Journal of

MECHANICS IN MEDICINE AND BIOLOGY

Volume 4 • Number 3 • September 2004

Evaluation of a Computer-Simulation Model for Human Ambulation on Stilts

C. S. Pan, K. M. Miller, S. Chiou and J. Z. Wu

Abstracts and table of contents are available at

www.worldscinet.com/jmmb.html

World Scientific

NEW JERSEY • LONDON • SINGAPORE • BEIJING • SHANGHAI • HONG KONG • TAIPEI • CHENNAI

Journal of Mechanics in Medicine and Biology
Vol. 4, No. 3 (2004) 283–303

World Scientific
www.worldscientific.com

EVALUATION OF A COMPUTER-SIMULATION MODEL FOR HUMAN AMBULATION ON STILTS

CHRISTOPHER S. PAN*, KIMBERLY M. MILLER, SHARON CHIOU and JOHN Z. WU
National Institute for Occupational Safety and Health
Morgantown, West Virginia, USA

Received 30 March 2004
Accepted 30 March 2004

Stilts are elevated tools that are frequently used by construction workers to raise workers 18 to 40 inches above the ground without the burden of erecting scaffolding or a ladder. Some previous studies indicated that construction workers perceive an increased risk of injury when working on stilts. However, no in-depth biomechanical analyses have been conducted to examine the fall risks associated with the use of stilts. The objective of this study is to evaluate a computer-simulation stilts model. Three construction workers were recruited for walking tasks on 24-inch stilts. The model was evaluated using whole body center of mass and ground reaction forces. A PEAK^{TM} motion system and two Kistler^{TM} force platforms were used to collect data on both kinetic and kinematic measures. Inverse- and direct-dynamics simulations were performed using a model developed using commercial software — ADAMS and LifeMOD. For three coordinates (X, Y, Z) of the center of mass, the results of univariate analyses indicated very small variability for the mean difference between the model predictions and the experimental measurements. The results of correlation analyses indicated similar trends for the three coordinates. Plotting the resultant and vertical ground reaction force for both right and left feet showed small discrepancies, but the overall shape was identical. The percentage differences between the model and the actual measurement for three coordinates of the center of mass, as well as resultant and vertical ground reaction force, were within 20%. This newly-developed stilt walking model may be used to assist in improving the design of stilts.

Keywords: Computer simulation; model evaluation; biomechanics; stilts; multi-body dynamics; direct-dynamics; inverse-dynamics; gait.

1. Introduction

The construction industry has the highest fatal and lost workday injuries among the major industry trades.[22] Two of the primary causes of on-the-job injury among construction workers are overexertion and falls. For the cases involving days away from work, the Bureau of Labor Statistics (BLS) reported that approximately

*Corresponding author: NIOSH, 1095 Willowdale Road, Morgantown, WV 26505, USA.
E-mail: cpan@cdc.gov; Tel.: 304-285-5978; Fax: 304-285-6047.

24% were the result of overexertion or repetitive motion in the construction industry, in 2000.[3,4] Falls accounted for approximately 22% of the cases involving days away from work in the construction industry.[4] For drywall installers and carpenters, 37% and 32% of injuries, respectively, are due to bodily reaction/exertion and falls individually.[5–7] One likely cause for the high incidence of overexertion and fall injuries among drywall installers and carpenters is their use of stilts and other elevated devices, such as scaffolding and ladders.[29,30] Like scaffolds, stilts are an elevated tool that raises workers above ground level to allow them to perform tasks on the ceiling or upper half of a wall. Construction workers can obtain an extra 1.5 to 3.5 feet of elevation using stilts without the burden of erecting scaffolding or a ladder. Stilts are composed of more than fifty small parts, and provide mobility for workers. Painters (or finishers) also use stilts to perform drywall finishing tasks, including drywall taping and sanding. Stilts have also been used widely for other elevated interior tasks such as painting, plastering, insulation installation, acoustical ceiling installation, or light-duty building maintenance. Schneider and Susi[32] hypothesized that the use of stilts may place workers at increased risk for knee injuries or increase the likelihood of trips and falls. Construction laborers, carpenters, drywall installers, and painters are responsible for almost 50% of fall-related injuries in the construction industry.[28] All of these four construction workforces use stilts, either regularly or irregularly, at worksites. A survey of drywall installers and carpenters indicated that workers perceived greater fall potential and physical stress when using stilts, compared to using other elevated tools such as scaffolds and ladders.[29,30]

The State of California and the province of Ontario, Canada, have established legislation prohibiting the use of stilts, as a preventive measure for occupational injuries. A variety of studies, which examined the safety of stilt use in construction workers, have noted that construction workers perceive an increased risk of injury when working on stilts.[29,30,38] Beyond the conceivable risks that might result from the use of stilts, however, no quantitative data exist to demonstrate the potential hazards associated with using stilts. In order to reduce the potential hazard associated with stilt use, several interested parties have made recommendations, based on their experiences and perceptions of safe operating parameters; for example, a training guideline[20] of the International Union of Painters and Allied Trades suggests that the recommended maximum safe height for the use of stilts is 24 inches for the drywall finishers and painters.

Multi-body system dynamic modeling and simulation have been used to evaluate crash incidents, fall scenarios, and other gait-related issues.[12,14–19,33–35] In 2002, using ADAMS and LifeMOD as a simulation platform, a human-stilts multi-body system (Fig. 1) was developed through a collaborative research project between the National Institute for Occupational Safety and Health (NIOSH) and Mechanical Dynamics Inc. It was hypothesized that this model of a human walking on stilts could assist researchers in evaluating adverse effects that might apply to the stilt users. Although computer simulations of human walking are well

Fig. 1. Multi-body model (ellipsoid representation) of the human-stilts system. The model was developed using LifeMOD^{TM} (MSC Software Inc.).

known,[1,2,8,26,27,31,35] there is no published literature on computer simulations of human walking on stilts.

This computer simulation model of human walking on stilts can be used to analyze the mechanical stability and to evaluate the joint reaction forces of the users during stilt walking, especially when results are analyzed along with findings from a current laboratory study with human subject tests at NIOSH.[25] Simulation results of the ground reaction force and center-of-mass found in stilt walking would be compared to those of normal walking to analyze the effects of the stilt walking on the joint loading and mechanical stability. The objective of this study is to evaluate this computer-simulation model in predicting trajectory of human center of mass and ground reaction forces during stilt walking. This newly-developed stilt walking model may be used to assess tripping hazards, struck by/against, and other traumatic injury scenarios, sudden starts/stops and pivots while using stilts, and most importantly, to assist in improving the design of stilts.

2. Materials and Methods

2.1. *Data collection*

Three male construction workers between the ages of 34 and 40 with at least 12 months of experience in the use of stilts were recruited for walking tasks on 24-inch stilts. They underwent a health-history screening before participating in the study. Subjects with the following conditions were to be excluded from this project: history of dizziness, tremor, vestibular disorders, neurological disorders, cardiopulmonary disorders, diabetes, chronic back pain, and individuals who have fallen within the past year resulting in an injury with days away from work. A health history that includes any of these conditions could influence the performance of the subject during testing. Subjects were questioned including height, weight, and shoe size. Then they changed into tight fitting black clothes and safety shoes so their clothing would not interfere with the testing equipment and data collection. Twenty-two reflective markers were placed on the subject's body (Fig. 2), and eight markers were placed on the stilts (Fig. 3). In Fig. 3, two markers (C and D) were used to identify the floor plate, one marker (B) was used to represent the shoe plate, and one marker (A) was used to replace the ankle marker once the subject donned the stilts. Subjects wore a bicycle helmet, safety shoes and a MSA vest-type harness that was attached to a Gantry system (Exonic Systems™, Pittsburgh, PA) that ran by remote control on rails. The lanyard of the harness was properly adjusted for each subject so that he would be protected but not restricted by the harness during the experiment. This harness system was manually operated by a hand-held control pendant, which incorporated a joystick. The gantry/harness system served as a safety control during the entire time of the experiment to reduce the chance of injury.

Before actual testing, the subject was asked to comfortably walk around on the 24-inch stilts to make sure that the springs were adjusted correctly to counterbalance his weight. Then, the subject was trained to perform a walking straight test on the 24-inch stilts across the lab (28.5 feet). The subject walked across two force plates (Model 9287, Kistler™ Instrumentation Corporation, Amherst, NY) with care, stepping on each plate only once, starting with the left foot.

While the subject was performing the test, six integrated cameras were used to capture the three-dimensional position of reflective markers, at a rate of 60 frames

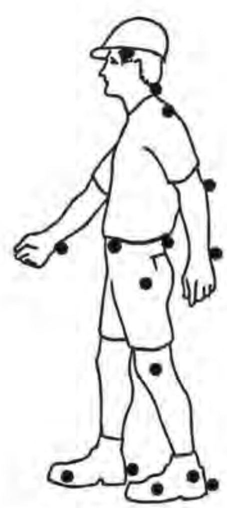

Fig. 2. Twenty-two reflective markers were placed on the subject's anatomical landmarks.

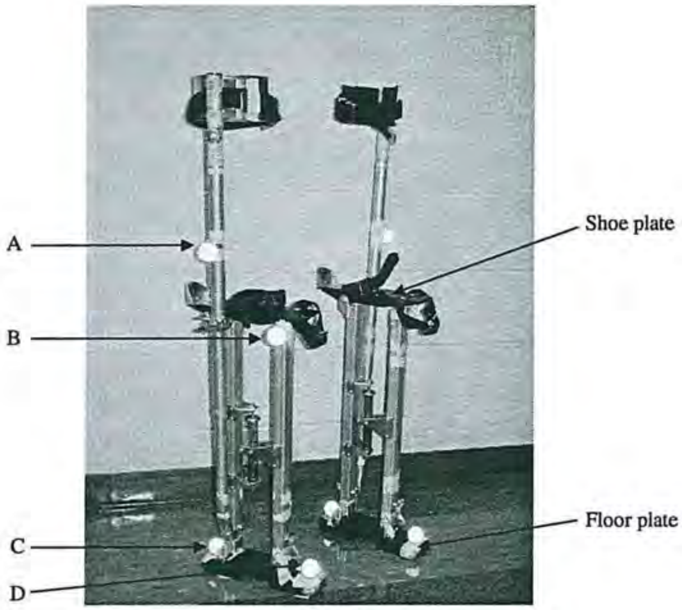

Fig. 3. Stilt used for the present study. Four reflective markers (A, B, C, D) were attached to each stilt for this study experiment. Marker A was used to replace the ankle marker once the subject donned the stilts. Marker B was used to represent the shoe plate. Markers C and D were used to identify the floor plate.

per second. A motion analysis system (Motus^{TM}, PEAK Performance Technologies Incorporated, Englewood, CO) was used for this kinematic measurement. The Motus system was also used to collect force data from the charge amplifiers (Model 9865B, Kistler^{TM} Instrumentation Corporation, Amherst, NY) that received the piezoelectric kinetic data from the force plates. The time-synchronized records of both kinetic and kinematic measures were saved in an output file.

2.2. *Model evaluation*

The position data for each marker were exported from the Motus system and imported into LifeMOD Biomechanics Human Modeler (version 1.5, Biomechanics Research Group Inc, San Clemente, CA), which is a plug-in to ADAMS (version 12, MSC^{TM} Software, Santa Ana, CA). ADAMS has the capacity to perform dynamic analysis of mechanical simulations, whereas LifeMOD Biomechanics Modeler adds the ability to incorporate human modeling into the simulation. The human model is composed of 15 segments and 16 joints (head, neck, upper torso, central torso, lower torso, upper and lower arms, upper and lower legs, feet). The LifeMOD Biomechanics Modeler software uses five anthropometric body size databases (GeBod data, and PeopleSize UK, USA, Japan and China) for segment weights geometry, and joint locations based on the gender, overall height, and overall weight of the simulated subjects. The joints used for this model were analytical representations derived from the Hybrid-III crash test dummy which included individual joint limits (forces with hysteresis), friction, damping, and non-linear stiffness profiles.[37]

An AutoCAD drawing of the stilts at a height of 24 inches was imported into LifeMOD Biomechanics Modeler and attached to the human model by six bushing forces. The bushing forces represented each of the straps used to attach the stilts to the persons' legs when preparing to walk on stilts. LifeMOD "motion agents" (0.7 inches in diameter) were attached to the model using a six-degree-of-freedom bushing force at the same locations as the physical markers attached to the subject during data collection. During the inverse-dynamics simulations the motion agents in the model were subjected to the time-histories of the position data collected using the Motus system. Since the motion agents were connected to the human model and stilt by bushing elements, the effects of the errors due to the relative movements of the markers to the human body skeletal system or stilts during motion can compromise measurement accuracy. Individual bushing stiffnesses were adjusted based on the relative accuracy of the specific motion target location permitting the more accurate target location to have a greater influence on the motion of the model. During the simulations, the angles and displacements of the joints are recorded in LifeMOD. After the inverse-dynamics simulations all the motion agents were deleted. One motion agent was re-created on the human model's back to generally stabilize the model during forward progression in the direct dynamics simulation. The recorded joint angles and displacements from the inverse-dynamics simulation were then used in a direct-dynamics simulation by including the histories in proportional-derivative servo-type controllers which generated the necessary joint torques to match the recorded motion. From the direct-dynamics simulations one can determine the body center of mass, segment displacements, contact forces, joint torques, and other characteristics of the entire model and each segment of the model. The model development procedure is summarized in Fig. 4.

The contact interactions of foot/ground and stilt/ground are modeled using nonlinear springs with exponential hardening:

$$F = A \cdot \Delta^n, \tag{1}$$

where F is the normal contact force, Δ is the relative displacement between the contacting bodies, and A and n are the stiffness parameters. When $n = 1$, the nonlinear spring, as defined in Eq. (1), is reduced to a linear spring.

In the model, the criterion of the relative slipperiness of the contact between the feet and ground or between the stilts and ground is defined as:

$$\frac{\sqrt{F_x^2 + F_z^2}}{|F_y|} \leq \text{COF}, \tag{2}$$

where F_x, F_y and F_z are the components of the ground reaction force in x (medial-lateral), y (vertical), and z (anterior-posterior) directions.

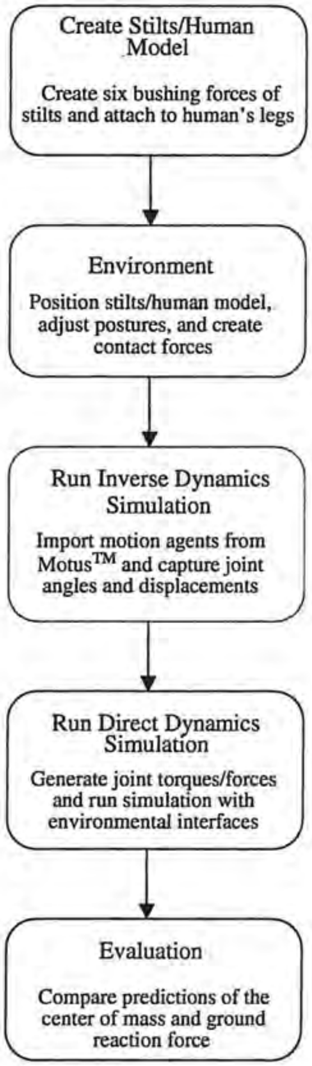

Fig. 4. A flow diagram of stilts model building process.

The maximum coefficient of friction (COF) between the feet and ground and between the stilts and ground was assumed to be 0.70 in the simulations. Test runs of the model simulations indicated that COF should be at least 0.7 to maintain mechanical stability during walking. This value appears reasonable based on our prior experimental data.[25]

The resultant and vertical components of the ground reaction force at the stilt foot as well as the center of mass of the model were used to evaluate the model. The forces and center of mass from the simulations were compared to the forces recorded by the Kistler force plates and the center of mass calculated from the data by the Motus$^{\text{TM}}$ system. The percentage difference between the two sets of data was determined using the data from the Motus Analog Acquisition Module as the point of comparison. Any percentage difference less than 20% was considered that the model predictions are satisfactory.[21]

For the center of mass, Pearson correlation analyses and actual differences were used to evaluate the trend of the changes for three coordinates. Univariate and multivariate analyses were additionally used to assess variability of the mean differences for three coordinates.

3. Results

3.1. *Center of mass*

One of the built-in functions of the Motus system is determination of the whole body center of mass. These data were compared to the whole body center of mass data of the model after the direct-dynamics simulation was completed. The time histories of the center of mass for the Motus system agree well with those obtained using LifeMOD Biomechanics Modeler (Fig. 5), with Pearson's correlation coefficients of 0.82, 0.88, and 0.99 for the X, Y, and Z coordinates, respectively. The relative (Fig. 6) and the absolute position differences (distance) (Fig. 7) between the two sets of data were calculated by:

$$\delta_{\text{relative}} = \frac{L_{\text{LifeMOD}} - L_{\text{Motus}}}{L_{\text{Motus}}} \times 100, \quad \text{in } X, Y, \text{and } Z \text{ directions,} \quad \text{and} \tag{3}$$

$$\delta_{\text{absolute}} = \sqrt{(X_{\text{LifeMOD}} - X_{\text{Motus}})^2 + (Y_{\text{LifeMOD}} - Y_{\text{Motus}})^2 + (Z_{\text{LifeMOD}} - Z_{\text{Motus}})^2}, \tag{4}$$

where L_{LifeMOD} and L_{Motus} represent the X, Y, and Z coordinates obtained using LifeMOD and Motus, respectively; δ_{relative} and δ_{absolute} are the relative and absolute differences.

In Figs. 5–7 and Tables 1 and 2, the X, Y, and Z labels refer to the coordinate system used by the LifeMOD Biomechanics Modeler software. The X coordinate is a horizontal axis that moves from the models' left to right (i.e. medial-lateral). The Y coordinate is the vertical axis with the positive direction pointing upwards (i.e. vertical). The Z coordinate is a horizontal axis that corresponds to the direction of walking, from the posterior to the anterior of the model. Results from univariate analysis showed that the mean differences between the LifeMOD and the Motus system were 0.14 cm, 5.40 cm, and −3.71 cm, in X, Y, and Z directions, respectively (Table 2). The variability of the mean differences for all three coordinates was relatively small as indicated in Table 2. Multivariate analysis results showed significant mean differences in Y and Z coordinates between the LifeMOD and the Motus system; however, the magnitudes of the differences were practically

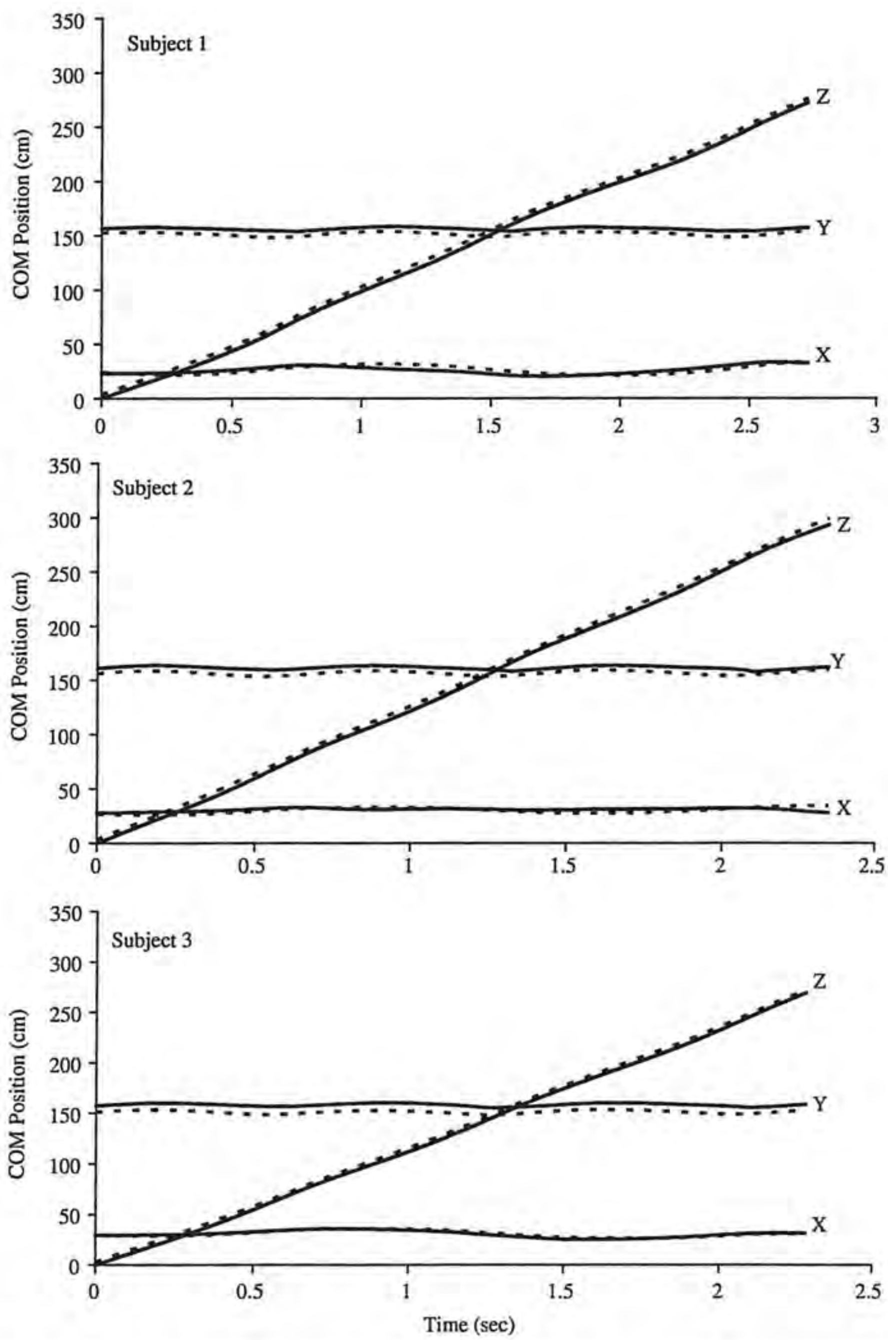

Fig. 5. Time-histories of COM positions. Solid lines represent model estimation values, dotted lines represent actual measurements from the PEAK motion system.

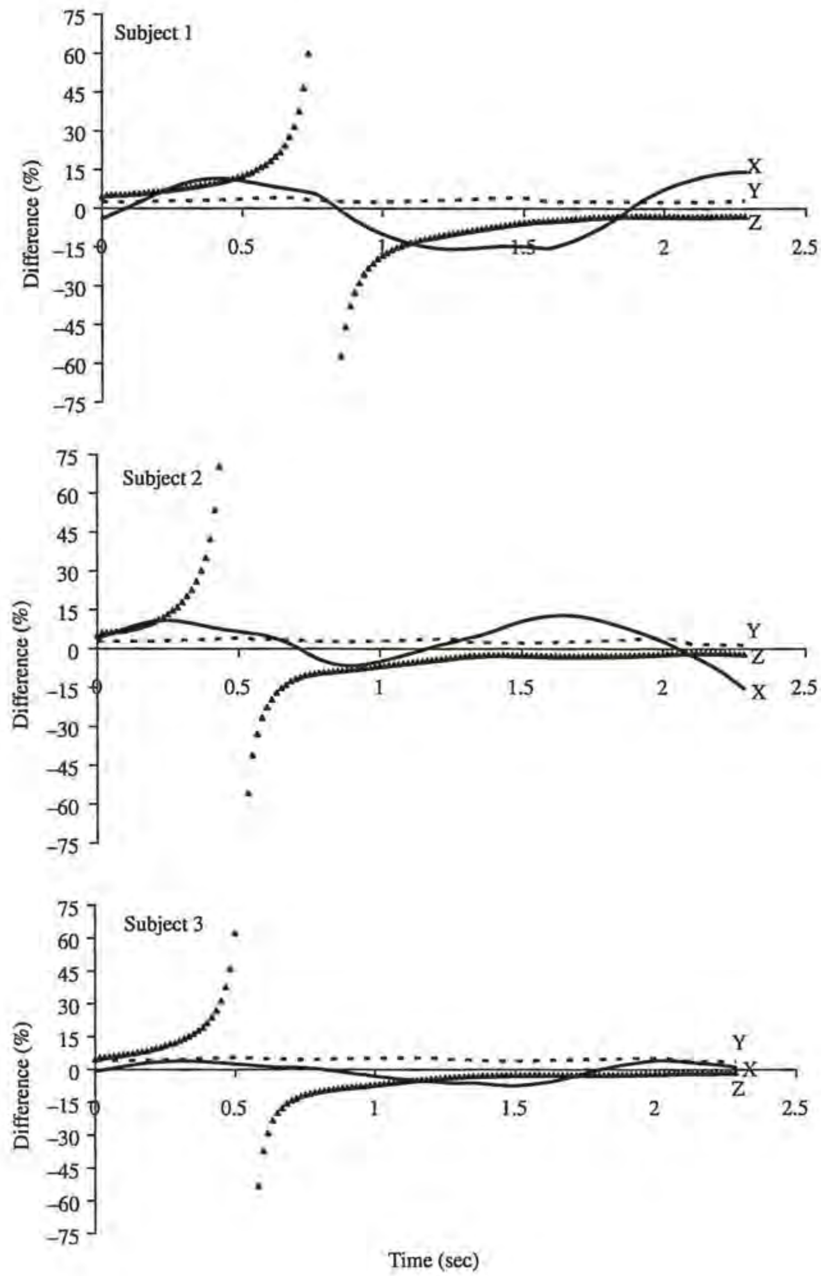

Fig. 6. Relative position differences of COM. X is the horizontal axis that goes across the model's body, from left to right (medial-lateral). Y is the vertical axis. Z is the horizontal axis that is perpendicular to the X-axis, going from the back to the front of the model (anterior-posterior).

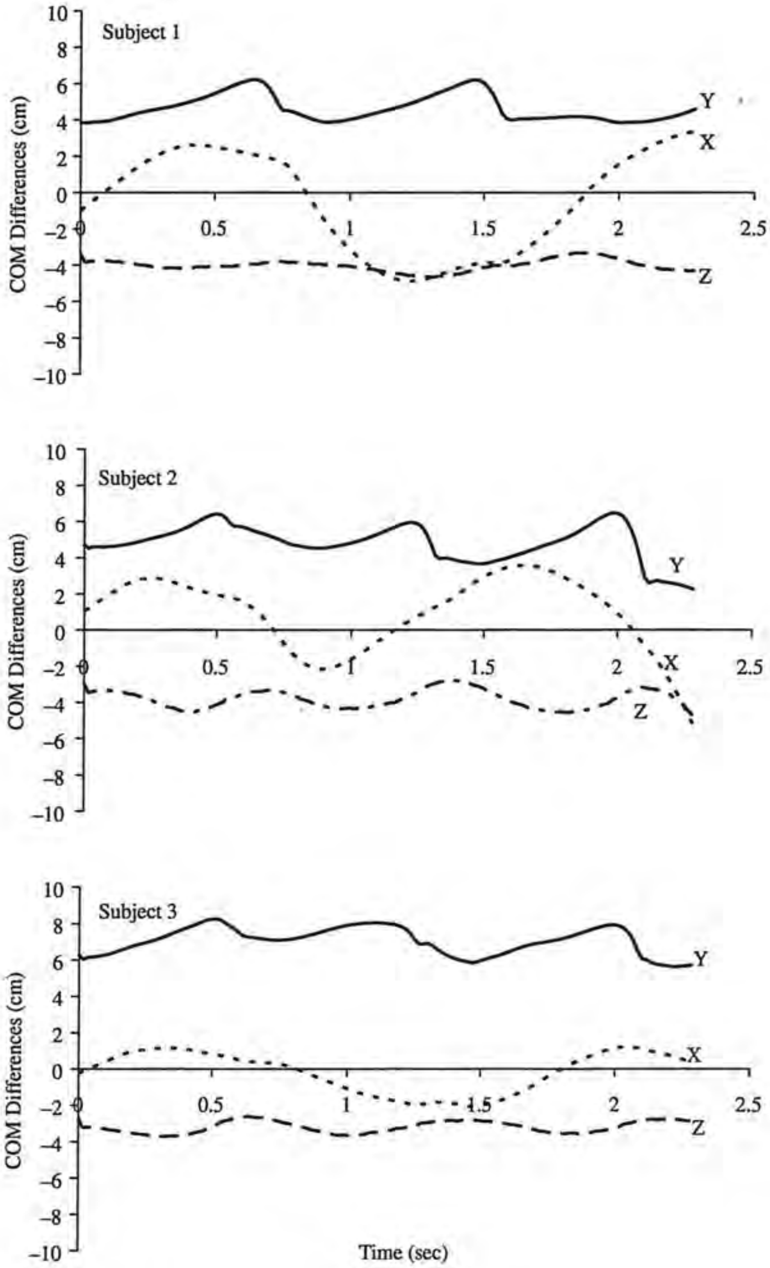

Fig. 7. Time-histories of absolute COM position differences for three coordinates. X is the horizontal axis that goes across the model's body, from left to right (medial-lateral). Y is the vertical axis. Z is the horizontal axis that is perpendicular to the X-axis, going from the back to the front of the model (anterior-posterior).

Table 1. Relative position differences of center of mass (COM) for X and Y coordinates. X is the horizontal axis and goes across the model's body, from left to right (medial-lateral). Y is the vertical axis.

	X		Y	
%	Average	Maximum	Average	Maximum
Subject 1	9.36	15.78	3.04	4.21
Subject 2	7.06	18.28	3.03	4.19
Subject 3	3.25	7.39	4.65	5.54

Table 2. Mean differences of COM and 95% confidence intervals for three coordinates. X is the horizontal axis that goes across the model's body, from left to right (medial-lateral). Y is the vertical axis. Z is the horizontal axis that is perpendicular to the X-axis, going from the back to the front of the model (anterior-posterior).

					95% CI[a]	
Variable (cm)	Mean	Std error	Minimum	Maximum	Lower[b]	Upper[b]
Difference in X Coordinate[c]	0.14	0.11	−6.25	3.57	−0.16	0.44
Difference in Y Coordinate[c]	5.40	0.07	2.18	8.26	5.21	5.59
Difference in Z Coordinate[c]	−3.71	0.03	−5.74	−2.60	−3.78	−3.64

[a]CI: Confidence Interval of mean difference.
[b]95% Simultaneous confidence intervals for the mean differences of X, Y, Z coordinates.
[c]Difference = LifeMOD measurement − Peak measurement.

negligible. There was no significant mean difference in the X coordinate (Fig. 7). The 95% simultaneous confidence intervals of the mean differences for all three coordinates were very narrow; i.e. −0.16 to 0.44 cm, 5.21 to 5.59 cm, and −3.78 to −3.64 cm for X, Y, and Z coordinates (Table 2), respectively.

3.2. *Ground reaction force*

The ground reaction force data collected from the Kistler force plates were compared with those predicted using the model after the direct-dynamics simulation was completed. Since each force plate was only stepped on once, and the subjects had to step on the first force plate with their left foot and the second force plate with their right foot, the ground reaction forces can be broken into left foot and right foot forces. For all three subjects, the forces in the vertical direction and force magnitudes obtained experimentally are compared with those predicted using the computer model, as in Figs. 8 and 9. The analysis indicates some discrepancies between the experimental measurements and theoretical predictions; however, the overall trends are consistent for all three subjects (Figs. 8 and 9). The average percent difference and average actual difference between the two sets of force data were also calculated and separated into left and right foot ground reaction forces (Tables 3 and 4). The percentage difference was calculated in the same method as center of mass data using Eq. (3).

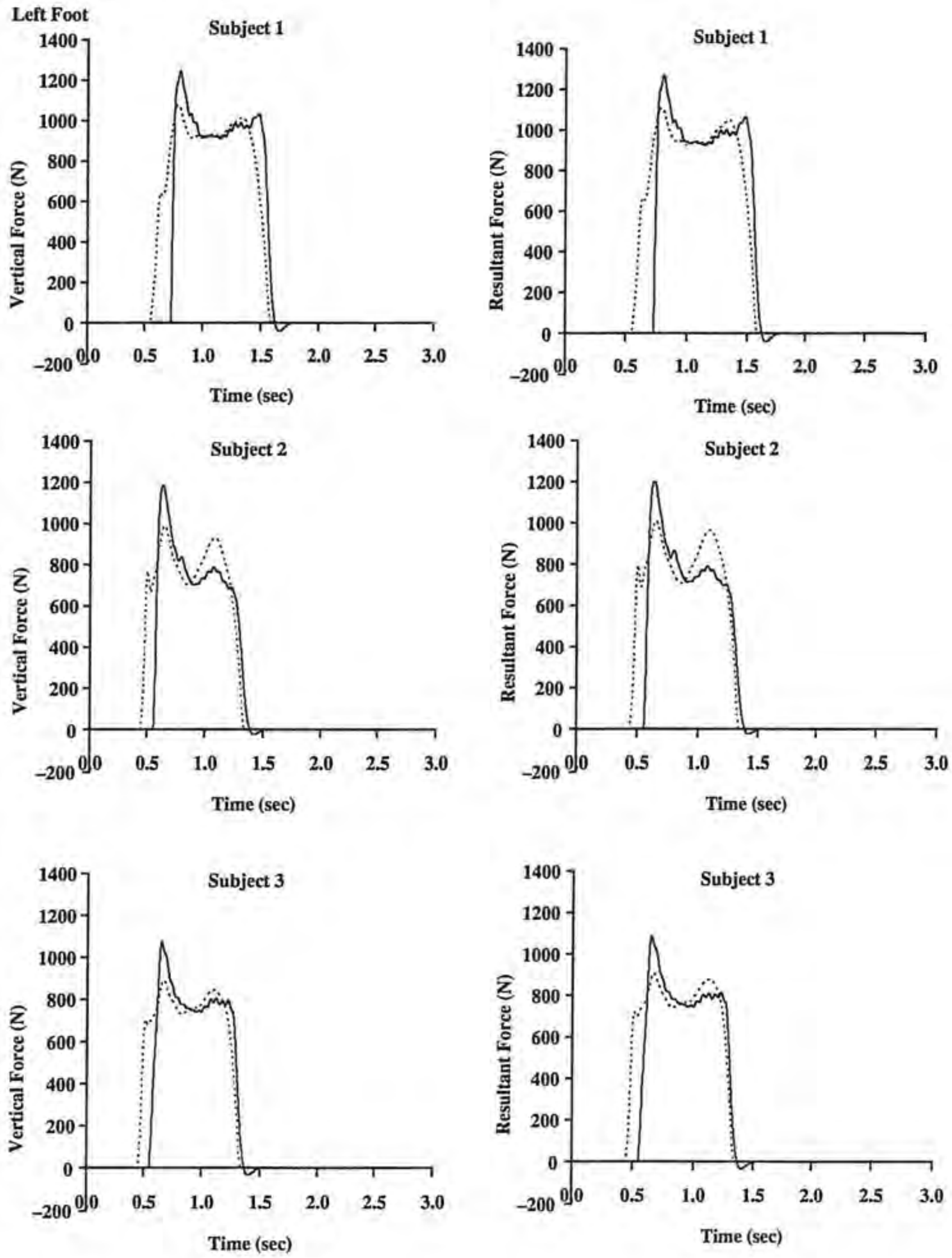

Fig. 8. Vertical component and resultant of ground reaction force (N) for left foot (solid lines represent model estimation values, dotted lines represent actual measurements from the PEAK motion system).

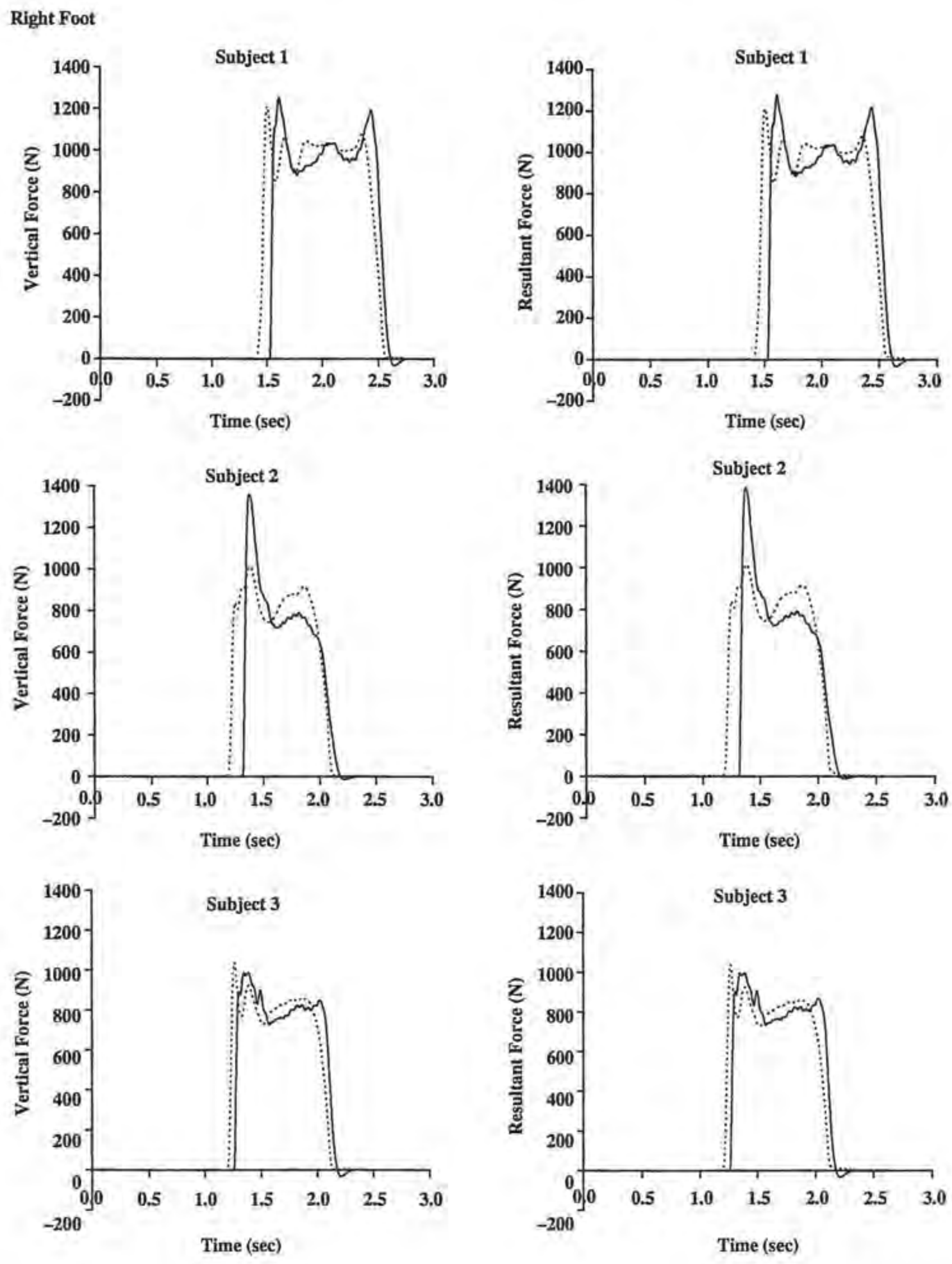

Fig. 9. Vertical component and resultant of ground reaction force (N) for right foot (solid lines represent model estimation values, dotted lines represent actual measurements from the PEAK motion system).

Table 3. Average percentage differences of ground reaction force.

Difference (%)	Right foot		Left foot	
	Vertical	Resultant	Vertical	Resultant
Subject 1	6.88	7.03	5.11	4.83
Subject 2	14.04	14.24	11.09	11.94
Subject 3	7.86	8.03	6.96	7.24

Table 4. Average absolute differences of ground reaction force (N).

Force (N)	Right foot				Left foot			
	X	Y	Z	Resultant	X	Y	Z	Resultant
Subject 1	16.73	101.37	31.49	102.71	18.95	58.14	51.82	56.49
Subject 2	20.86	130.11	28.96	131.13	16.95	58.00	62.72	61.47
Subject 3	4.89	71.71	28.60	72.73	28.69	41.99	30.92	41.95

4. Discussion and Conclusions

Walking on stilts has raised a number of concerns in the construction industry and the majority of the concerns are focused on slip, trip, and fall exposures. There is also a concern because workers perceive greater physical stress while on stilts, since stilts are rigid.[29,30] These concerns have lead researchers to look into the design safety of stilts. Do stilts pose an unnecessary risk to construction workers? Do stilts contribute to any chronic injuries and lead to long-term medical problems?

These questions can be examined through many means including human subject testing, mathematical modeling, subjective questionnaires, and computer modeling. However, computer modeling offers some advantages that the others do not. First of all, computer modeling generates more reproducible results[34] and is inexpensive compared to human subject testing because only one person is needed to develop and run the model, whereas with subject testing multiple people are usually required to be involved. Computer modeling also enables the user to easily implement many parameters to look at different variables, such as tripping hazards and joint forces. In this way a cause-and-effect simulation can be conducted efficiently.[12,33–35] A computer model can also be used to perform certain simulations that cannot be conducted at laboratories or work sites, such as actual tripping and falling.[35] Many other potential applications of computer modeling await development.[34]

The Motus™ center of mass (COM) definitions are specific to the spatial model set up in the trial. The body is divided into individual segments given proximal and distal endpoints. The location of the center of mass for each segment in terms of a percent distance from the proximal and distal endpoints was defined. Each segment's mass was expressed as a percentage of the whole body mass. These parameters typically come from body segment parameter data compiled by Dempster (1955)[10] or Clauser (1968)[9] or variations of them. Given the coordinates of the segment endpoints obtained through Motus, first the coordinates of the individual segmental centers of mass were computed. Then, their locations along with their mass contributions were used to determine the whole body center of mass (Motus^{TM}, PEAK Performance Technologies Incorporated, Englewood, CO). Other fundamental theories and studies[11,13,23,36] were also cited for this definition. Therefore, the Motus^{TM} model has been well-accepted for the evaluation of center of mass or other kinematics.

The model's calculated value for whole body center of mass is an important metric for this study. If the center of mass of the model is inaccurate, then any data collected using the model, such as joint or ligament forces and torques, would lead to a larger overall error. Plotting the center of mass of the LifeMOD model and the center of mass from the Motus (i.e., PEAK motion system) on the same graph, as shown in Fig. 5, shows that these two data look similar; there does not appear to be any large discrepancies between the two sets of data. The positive results of the correlation analyses indicated that these two data significantly correlated to each other and show the same trend. The pattern of the differences of the distance (or absolute position difference) also showed very similar results among three test subjects (Fig. 6).

In our study, the differences between the positions of COM predicted using LifeMOD and obtained using Motus were evaluated based on the relative difference. As shown in Table 1, the average difference for the X direction is between 3.25 and 9.36% with the maximum value not exceeding 20%. The Y direction values show even less difference. The average is between 3.03 and 4.65% with the maximum never exceeding 5.75%. Both of these results are below the tolerance threshold of 20% even though there is considerable more error in the X values than in the Y values. The main reason for the difference between the X and Y data is because the X values are at least four times smaller than the Y values. Since the values for the X data is much smaller than for the Y data, the denominator [see Eq. (3)] for the X data will be much smaller. Thus even if the absolute difference between the X data and the difference between the Y data is the same, the smaller denominator for the X data results in a larger relative position difference. Another possible reason for this is that when a person is walking with stilts he tends to kick his leg out, instead of bending at his knee and lifting the leg up, while bringing it forward to take a step. When modeling this activity, the model had a tendency to rotate on the planted foot because of the manner in which the stilts are modeled.

For the mean difference between the LifeMOD and Motus system, our 95% simultaneous confidence intervals indicated that the LifeMOD system consistently underestimated the values of the Z coordinate and overestimated the values of the Y coordinate (Table 2). The multivariate analysis results also showed no significant mean difference for the X coordinate, as indicated by a wide confidence interval with inclusion of zero. This could be due to a relatively larger variability in the X coordinate as compared to the Y and Z coordinates (Fig. 7 and Table 2).

The relative position difference of the center of mass in the Z direction is best understood by examining the graph (Fig. 6), since the average and maximum values can be misleading. The trends of the center of mass in the Z direction for all of the subjects and all of the tasks appear similar, with the majority of the relative position difference never exceeding 10%. However, there is a time when the relative position difference of the Z data is above 10%, as shown in Fig. 6; this happens between 0.4 and 0.7 seconds. There are no large discrepancies in the actual difference in the Z direction between the Motus and LifeMOD data (Fig. 7). Therefore, the large position difference is not due to a large difference between these two sets of data. Indeed, the large position difference is due to the placement of the point of origin. The origin of the axes is set at the near corner of the first force plate, but the person starts walking before that point and continues walking past that point. In Fig. 6, three subjects pass the origin at about 0.4–0.7 seconds. So even though the data may not differ by more than 2–3 cm during this short time frame, the Z values are so small, resulting in a small denominator, so that the slightest difference makes a large difference in the relative position difference calculation. Since this is considered to be mathematically trivial, the few outliers in this time interval can be excluded. After omitting out one or two of the outliers the average relative position difference of the data is consistently below 10%, which is well below the designated tolerance threshold (20%).

An estimation of ground reaction force is an invaluable indicator for the model projection, especially if that model is going to be used to determine forces or torques. For both right and left feet, the ground reaction forces obtained from computer simulations are similar in pattern to those measured via the force plates (Figs. 8 and 9). There are also some differences between them: the force responses predicted by the computer simulations lag behind those collected via the force plate. It appears that the contact duration of stilts on the ground in the computer simulation is smaller than that observed in the experiment. This could be due to the fact that when a subject is walking on stilts with a typically flat ground plate, he usually does not roll from heel to toe as in normal gait. However, in the computer simulation, the gait with stilts was modeled just like a normal gait, i.e. the foot rolls heel to toe. Another possible reason for the errors may be from the contact modeling in the simulations. To calculate the ground reaction force in LifeMOD, an ellipsoid-plane contact algorithm was employed which included stiffness/damping characteristics of the normal force and the frictional transverse force. For the rectangular stilt foot, two ellipsoids were attached to each stilt foot, one in the front half and one in the back half. This is a simplification of the actual physical contacts during stilt walking. Despite of all these differences between the computer models and actual subjects, the relative difference of the predicted ground reaction forces (Table 3) are within the desired threshold of 20%.

Examining Figs. 8 and 9, two horizontal forces apparently did not significantly change the trends of the force response patterns since vertical and resultant forces were identical. These similar patterns of force responses between the vertical and resultant forces are due to the corresponding horizontal force components being much smaller, compared to the vertical forces. Due to the small contributions of the horizontal forces, small discrepancies between two sets of data result in large percentage differences, as shown in Table 3 (note the actual differences between the Motus and LifeMOD data). Since the horizontal force component was found not to be predominant in our study, we analyzed the force differences only in the vertical direction (i.e. Y direction) and in magnitude (Tables 3 and 4). The good consistent direction of the percentage difference of the force in magnitude compared to that of the force in the vertical direction suggests that the corresponding difference for the horizontal force responses is small.

Our results showed that the average difference for the force magnitude is between 4.83 and 14.24% for all subjects and trials and that the average difference for the vertical force is between 5.11 and 14.04% (Table 3). For this part of the evaluation, only the force values, which were recorded with the floor plate of the stilt being in contact with the ground, were used. The reason for this is that when the floor plate (i.e. stilt foot) of the model is not on the ground the force readings are zero.

It can be seen from the comparison of the time-histories of the ground reaction forces predicted using LifeMOD™ with those measured experimentally (Figs. 8 and 9), that the predicted force is shifted forward by approximately 0.1 second relative to the measured force when the ground reaction force undergoes sudden changes, and the peak force values predicted via the model are greater than those measured experimentally. These differences are likely due to the inertial effects of motion upon soft tissues, which are distributed masses that are not rigidly connected to the body (wobbling masses). Liu and Nigg (2000)[24] indicated that these wobbling masses have remarkable influence on impact forces. The wobbling masses are not included in this current, preliminary model, and will be included in our refined model in the future. In the present study, a newly-developed computer simulation stilts model (i.e. using LifeMOD™) is compared with collected results from a motion system (i.e. PEAK Motus™) and force platform (i.e. Kistler™). The evaluation of the differences in model comparison was in term of two variables — center of mass and ground reaction force, which are the study focus and point of interest.

The PEAK Motus system uses its marker position data combined with force data from the Kistler force platform to generate joint dynamics through inverse dynamics algorithms. LifeMOD, on the other hand, utilizes both inverse and direct dynamic algorithms. Therefore, it has a refinement step that considers force effects on soft tissues and muscles. More importantly, LifeMOD has the ability to predict scenarios of extreme conditions. For example, LifeMOD can calculate joint forces and torques given a condition such as when the subject's walking speed is doubled. The PEAK Motus system can only calculate joint dynamics based on the monitoring of actual marker positions and tasks simulated in the laboratory.

In summary, a computer model of a person walking on stilts was evaluated within a 20% difference tolerance limit using three different subjects at a stilt height of 24 inches. Even though the model was evaluated there are some limitations to the model. The most significant limiting factor is that the ground reaction force for each foot could only be evaluated while the floor plate of the stilt foot was on the ground. This leads to questions about the usefulness and validity of any data collected for the swing leg, such as forces in the ankle when the foot is not on the ground. However, this difficulty can be overcome by comparing the testing data between human walking with and without stilts to see if there are differences between the two, and also by comparing the relative motion of the body segments and stilts to the motion recorded from the Peak system. The model will not produce any internal force in segments, such as force in the ankle, but it can aid in determining if there is an increase in specific localized joint stress during the use of stilts in the swing leg. Using this type of evaluated model of a person walking on stilts, researchers will be able to further examine whether stilt walking will result in an increase in joint loading of the legs or back. The model can also provide a useful tool to evaluate slips, falls, sudden stops/starts and tripping hazards associated with the use of stilts.

Acknowledgments

We would like to acknowledge the contributions of Amit Bhattacharya, Ronald Huston, and Ren Dong, who provided constructive suggestions in early stages of this study. We also thank Shawn McGuan, Matthew Ma, and Richard Current for their valuable involvements for various phases of the software design and development. We are grateful to Doug Cantis, Joyce Zwiener, Mahmood Ronaghi, and Tony McKenzie for their valuable assistance in data collection. We also want to express our gratitude to Tsui-Ying Kau for her contribution in carrying out the statistical analyses. Finally, special appreciation is extended to Paul Keane for his editorial review.

References

1. Anderson FC, Pandy MG, Three-dimensional computer simulation of gait, *ASME, Bioengineering Division (Publication) BED'* **42**:419–420, 1999.
2. Ashkenazy Y, Hausdorff JM, Ivanov PC, Eugene SH, A stochastic model of human gait dynamics, *Physica A: Statistical Mechanics and its Applications* **316**:662–670, 2002.
3. Bureau of Labor Statistics, Occupational injuries and illnesses: Counts, rates, and characteristics, 1995, US Government Printing Office, Washington, DC, 1998.
4. Bureau of Labor Statistics, Occupational injuries and illnesses: Counts, rates, and characteristics, 2000, US Department of Labor, Washington, DC, 2003.
5. Chiou S, Pan CS, Keane P, Traumatic injury among drywall installers — 1992 to 1995, *J Occup Environ Med* **42**:1101–1108, 2000.
6. Chiou S, Bhattacharya A, Succop PA, Evaluation of workers' perceived sense of slip and effect of prior knowledge of slipperiness during task performance on slippery surfaces, *AIHAJ* **61**:492–500, 2000.
7. Chiou S, Pan CS, Fosbroke DE, Identification of risk factors associated with traumatic injuries among drywall installers, in: Das B, Karwoski W (eds.), *Advances in Occupational Ergonomics and Safety*, IOS Press, Amsterdam, pp. 377–380, 1997.

8. Chung S, Hahn JK, in *Proc. Computer Animation Conference*, Animation of human walking in virtual environments, Geneva, Switzerland, pp. 4–15, 1999.
9. Clauser CE, McConville JT, Young JW, Weight, volume, and centre of mass segments of the human body, *AMRL Technical Report* (TR-69-70), Wright Patterson Air Force Base, Ohio, pp. 69–70, 1969.
10. Dempster WT, Space requirements of the seated operator (WADC-TR-55-159), Wright-Patterson Air Force Base, Ohio, Aerospace Medical Research Laboratory (NTIS No. AD-87892), 1955.
11. Enoka RM, Neuromechanical Basis of Kinesiology, Human Kinetics Books Champaign, IL, 1988.
12. Gerritsen KG, Van Den Bogert AJ, Nigg BM, Direct dynamics simulation of the impact phase in heel-toe running, *J Biomech* **4**:181–193, 1995.
13. Hay JG, Reid JG, Anatomy, Mechanics, and Human Motion, 2nd ed., Prentice-Hall Englewood Cliffs, NJ, 1988.
14. Huston RL, A summary of three-dimensional, gross-motion, crash-victim simulators, *Structure Mechanics Software Series*, University Press of Virginia, **1**:611–622, 1977.
15. Huston RL, Passerello CE, Harlow MW, UCIN vehicle occupant/crash-victim simulation model, in *Structural Mechanics Software Series*, Vol. 2, University Press of Virginia, pp. 131–150, 1978.
16. Huston RL, Huston JC, Harlow MW, A comprehensive, three-dimensional head-neck model for impact and high acceleration studies, *Aviat, Space Environ Med* **49**:205–210, 1978.
17. Huston RL, Crash victim simulation: Use of computer models, *Int J Ind Ergono* **1**:285–291, 1987.
18. Huston RL, Multibody dynamics — Modeling and analysis methods, *Appl Mech Rev* **44**:109–117, 1991.
19. Huston RL, Multibody Dynamics since 1990, *Appl Mech Rev* **49**:535–540, 1996.
20. International Brotherhood of Painters and Allied Trades, *Safety and Overview: Student's Guide* (Version 1), Washington, DC: International Brotherhood of Painters and Allied Trades, 1998.
21. Jansen MO, Van Den Bogert AJ, Riemersma DJ, Schamhardt HC, *In vivo* tendon forces in the forelimb of ponies at the walk, validated by ground reaction force measurements, *Acta Anat (Basel)* **146**:162–167, 1993.
22. Kisner SM, Fosbroke DE, Injury hazards in the construction industry, *J Occup Med* **36**:137–143, 1994.
23. Kreighbaum E, Barthels KM, *Biomechanics: A Qualitative Approach for Studying Human Movement*, 3rd ed., Macmillan Publishing Company, New York, 1990.
24. Liu W, Nigg BM, A mechanical model to determine the influence of masses and mass distribution on the impact force during running, *J Biomech* **33**:219–224, 2000.
25. Miller KM, Computer Simulation and Validation of Human Walking on Stilts, Unpublished Thesis, University of Cincinnati, 2003.
26. Mun JH, Freeman JS, Lim OK, Rim K, New three dimensional kinematic and dynamic model of lower limb, *ASME, Bioengineering Division (Publication) BED* **35**:233–234.
27. O'Riordain K, Thomas PM, Philips JP, Gilchrist MD, Reconstruction of real world head injury accidents resulting from falls using multibody dynamics, *Clin Biomech* **18**:590–600, 2003.
28. OSHA, Construction Accidents: The Workers' Compensation Database, 1985–1988, U.S. Department of Labor, April, 1992.
29. Pan CS, Chiou SS, Hsiao H, Becker P, Akladios M, Assessment of perceived traumatic injury hazards during drywall taping and sanding, *Int J Ind Ergon* **25**:621–631, 2000.

30. Pan CS, Chiou SS, Hsaio H, Wassell JT, Keane PR, Assessment of perceived traumatic injury hazards during drywall hanging, *Int J Ind Ergon* **25**:29–37, 1999.
31. Scheiner A, Ferencz DC, Chizeck HJ, Quantitative measurement of stability in human gait through computer simulation and Floquet analysis, *Annual International Conference of the IEEE Engineering in Medicine and Biology-Proceedings* **17**:1489–1490, 1995.
32. Schneider S, Susi P, Ergonomics and construction: A review of potential hazards in new building construction, *AIHAJ* **55**:635–649, 1994.
33. Stacoff A, Reinschmidt C, Nigg BM, Van Den Bogert AJ, Effects of shoe sole construction on skeletal motion during running, *Med Sci Sports Exerc* **33**:311–319, 2001.
34. Van Den Bogert AJ, Analysis and simulation of mechanical loads on the human musculoskeletal system: A methodological overview, *Exerc Sport Sci Rev* **22**:23–51, 1994.
35. Van Den Bogert AJ, Pavol MJ, Grabiner MD, Response time is more important than walking speed for the ability of older adults to avoid a fall after a trip, *J Biomech* **35**:199–205, 2002.
36. Winter DA, *Biomechanics and Motor Control of Human Movement*, 2nd ed., John Wiley & Sons, New York, 1990.
37. Wismans J, Hermans J, "MADYMO 3D Simulations of Hybrid III Dummy Sled Tests," SAE No. 880645, 1988.
38. WorkSafe, Why WorkSafe will not tolerate the use of stilts for construction work. Melbourne, Victorian Workcover Authority, WorkSafe Victoria, Australia, 2000.