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International Journal of Industrial Ergonomics 24 (1999) 157–171
International Journal of Industrial Ergonomics

Evaluation of a biomechanical simulation model for sagittal plane lifting

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Received 26 September 1997; received in revised form 15 December 1997; accepted 16 December 1997

Abstract

The kinetics and kinematics produced by a computerized dynamic biomechanical simulation model were examined and compared to those produced by actual human lifters. The purpose of the comparison was to demonstrate the accuracy of the simulation in predicting stresses imposed on the human body during the performance of a lifting task. The simulation model was shown to predict quite well under different task conditions (range of lift, weight of load, size of box, and gender of lifter.) Use of the simulation model is advocated for evaluations of lifting performed under a variety of conditions. Although highly correlated, the simulation tended to overestimate the kinetics and kinematics. The results provided in this study demonstrate that the simulation model can be an effective alternative for lifting task analyses. Through use of the simulation model, the tedious, time-consuming and costly data collection step required for lifting analyses can be eliminated so time and effort can be spent more productively on evaluation and design.

Relevance to industry

Manual material Handling, particularly lifting, is a major cause of work-related injury. Biomechanical evaluations are used to recognize dangerous work conditions and to implement safer lifting conditions. Although lifting is a dynamic activity and static evaluations underestimate stresses, static evaluations will continue to be used due to their simplicity until an efficient dynamic biomechanical method becomes available.

Keywords: Dynamic biomechanical models; Sagittal plane lifting; Kinematics; Kinetics

1. Introduction

The lifetime prevalence of low back pain (LBP) is estimated at approximately 75%. Although the precise cause of LBP is unknown in 80–90% of the cases, mechanical loading is speculated to be the most important factor (Pope and Novotny, 1993). Manual material handling (MMH), particularly lifting, poses a risk to many and is considered a major cause of work-related LBP and impairment (Waters et al., 1993). Approximately one-third

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0169-8141/99/$-see front matter
PII: S0169-8141(98)00011-0

of all jobs in industry involve MMH. The need for MMH in the future is not expected to decline, but instead increase due to the need for flexibility (Kuorinka et al., 1994). Recognizing that MMH, particularly lifting, is and will continue to be a problem for some time, the practical reasons for studying it are to recognize potentially dangerous work conditions, identify factors causing risks, and to recommend and implement conditions for safer lifting.

It is difficult to predict why certain workers are injured while others are not under similar biomechanical loading, but overexertion in the form of job demand exceeding individual worker capabilities is considered a primary cause (Garg, 1982). In order to investigate the biomechanical risks of a particular lifting task, it is necessary to collect displacement-time data at the workstation or a laboratory mockup. This can be difficult and costly to perform. Through the use of biomechanical simulations, these steps may be greatly reduced and eventually eliminated.

The primary goal of the research presented here was to examine the performance of a computerized dynamic biomechanical simulation model of lifting to evaluate its accuracy in simulating lifting motions of humans. Simulation predictions of select kinematic and kinetic variables were compared to those of actual human subjects who performed lifts in experimental sessions. Lee (1988), Hsiang (1992) and Lin (1995) evaluated angular displacement predictions from prior versions of the simulation model. In the interest of reducing work-related injuries, ergonomists evaluate stresses on the body during task performance. Thus, a demonstration of the accuracy of the simulation model in predicting forces and moments was necessary for the model to be considered of practical use.

1.1. Biomechanical models

Biomechanical models are used to quantify the effects of a physical activity, such as lifting, by estimating the forces and moments acting on the musculoskeletal system as an activity is performed. The main purpose of biomechanical modeling is to reduce work-related injuries of the musculoskeletal system (Ayoub, 1992). Biomechanical models may be static or dynamic. The stresses in static models are a function of the load held in the hands, body segment weights and the posture of the body, while the stresses in dynamic models include inertial effects in addition to the static stresses.

Static models have been studied and applied more extensively because of several advantages they have over dynamic models including: static models require relatively simple logic and task data; and more practical use has been made of static models because of the availability of population muscle strength data (Garg et al., 1982). The main disadvantage of static models is that since inertial effects are ignored, stresses on the musculoskeletal system can be vastly underestimated (Ayoub, 1993). Dynamic models more accurately represent the stresses produced during dynamic activities, however, the major limitations in using dynamic models, such as those developed by Ayoub and El-Bassoussi (1978), Freivalds et al. (1984), Chen and Ayoub (1988) and Morris and Trimble (1992), are the difficulties in collecting and processing data and comparing predicted stresses with safe, allowable stresses (Garg et al., 1982).

Static model predictions have been compared to dynamic model predictions by several investigators who concluded that static models can grossly underestimate the stresses of a dynamic activity. Freivalds et al. (1984) reported dynamic effects increased the static load by as much as 40% of its weight. Garg et al. (1982) found that peak dynamic joint moments were two to three times greater than peak static moments. Leskinen (1985) compared static and dynamic estimates of peak L5/S1 compression and found the increase in spinal compression due to dynamic effects ranged from 33% to 60% depending on the lifting technique used by the subjects. McGill and Norman (1985) reported that peak dynamic L4/L5 moments were 19% higher on average up to a maximum of 52% larger, indicating that in many tasks, static analyses may severely underestimate the demands of a dynamic lift. These studies suggest that static models are inadequate when analyzing dynamic tasks, particularly since the effect of introducing inertial effects into the static model may vary with posture, lifting technique and the point in time at which the lift is examined. A need exists for a less laborious dynamic biomechanical method of evaluation such as simulation modeling.

1.2. Computer simulation modeling

Dynamic biomechanical models using the inverse dynamics approach to estimate kinetics and kinematics require inputs of the body's displacement–time profiles from film data. Since collection of displacement–time information requires a considerable amount of effort, an alternative is to use a biomechanical computer simulation model to predict human motion.

According to Vaughan and Sussman (1993), computer simulation models have several advantages over real-world experimentation. Simulation models are safe to experiment with since real human subjects do not perform any of the hazardous tasks; tasks which pose a risk can be identified and layouts can be reconfigured for further evaluation. Simulations save time because many different conditions can be run relatively quickly, and can save money because it is not necessary to physically build workstations and pay subjects. Simulations can also be used in an exploratory manner to investigate the influence of a single variable or combination of variables, and to predict optimal motion patterns (Miller, 1976).

A variety of human postures and motions have been modeled in the sports, ergonomics and medical literature. A simulation model to predict dynamic lifting motions in the form of angular displacements at the major joints of the body was developed for sagittal lifting activities (Ayoub, 1992, 1989; Hsiang, 1992; Lee, 1988). The simulation model was formulated as an optimization problem which assumed that the body performed a lifting motion, particularly in high exertion tasks, according to some internal criterion such as the minimization of muscular effort.

Lee (1988) developed a two-step solution approach using the Trajectory Approximation in State Spaces and Dynamic Programming search methods. In the first step, an initial solution was generated to satisfy the static constraints of the model including the range of joint motion, the geometry of the workstation and the voluntary muscle strengths at the joints. This initial solution was the set of five joint trajectories, each predicted at twenty discrete points in time. In the second step, dynamic programming was used iteratively to refine the paths while minimizing the objective function. The simulated angular displacements of the joints were compared to four male subjects and Lee reported significant agreement between the simulation and experimental data. However, when the joint moments were compared in a small subset of the data, significant discrepancies in the magnitudes of the moments were observed. The simulation loaded the stronger lower extremities more heavily and extended the hip joint earlier and faster than actual humans did in the experimental data.

Further work on the model led Hsiang (1992) to use eighth-order polynomials to describe the continuous angular displacement trajectories of the body joints. The optimization problem became one of finding coefficients of the angular displacement polynomials which minimized the objective function. Hsiang (1992) validated the simulation by comparing the angular displacements of the model predictions with those of human subjects. Statistical analyses indicated some discrepancies in the simulation model predictions indicating that the motions of the shoulder and ankle joints were predicted less well than the other joints. Graphically, the envelope patterns of the simulations generally overlapped the actual data (Ayoub and Hsiang, 1992). However, differences occurred in the load trajectories, e.g., actual subjects tended to move the load closer to their bodies at the start of the lift than predicted by the model. Phase shifts in time were also observed in the predicted motion with respect to the actual motion.

Further refinement of the dynamic simulation model of lifting was performed by Lin (1995) and is reported in Ayoub et al. (1998). An objective function was selected based on the hypothesis that the body will minimize its efforts, subject to a set of constraints related to the capability of the body, the geometric layout of the task, and the kinematics and kinetics of the motion. It was assumed that the lifting trajectories, i.e., the angular displacement-time profiles, of the five joints are selected such that they minimize the following objective function:

$$\int_{t=0}^{T} \sum_{j=1}^{5} \left(\frac{Mj(t)}{Sj(t)}\right)^2 \mathrm{d}t$$

where:

$M_j(t)$ are the joint moments at time t,
$S_j(t)$ are the moment strengths of the joints at time t (Stobbe, 1982),
j is the joint identification number,
T is the total time of the lift.

Inputs to the simulation model include: initial and final postural configurations of the body as defined by the angular position of the joints, the total time to perform the lift, the height and weight of the subject, the weight of the load and the geometric dimensions of the container. The problem for the optimization process becomes one of distributing the workload to the five joints according to the resources each joint has available. The GRG2 algorithm, a finite improvement algorithm used to solve nonlinear optimization problems with bounded constraints (Lasdon and Waren, 1979), was used for the optimization process. Based on the recognition that each joint has a characteristic angular displacement–time profile, the algorithm was used to determine the normalized profile for each joint (Lin, 1995).

2. Methods

The performance of the lifting simulation model was compared to data collected from human subjects. The data produced by the simulation model will be referred to as "predicted" and the data from the human subject will be referred to as "actual". A total of 720 lifts were analyzed, 360 "actual" human lifts and 360 "predicted" simulated lifts, each modeled under conditions representing the actual lifts. The data were processed through the following model.

The biomechanical model was a two-dimensional, whole-body representation describing the motion of a symmetrical lift performed in the sagittal plane. The body was represented by five rigid links (lower leg, upper leg, trunk, upper arm and forearm with hand) with pin-centered joints at the ankle, knee, hip, shoulder and elbow. The configuration of the body was described in terms of the rotational angles of each joint and the segment link lengths. Anthropometric data including segment lengths, weights, centers of mass and inertial properties were from Winter (1990). For a detailed description of the model, see Ayoub et al. (1998). Kinematic variables (linear and angular displacements, velocities and accelerations) were determined using the finite difference method equations:

$$x_i = \frac{x_{(i+1)} - x_{(i-1)}}{2\Delta t}$$

$$\ddot{x}_i = \frac{x_{(i+1)} - 2x_i + x_{(i-1)}}{\Delta t^2}$$

where

x_i refers to linear or angular displacement at time i,
x_i is the linear or angular velocity,
x_i is the linear or angular acceleration.

The kinetic variables (reactive forces and moments) were calculated from the following equations of motion:

$$R_j = m_j a_j + R_{(j-1)} + m_j g$$

$$M_j = I_j \ddot{\theta}_j + M_{(j-1)} + r_{(j-1,\mathrm{cm}j)} \times R_{(j-1)} - r_{(j,\mathrm{cm}j)} \times R_j$$

where

R_j is the force vector acting at joint j,
M_j is the moment vector acting about joint j,
$r_{(j,\mathrm{cm}j)}$ is the linear position vector from the link center of mass to the joint,
a_j is the linear acceleration vector of the center of mass of link j,
$\ddot{\theta}_j$ is the angular acceleration vector,
g is the gravity vector,
m_j is the mass of link j, and
I_j is the moment of inertia of link j.

2.1. Actual data

Ten subjects, five male and five female, were recruited from the university student population of

Table 1
Subject characteristics

Subject	Sex	Age	Height (m)	Weight (kg)	Shoulder Height (m)	Knuckle Height (m)	MAWL (kg)
1	F	22	1.71	59.5	1.25	0.62	11.1
2	F	28	1.70	69.1	1.28	0.63	11.6
3	F	25	1.62	50.9	1.18	0.53	9.3
4	M	27	1.75	69.5	1.29	0.63	21.1
5	F	21	1.73	70.4	1.27	0.61	11.1
6	M	19	1.83	80.9	1.35	0.64	27.0
7	M	23	1.68	79.1	1.26	0.63	22.0
8	F	29	1.57	80.9	1.11	0.51	9.5
9	M	27	1.76	65.0	1.29	0.62	20.5
10	M	19	1.93	90.9	1.46	0.69	21.5

Texas Tech University. Table 1 contains subject characteristic data. Each subject participated in a psychophysical experimental session to determine their maximum acceptable weight of lift (MAWL), six practice/training sessions and one data collection session in which data from twelve task conditions were collected (2 ranges of lift × 2 box sizes × 3 weights).

The psychophysical experimental session was used to determine the maximum amount of weight each subject was capable of lifting in the most physically demanding of the twelve task conditions, e.g., lifting the larger box from floor-to-shoulder height. For one hour, each subject lifted the box and adjusted the load once every five minutes to find their one-time MAWL. The most physically demanding of the twelve task conditions was used as a baseline to set standard weights for all subjects and range of lift by box size combinations. The three standard weights, identified as light, medium and heavy, for the remaining experimentation were set such that the heaviest weight required for the males to lift was less than or equal to the minimum MAWL selected by the five males, and likewise for the five females. The medium and light weights were 2.3 and 4.5 kg less than the heavy weight, respectively.

The practice/training sessions were used to familiarize the subjects with the twelve lifting task conditions and the procedures used during the data collection session. Each subject performed twelve hours of practice lifting, split into six two-hour sessions, such that each subject received one hour of practice in each condition. Each subject performed 360 practice lifts, or 30 trials per task condition.

The experimental data collection session was used to collect the "actual" data for comparison with the simulation model. For each subject, three replications of each of the twelve task conditions (2 ranges of lift × 2 box sizes × 3 weights), a total of 36 trials, were collected. The trials were completely randomized within each subject using the PROC PLAN routine (SAS/STAT, 1990). Each trial was set up in the following manner. The shelf height was adjusted to the appropriate height (knuckle or shoulder height) based on the anthropometry of the subject. Each subject selected a foot position which permitted them to lift the box onto the shelf without moving their feet throughout the lift; the initial position of the box was controlled to this location. The box was loaded with the appropriate light, medium or heavy weight. Reflective joint center markers were placed on each subject at the knuckle, wrist, elbow, shoulder, hip, knee and ankle. The subject was instructed to position his feet as marked, then he reached down and grasped the handles of the box and temporarily held his initial posture. The initial posture was self-selected by each subject as the start of a freestyle lift. The data collection process was triggered to begin, then the subject was verbally instructed to lift the box off the floor and put it onto the shelf. Each trial was completed in a similar manner.

2.2. Experimental equipment

Two plywood boxes with metal handles, each weighing 4.5 kg, were used in the experiment. The small box measured 30.5 cm × 30.5 cm × 30.5 cm and the large box measured 45.7 cm × 30.5 cm × 30.5 cm, where the 45.7 cm side was the length of the box in the sagittal plane.

The height of the shelf, which was set according to the knuckle or shoulder height of each subject, was set on a lifting/lowering machine which had an adjustable shelf. In order to identify the beginning and end of the lift, the box was wired with a circuit such that when the box was lifted off the floor, the circuit was broken and the time of liftoff marked on the data. When the box was placed on the shelf, the circuit was completed and the end of the lift was marked on the data.

A Motion Analysis System was used to record each of the actual lifting trials. Hardware included a Motion Analysis VP-320 video processor, three 180 Hz video cameras and a Sun Sparc Station. Software included the Motion Analysis Video-Analog Data Collection Program and the Expert Vision Data Analysis Program. The data were collected at a rate of 100 Hz.

The 360 "actual" human lifts captured with the motion analysis system were tracked and digitized to obtain the x- and y- spatial coordinates of each reflective marker on the subjects. Data collected during this experiment are referred to as the "actual" data. For each "actual" lift, corresponding inputs (initial and final postural configurations of the body, total time to perform the lift, height and weight of the subject, weight of the load, and geometric dimensions of the container) were supplied to the simulation model to produce a matching "predicted" lift.

2.3. Response variables

To demonstrate that the computerized dynamic biomechanical simulation model is a useful tool for the design of lifting tasks, it must be shown to accurately predict parameters commonly used in ergonomic evaluations. Select response variables including the distance traveled by the load, peak static and dynamic compressive forces, objective function value, the total net muscular work, and peak moment at each joint were estimated for both "actual" and "predicted" lifts using the biomechanical model presented above. The objective of these analyses was to determine whether motion predicted by simulation model differed from actual human motion. Given that differences existed, the secondary objective was to determine their magnitudes and the effects of the independent variables. The following response variables were evaluated:

1. T = travel distance of load (m) = $\int_{t=0}^{T}(\sqrt{(x_t - x_{t-1})^2 + (y_t - y_{t-1})^2})\,dt$,
2. F_S = peak static L5/S1 compressive force (N) = Max $F_S(t)$,
3. F_D = peak dynamic L5/S1 compressive force (N) = Max $F_D(t)$,
4. OFV = objective function value = $\int_{t=0}^{T}\sum_{j=1}^{5}(M_j(t)/S_j(t))^2\,dt$,
5. W = total net muscular work (J) = $\int_{t=0}^{T}\sum_{j=1}^{5}|M_j(t)\theta_j(t)|\,dt$,
6. M_j = peak moment at each joint (N m) = Max $M_j(t)$

The distance traveled by the load was used to evaluate the motion of the load. By examining this variable, different strategies used to complete a lift may become apparent. For example, the distance traveled by the load may be minimized in an efficient lift (Fogleman, 1993).

Compression on the L5/S1 segment was evaluated because it is the site which has the potential to experience the largest moment in lifting and is one of the most vulnerable areas to force-induced injury (Chaffin and Park, 1973; Waters et al., 1993). The peak static compressive force was evaluated because ergonomists commonly judge the risk associated with a lift by comparing it with the 3.4 kN criterion recommended by NIOSH (Waters et al., 1993). While static estimates are the current criterion of choice, dynamic values more accurately estimate the stresses occurring in a dynamic lift, thus the peak dynamic compressive force was evaluated. Peak dynamic compressive forces may be 33–60% (Leskinen, 1985), up to as much as 300% (Garg et al., 1982), larger than static forces; however, a standard criterion value indicative of an acceptable level of risk has yet to be adopted.

The objective function value and the total net muscular work may be considered measures of the effort of the whole body over the entire duration of the lift. Although the objective function has been used to produce lifting simulation results, the values have not been documented in prior studies (Lee, 1988; Hsiang, 1992; Lin, 1995). The total net muscular work was used to assess the loading of all the body joints by Gagnon and Smyth (1991) who were concerned that evaluation of a single joint, namely the L5/S1 joint, may fail to identify dangerous loads at sites in the body other than the one under evaluation.

Where differences between the simulation and actual data exist in the variables describing whole body motion, it was necessary to examine each joint in turn to identify the discrepancies. The peak moments of each joint were used to investigate the contribution of each joint to the lift. The objective function of the simulation model utilized the ratio of dynamic moments to the static strength. There is concern that the static strength overestimates the strength available during a dynamic activity and may affect the results of the simulation, thus an analysis of the peak moments was performed.

2.4. Methods of analysis

Upon examining scatter plots of the predicted values vs actual values and observing linear trends, simple linear regression models of the form ($Actual\ Value = Intercept + Slope \times Prediction$) were produced for each response variable. Slopes with values close to unity were indicative of signicant correspondence between the predicted and actual values; deviations from unity were indicative of prediction errors. The coefficient of determination (R^2) values were used as an indication of the fit of the simple linear regression. Although the predicted and actual variables may be highly correlated, the magnitudes of the variables may differ. To test for differences between the predicted and actual values of the response variable, paired *t*-tests were performed on the prediction errors of the response variables. Mean errors significantly different from zero indicate that the simulation model either over- or underpredicted actual values.

Differences were further examined through ANOVA tests which considered main effects and appropriate interactions between gender (male, female), range of lift (floor-to-shoulder, floor-to-knuckle), size of box (30.5 cm, 45.7 cm in the sagittal plane), weight of load nested under gender (male 11.5, 16.0, 20.5 kg; female 4.5, 6.8, 9.1 kg), and subject nested under gender which served as a blocking variable. Strategies adopted by lifters may vary with changes in task variables such as the weight of the load, the size of the box or the range of lift. These were investigated to determine how they affected model predictions, e.g., whether the model was capable of adapting to different conditions as humans adapt. For example, consider the effect of changing the weight of the load. It was assumed that the simulation model distributed the load to all the joints based on their relative capacity. While an actual subject may distribute a heavy load in this manner to reduce the risk of overstressing the individual joints, the subject may not necessarily need to distribute a light load. In such a case, the simulated lifts will agree with the actual lifts when the load is heavy, but disagree when the load is light. By examining differences in simulation predictions and human data, specific areas for improvement could be identified for the purposes of refining the simulation or recognizing its limitations.

3. Results

For each response variable, Table 2 contains the slope and coefficient of determination (R^2) for the simple linear regression equations, the mean and standard deviation of both the predicted and actual lifts, and the mean errors and their significance. In general, the actual lifts were predicted quite well by the simulation model as indicated by the slopes and R^2 values in Table 2.

The distance traveled by the load, was predicted well by the simulation model although the simulation model increased the travel distance by 0.137 m. The weight of load and range of lift significantly affected the errors; predicted travel distances were larger in floor-to-shoulder lifts and errors increased with increased weights of lift.

Table 2
Simple statistics for response variables

Response variable	Slope	R^2	"Predicted" Mean (SD)	"Actual" Mean (SD)	Mean Prediction Error
T (m)	0.775	0.760	1.49 (0.345)	1.35 (0.299)	0.137***
F_S (N)	0.960	0.984	4772.1 (1020.4)	4760.8 (987.6)	11.3
F_D (N)	0.781	0.798	5987.9 (1238.4)	5691.1 (1082.1)	296.9***
OFV	0.738	0.859	1.332 (0.744)	1.015 (0.593)	0.317 ***
W (J)	0.890	0.818	268.5 (93.52)	250.0 (92.02)	18.5 ***
M_E (N m)	0.792	0.905	33.37 (15.46)	27.69 (12.88)	5.67 ***
M_S (N m)	0.852	0.928	53.33 (24.56)	44.90 (21.74)	8.43 ***
M_H (N m)	0.769	0.849	144.5 (34.44)	136.0 (28.75)	8.58 ***
M_K (N m)	0.642	0.806	119.9 (46.42)	101.6 (33.19)	18.3 ***
M_A (N m)	0.384	0.469	146.1 (58.40)	112.2 (32.74)	33.83 ***

***$p < 0.001$.

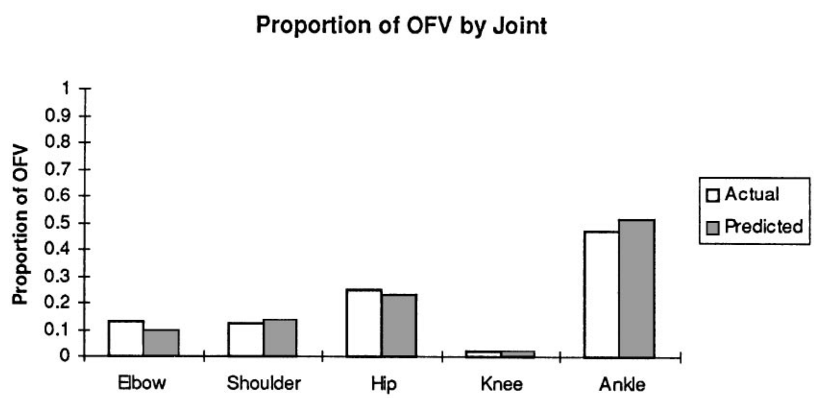

Fig. 1. Proportion each joint contributes to the objective function value.

The peak static compressive force was predicted well by the simulation as demonstrated by the high correlation and insignificant mean error (11.3 N). Although correlated, the peak dynamic L5/S1 compressive force was significantly overpredicted by the simulation model. Fig. 2 contains the compressive force trajectory throughout six lifts, three actual and three predicted, for a female lifting the medium weight in a small box over the floor-to-shoulder range. The mean error between the actual and predicted peaks was 296.9 N (5.2%). The errors were significantly larger in floor-to-knuckle range lifts than in floor-to-shoulder lifts, increased with increased weight of lift, and were larger when the small box was lifted.

With respect to the objective function value, on average the simulation's OFV was larger by 0.317. Fig. 1 contains the contribution of each joint to the value of the objective function for both predicted and actual lifts. The main effects range of lift, weight of the load and size of the box and interactions *gender* × *range* and *gender* × *range* × *box* were significant. Prediction errors were larger in floor-to-knuckle range lifts than in floor-to-shoulder, increased with increased weight of the load and were larger when the large box was lifted. The fact that the errors for the female floor-to-knuckle lifts were smaller than the floor-to-shoulder, whereas the opposite was true for the male lifting ranges, was responsible for the *gender* × *range* interaction.

The significant three-way interaction, *gender* × *range* × *box* was the result of the errors in the male floor-to-knuckle lifts which were at times larger for the small box than the large box; whereas they were smaller with the small box in the remaining conditions.

The model predictions of total net muscular work were highly correlated with their corresponding actual values although overpredicted by 18.5 J (7.4%). The effects gender, range of lift and *gender* × *range* were significant. The prediction errors were larger for males than for females, and larger in floor-to-knuckle range lifts than in floor-to-shoulder. The relatively larger errors of the male floor-to-knuckle lifts, as compared to the females in some conditions, were responsible for the *gender* × *range* interaction.

The peak moment at each joint and the relative occurrence time of the peak moment were determined for both the predicted and actual lifts. Figs. 3–7 contain three replications each of actual and predicted moment trajectories of the elbow, shoulder, hip, knee and ankle for a female lifting the medium weight, small box over the floor-to-shoulder range. The simple linear regression for the peak moments at the joints provided in Table 2, with slopes approaching unity and large R^2 values indicate that the model predictions corresponded well with the actual peak joint moments although the prediction errors were significant at each joint.

The simulation overpredicted the peak elbow moment by 5.67 N m and errors were larger for males than for females, larger in floor-to-knuckle range lifts than in floor-to-shoulder, and increased with increased load weight. Two range of lift interactions were also present. The relatively larger errors in the male floor-to-knuckle lifts, especially when lifting a heavy weight, were responsible for the *gender* × *range* and *range* × *weight* interactions. With respect to the time of occurrence of the peak elbow moment, the actual peaked near the end of the lift at 94% of the total lifting time on average and the predicted peaked 9% earlier, at 85% of the total time.

The mean peak shoulder moments were overpredicted by 8.43 N m. The errors were significantly larger for males than for females and increased with increased weight of load lifted. One interaction, *range* × *weight*, was significant due to the larger errors appearing in heavy floor-to-knuckle lifts as compared to the remaining combinations of range and weight. With respect to the time of peak occurrence, the actual peaked near the end of the lift on average at 96% of the total lifting time and the predicted peaked 5% earlier, at 91% of the total time.

The mean peak hip moments were overpredicted by 8.58 N m. The errors were larger in floor-to-knuckle range lifts than in floor-to-shoulder lifts

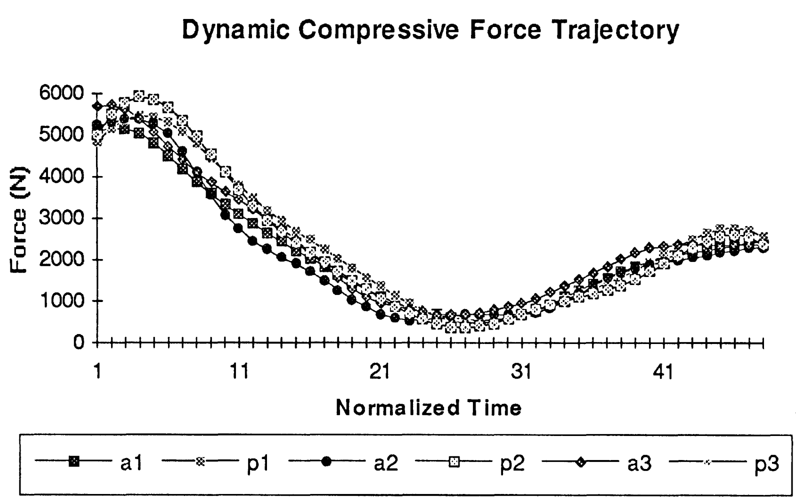

Fig. 2. Trajectory of dynamic compressive forces.

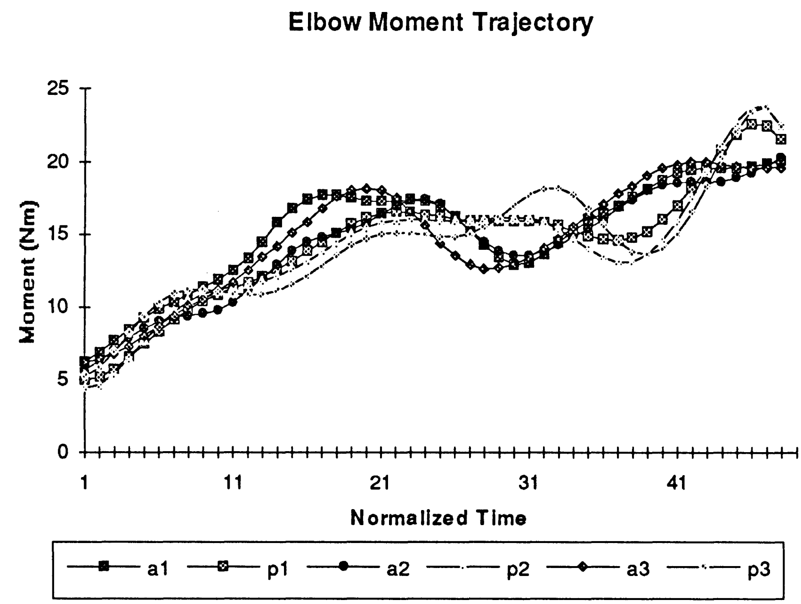

Fig. 3. Trajectory of elbow moments.

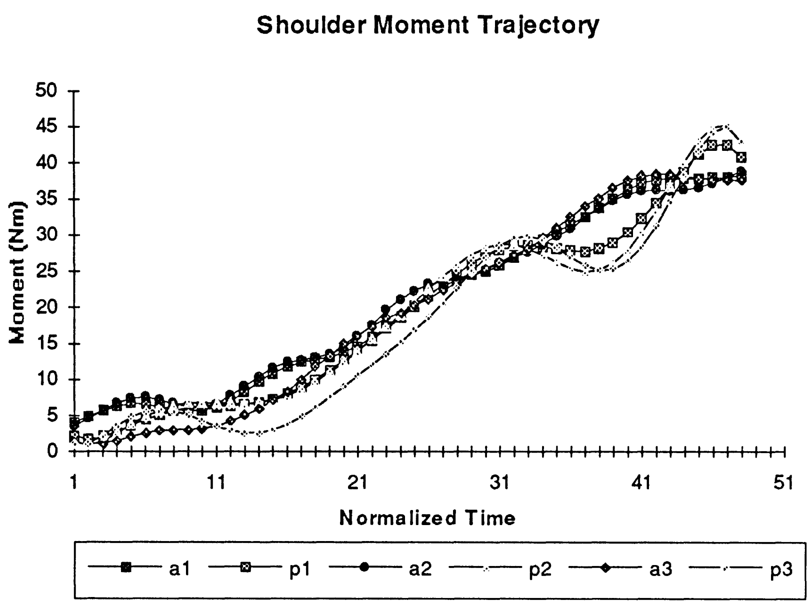

Fig. 4. Trajectory of shoulder moments.

and increased with increased weight of load. Two interactions were significant: *gender × box* and *gender × range × box*. The relatively large over-prediction errors when males lifted the large box as compared to the small box, particularly in floor-to-knuckle lifts, vs the fact that on average the errors for females were smaller since the model tended to underpredict rather than overpredict hip

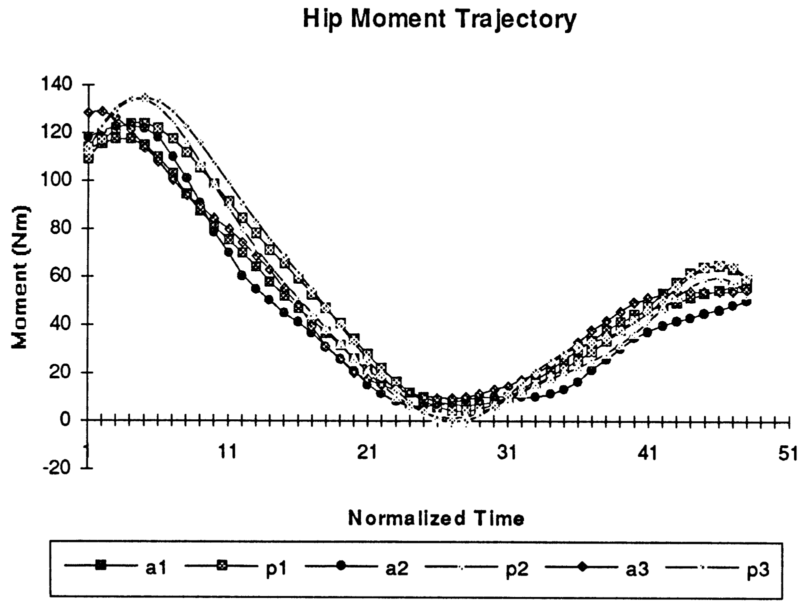

Fig. 5. Trajectory of hip moments.

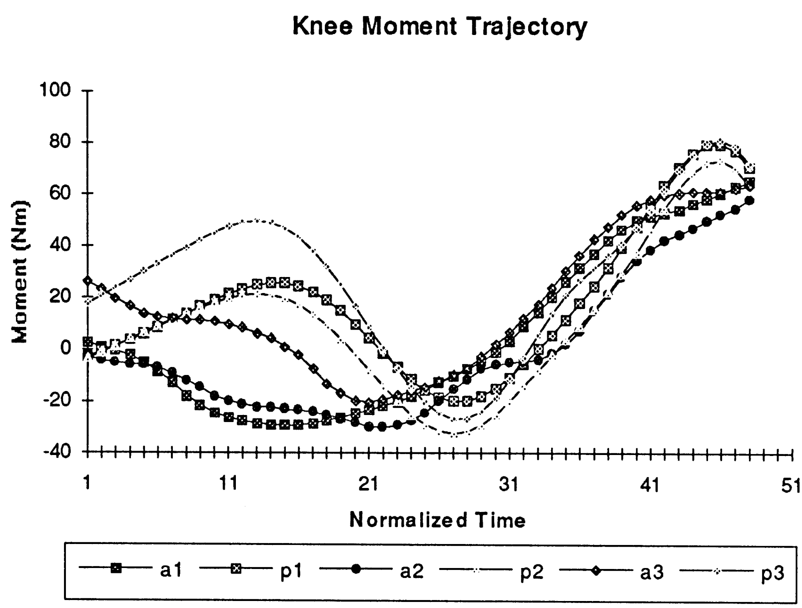

Fig. 6. Trajectory of knee moments.

moments when females lifted the large box, were responsible for these interactions. With respect to the time of peak occurrence, the actual peaked on average at 36% of the total lifting time and the predicted peaked 2% later, at 38% of the total time on average.

The peak knee moments, though less well predicted than the elbow, shoulder and hip peak

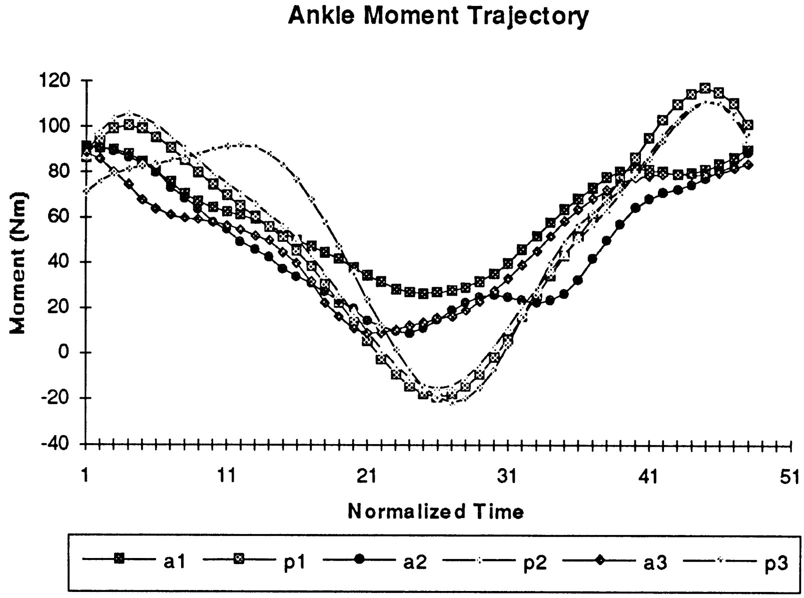

Fig. 7. Trajectory of ankle moments.

moments were still highly correlated with the actual knee moments and like the previous joints were overpredicted by the simulation (18.37 N m). Prediction errors were larger for males than for females; larger in floor-to-knuckle range lifts than in floor-to-shoulder lifts and a *gender* × *range* interaction occurred since females had approximately equal errors in both ranges of lift; whereas males had larger errors in the floor-to-knuckle range than in the floor-to-shoulder range. With respect to the time of peak occurrence, the actual peaked on average at 94% of the total lifting time and the predicted peaked 4% earlier, at 90% of the total time.

Peak moments at the ankle joint were more poorly predicted than at the other four joints; the simulation model overpredicted by an average 33.83 N m. The prediction errors were similar to the knee errors: larger for males than for females and larger in floor-to-knuckle range lifts than in the floor-to-shoulder lifts, and significant *gender* × *range* interaction since both ranges of lift had approximately equal errors in females; whereas, the males had larger errors in the floor-to-knuckle range than in the floor-to-shoulder range. When evaluating the timing of the peak, the actual peaked on average at 84% of the total lifting time and the predicted peaked 8% earlier, at 76% of the total time.

4. Discussion

In order for the computerized dynamic biomechanical simulation model to be a useful tool for the design of lifting tasks, it needed to be shown to accurately predict response variables commonly used in ergonomic evaluations. The response variables evaluated in this study were selected for their importance to ergonomists designing and evaluating lifting tasks and their potential to identify problems in the simulation model.

The predicted distance traveled by the load was correlated with the actual distance. The errors in the predicted travel distance were larger in floor-to-shoulder lifts than in floor-to-knuckle lifts, and the errors increased with increased weights of lift. The larger errors in floor-to-shoulder range lifts could be explained by the cumulative effect in the deviation of the path accruing over a longer distance. However, the increase in errors with increased weight of load was neither expected nor desirable since it would require the lifter to exert more effort to move the load over a longer distance and this effect intensifies as the weight increases. The simulation model increased the distance traveled by the load by an average of 0.137 m, a result contrary to expectations since the distance traveled by the load was hypothesized to be minimized in efficient lifts. The simulation model was assumed to minimize effort and/or increase efficiency, thus it was expected to decrease the travel distance. Instead the simulation increased the path of the load by 10.4% on average. The optimization routine may be responsible for this contradiction It was assumed that the desired global optimum was reached in each run; however, local minima may have been used instead.

The peak static L5/S1 compressive force predicted by the simulation model corresponded very well with the actual peak force values and the mean prediction error was insignificant. The predicted peak dynamic L5/S1 compressive force also corresponded quite well with the actual peak values, although it was overpredicted. With respect to the time of peak occurrence, the actual dynamic compressive force peaked on average at 17% of the total lifting time and the predicted peaked an insignificant 3% later, at 20% of the total time. As compared with the results of other researchers, the average actual (5691.1 N) and predicted (5987.9 N) peak dynamic L5/S1 compressive forces estimated in this study were within the ranges reported in studies performed under similar lifting conditions. For example, Troup et al. (1983) reported values of 5765 and 6039 N when subjects used a back lift and leg lift, respectively, to lift a 15 kg load; and under like conditions, Leskinen (1985) reported 6365 and 5866 N, respectively.

The predicted objective function values of the simulation model were highly correlated with the actual values of the objective function, although the simulation overpredicted. This result casts some doubt on the effectiveness of the outcome of the optimization. The simulation is set up to minimize the objective function, subject to a set of constraints. However, the fact that the actual human lifters selected motion patterns with smaller objective function values than the simulation provides more evidence that the optimization routine did not reach global minima as hypothesized above. Although the objective function has been used to produce lifting simulation results (Lee, 1988; Hsiang, 1992; Lin, 1995), the values had not been documented in prior studies.

The model predictions of total net muscular work were highly correlated with their corresponding actual values but overpredicted by 7.4% increase. The relative amounts of work performed by the joints were all highly correlated with their corresponding actual values. In general, the simulation model predicts the total net muscular work quite well in a manner consistent with the actual work performed.

The predicted peak moments at the elbow, shoulder, hip and knee were highly correlated with the actual peak moments as indicated by the closeness of their simple linear regression slopes to unity and large R^2 values. The peak ankle moments were less correlated than the other joints. Although well correlated, the simulation model tended to overpredict the peak moments by 5.67, 8.43, 8.58, 18.4 and 33.8 N m at the elbow, shoulder, hip, knee and ankle joints, respectively. An examination of the errors between the actual and predicted moments showed their magnitudes were affected most strongly by the gender of the lifter, the range through which the box was lifted and the weight of the load. The predictions for females were more accurate than for males. The peak moments when lifting over the floor-to-shoulder range were better predicted than the floor-to-knuckle range. With respect to the time of peak moment occurrence, the actual peak at the hip occurred at 35% of the total time of the lift and the predicted peaked 2% later. In all other joints the actual moments peaked on average between 85–95% of the total time and the predicted moments peaked 4–8% earlier than the actual.

To verify that the experimental and simulated data were within recognized ranges, some data was taken from the literature for comparison with the hip and knee moments. In this study, the predicted peak hip moments, for one hip, ranged from 81.9 to 257.7 N m, with an average value of 144.5 N m, while the actual ranged from 77.4 to 220.0 N m with an average of 136.0 N m. Garg et al. (1982)

reported peak hip moments of 452 N m when subjects lifted a 29.4 kg load and 475 N m for a 33.1 kg load. Troup et al. (1983) reported values of 346 and 263 N m when subjects used a back lift and leg lift, respectively, to lift a 15 kg load. Schipplein et al. (1990) reported peak hip moments for one hip of 119 and 185 N m for loads of 5.1 and 25.5 kg, respectively. In comparison, the peak hip moments found in this study fell within the range established by these researchers. In the current study, the predicted peak knee moments for one knee ranged from 46.7 to 346.7 N m, with an average value of 119.9 N m; while the actual ranged from 32.3 to 205.9 N m with an average of 101.6 N m. Garg et al. (1982) reported peak knee moments of 260 N m when subjects lifted a 29.4 kg load and 287 N m for a 33.1 kg. Schipplein et al. (1990) reported peak knee moments for one knee of 13 and 53 N m for loads of 5.1 and 25.5 kg respectively. In comparison, there was a small subset of predicted peak knee moments that were larger than the values established by other researchers; however, the actual peak knee moments and the majority of the predicted peak knee moments fall within the established range.

5. Conclusions

In the analysis which compared simulation predictions to estimates for actual human lifts, it was determined that the simulation tended to overpredict the values on all parameters including the distance traveled by the load, peak static and dynamic L5/S1 compressive forces, objective function values, net muscular work and peak moments at the joints. The groundwork for the simulation model was based on the assumption that the objective function would lead to efficient lifts which would minimize these effects, the stresses experienced by the lifter. The fact that the simulation model overpredicted the response variables was contrary to this hypothesis since the predictions, in effect, increase the stresses experienced by the simulated lifters. While this result appears undesirable in that the model does not appear to predict "optimal" lifting motions, it was not totally undesirable since the predictions could be considered to be conservative estimates with a built in margin of safety. Otherwise, if it is desirable to decrease this error and bring the predictions closer to actual values, the high linear correlations between the actual and predicted data suggest that the predictions can be corrected through use of simple linear regression equations.

Acknowledgements

This study was conducted at the Texas Tech University and was supported by a fund from NIOSH (No. RO1-OH02434).

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