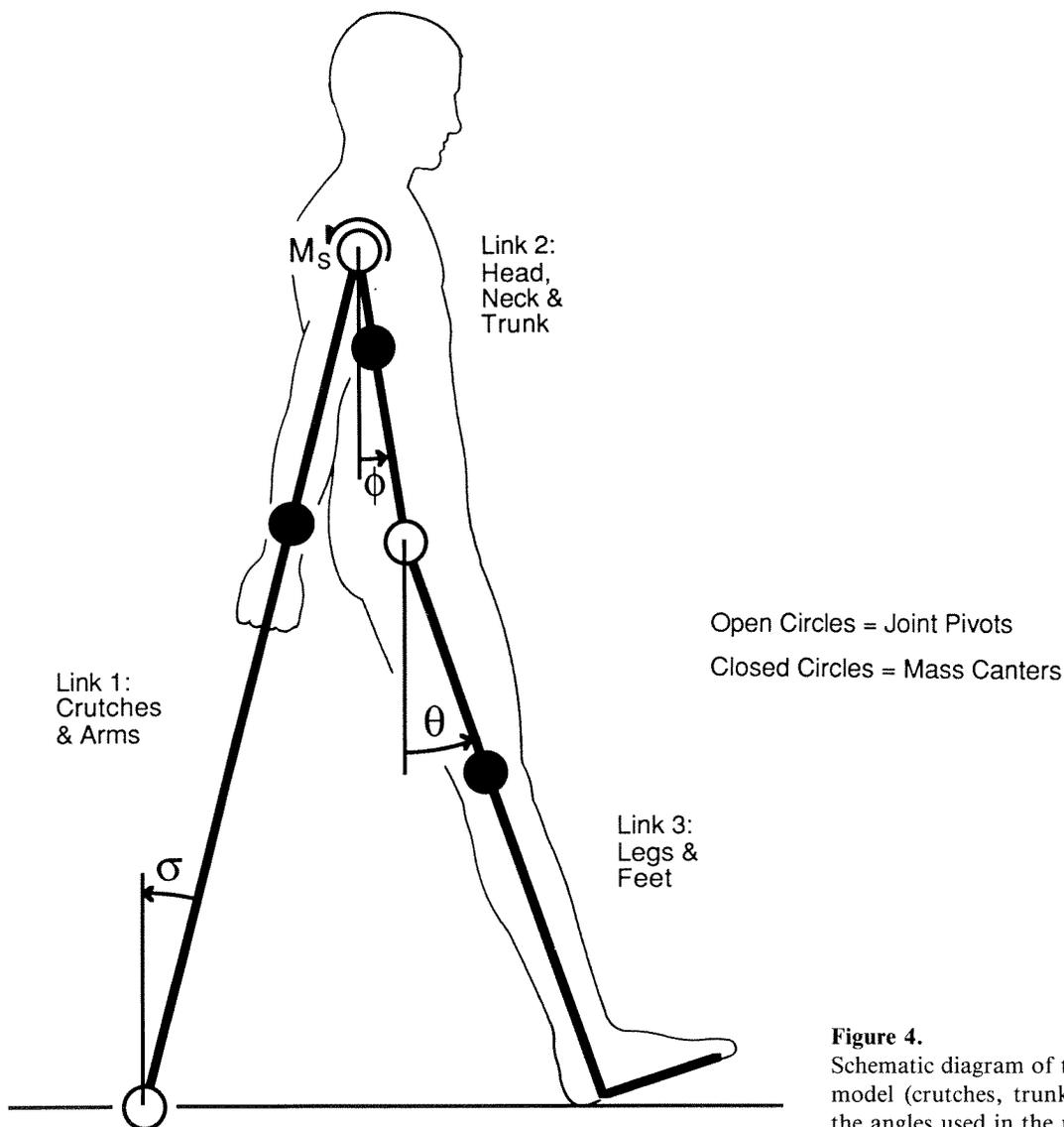


Figure 4.
Schematic diagram of the 3 pendular links used in the model (crutches, trunk, and legs), and definition of the angles used in the model.

motion deviates in both magnitude and phase. The poor correspondence between the measured and modeled data indicates that the fully conservative assumption is not a good one. Nevertheless, there are some interesting implications that come out from the presentation of this model.

For our subject, the legs were observed to swing to a particular angular orientation in the first 60 percent of the body-swing phase, and then maintain a nearly constant orientation with respect to the vertical until the heels strike the ground. Initially, the solution of the conservative model corresponds closely to the observed angular movement of the legs, but rather than being restricted to the initial 60

percent of the body-swing phase, the modeled leg motion continues to progress throughout the entire phase. This results in a large angular deviation between the measured data and the modeled response by the time heels-strike should actually occur. The trunk was observed to swing slightly backward with respect to the vertical in the first 50 percent of the body-swing phase and then swing through in the final half of this phase. In contrast, the response of the conservative model is for the trunk to immediately begin swinging forward at the shoulders, and to continue with this progression throughout the simulation. For the arms-crutches segment, the observed angular velocity was approxi-

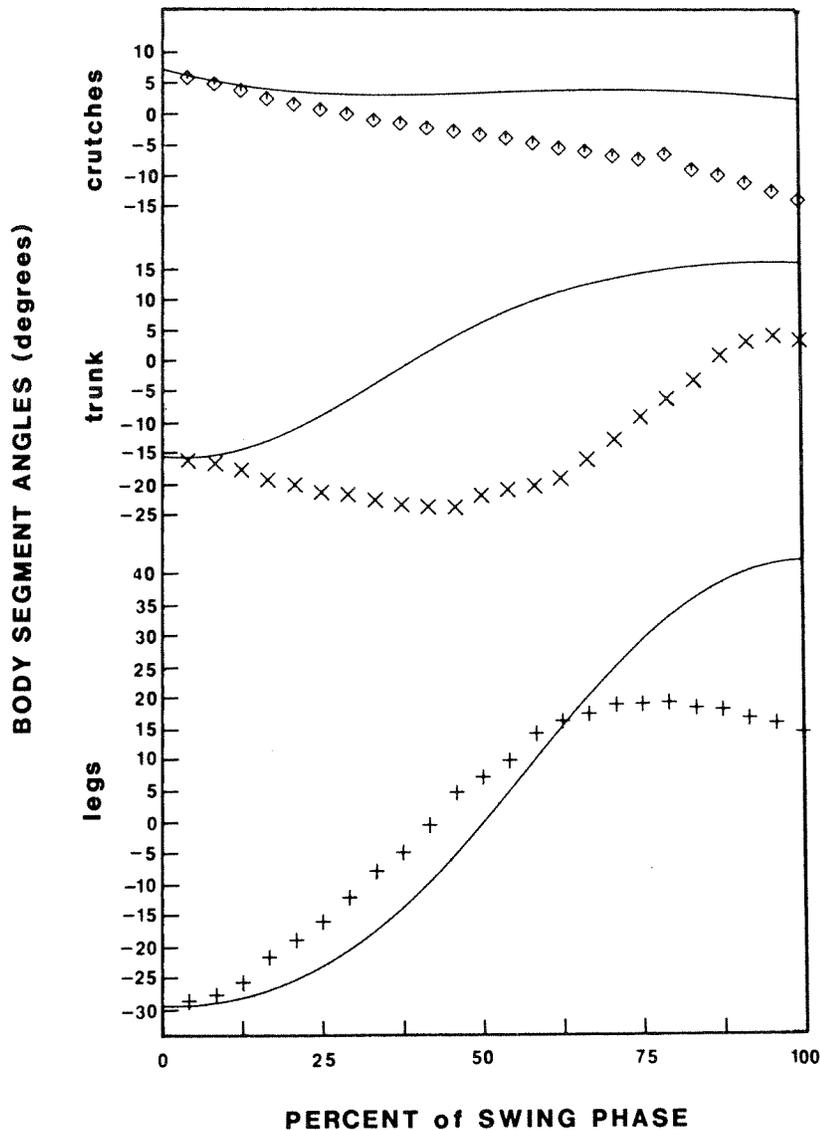


Figure 5.

Solution of the pendular model when all pivots are frictionless and unrestricted. Continuous curves are the modeled solutions and points are measured data. Plots begin at toes-off and end at heels-strike.

mately constant from the time of toes-off until heels-strike. For the conservative model, however, the arms-crutches show only about one-fourth of the angular displacement as that measured from the subject, and additionally includes a slight oscillation. In spite of these differences between the measured data and the simulated response of the model, it is important to note that the fully conservative system did progress from toes-off to an acceptable heels-strike position (heels forward of the crutch tips and just above the ground). This acceptable heels-strike position occurs prematurely, relative to the normal body-swing time. Hence, the normal distance traveled in the body-swing phase

would be reduced by approximately 25 percent, and the amount of time spent in the body-swing phase would be reduced by 33 percent.

In the next phase of the modeling, the assumption of a 3-link pendular system was preserved, but the restriction to a fully conservative system was relaxed. No torques can exist at the pivot of the crutch tips, and our subject was unable to produce muscular torques at the hips. It therefore seemed reasonable to assume that the deviation of this conservative model from the measured data might be due primarily to muscular torques acting at the shoulder pivot. The pendular model was modified by reworking the equations using the shoulder angle, δ ,

which was defined as the angle between the trunk and the crutches ($\delta = \phi - \sigma$). This modification of Model 1 gives the following equations of motion:

$$\begin{aligned}
 -(M_L * L) * g * \sin(\theta) &= (I_L + M_L * L^2) * \ddot{\theta} \\
 &+ (M_L * B * L) * (\ddot{\delta} + \ddot{\sigma}) * \cos(\delta + \sigma - \theta) \\
 &- (M_L * B * L) * (\dot{\delta} + \dot{\sigma})^2 * \sin(\delta + \sigma - \theta) \\
 (M_C * A + M_T * C + M_L * C) * g * \sin(\sigma) - \\
 (M_T * T + M_L * B) * g * \sin(\delta + \sigma) &= (I_C + I_T + M_C * A^2 - M_T * C^2 + M_T * T^2 - M_L * C^2 + M_L * B^2) * \ddot{\sigma} \\
 &+ (I_T + M_T * T^2 + M_L * B^2) * \ddot{\delta} \\
 &- (M_T * C * T + M_L * C * B + M_L * C * L) * (\ddot{\delta} + 2\ddot{\sigma}) * \cos(\delta) \\
 &+ (M_T * C * T + M_L * C * B + M_L * C * L) * (\dot{\delta}^2 + 2(\dot{\delta} * \dot{\sigma})) * \sin(\delta) \\
 &+ (M_L * B * L) * \dot{\theta} * \cos(\delta + \sigma - \theta) \\
 &+ (M_L * B * L) * \dot{\theta}^2 * \sin(\delta + \sigma - \theta)
 \end{aligned}$$

The solution of these equations not only requires initial conditions, but a function $\delta(t)$. This function $\delta(t)$ was determined by a least squares cubic spline curve fit of the measured data, which was continuous through the second derivative. The smoothing allowed input data to be provided at the 1/3000 second step interval used in the solution of the model, and also allowed reasonable differentiation of the data. Continuity on the second derivative is required, since the equations of motion include both the first and second derivative of the shoulder angle. Using this scheme, the torque at the shoulders need not be computed directly. Rather, the effects of these torques are provided for by the time-varying function of δ and its derivatives. In **Figure 6** the solution to this forced pendular model is plotted along with the measured data. The patterns of motion for this second model correspond to the patterns of the measured motion. The magnitudes of angular motion deviate from the measured values by an increasing amount as the cycle progresses, with the legs angles having the best representation of actual motion and the crutches angles having the poorest.

In an effort to gain insight into the effects of various measurement inaccuracies on the solution of the models, the link parameters and initial conditions were varied one at a time during repetitive solutions of Model 2. This provided a form of sensitivity analysis. The results of this analysis

relative to the unperturbed, forced pendular model, are listed in **Table 2**. The greatest effect on the solution of the model occurred when the initial angular orientation of the links were varied. When the initial angular positions were varied by 5.5 degrees, the following maximum change in the

Table 2.

Effects of variations in the body segment parameters and initial conditions on the solution of the pendular model.

MAGNITUDE OF PERTURBATION*	MAXIMUM ANGULAR VARIATION (%)		
	Legs	Trunk	Crutches
$\Delta\theta_0 = 5\frac{1}{2}^\circ$	10.50	0.75	0.50
$\Delta\sigma_0, \Delta\phi_0 = 5\frac{1}{2}^\circ$	4.50	26.50	24.50
$\Delta\theta_0 = 10\%$	0.50	0.25	0.25
$\Delta\sigma_0, \Delta\phi_0 = 10\%$	0.50	7.00	7.50
$\Delta A = 10\%$	2.75	1.50	1.50
$\Delta L = 10\%$	3.50	2.75	3.00
$\Delta T = 10\%$	2.75	1.25	1.25
$\Delta C = 10\%$	2.75	0.50	0.75
$\Delta B = 10\%$	5.50	3.50	3.50
$\Delta M_C, \Delta I_C = 10\%$	2.75	1.50	1.50
$\Delta M_T, \Delta I_T = 10\%$	2.75	2.00	2.25
$\Delta M_L, \Delta I_L = 10\%$	2.75	1.50	1.50

*Percentages are based on the initial values at toes-off.

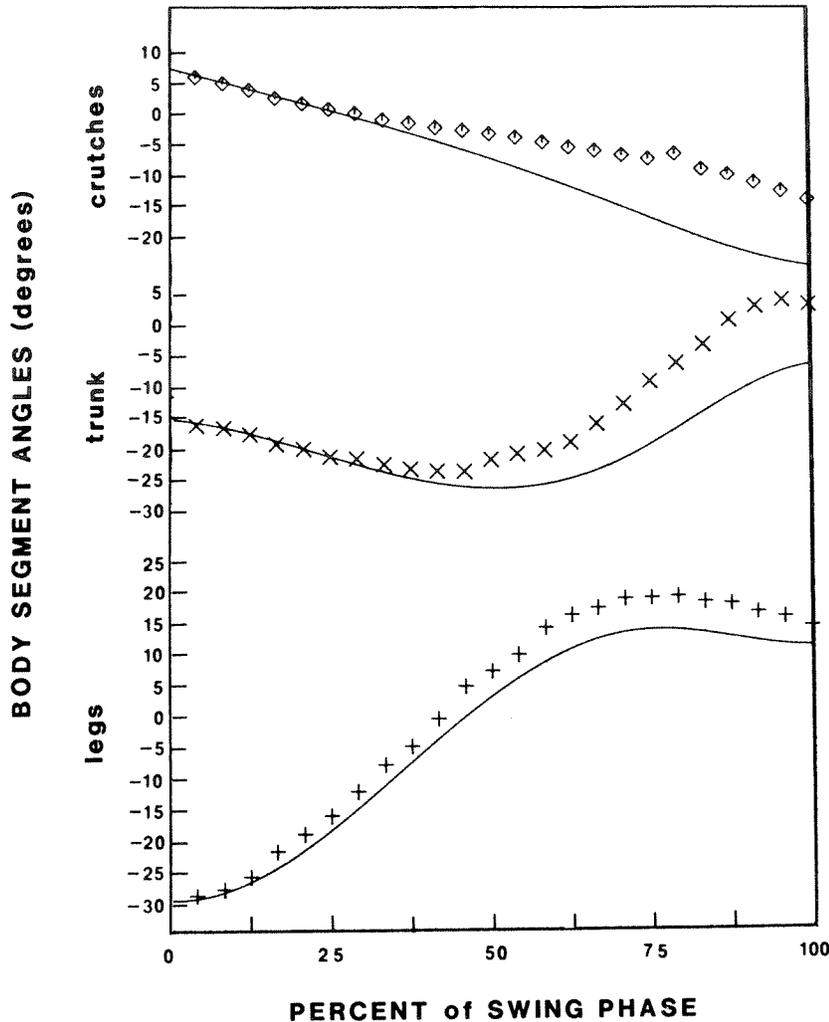


Figure 6.

Solution of the pendular model when the shoulder angle is forced to conform to the measured pattern of movement. Continuous curves are the modeled response and discrete points indicate the measured data. Plots begin at toes-off and end at heels-strike.

output curves resulted: 10.5 percent when initial leg orientation was shifted, and up to 26.5 percent when initial crutch and trunk orientation was shifted. Varying each of the following link parameters by 10 percent resulted in the corresponding maximum change in the output curves: initial angular velocity, 7.5 percent; position of the mass center on the link, 3.5 percent; length of the link, 5.5 percent; and mass/moment of inertia of the link, 2.75 percent. In general, the output response was more sensitive to the initial angular positions of the links than to the other link parameters.

Inaccuracies in the measured kinematic data have contributions from various sources: transcription and measurement error in recording the position of the individual joint markers, placement error in the positioning of joint markers on the subject, and error introduced when the actual physical system is reduced to a simplified system for modeling. Tran-

scription and measurement was estimated to be accurate to within 2 degrees. The other 2 sources of inaccuracy, marker placement and model representation, do not change the resolution of the measurements, but instead, result in baseline shifts of the 3 measured angular quantities (i.e., an angular offset of all the measured angle values for a given link).

Segment angles as measured in this study were angles, with respect to the vertical, of a line connecting the proximal and distal joint markers for a given segment (i.e., crutch tip and shoulder marker, shoulder and hip marker, hip and ankle marker). If a joint marker were misplaced, the angle of the wrong line would be measured. The angular change between successive points in the data set would be proper, but the measured angles would be offset by a constant amount equal to the angle between the line that was measured, and the line that should have been measured. Malplacement of a

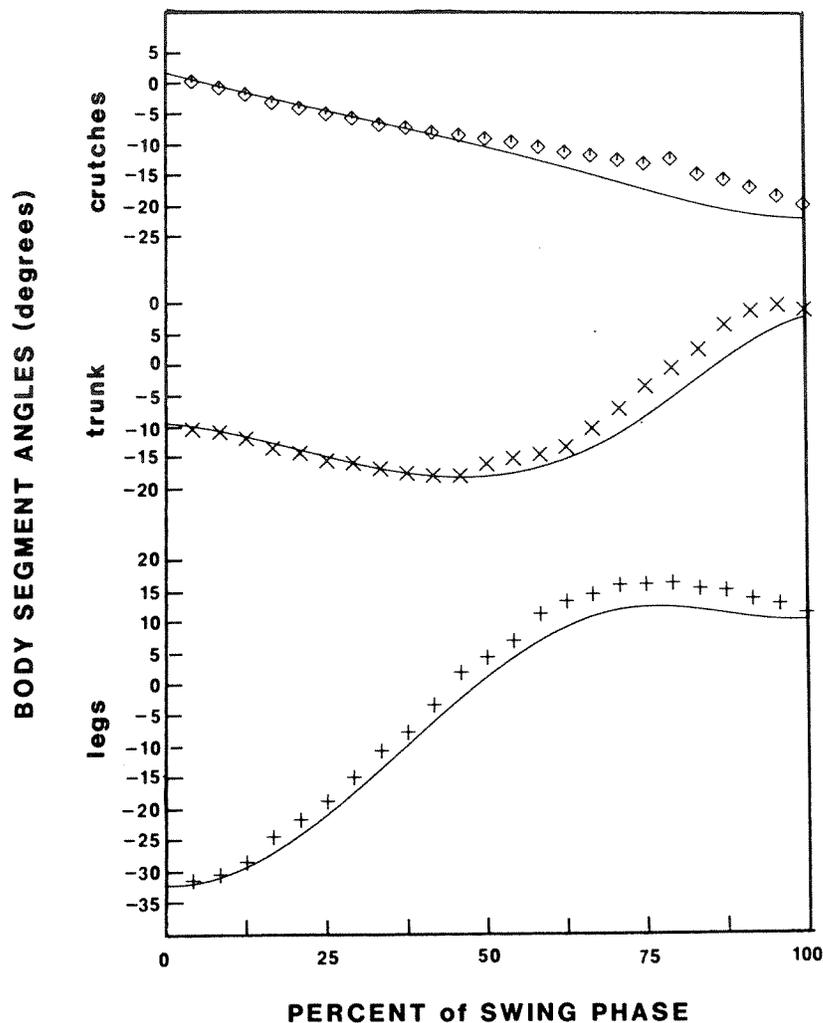


Figure 7.

Solution to the pendular model with controlled shoulder angle after small adjustments to the initial positions of the 3 links of the model, and with corresponding baseline shifts of the measured data. Curves represent the modeled response and the points correspond to the measured data. Plots begin at toes-off and end at heels-strike.

joint marker by a few centimeters can result in baseline angle shifts of 2 to 5 degrees. For a long link, like the crutch, up to 2 degrees offset can result: for a short link, like the trunk, up to 5 degrees offset can result. The polycentric nature of the joints (particularly the shoulder joint) could exacerbate this problem, since accurate marker placement is difficult. In the modeling of the physical system, each segment mass center was assumed to lay on a line connecting the proximal and distal joint pivots for that link. Deviations from this ideal could result in additional baseline shifts. This is noteworthy, in view of the fact that errors in initial angular positions (i.e., baseline values) have the greatest effect on the solution of the model.

With this in mind, small baseline shifts were made to the measured data and Model 2 was again solved

with the goal of bringing the modeled response into closer correlation with the measured data. Outputs using baseline shifts of -3 degrees for the legs-angles; 6 degrees for the trunk-angles; and -6 degrees for the crutches-angles are plotted in **Figure 7**. These small angle shifts bring the measured and modeled data into close agreement. The curve shapes are quite similar and the magnitude differences are within 5 degrees of measured values. Stick diagrams representing the actual measured motion and the solution of this model are shown in **Figures 8a** and **8b**. While some variation between actual and modeled motion is evident, the similarities are pronounced.

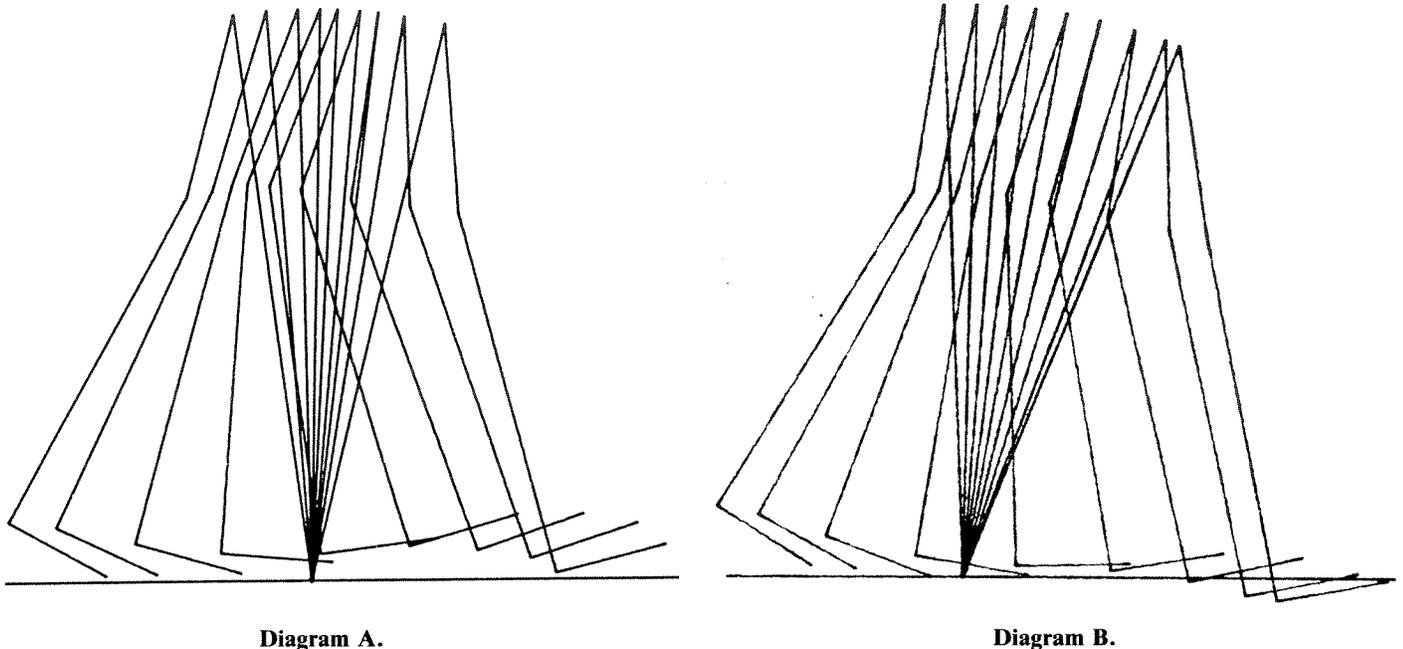


Figure 8.

Diagram A. The measured data is presented as stick figures at successive points in the body-swing phase. Progression is from left to right, beginning at toes-off and ending at heels-strike. The time between images is 1/10 second. **Diagram B** is a similar representation of the output response, but from the final mathematical model. This sequence also begins at toes-off and runs for the same period of time as in Diagram A.

DISCUSSION

We have performed a kinematic analysis of paraplegic swing-through crutch gait and used this analysis in the development and evaluation of a mathematical model of the body-swing phase. Comparison between the modeled and observed kinematics gives insight into the mechanisms of control used in the body-swing phase, and provides quantitative insight into the mechanical energy demand of the body-swing phase. Additionally, an examination of the overall gait style gives clues to possible reasons why swing-through crutch ambulation is such an exhausting activity.

Our first model was a simple, 3-link pendular model with freely swinging joints at the crutch tips, shoulders, and hips. When compared to the observed motion patterns, this model exhibited minimal forward rotation of the crutches, a reduced stride length and, if allowed to swing for the normal body-swing time, a final body orientation that was inappropriate for heels-strike. However, after 67 percent of the normal body-swing time, the body is in a reasonable orientation for heels-strike to occur. This indicates that given the initial conditions at

toes-off, there is sufficient energy in the system to carry it to a heels-strike position without the input of any mechanical work. While the kinetic energy at toes-off is sufficient to get to heels-strike, a conflict exists between the modeled response and the physical environment, since the toes pass below the surface of the walkway, indicating that the body must be elevated slightly during the body-swing phase. Also, when the body is in a position to accept heels-strike, the heels are slightly above the ground, indicating that the body must be lowered to the ground for heels-strike. These 2 incompatibilities with the physical environment imply that if this pendular model is representative of the physical system, some mechanical work must be applied during the body-swing phase.

In the second model, the 3-link pendular system had a forced input at the shoulder. The forcing function used in this model represented muscular control of the shoulder angle and was applied in such a way that the modeled shoulder angle corresponded to the observed shoulder angle at each time step in the body-swing phase. The motion patterns of this model more closely matched the observed kinematics than when the shoulder was allowed to

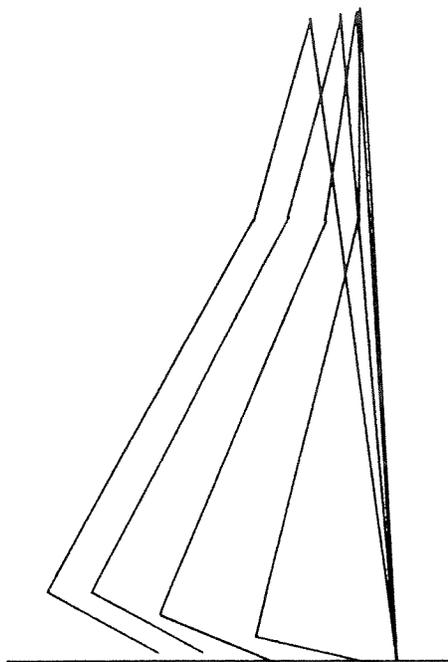


Diagram A.

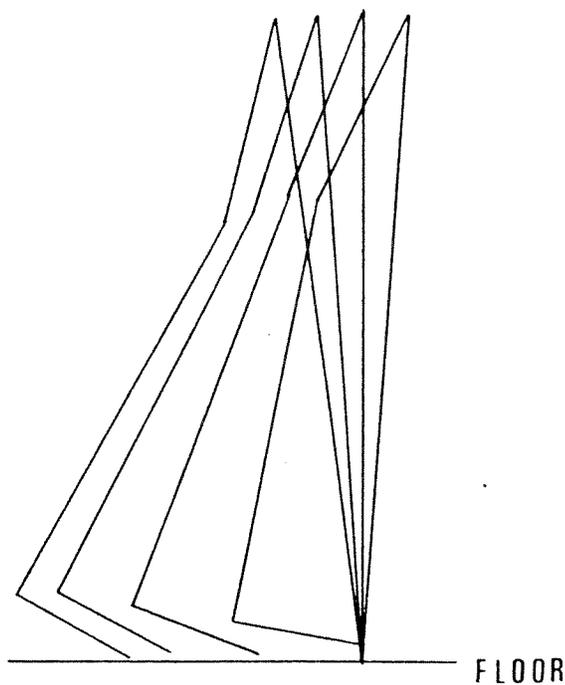


Diagram B.

Figure 9. The first half of the body-swing phase, beginning at toes-off, is represented using stick figures. **Diagram A** is the solution of the pendular model with all pivots frictionless and unrestrained. **Diagram B** is another solution, using the same initial conditions, but with the shoulder angle kept fixed.

swing freely. There was a smooth progression from toes-off to heels-strike and the body maintained a normal orientation throughout that phase. In the first half of the body-swing phase, shoulder angle control was essentially a restriction of shoulder rotation, which caused the trunk to pivot about the crutch tips rather than about the shoulder. In such a configuration, the level of the hips are elevated as the body pivots over the crutch tips and reach a maximum height when the hips are aligned vertically above the crutch tips. In addition, the restriction of shoulder rotation causes swing-through of the legs alone. The resulting hip flexion after toes-off results in an effective shortening of the body length. The maintenance of a fixed shoulder angle, therefore, provides a mechanism for increasing ground clearance after toes-off, as illustrated in **Figure 9**. When, as in Model 1, the shoulder was allowed to pivot freely, the body swings at both the hips and shoulders as soon as the toes lift off. As the trunk pivots about the shoulder, the height of the hips are lowered, while at the same time, the extending hips cause a lengthening of the body and a reduction in ground clearance. The imposed control of shoulder angle in the system simulation was sufficient to eliminate the ground interference problem and also acts to increase the stride length.

The physiological effort required to control the shoulder angle during the body-swing phase is unlikely to account for the significant metabolic energy expenditure during this same period. A consequence of operating a mechanical system by physiological means is that metabolic "work" must be provided during activities where no mechanical work is required; the muscles are actively used to control the position and compliance of the joints even when there is no joint motion. The legs are normally well-suited for weight bearing, having stable joints and large muscles with large lever arms. By contrast, the arms and shoulders are poorly-suited for supporting the body. The crutches are weight bearing for about 48 percent of the gait cycle in swing-through crutch ambulation. Supporting the body with the arms requires that stabilizing forces be provided to the wrists, elbows and shoulders. These joints have multiple, independent degrees-of-freedom and are served by relatively small muscles which are heavily taxed when weight bearing. One strategy for reducing the effort required when ambulating in the swing-through style would be to

support the body weight with a mechanical structure, rather than through the arms and shoulders. The results of the pendular model simulation suggest that any such device should not interfere with the individual's ability to control the angle of the shoulder while in the body-swing phase. Of the commonly prescribed crutch types, the axillary crutch, when used properly (i.e., no weight bearing through the underarm pad) provides no stability to the arm. The forearm crutch may provide some stability to the wrist; and the Canadian crutch, which bears on the brachial muscles, provides some stability to both the elbow and the wrist. Not surprisingly, the metabolic energy consumption is reported to be lowest for the Canadian crutch, highest for the axillary crutch, and in-between for the elbow crutch (15). The saddle crutch, designed by Joll (9), eliminates the arms and shoulders from weight bearing entirely. With this device, the individual sits on a saddle which is suspended by straps from the tops of the crutches, like a playground swing. The weight is borne through the seat, and the arms are merely used for controlling the motion.

Another area of energy expenditure apparent from observation, happens during the transition between the stance and swing phases. The period of weight transfer occurs prior to toes-off and hence was not included in the modeling effort. In this transition phase, weight bearing is transferred from the legs to the arms, and hence there are some of the same weight bearing issues that are present in the body-swing phase. More importantly, the body-swing phase must be initiated by an active lifting of the toes off the ground. For many crutch ambulators, the toes can be lifted by bending of the knees. However, for the paraplegic ambulator, a different strategy is used: the entire body is lifted by elbow extension and shoulder depression to initiate toes-off. This lifting is continued after toes-off, presumably to gain additional ground clearance for the toes. In essence, the paraplegic crutch ambulator must perform 1 push-up with each stride. This push-up activity has a second drawback in that a more stabilizing crutch, like the Canadian crutch, cannot normally be used by the paraplegic, since the required elbow extension would lift the arm out of the cuff and negate the benefits of that crutch design. Body elevation, using shoulder depression, is still possible with the Canadian crutch, but the amount of elevation is reduced from when the

elbows can be extended as well. A device like the saddle crutch would completely eliminate the possibilities for pushing-up the body, since either elbow extension or shoulder depression would result in lifting the individual off the seat of the crutch.

Some criteria for design are suggested, based on this study: 1) Any crutch design should not interfere with the ability of the individual to control the angle of the shoulder in the swing phase. 2) The individual must be able to initiate the body-swing phase by lifting the toes off the ground somehow. 3) Methods of relieving the joints of the arms and shoulders from load bearing should reduce the metabolic energy expenditure during the body-swing phase. 4) There must be sufficient clearance between the feet and the ground to prevent interference during the body-swing phase.

The Canadian and saddle crutches are interesting designs which satisfy the design goal of arm stabilization, but at the same time, interfere with another one of the design goals that cannot be violated, i.e., ground clearance. If a mechanism for toes lift-off and ground clearance could be provided for by a mechanism in the orthoses (perhaps assisted by functional neuromuscular stimulation [FNS] of the ankle or knee), then these stabilizing crutch designs might have application for the paraplegic. The final design solution, it would seem, may require a hybrid system consisting of a specialized orthosis and a stabilizing crutch; or FNS and a stabilizing crutch, which when combined, can reduce the difficulty of crutch-aided gait, while providing for functional ambulation by paraplegic persons.

CONCLUSIONS

1. The body-swing phase of swing-through paraplegic crutch ambulation has been successfully modeled as a pendular system with muscular control of the shoulder angle. During the first half of the body-swing phase, the muscular control is primarily restriction of shoulder joint rotation. This contributes to increased ground clearance, as well as controlling body orientation and the timing of the body-swing phase.

2. Areas of obvious energy demand are the stabilization of the joints of the arms and shoulders during the body-swing phase, and the active lifting of the toes to initiate the body-swing phase. Those

interested in designing devices which may reduce the energy demand of swing-through crutch ambulation for paraplegics may want to consider hybrid mechanisms which address these two issues.

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APPENDIX

In this section, *Legs* refers to the pendular link composed of both legs with KAFOs; *Trunk* refers to the pendular link representing the head, neck and trunk; and *Crutches* refers to the pendular link of both arms with crutches.

SYMBOL	MEANING
M_L, M_T, M_C	Mass of the Legs, Trunk and Crutches.
I_L, I_T, I_C	Moment of inertia of the Legs, Trunk and Crutches about their respective mass centers.
θ, ϕ, σ	Angle of the Legs, Trunk and Crutches with respect to vertical.
$\dot{\theta}, \dot{\phi}, \dot{\sigma}$	Angular velocity of the Legs, Trunk and Crutches.
$\ddot{\theta}, \ddot{\phi}, \ddot{\sigma}$	Angular acceleration of the Legs, Trunk and Crutches.
$\delta, \dot{\delta}, \ddot{\delta}$	Shoulder angle ($\delta = \phi - \sigma$), angular velocity and acceleration.
L	Distance from the hip to the Legs mass center.
T	Distance from the shoulder to the Trunk mass center.
B	Length of Trunk (link 2).
C	Length of Crutches (link 3).
A	Distance from crutch tip to Crutches mass center.
g	Gravitational acceleration.

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