A Multisegment Computer Simulation of Normal Human Gait

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***Abstract*—The goal of this project was to develop a computer simulation of normal human walking that would use as driving moments resultant joint moments from a gait analysis. The system description, initial conditions and driving moments were taken from an inverse dynamics analysis of a normal walking trial. A nine-segment three-dimensional (3-D) model, including a two-part foot, was used. Torsional, linear springs and dampers were used at the hip joints to keep the trunk vertical and at the knee and ankle joints to prevent nonphysiological motion. Dampers at other joints were required to ensure a smooth and realistic motion. The simulated human successfully completed one step (550 ms), including both single and double support phases. The model proved to be sensitive to changes in the spring stiffness values of the trunk controllers. Similar sensitivity was found with the springs used to prevent hyperextension of the knee at heel contact and of the metatarsal-phalangeal joint at push-off. In general, there was much less sensitivity to the damping coefficients. This simulation improves on previous efforts because it incorporates some features necessary in simulations designed to answer clinical science questions. Other control algorithms are required, however, to ensure that the model can be realistically adapted to different subjects.**

***Index Terms*—Computer simulation, gait.**

I. Introduction

In the field of lower limb prosthesis design a full-body, kinetically-driven computer simulation of human gait has the potential to be an invaluable tool. With an ideal simulation, design questions could be answered using the computer model, reducing dependency on costly prototypes. Before such a practical simulation can be realized, however, it is necessary to develop a basic simulation of healthy human gait. The intent of this project was to develop a computer simulation of a walking subject incorporating features that will ultimately be necessary for use in an applied setting (e.g., as a design tool). The most obvious criterion for success with any simulation is that it should predict a reasonable kinematic pattern; for gait this means including both single and double support phases. A feature that will be important for the eventual application of a simulation is that *a priori* knowledge of a specific movement trial should not be required. Ground reaction forces should not be used to drive the simulation; nor should segments be constrained to follow predetermined trajectories. Furthermore, a reasonable representation of the physiological system must be used; this influences both the number of segments chosen for the model and the nature of the joints between them. The muscle forces or joint moments used to drive the model must approximate the physiological counterpart.

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Simulations have been created previously to achieve diverse goals, both clinical (e.g., [1] and [2]) and otherwise (e.g., [3]). Of the simulations reported to date, however, none have achieved the ideal described above. Some investigators have focused on only one part of the gait cycle [4], [1], [5]–[7]. Dealing with only one phase of the gait cycle avoids the numerical difficulties which can arise as a foot makes the transition from no loading (swing phase) to full loading (stance phase). Many investigators have not included a foot segment [8], [9], [10]; still others have fastened the stance leg foot to the ground [11], [3], [12]–[14], [2], [15]. This strategy also avoids a free transition from swing to stance. For a simulation to be truly useful in practical applications, it must be able to move smoothly through this transition.

Much work has been done on two-dimensional simulations [11], [9], [12], [13], [7]. Although in walking the movement of each individual limb is primarily planar, the limbs are separated from each other and as a result, the pelvis undergoes some degree of movement in both the frontal and horizontal planes. If this movement is ignored, it is unreasonable to expect good kinematic results for both limbs simultaneously. The pelvic movement is also critical in minimizing the amount of motion of the upper body during the walking cycle.

Some investigators have considered only the lower limbs, modeling the rest of the body as a point-mass [16], [3], [12]–[14]. Regulation of the torso's orientation is an important feature in maintaining balance during locomotion and the control of the large inertia of the torso should not be ignored [15], [32].

Other investigators have used predetermined trajectories of one or more segments as feedback, as a constraint or as part of an objective function for an optimization algorithm [17], [8], [18], [19], [9], [10], [2]. While this is a very appealing way of ensuring reasonable kinematics, it does require *a priori* knowledge of a specific walking trial. If the trajectory is included as a hard constraint or as feedback, it limits the use of the simulation in investigating the response to a novel situation (i.e., in response to an external perturbation). It does not allow the simulated body to respond in the same manner as the human body would. This is less of a concern when the trajectory is incorporated as part of an optimization scheme. Whether or not the physiological system operates in such a fashion, however, is still open to debate.

The work that perhaps comes closest to the ideal gait simulation is that by Meglan [20]. He modeled the body as

13 linked segments with no constraints on the trajectories of any segment. Passive joint constraints limiting the range-of-motion were applied at each joint. In addition, equations governing the passive motion of each joint (i.e., the stiffness at the joint) were used. The driving forces were the resultant joint moments at each joint taken from an inverse dynamics analysis. The foot was modeled with nonlinear spring/damper contact elements arranged over the sole of the foot. Although there were many commendable features about Meglan's work, as a gait simulation it was not fully successful because no simulation run was able to complete even a partial gait cycle.

The purpose of the current project was to develop a computer simulation of gait that would be as close to ideal as possible. The goal was to minimize the compromises made in representing the physiological system while still ensuring an accurate prediction of the kinematics. This is the first step in developing a simulation that can ultimately be used to solve questions concerning both healthy and pathological gait.

II. Methods

A. Data Collection and General Model

The data collection was undertaken as part of another study independent of the current simulation project [21]. Details of the data collection and reduction have been previously reported (cf., [21] and [22]) and so are only briefly described here. Three orthogonally placed CCD cameras and two force plates were used to perform three-dimensional (3-D) gait analyses of healthy, young subjects. The body was modeled as a linked chain of nine segments: feet, shanks, thighs, pelvis, trunk (L3/L4 to C1/T7 spinal level) and head. Each video record was used to identify two-dimensional coordinates for specific points on the body with less than 1 mm rms error. The 3-D coordinates were then reconstructed using multiple sets of the two-dimensional (2-D) data. These data were smoothed (fourth-order Butterworth filter) and further kinematic data (angles, velocities, accelerations) were calculated. Inverse dynamic techniques were used to calculate resultant moments at each joint, combining the measured external forces with the kinematic data. Thus, 3-D moments were calculated at the ankles, knees, hips, and the joints at L3/L4, and the neck (C7/T1). The simulation was developed using the data from one trial of one subject.

The foot was modeled as two segments with a revolute joint between them: the forefoot, including all the toes, and the main part of the foot, stretching back from the metatarsal-phalangeal (mp) joints. The torso model was also modified for the simulations. The spherical joints between the pelvis and the trunk and between the trunk and the head were locked, i.e., constraints were added to prevent all motion at either joint. In this way, a combined head-arms-torso segment was created and six degrees of freedom (DOF) were removed from the model. Therefore, the model configuration used for the simulation included nine segments organized in a manner that differed from the configuration used in the inverse dynamics analysis.

For the simulation, the hip joints were modeled as spherical joints allowing full rotational DOF. The knee joints were modeled as revolute joints with the fixed axis set parallel to a line joining the femoral condyles. The ankle joints were modeled as universal joints, allowing plantarflexion/dorsiflexion and inversion/eversion. The two axes were set perpendicular to each other, passing through a common point (the ankle joint center). The mp joints on each foot were treated as one revolute joint at the location of the mp joint for the great toe. The joint axis ran transversely, perpendicular to the long axis of the foot and parallel to the ground. The hip, knee and ankle joint centers were located in the same segments, with the same local (segment-based) coordinates as in the inverse dynamics calculations. The final simulation model had 20 DOF.

Individual muscles were not included in the simulation because of the increased complexities and difficulties involved in using muscles as force actuators. Instead, torque actuators were used at all the joints with the exception of the mp joints in the feet. The torques applied were the resultant joint moments from the inverse dynamics analysis of the subject's gait data. An Akima spline was fit to the torque/time data sets to ensure continuous functions for all the torque actuators [23]. The mp joints were modeled as passive joints only: a torsional spring and damper were used to supply torque based on the relative angular position and velocity at the joint.

The simulations were all developed and run on a DEC 5000 machine using ADAMS, a multibody dynamic analysis software package from Mechanical Dynamics Inc. (Ann Arbor, MI).

B. Controls

Many authors have shown that there is a relatively small but significant amount of error present in displacement data gathered by the motion analysis systems commonly used to study gross human movements (cf., [26]–[28]). It has also been shown that even small amounts of measurement error have disastrous effects on the ability of simulations to predict the original displacement pattern [29], [30]. These two factors, taken together, mean that a full stride simulation, dependent only on the initial conditions, system description, and torque actuators at each joint is, for all intents and purposes, impossible. Some mechanism(s) must be included to prevent any given error from significantly degrading the simulation run.

The preferred type of controller for a simulation of a biological system is one which mimics the physiological controlling factors. Unfortunately, although several potential feedback mechanisms have been identified in the human body, the strategy by which the neural system integrates the feedback into a coordinated pattern of muscle contractions in a complex movement remains elusive. Lacking the precise knowledge of the physiological control strategy, one can, however, identify features of gait which the body must somehow incorporate in any movement plan for walking. Two fundamental features are 1) the upper body must remain close to the vertical [32], [33], and 2) the movement at each joint must remain within its physiological range of motion (ROM).

Given that in the current project, the torque actuators are at least one step removed from the actual muscles (and hence

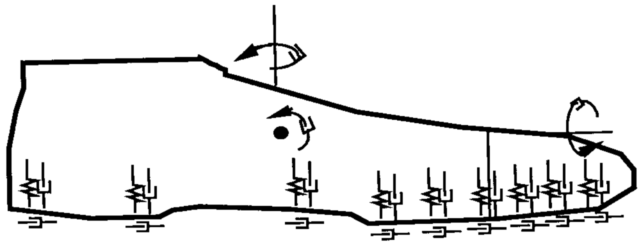


Fig. 1. Diagrammatic representation of the foot model used in the simulation. Note that the horizontal dampers in the frontal plane are not visible in this view.

from muscle-related feedback mechanisms) and that the true (physiological) control strategy is unknown, it was decided to strive for mathematically simple control strategies that would yield a relatively normal walking pattern. In accordance with the two standard gait features noted above, two types of simple "controls" were identified for use in the simulation: trunk controls and ROM controls. To keep the trunk vertical, torsional spring/damper systems were incorporated about the right/left (R/L) and the anterior/posterior (A/P) axes of the stance leg hip joint. Essentially, if the trunk deviated from the vertical in any direction, a restorative moment proportional to both the angular deviation and the trunk angular velocity was applied at the stance leg hip joint. A similar controller was used to prevent the trunk from rotating too far about the vertical axis. It seems reasonable to place some limits on the rotation because, in general, for normal walking, the torso faces anteriorly. At the ankle and knee joints torsional spring/damper systems were used as ROM controls to provide physical stops at the ends of the range of motion. Such a stop prevented hyperextension at the knee as well as unnatural ankle plantarflexion or dorsiflexion.

It was decided that the trunk controls and the ROM controls would be the basic controls applied; if others were needed, simple linear springs and/or dampers, based on the relative angular positions and velocities, would be applied at the appropriate joints. The parameters for all the controls were established by running the simulation for progressively longer periods of time and identifying the torque adjustments necessary. The only exception was for the ROM limits, where approximate ranges were taken from standard physical therapy texts.

C. Foot Model

A foot model, based on visco-elastic contact elements, was used to calculate the ground reaction forces. This has been previously described in detail [24], [25] and will only be summarized here. The foot model was developed by treating the right foot as an isolated segment. Ankle joint forces and moments were used to drive a simulation which encompassed a full stance phase, from a few time steps prior to heel contact extending to a few frames after toe-off. The philosophy used in developing the foot model was similar to that used in developing the rest of the model: try to find a relatively simple formulation which will produce acceptable kinematics and which has at least potential for use with different data sets.

A total of nine "contact" elements, arranged down the midline of the foot, were used; three of these were on the forefoot segment (Fig. 1). The vertical ground reaction forces resulted from linear spring/damper systems in each of the contact elements. The shear forces were made linear functions of the velocity of the contact points. Initial attempts using a smaller number of elements were unsuccessful; as were attempts in which a wide variety of off-center placements for the elements were tested. Heavy torsional damping was used to control movement about both the vertical and anterior/posterior (A/P) axes to counteract the effects of the midline arrangement.

D. Assessment of the Simulation

Note that at each joint, the torques calculated from the inverse solution were applied throughout the simulation. The corrective torques from the "controllers" were applied over and above the resultant joint torques. To give some basis for assessment as to the magnitude of the corrective torques, the resultant joint moments for the entire 24 walking trial data set collected as part of the study by Eng and Winter [21], were examined. These data were normalized to a standard time pattern where two successive right heel contacts occurred at 0% and 100% of the stride time, and right foot toe-off occurred at 60% of the stride time. The moments were normalized to body mass. The mean and standard deviation of these normalized data were calculated. In this way, the adjusted joint moments used in the simulation could be compared with moments taken from a broad data set.

Once the control parameters that yielded an acceptable walking pattern had been identified, several analyses were run to test the sensitivity of the model to changes in these parameters. The effect of increasing and decreasing each parameter by 5, 10, 20, and 30% of its value was examined.

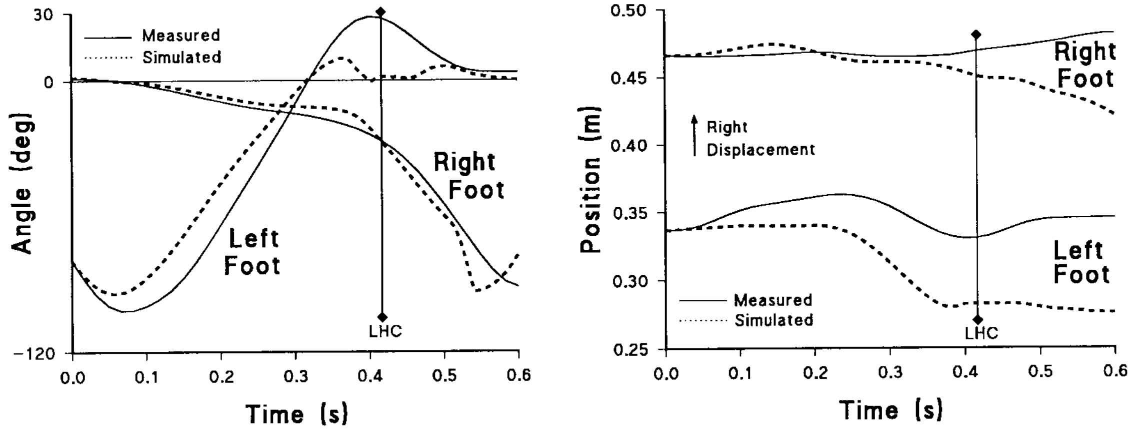
III. Results

A. Simulation Behavior

In general, the kinematics predicted by the simulation were reasonably close to the original measured kinematics for a 550 ms period, which is slightly more than one step (Figs. 2–5). The simulation ran from the start of the foot-flat portion of right foot stance to roughly the same point in left foot stance. After 550 ms, the deviations between the simulated movement and the original movement became too large to be considered acceptable. Clearly, this is a subjective evaluation: it is based on examination of the displacement, velocity and GRF patterns as well as on viewing the simulation graphics.

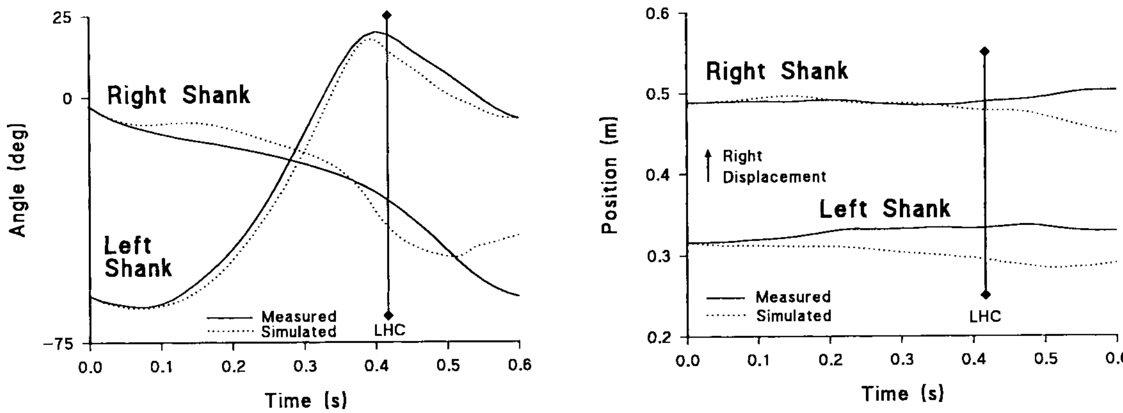
The first group of results presented allows comparison of values predicted in the simulation to the comparable values from the gait analysis (Figs. 2–6, Table I). The angles presented are projection angles: the long axis of the segment was projected onto the applicable plane and the rotation around the axis normal to the plane was calculated. In all the graphs "measured" refers to values from the inverse dynamics analysis, whether or not they were directly measured.

The right foot ground reaction forces were also close to the measured values (Fig. 6). The exception occurred at the time of left heel contact, when there was a sharp change in both the A/P force and the vertical force under the right foot. This



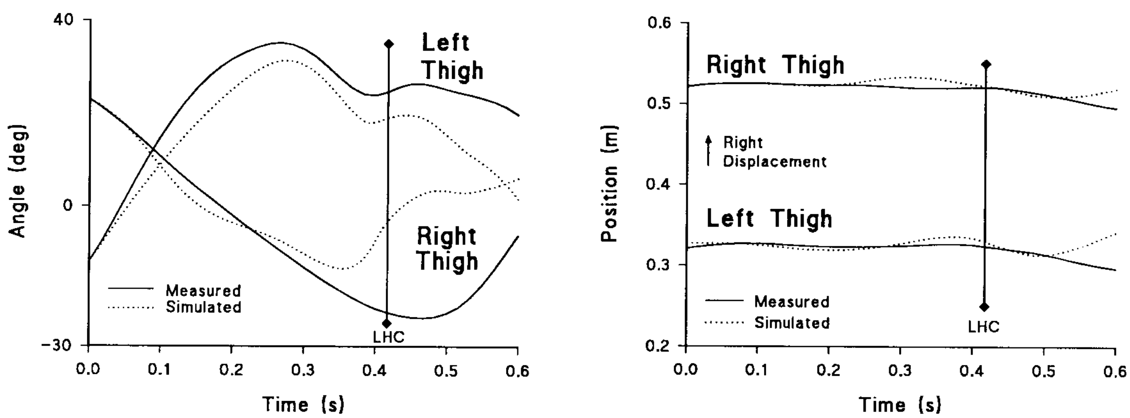
(a) (b)

Fig. 2. Orientation and position of the feet. (a) The sagittal plane angle. (b) The right/left position of the segment center of mass. At time = 0 s the right foot has just begun the foot-flat portion of stance. LHC = left heel contact for the measured data. LHC for the simulation occurred at $t = 0.4$ s.



(a) (b)

Fig. 3. Orientation and position of the shank segments. (a) The sagittal plane angle. (b) The right/left position of the segment center of mass. At time = 0 s the right foot has just begun the foot-flat portion of stance. LHC = left heel contact for the measured data. LHC for the simulation occurred at $t = 0.4$ s.



(a) (b)

Fig. 4. Orientation and position of the thigh segments. (a) The sagittal plane angle. (b) The right/left position of the segment center of mass. At time = 0 s the right foot has just begun the foot-flat portion of stance. LHC = left heel contact for the measured data. LHC for the simulation occurred at $t = 0.4$ s.

heel contact was not as smooth in the simulation as it was in the measured case. The left foot trajectory did not remain as high above the floor as it should have been during swing and as a result, the spring/damper closest to the heel was activated prematurely (i.e., it dropped below the resting length of the spring). The effect of this was to cause the ankle to undergo less dorsiflexion and to plantarflex too early [see Fig. 2(a)], compressing the other springs too rapidly. The "bouncing" pattern seen in the ground reaction forces (Fig. 6) was the result.

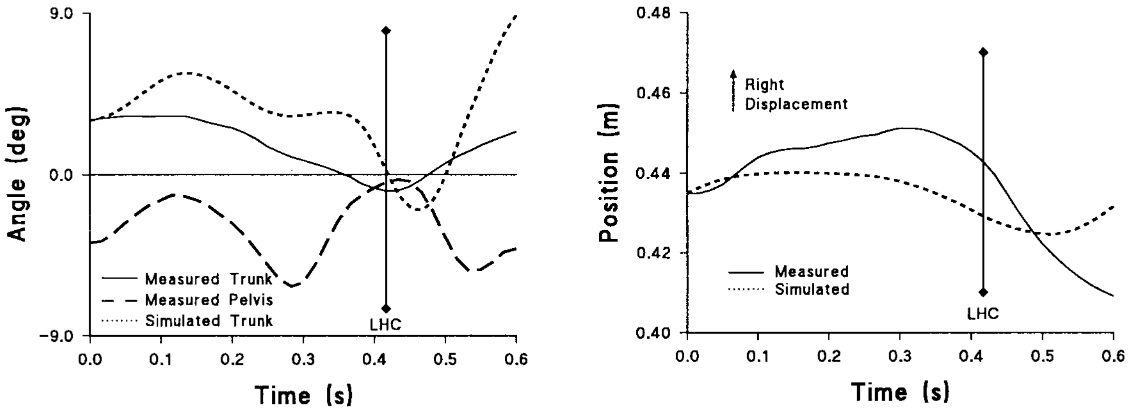


Fig. 5. Orientation and position of the trunk. (a) The sagittal plane angle. (b) The right/left position of the segment center of mass. At time = 0 s the right foot has just begun the foot-flat portion of stance. LHC = left heel contact for the measured data. LHC for the simulation occurred at $t = 0.4$ s.

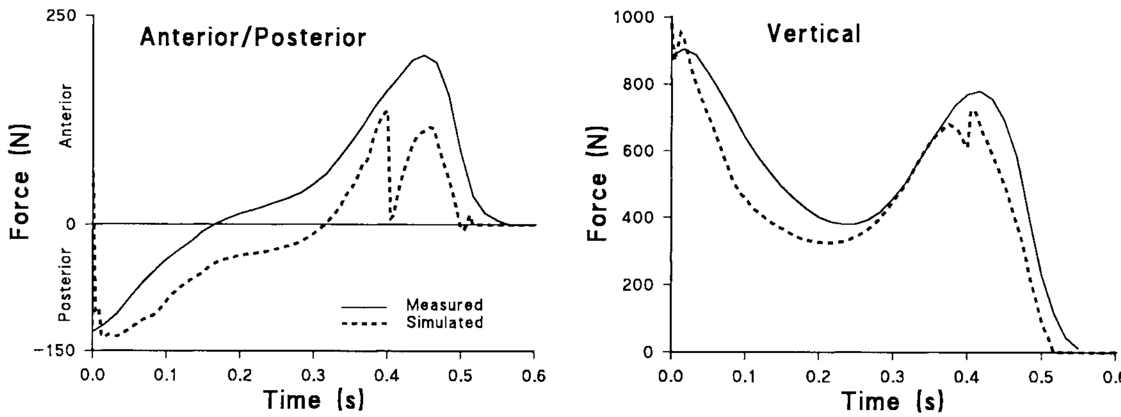


Fig. 6. Ground reaction forces under the right foot. (a) Anterior/posterior. (b) Vertical. At time = 0 s the right foot has just begun the foot-flat portion of stance. LHC = left heel contact for the measured data. LHC for the simulation occurred at $t = 0.4$ s.

TABLE I
RMS DIFFERENCES BETWEEN THE MEASURED AND SIMULATED FORCE AND DISPLACEMENT CURVES FOR A 550 ms PERIOD

	CM displacement (m)			Projection Angles (deg)		
Segment	x	y	z	θ_x	θ_y	θ_z
R. Foot	0.033	0.007	0.023	57.7	10.9	5.4
R. Shank	0.033	0.011	0.019	5.3	6.8	3.7
R. Thigh	0.034	0.024	0.008	3.7	3.5	13.3
L. Foot	0.117	0.025	0.044	26.0	4.9	13.6
L. Shank	0.105	0.015	0.032	5.0	6.5	3.6
L. Thigh	0.060	0.014	0.012	4.9	7.6	8.0
Trunk	0.064	0.016	0.011	3.9	13.4	2.6
Force	x (N)		y (N)		z (N)	
R. GRF	65.7		111.1		25.3	
L. GRF	43.2		171.1		39.1	

axes: x - anterior/posterior; y - vertical; z - right/left.

The root mean square (rms) differences between the measured and simulated curves for all the displacement and force values vary considerably between variables (Table I). Part of the high values for the trunk stem from the fact that in the simulation, the trunk and pelvis were locked together, whereas in the measured case they were not. Another fact important when interpreting these results is that although projection angles are somewhat easier to interpret than Euler angle sequences, in certain configurations (i.e., when the long axis is almost perpendicular to the plane of projection), projection angles are very sensitive to errors. In the current project, this is of concern particularly for the frontal plane angle (θ_x) of the feet and the horizontal plane angle (θ_y) of the trunk.

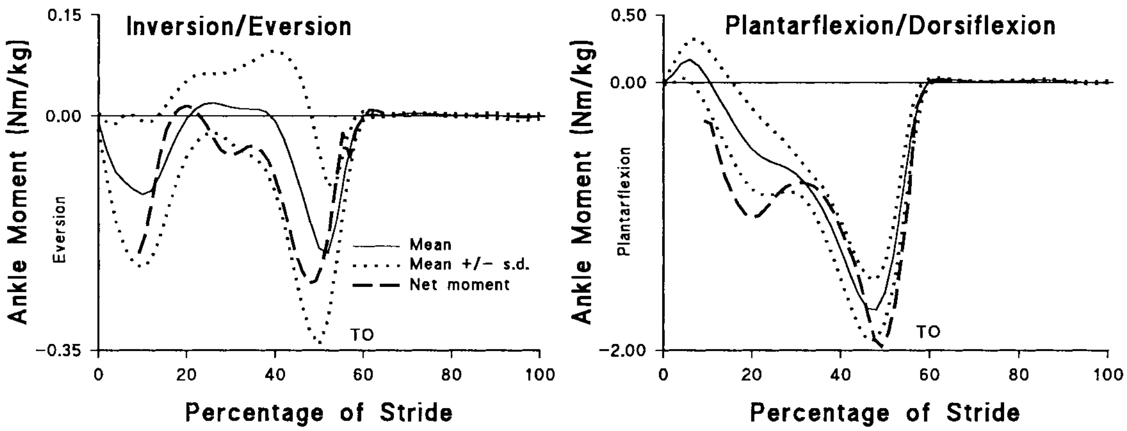
B. Controls

The general form of all the controllers was as follows:

$$\text{if} \quad \theta > \theta_o \quad \text{then} \quad T_S = k(\theta - \theta_o) \quad \text{else } T_S = 0$$
$$\text{if} \quad \theta > \theta_o \quad \text{then} \quad T_D = c\omega \quad \text{else } T_D = 0$$

where θ is the angle used by the controller, θ_o is the angle at which the restorative torque is first applied, T_S is the torque contributed by the torsional spring, k is the spring stiffness, T_D is the torque contributed by the damper, c is the damping coefficient and ω is the rate of change of θ. The total restorative torque about a given axis was the sum of T_S and T_D.

To avoid sharp changes in the restorative torques at those points when dampers became active or inactive, a third-order



(a) (b)

Fig. 7. Moments for the right ankle. The net moment from the simulation [the driving moment + the controller moment(s)] is plotted with the mean resultant joint moment ±1 standard deviation calculated from 24 walking trials. Right heel contacts occurred at 0 and 100% of stride. Right toe-off (TO) was lined up at 60%. The simulation curve is shorter because the 600 ms simulation was less than a full stride.

TABLE II
TRUNK CONTROLLERS: TORSIONAL SPRINGS AND DAMPERS APPLIED AT THE STANCE LEG HIP JOINTS TO MAINTAIN A NATURAL TRUNK POSITION

	Frontal Plane	Horizontal Plane	Sagittal Plane
Minimum angle for torque to be applied	± .001° from the vertical	± 5° from neutral	± .001° from the vertical
Spring constants	R 800 Nm/r L 900 Nm/r	100 Nm/r	RD 1200 Nm/r RS, L 600 Nm/r
Maximum damping coefficients	R 5 Nms/r L 35 Nms/r	10 Nms/r	RD 20 Nms/r RS,L 15 Nms/r
Angular deviation for full damping	0.1°	0.1°	0.1°

RS = Right leg, single support; RD = Right leg, double support; L = Left leg.
Neutral = normal stance.

polynomial was used to model the damping coefficient. That is

$$\text{For}\quad \theta \le \theta_0, \quad c = 0$$
$$\text{For}\quad \theta_0 \le \theta \le \theta_1$$
$$c = c_{\max} \times \left[\frac{\theta-\theta_0}{\theta_1-\theta_0}\right]^2 \times \left\{3 - 2 \times \left[\frac{\theta-\theta_0}{\theta_1-\theta_0}\right]\right\}$$
$$\text{For}\quad \theta \ge \theta_1, \quad c = c_{\max}$$

where θ is the angle used by the controller, θ_0 is the minimum angular deviation at which a restorative torque is applied, θ_1 is the angular deviation at which full damping is reached, c is the damping coefficient and $c_{\max}$ is the maximum value of the damping coefficient.

It should be noted that the step function was applied over very small ranges, i.e., $(\theta_1 - \theta_0 \ll 1°)$. Thus, for a large percentage of the time when a damper was active, the damping coefficient was constant $(c_{\max})$, providing linear damping.

The torsional spring/dampers used to control the trunk were set as functions of the orientation and angular velocity of the trunk principal axes with respect to the global reference system (Table II).

The ROM controllers were mathematically similar to the trunk controllers with two important differences (Table III). The angles used were the relative angles between the adjacent segments as opposed to the projection angles of a given segment. The second difference was that the angles at which the torsional spring/damper systems first became active were much greater. In this way, there was a much bigger range throughout which no restorative torques were applied. Therefore, these controllers did not always contribute torques: If the joint angle remained within the allowable ROM, no restorative (controller) torque was calculated.

In addition to the controls described above, it was found necessary to apply other correcting torques to prevent non-physiological movements. With the exception of the swing phase ankle control, all of these controllers were torsional dampers only (Table IV).

Another issue important in the assessment of the simulation is how much alteration in the driving torques was required to produce the kinematics seen. For this, the joint moments from the original 24 trial data set were used. The net ankle

TABLE III
ROM CONTROLLERS: TORSIONAL SPRINGS AND DAMPERS USED TO PREVENT JOINTS FROM EXCEEDING NORMAL END LIMITS

	Knees (F/E)	Ankles(PF/DF)
Range through which no torques were applied	2° to 145° F*	45° PF to 50° DF†
Spring constant	400 Nm/r	600 Nm/r
Maximum damping coefficient	5 Nms/r	10 Nms/r
Angular deviation for full damping	.01°	.01°

*0° = full extension; †neutral = normal stance
F = flexion; PF/DF = plantarflexion/dorsiflexion.

TABLE IV
EXTRA CONTROLLERS: TORSIONAL SPRINGS AND DAMPERS REQUIRED TO ACHIEVE NORMAL GAIT

Joint	Motion	Damping (Nms/r)	Stiffness (Nm/r)
Ankle	Inversion/ Eversion	Left swing: 0.2 Left stance: 5.0 Right swing: 0.5	none
	PF/DF*	Swing: 0.08 Left stance: 1.0 Right stance: 5.0	Swing: 1.0 Stance: none
Knee	Flexion/ extension	Left: 0.01 Right double support: 6.0	none
Hip	Int/Ext†	Swing: 0.5	none

*plantarflexion/dorsiflexion; †internal/external rotation

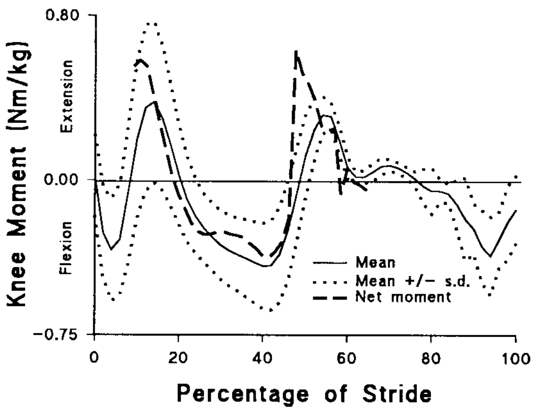


Fig. 8. Moments for the right knee. The net moment from the simulation (the driving moment + the controller moment) is plotted with the mean resultant joint moment ±1 standard deviation calculated from 24 walking trials. (See Fig. 7 caption for timing details.)

torques were generally within ±1 standard deviation of the 24 walking trial mean value because the ankle joint controllers added relatively small amounts of torque (Fig. 7). This was not the case at the knee joint (Fig. 8). During the late stance (double support) period, the net torque was well beyond the ±1 standard deviation for the group mean. The net hip torques followed a similar pattern to the group mean driving moments, but the magnitudes were such that they exceeded the ±1 standard deviation boundaries at several points (Fig. 9).

C. Sensitivity Analyses

Although sensitivity analyses were run on almost all the control parameters, the sheer volume of data generated make it impossible to present all the results graphically. Accordingly, rms differences are presented comparing the original simulation output with that of ±5% sensitivity runs (Tables V–VII). It should be noted that in some cases, the sensitivity runs were for shorter periods of time than the 550 ms of the original simulation. The simulations were ended when unnatural behavior occurred and/or when the integration algorithm failed to converge.

The model proved to be quite sensitive to changes in the trunk control parameters, particularly the spring stiffness values (Table V). Normal push-off was completely disrupted when alterations were made in either the frontal or sagittal plane springs. The model did show changes when the dampers used in the trunk controllers were changed, but these were considerably less dramatic.

There was some sensitivity to changes in the ROM controller at the left knee (Table VI). This controller was used to generate a brief torque to prevent the left knee from hyperextending immediately following heel contact. The other ROM controllers were much less sensitive. The set of sensitivity analyses run to test the sensitivity to changes in the passive torque generator at the mp joint indicate the extreme sensitivity of the model to changes in the spring constant (Table VI).

In general, the “extra” controllers were not very sensitive to changes in the damping coefficients with the exception of the plantarflexion/dorsiflexion damper at the right ankle during stance (Table VII).

TABLE V
TRUNK CONTROLLER SENSITIVITY: RMS DIFFERENCES BETWEEN THE ORIGINAL SIMULATION CURVES AND THOSE FROM ±5% SENSITIVITY RUNS

Plane	Coefficient	Right Foot * (deg) + 5 %	Right Foot * (deg) - 5 %	Trunk * (deg) + 5 %	Trunk * (deg) - 5 %	Time (s)
frontal	stiffness	1.62	6.40	0.06	0.31	0.4
	damping	0.17	0.20	0.01	0.02	0.4
horizontal	stiffness	1.03	1.33	0.08	0.08	0.4
	damping	0.09	0.15	0.01	0.01	0.4
sagittal	stiffness	14.38	1.52	0.63	0.21	0.4
	damping	0.80	0.73	0.05	0.06	0.4

* Angle in the sagittal plane.

TABLE VI
ROM CONTROLLER AND mp JOINT SENSITIVITY: RMS DIFFERENCES BETWEEN THE ORIGINAL SIMULATION CURVES AND THOSE FROM ±5% SENSITIVITY RUNS

Joint	Coefficient	Right Foot * (deg) + 5 %	Right Foot * (deg) - 5 %	Trunk * (deg) + 5 %	Trunk * (deg) - 5 %	Time (s)
L. Knee	stiffness	0.09	30.00	0.02	1.08	0.55
	damping	0.11	0.16	0.02	0.02	0.55
R. Ankle	stiffness	0.14	0.20	0.02	0.01	0.55
	damping	0.15	0.15	0.02	0.02	0.55
R. Mp.	stiffness	9.28	12.11	0.13	0.18	0.50
	damping	9.86	14.20	0.15	0.32	0.50

* Angle in the sagittal plane.

TABLE VII
EXTRA CONTROLLERS SENSITIVITY: RMS DIFFERENCES BETWEEN THE ORIGINAL SIMULATION CURVES AND THOSE FROM ±5% SENSITIVITY RUNS

Joint	Motion	Right Foot * (deg) + 5 %	Right Foot * (deg) - 5 %	Trunk * (deg) + 5 %	Trunk * (deg) - 5 %	Time (s)
R. Ankle	Inv/Ev	0.04	0.01	0.00	0.01	0.55
	PF/DF	10.44	30.57	0.41	1.61	0.55
R. Knee	F/E	0.27	0.16	0.00	0.01	0.50
R. Hip.	Int/Ext	0.02	0.00	0.00	0.00	0.55

* Angle in the sagittal plane; Inv/Ev = inversion/eversion
PF/DF = plantarflexion/dorsiflexion; F/E = Flexion/extension
Int/Ext = internal/external rotation

IV. DISCUSSION

The simulation model presented here was successful in that the predicted kinematics, ground reaction forces and even the adjusted moments approximated normal walking for a 550 ms period. Moreover, both single and double support phases have been modeled, including a continuous transition through heel contact. In this way it has moved us closer to the ideal simulation, that is, the one that will be suitable to address clinical issues such as prosthesis design. It is clear, however, that the model’s limitations are such that it still falls short of the ideal.

Few previous studies have modeled so many segments with so many degrees of freedom. It is clear that for the simulation to be useful, it must use a realistic model. In particular, three degrees of freedom at the hip joints are important because the additional movements (adduction/abduction and internal/external rotation) allow the torso to maintain its smooth forward progression, with minimal twisting and/or rising and falling. The rms differences between the measured and simulated curves (Table I) indicate that allowing these DOF was somewhat successful: the trunk linear translation patterns were very similar, although there were greater differences in the twist angle (θ_y).

Most previous models have fixed the foot to the ground in some fashion for some portion of the gait cycle. As a

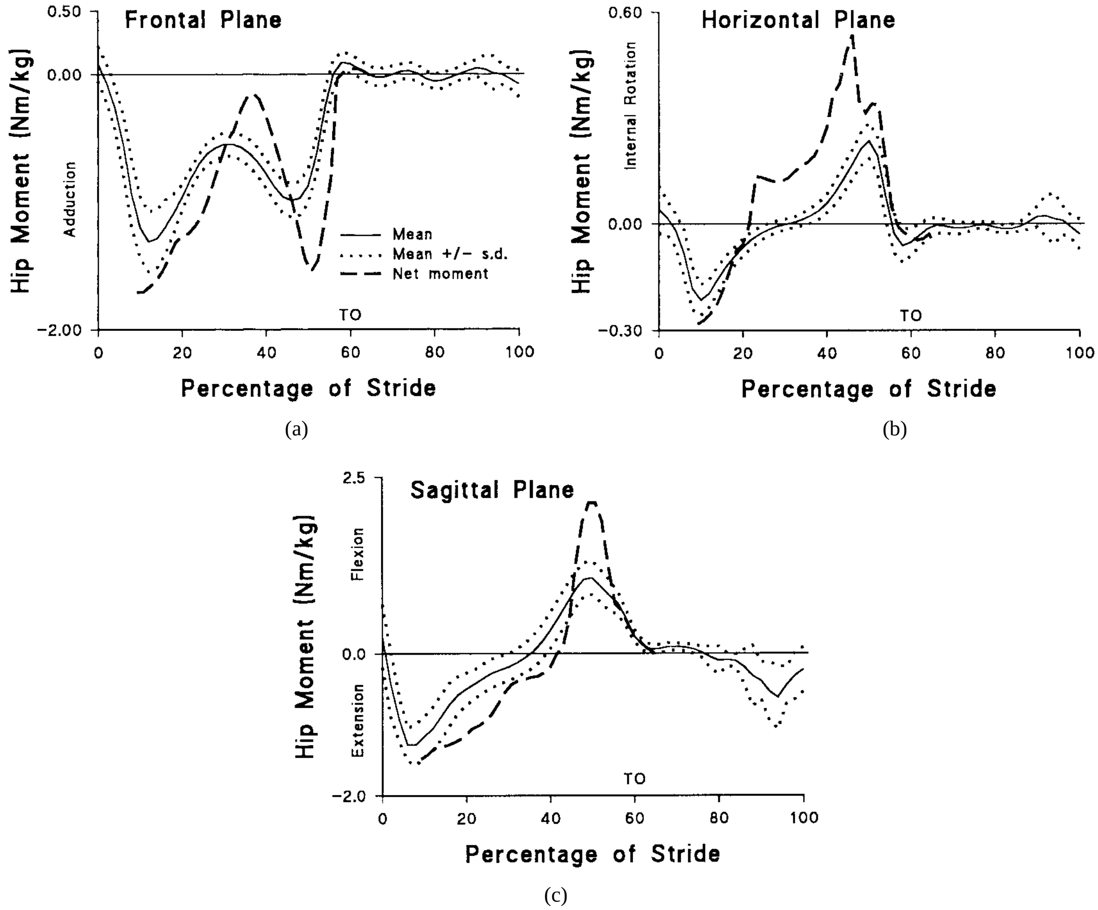


Fig. 9. Moments for the right hip. The net moment from the simulation (the driving moment + the controller moment) is plotted with the mean resultant joint moment ±1 standard deviation calculated from 24 walking trials. (See Fig. 7 caption for timing details.)

result, few simulations have modeled the transition phases (heel contact and toe-off), crucial points in the gait cycle. The use of visco-elastic contact elements in the current simulation was the critical feature that allowed impact to be successfully modeled. Being able to model impact is vital for an applied gait simulation tool: foot placement at heel contact is an important strategy for maintaining balance while walking [32]. Thus, if one wants to be able to include considerations of balance, the simulation must be able to make the transition from swing to stance. Furthermore, several gait pathologies manifest themselves in the push-off (propulsive) phase at the end of stance. It would be very limiting not to include this in an applied tool as improved propulsion is often the goal of rehabilitation strategies.

A major limitation of the model is apparent in the sensitivity results (Tables V–VII). When the model is found to be very sensitive to a given parameter, it implies that a critical value of the parameter has been found. If this critical value proves to be applicable to a large segment of the population, then it is of great value. Unfortunately, the high variability of biological systems, combined with the errors inherent in the measurement and computational processes, make this an unlikely scenario. It seems reasonable to expect that, if this model was applied to data from different subjects, the parameters with low sensitivity values would transfer more directly than those with high sensitivity values.

Nonetheless, what can be learned from the sensitivity results is that simple dampers can provide a great deal of "control" to linked, rigid-body models, without being extremely sensitive to the exact coefficient value. Caution must be taken however, to avoid pure step changes in the damping forces. Including damping at many joints in a model is not inconsistent with the biological system: some light damping must be present in the joints of the body, given the synovial fluid and the visco-elastic nature of tissues such as ligaments and cartilage.

Although other simulations have met the criterion of predicting reasonable kinematics, this project differs in that it did not force the model to fit a pattern of pre-determined trajectories. It can be argued that there is nothing wrong with using predetermined trajectories and certainly the gait simulation literature attests to the fact that a reasonable level of success can be achieved with closed-loop feedback systems based on such trajectories (e.g., [19]). Others (e.g., [2]) have had success with an optimization pattern which minimized the deviations in segment displacement patterns from prescribed trajectories. The fact remains, however, that these techniques require very specific *a priori* knowledge of the movement pattern.

The first philosophical argument against using prescribed trajectories is that if a novel situation were to be introduced to this type of simulation, the response would be always directed toward maintaining the normal walking trajectories. This is not how the human body operates: if loss of balance is threatened, for example, the swing leg trajectory can be sharply altered from its normal pattern to prevent an uncontrolled fall. The overall walking pattern may continue, but this is achieved through very different joint and segment displacement patterns. This is not to say that some form of general trajectory planning does not occur. Studies of obstacle avoidance strategies (cf., [34]–[36]) clearly show that some pre-planning occurs when adequate time is available. These same studies, as well as gait perturbation studies (cf., [37]), however, also indicate that when necessary, pre-planned trajectories can be superseded.

A further argument against prescribed trajectories is that data show that even for one subject, there is stride-to-stride variability in the normal gait displacement patterns [38]. So it is unrealistic to expect (or constrain) a given segment to be at one unique angle at a given point in time.

These disadvantages in using prescribed trajectories must, however, be weighed against the fact that a pure open loop strategy is unlikely to work because of the error factor [29], [30]. Somewhere in between these two extremes lies a strategy that is both realistic and pragmatic. Such a strategy will come about from consideration of what features characterize *all* human walking, not just one trial of one individual. These include the following:

- the upper body must remain close to the vertical [32], [33];
- the movement at each joint must remain within its physiological ROM;
- a stance leg must provide support at all times;
- there must be forward progression with an alternating pattern of leg support;
- the swing foot must clear the ground until the body is suitably positioned for weight transfer.

The controllers used in the current project specifically addressed the first two items on this list. No specific strategy was adopted to deal with the remaining ones—the "extra" controllers accomplished what was needed but not as part of a control algorithm. The development of such an algorithm would offer substantial improvement to the simulation.

In summary, a modeling approach for a simulation was taken that differed quite markedly from what has appeared previously in the literature. It was reasonably successful in meeting the immediate aim of simulating normal gait but has not yet achieved the ultimate goal of an applied research tool.

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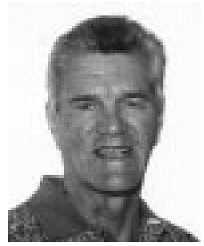

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