FROM BIOMECHANICAL MODELING TO BIOMECHANICAL SIMULATION

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INTRODUCTION

Biomechanics of human movement has been defined by many investigators over the years. In 1966, Drillis and Contini defined biomechanics as the science which investigates the effects of the internal and external forces on bodies whether in movement or at rest. Since then, many other researchers reported similar definitions such as Winters (1979), Frankel and Nordin (1980). Biomechanics uses laws of physics and engineering principles to describe body movements and the forces acting on it. Biomechanics is a multidisciplinary area which requires the combined knowledge from the biological, physical and behavioral sciences. There are many contributing disciplines that are called upon biomechanical analysis, these are: (1) engineering mechanics, (2) engineering anthropometry, (3) kinesology, (4) anatomy, (5) nuromuscular physiology.

Occupational biomechanics, a more recent term, is concerned with the application of mechanics to the Man-Task-Environment-system to reduce mechanical stresses on the worker's musculoskeletal system while maintaining high performance levels. Therefore occupational biomechanics can be considered an applied division of biomechanics (see Figure 1). Occupational biomechanics is quite useful in: (1) understanding motion patterns required by jobs, (2) estimating the kinematics and

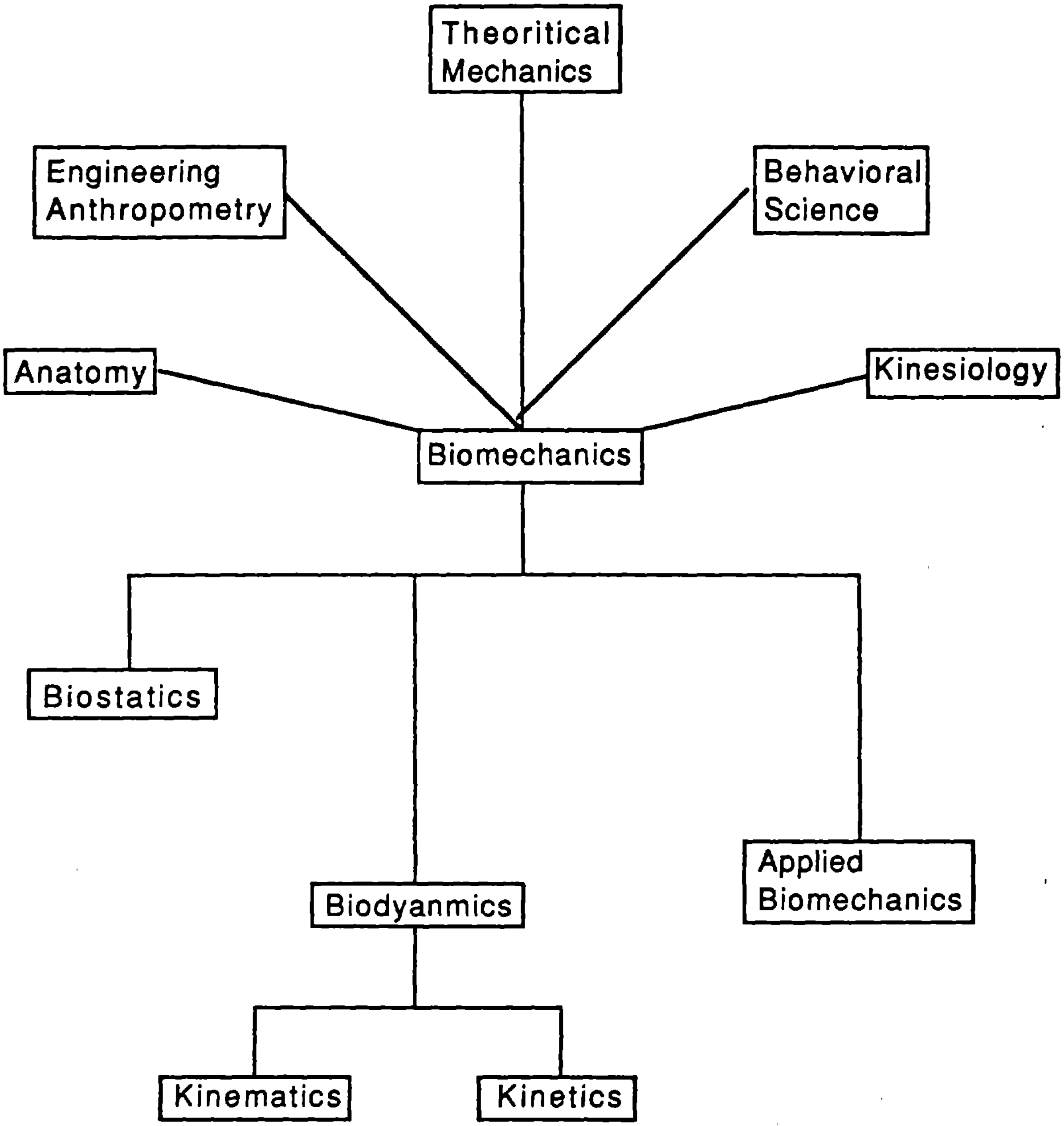


Figure 1. Divisions of Biomechanics

kinetics of these movements, and (3) estimating the stress imposed on various body parts as a result of job performance. In investigating the industrial job related workloads, it is important to measure distribution of loads and stresses among various body segments and tissues. For example, through biomechanical evaluations of the job related stresses imposed on a worker, a potential means of reducing the high incidence rates of manual materials handling (MMH) injuries in industry can be realized. Because of the large number of biomechanical studies and models generated in the area of manual handling, especially lifting, this paper will focus on modeling and simulation of lifting activity.

BIOMECHANICAL MODELS

Biomechanical models were an inevitable result of the investigation of body movements and the kinematics and kinetics of these movements. These models are a representation of the actual system in order to understand the system behavior. Quite often gross simplification and assumptions are made. By constructing a model and comparing the model's behavior with the behavior of the actual system, such as in the case manual lifting tasks, we may gain an insight into how the system functions and the interactions between its components.

Therefore through biomechanical modeling of the human activities, the biomechanical stresses imposed on the body can be estimated. A wide variety of models both static and dynamic have been developed to study the stresses of manual materials handling (MMH) activities.

Biomechanical models in general date back to the work of Braune and Fischer (1889) while studying soldiers while carrying loads. Another pioneer in the development of biomechanical models and related material is Dempster (1955) who described mass and inertia properties of US military population in various postures and motions. Although biomechanical model developments were progressing, it was not until the mid-50's and '60's when more focus and effort was placed on the development of more sophisticated multilink biomechanical models. Rapid development of these models can be attributed in part to the availability of both high speed computers and motion tracking equipment. As a result of these developments several biomechanical models for lifting tasks were developed to estimate stresses on the various body segments and especially the lumbar spine.

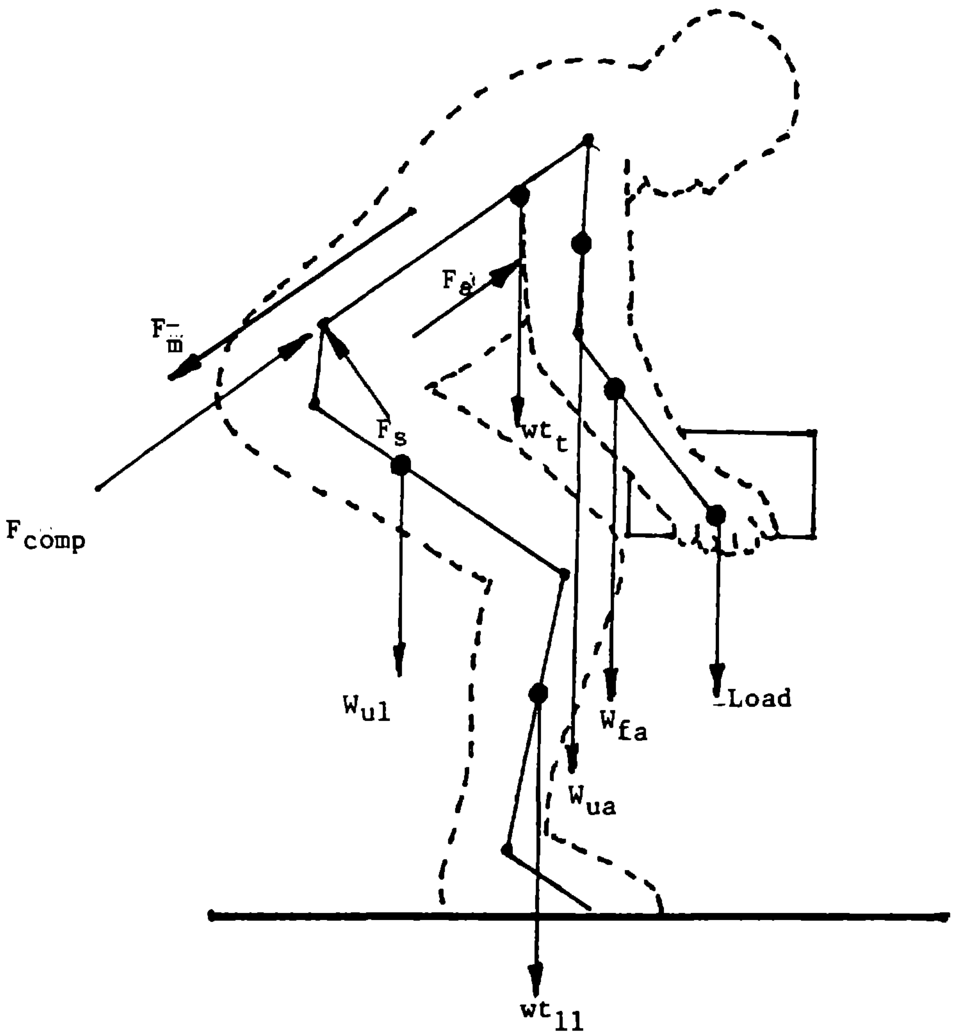
Several two and three dimensional static and dynamic biomechanical models have been developed to determine stresses on manual handling tasks. Some of these static models include Chaffin (1969), Martin and Chaffin (1972), Garg and Chaffin (1975), and Andersson, et al (1985). Dynamic models include those developed by Fisher (1967), Troup (1977), Ayoub and El-Bassoussi (1978), Garg, et al (1982), Lesken, et al (1983), Bejjani, et al (1984), Frievalds, et al (1984), McGill and Norman (1985), and Ayoub, et al (1986).

Most of the above mentioned models were developed using a single muscle equivalent to account for internal trunk muscle forces and resulting compressive and shear forces. Based on the rationale that individual muscle models are of limited practical value due to mechanically undetermined systems, and a precise relations between the mechanical and the electric output of muscle is uncertain as reported by Ortengren and Andersson (1977). The two-dimensional models appear to be satisfactory in analyzing two-handed symmetric sagittal plane exertions.

Three-dimensional static models of the trunk show that for symmetrical sagittal plane lifting activities, only the erector spinae muscles are active (Bean, et al., 1988, Schultz, et al., 1982). But for tasks that involve asymmetrical lifting, many of the lumbar trunk muscles are recruited. More contemporary models of the back include Schultz and Andersson (1981), Gracovetesky, et al (1981), Schultz, et al (1982), Jager (1987), McGill and Norman (1986), Bean, et al (1988), Chen and Ayoub (1988). These are three-dimensional biomechanical models based on several muscle groups to more accurately reflect the muscle activities and compression and shear loads on the spine. Examples of typical two and three-dimensional models of the trunk are shown in Figures 2 and 3.

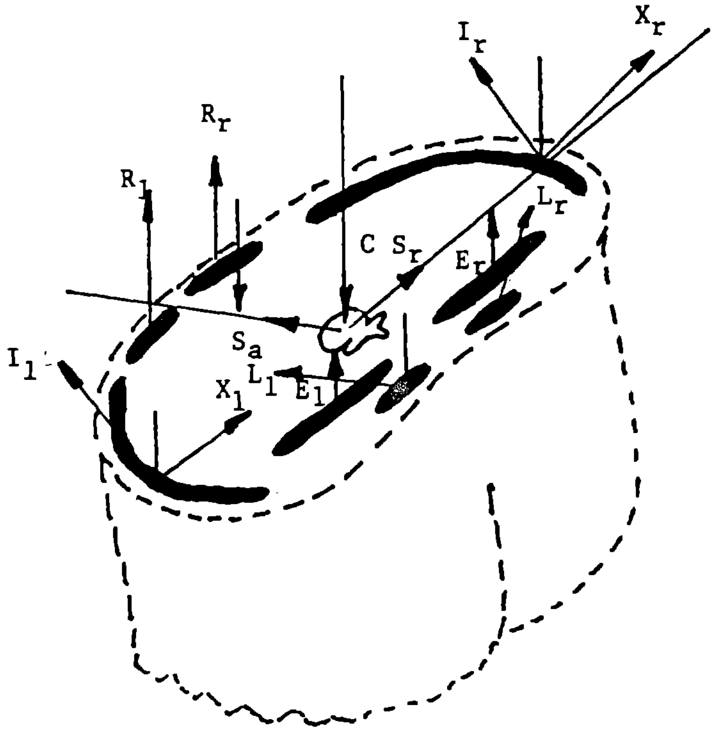
Because of the unknown internal forces and fewer equations of equilibrium, many three-dimensional models of the trunk are statically indeterminent. Therefore, additional assumptions have been made to estimate internal muscle forces. These assumptions need to be validated against experimental measurements or other means such as myoelectrical activities, intradiscal pressure. Such assumptions include:

1. Assume "zero antagonist" activity and the the muscle acts only in tension. Based on this assumption, all the rectus abdominus are given a value of zero. This will tend to reduce the number of unknowns; hence the system can be determinant.



F_m = Erector Spinar Force
F_{comp} = Compressive Force
F_a = Abdominal Pressure Force
F_s = Shear Force
w_{fa} = Weight of Forearm
w_{ua} = Weight of Upper Arm
w_t = Weight of Trunk
wt_{ul} = Weight of Upper leg
wt_{ll} = weight of Lower Leg

Figure 2. Two Dimensional Model



R = Rectus Abdominus (r = right, l = left)
X = External Oblique (r = right, l = left)
E = Erector Spinae (r = right, l = left)
L = Latissimus Dorsi (r = right, l = left)
I = Internal Oblique (r = right, l = left)
C = Compressive Force
P = Abdominal Pressure Force
S = Shear Force (a = anterior, r = lateral)

Figure 3. Three Dimensional Model of Trunk

This, however, is not a good assumption and has not been supported by experimental data (Schultz, et al. 1981; 1982).

2. Use of optimization techniques such as linear programming to determine those internal muscle forces. Using "minimum compression" force on the spine; as the objective function, Schultz, et al. (1981; 1982) found that such techniques produced good agreement between computed muscle tension and measured magnitude of electromyographic (EMG) data.

Dynamic vs. Static Models

Due to the complexity of dynamic biomechanical analyses, as well as the limited dynamic muscle strength data to compare with the task produced forces and moments at various body joints (Garg, et al., 1983), assessment of the stress of lifting on the musculoskeletal system has most frequently been done with the aid of static models. The comparison of the differences between the dynamic and static analyses has been studied by several investigators. Many lifting motions appear to have substantial inertia components and as a result in biomechanical analysis, body dynamics need to be considered when the inertial forces and inertial moments produced are significant when compared with the forces and moments needed for equilibrium (Schultz, 1981). The important factor for using dynamic modeling is the fact that jerking of the load may be necessary by the worker. Such jerking of a load produces inertia forces resulting in momentarily but potentially high overloads on the back structures that are not identifiable in static analyses (McGill and Norman, 1985).

Wood and Hayes (1974) determined the load on the spine using both back and straight leg lifting techniques. The lifting motion was purposely kept slow and simple in order to reduce the dynamic effects. Despite the relatively low accelerations, the statically derived values of L4/L5 torque were considerably lower than the corresponding dynamic values.

Leskinen, et al. (1983) also used a biomechanical sagittal plane model to evaluate lumbosacral compression and stresses on the musculoskeletal system. Comparing the peak L5/S1 compressions in the leg lift and the back lift with the data interpolated from Garg and Herrin (1979), they reported that their data were about 70% and 100% higher, respectively. They also noted that the reasons for these differences were due to the dynamic effects and the intra-abdominal pressure.

McGill, et al. (1985), compared the low back moments during lifting when determined dynamically and statically. They found that the dynamic model resulted in peak L4/L5 moments 19% higher on the average, with a maximum difference of 52%, than those determined from the static model.

The comparison between the dynamic maximum compressive force and static maximum compressive force on L5/S1 showed that the values of the dynamic case were always larger than the static case (Kim, 1990) (See Table 1). Significant differences were found ranging from 4 to 40%. On the average, the value of the maximum compressive force for the dynamic was about 120% of the maximum compressive force of the static case. Similar results were reported by Marras, Nongsam, and Rangarajulu (1986). They claimed that the introduction of small amount of velocity into the compressive force analysis increases the compressive loading by almost 40%.

Table 1. Comparison of Dynamic and Static Maximum

Compressive Force of Subject 2

RANGE	FREQ	WT (KG)	DYNAMIC	STATIC	DYNAMIC/STATIC
FK	2	0	267.71	200.76	133.35%
FK	2	10	433.81	338.61	128.11%
FK	2	20	536.53	463.00	115.88%
FK	2	30	620.58	529.49	117.20%
FK	2	40	722.45	712.78	101.36%
FK	8	0	281.97	204.83	137.66%
FK	8	10	480.29	349.28	137.51%
FK	8	20	537.21	438.10	122.62%
FK	8	30	683.20	612.36	111.57%
FK	8	40	663.26	565.57	117.27%
FS	2	0	256.83	202.52	126.82%
FS	2	10	407.58	345.38	118.01%
FS	2	20	500.12	437.71	114.26%
FS	2	30	632.74	580.00	109.09%
FS	2	40	659.50	632.38	104.29%
FS	8	0	259.11	205.52	126.08%
FS	8	10	466.44	331.97	140.51%
FS	8	20	571.21	445.66	128.17%
FS	8	30	665.38	555.94	119.69%
KS	2	0	172.83	135.57	127.48%
KS	2	10	334.19	294.92	113.32%
KS	2	20	516.72	469.49	110.06%
KS	2	30	489.33	474.50	103.13%
KS	2	40	664.09	633.99	104.75%
KS	8	0	148.65	117.23	126.80%
KS	8	10	305.89	303.65	100.74%
KS	8	20	498.64	477.69	104.39%
KS	8	30	625.14	594.23	105.20%

SIMULATION OF HUMAN MOTION

Humans perform physical activities in a variety of ways. These would depend upon individual anatomical structure, physiological functions, psychomotor control pattern, and associated pathological or chronological changes (Chao, 1986). In addition, many of our physical activities are task-oriented, varying drastically depending upon our personal habits, training and motivation. Interest in the behavior of the skeletal, muscular, and neural control subsystems has grown to such an extent that numerous efforts have been dedicated simulation of these biosystems and have obtained fruitful results.

In the search for such "optimality" in human motion, theoretical attempts to formulate and demonstrate a "minimal principle" has been provided by Nubar and Contini (1961). In addition, a modeling approach (with experimental data support) based on a minimization principle can be found in Chow and Jacobson (1971); Seireg and Arvikar (1975); Ayoub (1974); Petreno (1972); Muth, et al. (1976); Hatze (1976); Crowminshield and Brand (1981); Redfield and Hull (1986); and Marshall, et al. (1985). As a result, the use of optimization, prediction, and quantitative hypothesis testing, the mathematical modeling approaches has generated results within an acceptable degree of accuracy.

The Optimal Principle in the Modeling of Human Lifting

The feasibility of using a principle of optimality in the modeling of load lifting are discussed in three points:

1. Considering the minimal principle postulated by Nubar and Contini, in a situation such as infrequent load lifting by

experienced industrial workers, it is very appealing to think that the body selects a pattern of motion to minimize stress imposed on it. Muth, et al. (1976) developed a lifting model using the minimum principle. Seireg and Arvikar (1975) used a similar principle to address the problem of finding moments of individual muscles producing walking motion.

2. Crowninshield and Brand (1981) and Rohrle, et al. (1984) has proven to be a promising approach to the indeterminant optimal control problem. The general formulation can be summarized as follows:

Minimize $f(w_1, w_2, - - - w_n)$,

Subject to $g(w_1, w_2, - - - w_n) = 0$,

and $b_i <= w_i <= a_i$ $(b_i >= 0)$ $i = 1, 2, ..., n$.

where f is a cost function that can be either linear or nonlinear. The function g represents the equations of motion and other equality constraint relationships based on anatomy and dynamic characteristics of the task. The w_i stands for the moments at joints. The w_i are also subject to inequality constraints.

3. The critical issue in the modeling of human lifting is the determination of a realistic objective function (cost function). The ability of several criteria to predict motion patterns (i.e., the body segment trajectories) has been examined. The criteria were: the sum of the muscle effort at joints (Nubar and Contini, 1961; Seireg and Arvikar, 1975) in gait analysis; Lee (1988); the angular jerk and the sum of the segmental mechanical energies (Marshall, et al., 1985) in gait analysis; the sum of the square of mechanical work done at ankle joint

(Muth, et al., 1976) in lifting analysis; the joint moments and muscle stress (Redfield and Hull, 1986) in bicycling simulation.

Studies of Human Motion via Optimal Programming

The modeling, simulation, and optimization of the dynamics of the human musculoskeletal system are a real challenge to ergonomists, control engineers, and mathematicians. Chow and Jacobson (1971) formulated an optimization model to describe human gait. The performance criterion was the minimization of mechanical work done at the hip and knees during normal walking.

Ayoub (1974) proposed a biomechanical model to predict the path of motion for the arm while performing a simple task. The performance criterion for this model was the minimization of mechanical energy used.

Seireg and Arvikar (1975) tackled the problem of optimal configuration of muscular forces about the hip, knee, and ankle during normal walking. The model, which determined the activity level of muscles at each joint, was formulated as a linear programming problem.

Muth, et al (1976) described a lifting task as an optimization problem with a non-linear objective function subject to a set of linear constraints. The objective function was expressed as the time integral of torques acting on the ankle joint while lifting. Model constraints were determined by the limitations imposed by the physical characteristics of the human body, the task and the workplace.

Pedotti, et al. (1978) studied the muscular-force optimization problem in human locomotion by formulating four biological optimization criteria for the muscular forces, and comparing the experimental EMG data from the muscle with the muscle force patterns obtained computationally under the various performance criteria.

Marshall, et al (1985) examined the ability of seven optimization criteria to predict the segment kinematics and center of mass trajectory of a normal subject walking at his preferred pace. A nonlinear optimal control program was formulated and a generalized simulation package was used to predict the movement patterns. The results indicated that functions involving either the sum of the joint torques, the angular "jerk", or the sum of the segmental mechanical energies predicted the segmental kinematics and center of mass trajectory most accurately.

Simulation Models Development

To discuss details of human simulation studies, I like to focus on a single study currently in progress at Texas Tech. The study's objective was to simulate human lifting motion trajectories.

Generally in studies of human motion simulation, the research activities are divided into three phases (see Figure 4).

1. Pre-model development analysis: During this phase, data on lifting kinematics were collected to identify ranges of motion and angular acceleration for each joint (range of lift and container size). In addition, dynamic joint strength tests are conducted to obtain the maximum joint strength values at the joints of interest.

2. Model development: A mathematical model was developed to generate lifting trajectories. The lifting task is presented

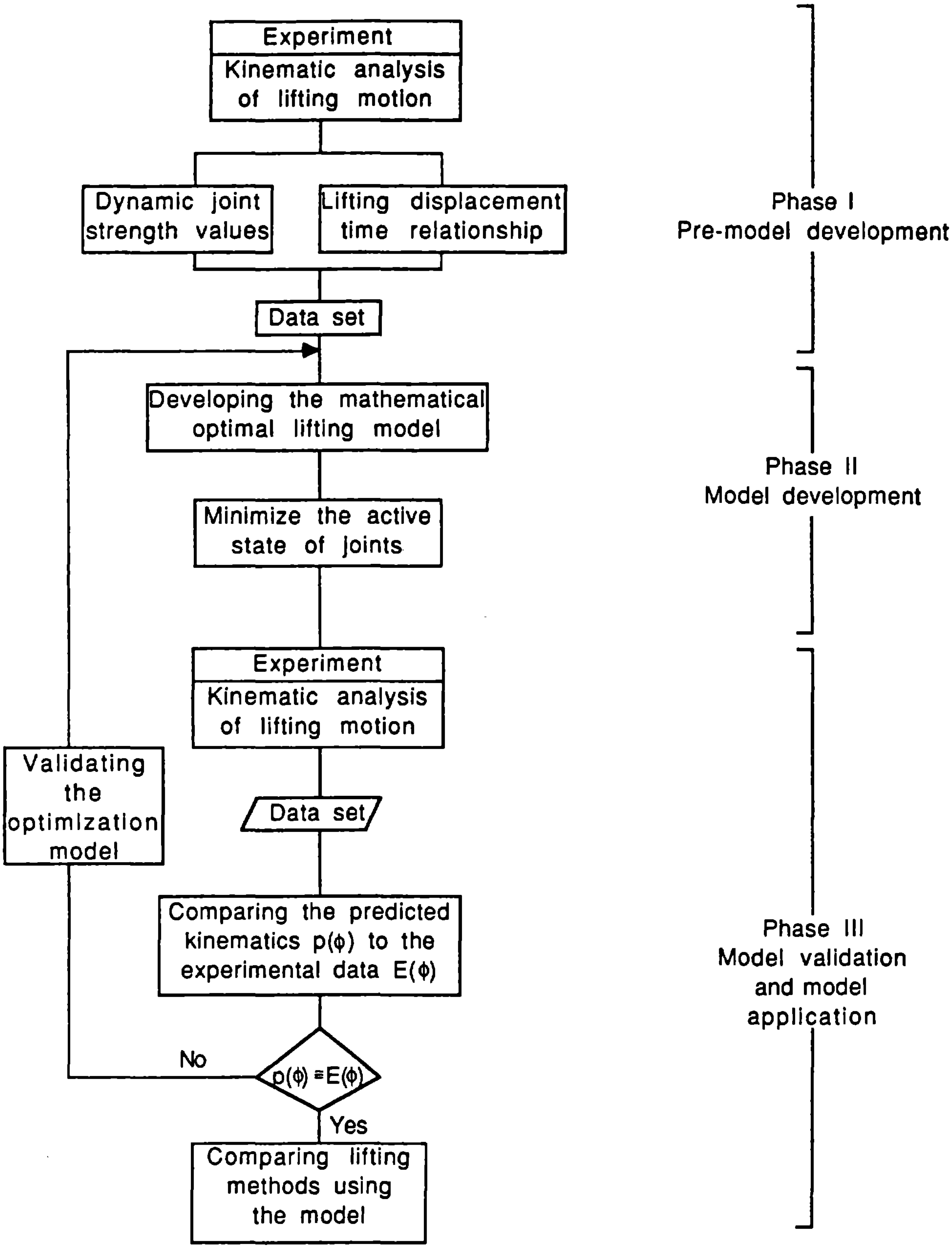


Figure 4. The Flow of Study Activities

as a non-linear programming problem with a non-linear objective function subject to linear, as well as non-linear constraints.

3. Model validation and model applications phase: Lifting tasks were predicted using the model. Then, a verification procedure was conducted to justify the homogeneity between the experimental (measured) trajectory of the joint and the corresponding trajectory generated by the model.

Simulation Model Development

In a recent study by Lee (1988) and Ayoub, et al. (1989), a simulation model was developed for sagittal lifting activities utilizing five joints. The mathematical form of the objective function and constraints are:

1. Objective Functions:

opt_obj = MIN

$$(torque(t,j)/max\_stren(j))^2 \ dt. \qquad (1)$$

The objective function (eq. 1) or the cost function is the minimization of the time integral of the sum of the square of the active state of each joint. Use of the ratio as the criterion emphasizes that the optimization process distributes moments to the joints according to their relative abilities. The "square" term provides for a heavier penalty for large deviants (compared to linear) in the minimization process.

2. Nominal Constraint Set:

The constraints are: the constraints imposed by the initial and final joint angles, the constraints imposed by the joint mobility, the constraints imposed by the reaching envelope, the constraints imposed by the workstation, the constraint of the angular acceleration, the constraint of the rate of change of acceleration, the constraint due to muscle strength, and the constraint on the limit on the center of mass of the body and load combination.

Solution Methods

A model such as that described above differs from the traditional nonlinear programming problem in that it requires the minimization of a time integral of a function. In effect, the model is required to "pick" a motion path which results in minimizing the objective function. The method employed to solve these continuous, multi-variable, multi-period, and nonlinear optimization problem involves the use of a dynamic programming procedure and a coarse grid search technique.

From the optimization point of view, the following observations regarding the model can be made:

1. The model has a quadratic objective function and linear as well as non-linear constraints.

2. The problem is very large in size and is characterized by the number of stages and state variables, as well as the number of coupling, non-coupling, time invariant, and time variant constraints. The formulation results in a dynamic programming problem with states represented by a five dimensional vector. The number of stages chosen depends upon the precision of the integration approximation required.

3. The problem was solved using the 'Trajectory Approximation in State Spaces' technique (Durling, 1964). Computationally, the problem is decomposed into two parts for solution. The first part is to generate initial joint

trajectories: the second part is to improve the trajectories to arrive at an optimal or (near-optimal) result (see Figure 5).

Model Application

In an attempt to apply the model to a lifting task, a lifting experiment was performed. The task performed was lifting from the floor to knuckle height (approximately 30" above the floor) and lifting from the floor to shoulder height (approximately 50" above the floor). Two sizes of containers were lifted 24 x 12 x 12" and 24 x 18 x 12" while the weight lifted was the subjects maximum acceptable weight of lift (MAWL) determined psychophysically. Five subjects performed the lifting tasks while being photographed with the Motion Analysis System. Five joints were tracked by the system. These were the ankle, the knee, the hip, the shoulder, and the hand. In addition the center of mass of the container was also monitored.

To apply the model the following inputs were provided:

(a) The initial and final position of the body based on the initial and final locations of the load,

(b) Strength data for the subject,

(c) Weight of container (the load), and

(d) The physical constraints dealing with range of motion, reach envelope, max accelerations, and first derivative of the acceleration with respect to time, and body balance.

When the model was applied to lifting tasks, the model generated joint motion patterns were compared with actual motion patterns. Figure 6 shows a sample of the results of this application. Based on the results of several applications of the

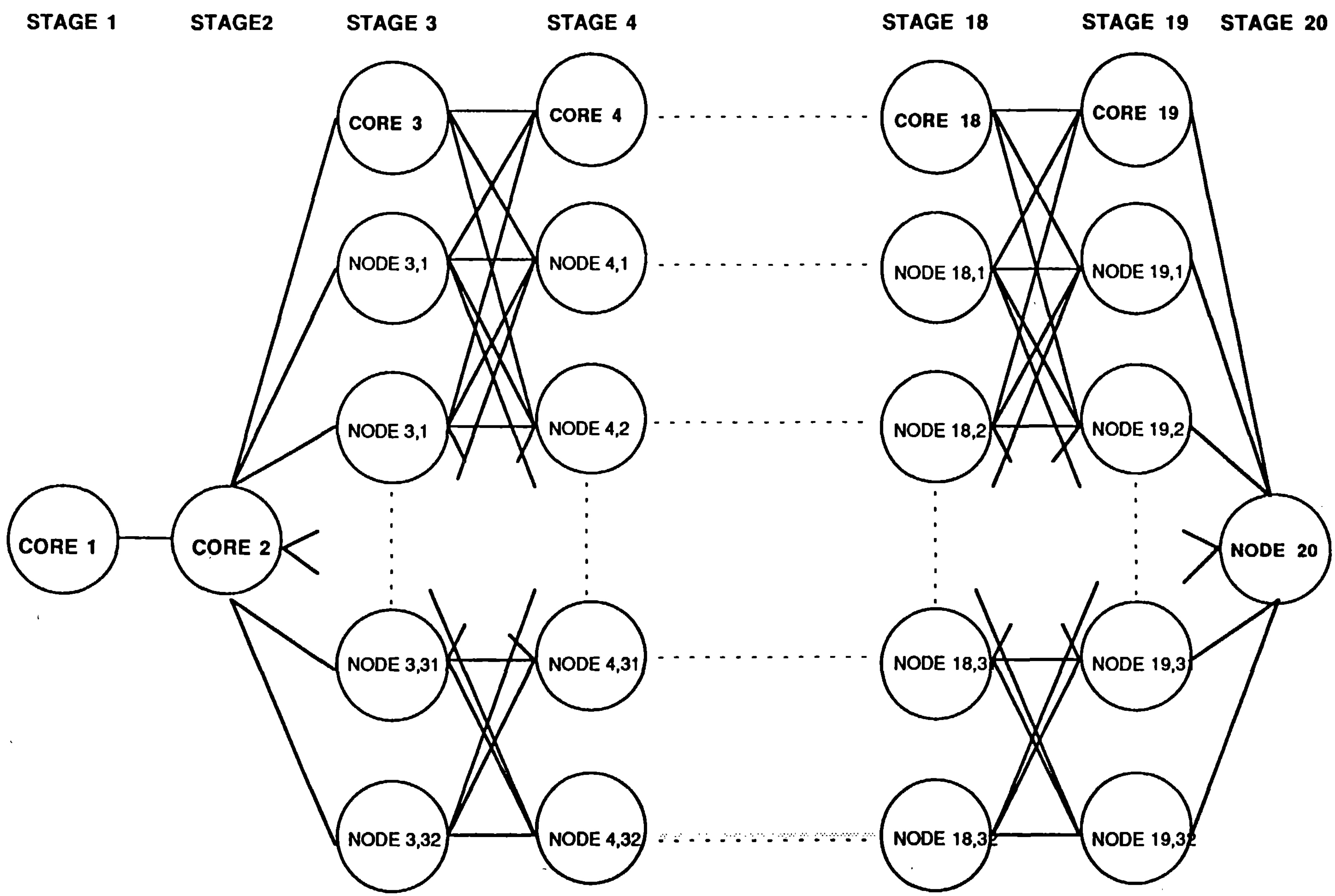
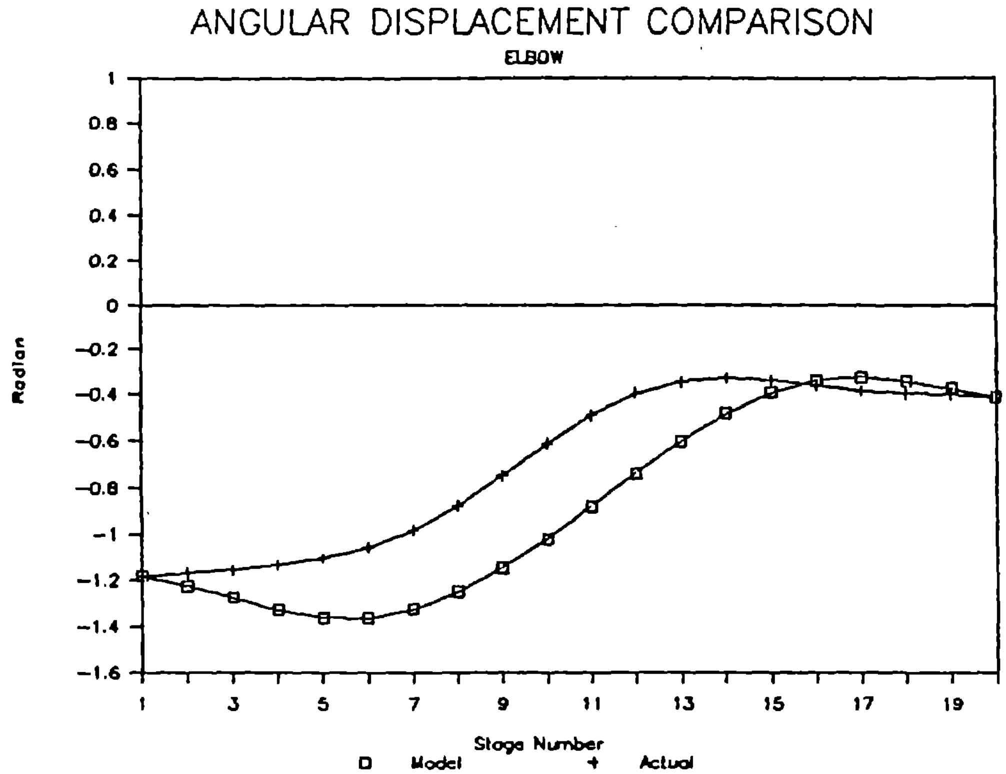


Figure 5 - Network For The Model



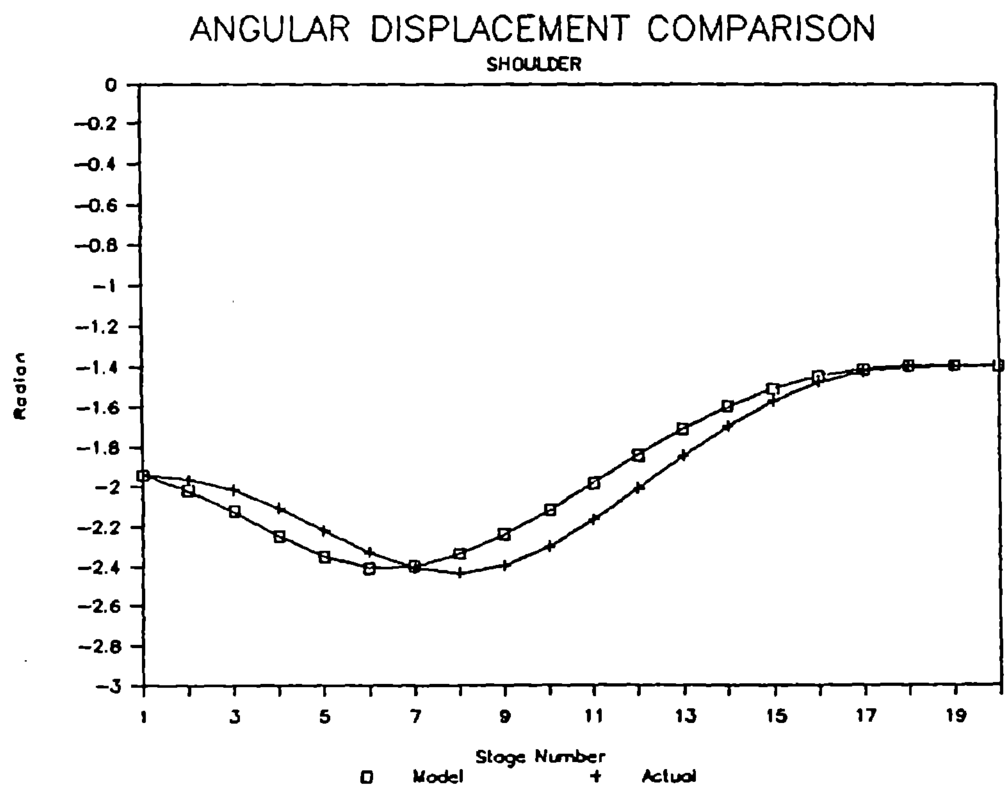
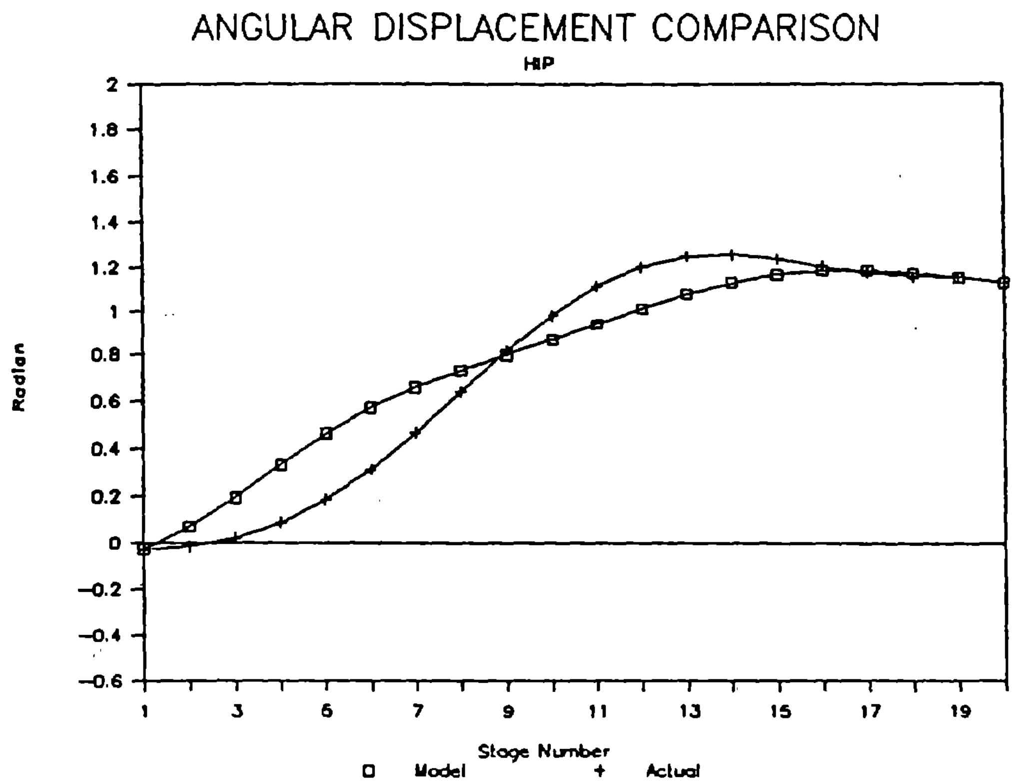


Figure 6 - Displacements of the joints and the load



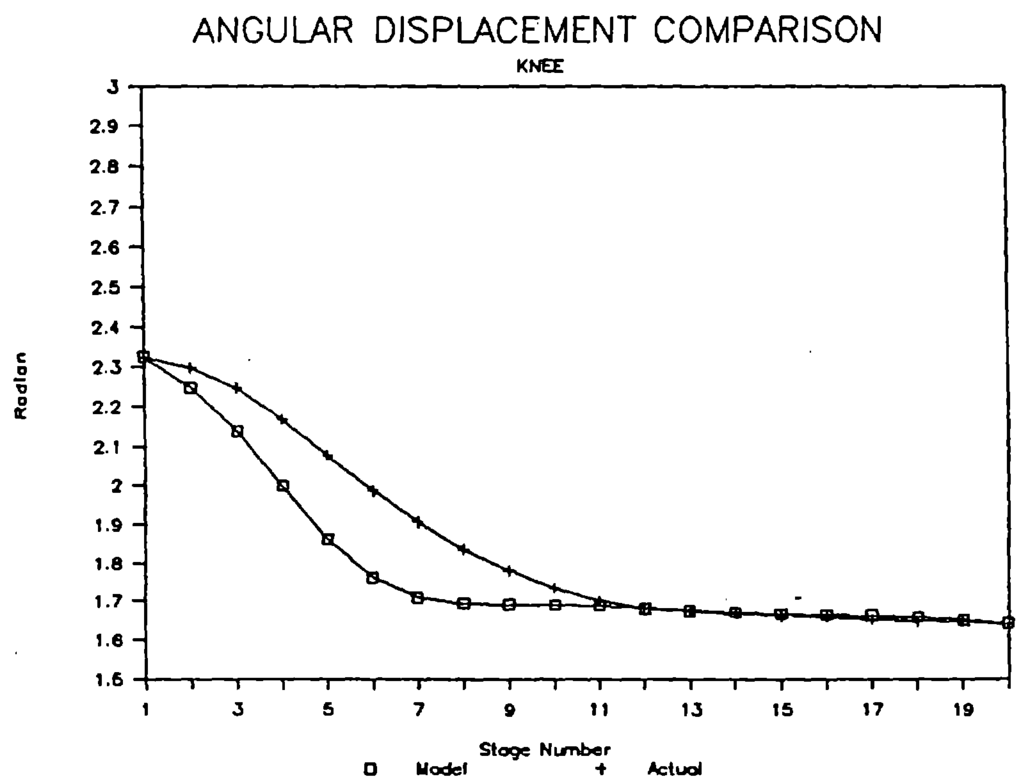


Figure 6 - continues..

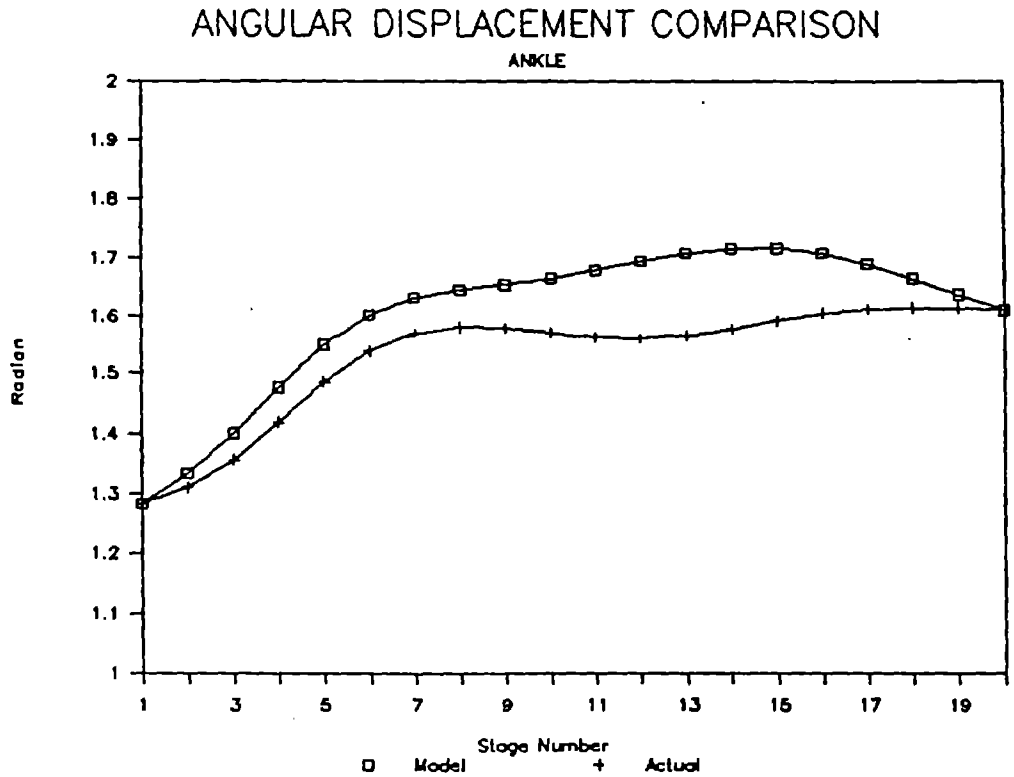


Figure 6 - continues..

model, one may conclude that it may be feasible to develop simulation models to predict the path of motion body joints in two dimensions while performing a task. Although the paths generated by the model are not identical to the actual paths, the accuracy of prediction considered adequate enough with the majority of mean square error constituted by the random error component. However, there is room for improvement (Lee, 1988).

SUMMARY

With regards to biomechanical modeling, several models exist. These can be divided into static models and dynamic models. These in turn can be divided into two-dimensional and three-dimensional models. The objective in biomechanical modeling is to develop a model of the body or segment of the body which can accurately predict the kinematics and kinetics of task performance and hence the risk of injury. Such a model must be provided with adequate data such as information from EMG studies, muscle strength studies, anthropometric measurements, external forces applied to the body, displacement-time information, internal muscle force vectors, passive tissue data, etc. By examining this list such models do not now exist, but information needed to develop such models are now being pursued by several investigators. The ultimate goal is to develop three-dimensional dynamic models.

With regards to human simulation models, very few models exist which predict the motion pattern a worker may or should follow in the performance of a task. These simulation models differ from biomechanical models in that biomechanical models require input information about displacement-time relationship, while simulation models have the displacement-time relationships as their output. These simulation models depend on optimization theory and techniques to solve problems about the motion patterns required in the performance of a task to minimized a specific cost function or functions subject to a set of constraints.

Both types of models (biomechanical and simulation models) can be combined to provide the ultimate goal of developing a model which can, not only estimate accurately the forces acting on the body, but also predict the motion patterns which will minimize such stresses on the body.

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16. Abstract (Limit: 200 words) The modeling and simulation of manual lifting were described and discussed. Several biomechanical models have been developed. The objective of biomechanical modeling has been to develop a model of the body or segment of the body which can accurately predict the kinematics and kinetics of task performance and hence the risk of injury. Biomechanical models can be divided into two dimensional and three dimensional models and can be either static or dynamic. These models require the input of data from electromyographic studies, muscle strength studies, and anthropometric measurements, as well as information on external forces applied to the body, displacement time, internal muscle force vectors, passive tissues, and other facts which will enable the researcher to achieve good results from the models. Very few human simulation models exist which predict the motion pattern a worker may or should follow in the performance of a task. These simulation models differ from biomechanical models in that biomechanical models require information about displacement time relationships as input, while simulation models have displacement time relationships as their output. According to the author, biomechanical and simulation models can be combined to provide the ultimate goal of developing a model which can estimate accurately the forces acting on the body and predict the motion patterns which will minimize such stress on the body.

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