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# A COMPUTERIZED BIOMECHANICAL MODEL–DEVELOPMENT OF AND USE IN STUDYING GROSS BODY ACTIONS*†

DON B. CHAFFIN

Department of Industrial Engineering, The University of Michigan, Ann Arbor, Mich. 48104, U.S.A.

**Abstract**—Gross body actions involved in heavy industry, e.g. lifting and carrying materials, are often the cause of injury to the musculoskeletal system. A computer model is developed which treats the human body as a series of seven links from which reactive forces and torques are computed at each articulation during various simulated materials handling tasks. In addition, an analysis of shearing and compressing forces at the lower lumbar spine is included. The assumptions of the present model are presented, along with a discussion of future models.

IT HAS been reported by Troup (1965) that a relatively large portion of industrial injuries (as great as 12 per cent) are back disorders resulting from a lifting task. It is also evident from papers by Tichauer (1965), Raof *et al.* (1960) and Davis *et al.* (1965) that the estimation of stresses on various parts of the musculoskeletal system during lifting activities will require a complex methodology which takes account of such factors as (1) instantaneous positions of the extremities and trunk, (2) curvature changes in the spine, (3) strength variations within different muscle groups and people, and (4) abdominal pressure effects.

It is the intent of this paper to describe a computerized biomechanical model which can be used to estimate the forces and torques that are created at six major articulations of the body, (i.e. wrist, elbow, shoulder, hip, knee, and ankle) as well as at the fourth lumbar through the first sacral spinal vertebrae of a person who is performing a weight handling task. This particular model is used as an example of the types of models that are now practical due to the relative ease of use and computational speed of today's digital computers.

## WHOLE-BODY COMPUTERIZED BIOMECHANICAL MODELS–BACKGROUND

It has been nearly 80 years since Braune and Fischer (1890) published their data regarding the mass distribution for the various body segments. Since then, fundamental extentions by Dempster (1955) and Drillis and Contini (1966) have resulted in better estimates of (1) the location of the mass centers-of-gravity, (2) the link lengths, and (3) the magnitudes of the moments-of-inertia of the various body segments.

However, it was not until the widespread use of the commercial high speed digital computer that this type of data could be easily used in developing analytical models to study the mechanics of the human body. The digital computer has provided a computational capacity which, in turn, has fostered the development of several different types of biomechanical models. Some of these models have been formulated to determine the whole-body center-of-gravity location when the body is placed in various configurations (Hanovan, 1964). This type of model is used in determining body movements if mechanically unrestrained and acted upon by changing

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momentum or gravitational forces, such as in a vehicle accident or when falling.

Another type of computerized biomechanical model primarily estimates the forces and torques at various articulations of the body during voluntary actions (e.g. lifting, running, or throwing). An early example of this type is a two-link model of the arm developed by Pearson *et al.* (1961). The intent of this model is to compute the forces and torques at the elbow and shoulder during a sagittal plane motion of the arm-forearm-hand aggregate. Figure 1 is an illustration of the mechanical analogue of the arm used in this type of model. Stroboscopic photographs of the various arm motions of interest are taken to determine the instantaneous positions, velocities, and accelerations of the arm segments. This 'activity' data along with the anthropometric dimensions of the segment lengths and weights, provides enough input information to compute the stress levels (in forces and torques) at the elbow and shoulder, thus providing a means to achieve a better understanding of both the complex muscle actions required for control of the arm, and the resulting strain at the articulations.

An extension of the Pearson Arm Model was developed by Plagenhoef (1966). Again, photographic data is used to describe the body configurations during the relevant task. This spatial information, combined with additional anthropometric dimensions regarding the length of the arm, trunk, and leg segments, and the total weight of the subject, provide adequate information to compute the forces and torques at the elbow, shoulder, hip, and knee during various physical activities performed in the sagittal body plane.

## THE PRESENT SSP MODEL–BACKGROUND

The recent efforts of this author have been to extend the biomechanical model of Plagenhoef

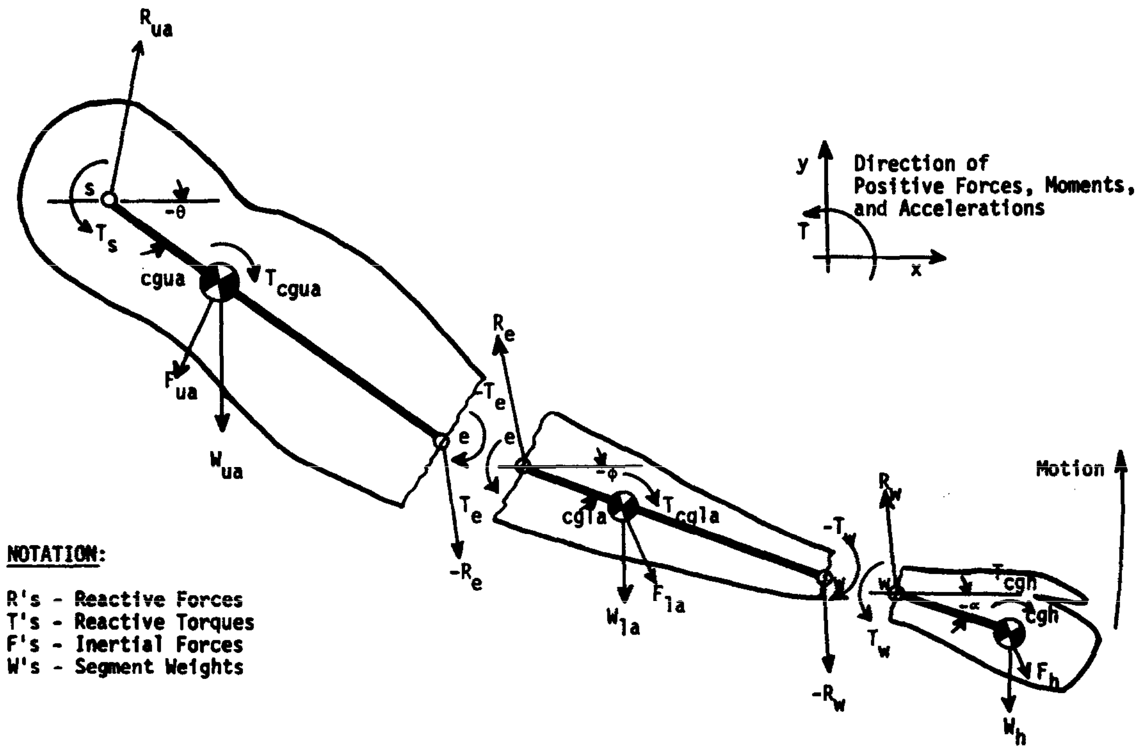



Fig. 1. Three segment human arm for analysis of sagittal plane movements per Chaffin (1967).

to include (1) an estimate of the stress in the lower lumbar spine, (2) the addition of external loads on the hands (e.g. in a materials handling task), and (3) an evaluation of the effects of various muscle group strengths on human performance. In accomplishing this, several different computerized biomechanical models have evolved. The following is a description of one of these models, along with a presentation of a few of the more interesting results that have been obtained from using the model to study various whole-body activities.

The model has been referred to as the Static Sagittal Plane model, abbreviated SSP, by the research group that has used it over the last two years. As the name infers, this particular model has been developed to evaluate various 'static' situations, such as when one is holding a weight, or pushing or pulling on a non-moving container. In addition to these applications, the model can be used to analyze 'slow' moves by formulating the input data to describe a sequence of static positions with very small changes in each successive position. In making this type of pseudo-dynamic analysis, it must also be assumed that the effects of acceleration and momentum are negligible.*

The SSP model is also restricted to symmetric sagittal plane activities, thus a rotation or lateral deviation cannot be analyzed. A complete relaxation of this latter restriction is expected in the near future through the development of a three-dimensional model. Two of the greatest problems in this more advanced model are (1) to present spatial data describing the position of each body segment in both time and three-dimensional space, and (2) to intuitively understand the complex vector representations of forces and torques that result from this three-dimensional model of the human body. Computerized graphical displays are being developed to reduce this latter problem.

## THE SSP MODEL – WHAT IS INCLUDED AND WHY

The static sagittal plane model develops the estimates of the forces and torques at each of the major articulations of the extremities in much the same manner as the Pearson and Plagenhoef Biomechanical Models. In essence, this requires that the body be treated as a series of seven solid links which are articulated at the ankles, knees, hips, shoulders, elbows, and wrists.† Each of the links in the model is considered to have a mass whose estimate is based on the proportionality constants presented by Drillis and Contini (1966). The distribution of the mass within each link is based on the data of Dempster (1955). The link lengths are established from over-the-body measurements, with the reference landmarks described by Dempster (1955). Specifically, the body measurements needed as input data are: body stature, body weight, center-of-gravity of the hand to wrist distance, lower arm length, lower leg length, foot length, and elbow height when standing. From these, the link lengths (i.e. the straight line distances between the articulation points-of-rotation) are estimated based on the empirical relationships developed by Dempster (1964).

The task under analysis by the SSP model is described by two types of data. First, any external force that may be exerted against the hands is measured and entered into the program as a vector acting at the center-of-gravity of the hands. For example, if a person is holding a 10-lb box it is entered into the program as a 10-lb force acting downward (i.e. a force acting in respect to some

*A dynamic version of this same model which includes estimates of acceleration and momentum has recently been developed by Fisher (1967).

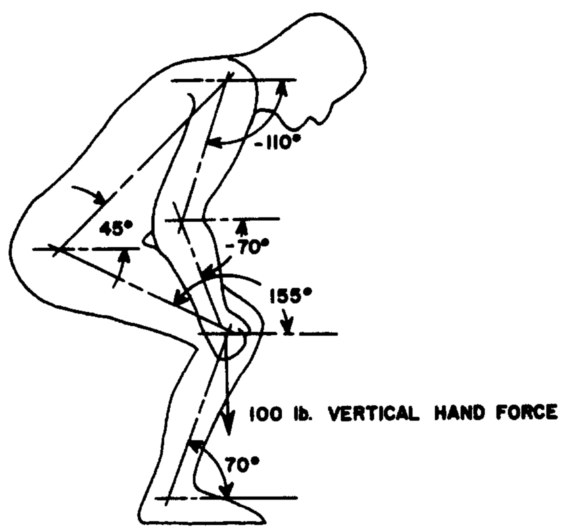
†The computational techniques of the model have been described previously, and in the interest of conserving space are not presented here. It is suggested that if interested the reader review Plagenhoef (1968), Pearson *et al.* (1961), Williams and Lissner (1962), or Chaffin (1967).

defined reference axis). The second type of information required to describe the activity is the position of the body. To obtain this data it is necessary to measure the articulation angles from either a lateral photograph of a person who is in the position of interest, or from an articulated drawing board body template placed in the 'Task' position.

The above data is sufficient for the model to compute the torques and forces at each of the six major articulations of the body. As an example of the insight that can be gained from the model, the effects of various body sizes is clearly indicated by examining the articulation torques presented in Fig. 2, assuming, (1) dimensions ranging from the smallest 5 per cent to the largest 5 per cent of the general male population in the United States and (2) the lifting position indicated.

The addition of a specific spinal stress evaluation was based on prior studies by others. These studies have disclosed that compressive forces on the spinal column can cause end-plate fractures of the vertebral bodies, Armstrong (1965), Morris *et al.* (1961), Nachemson (1962) and Perey (1957). Furthermore, degeneration of the intervertebral discs and resulting herniations have been attributed by both Perey (1957) and Gordon (1961) to the compressive forces to which they are subjected in daily activities. Thus, it was believed that an estimate of the compressive forces created at the various disc/vertebral end-plate interfaces during specific physical tasks would provide an important contribution towards the further understanding of low-back injury as a limit to human performance.

The particular estimations of compressing and shearing forces at the *lower lumbar spine* were chosen for inclusion in the model based on statistics regarding back disorders. These




| Male population dimensions* (%) | Resultant torques (kg-cm) CW = + | | | | | |
|---|---|---|---|---|---|---|
| | Ankle | Knee | Hip | Shoulder | Elbow | Wrist |
| Small 5 | 247 | −1041 | 2445 | 25 | 521 | 106 |
| Average 50 | 299 | −1258 | 2823 | 24 | 554 | 111 |
| Large 95 | 346 | −1620 | 3377 | 20 | 589 | 118 |

*Averaged dimensions from Damon *et al.* (1966) and Dempster (1955).

Fig. 2. Example effect of body size on articulation torques when lifting.

disclosed that between 85 and 99 per cent of all serious back injuries, (i.e. disc herniations) occur at the L4/L5 and L5/S1 levels, (Krusen, 1965; Smith, 1944; and Armstrong, 1965).

To perform the analysis of the lumbar spinal stress, the following concepts were included in the SSP model. First, the geometry of an average erect spinal column was developed from the dimensions of Fick (1904) and Lanier (1939). This resulted in the column depicted in Fig. 3. The dimensions of this average male column were proportionally scaled in the model, based on the hip-to-shoulder distance, to enable the study of smaller or larger individuals.

The curvature change for the column during sagittal rotation of the hips was assumed from the data of Dempster (1955) which disclosed that for the first 27° of trunk flexion the pelvis does not rotate (i.e. the rotation is in the lumbar spine) and for each additional degree of trunk rotation the pelvis contributes about two thirds of a degree. Also, it was assumed that 22 and 29 per cent of the lumbar rotation occurs at the L5/S1 and L4/L5 discs :espectively, based on data by Davis (196 ), Allbrock (1957), Lindahl (1966) and Rolander (1966).

The contribution of intra-abdominal pressure in relieving compression on the lumbar spine was estimated from the data of Morris, Lucas and Bresler (1961). The procedure required correlating their pressure data with the torque and angle estimated at the hip of the subjects used in their experiments. The least-squared error expression developed from this procedure is:

$$\begin{bmatrix}\text{Abdominal}\\ \text{pressure}\end{bmatrix} = 10^{-4}\,[0{\cdot}6516 - 0{\cdot}005447\,(\text{Hip angle})]\,[\text{Hip torque}^{1{\cdot}8}]$$

where:

Abdominal pressure is in mm of Hg, with a maximum limit of 150 mm Hg. Hip angle is in degrees from the erect position. Hip torque is in kg-cm.*

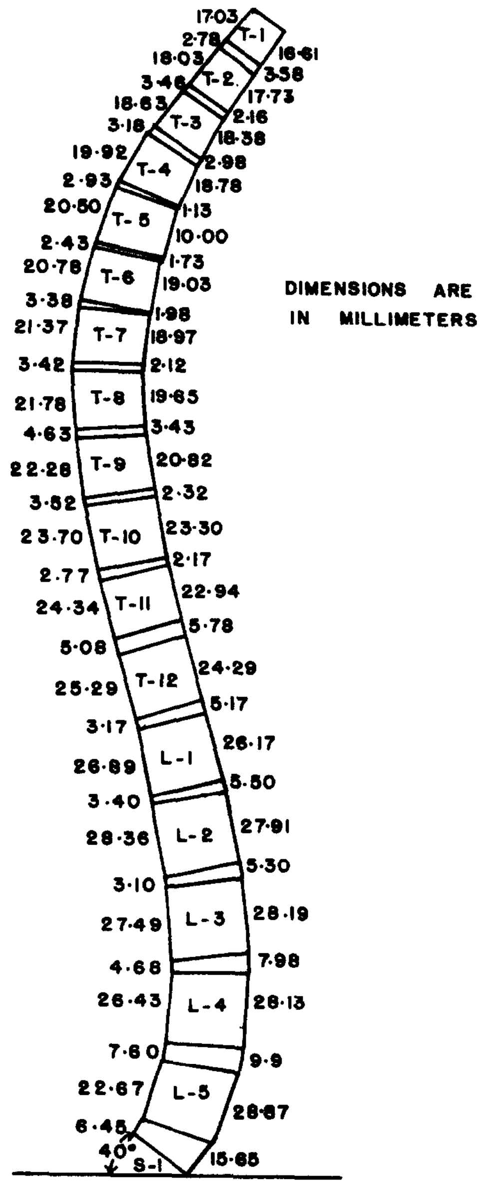



Fig. 3. Average spinal column as presented by Fisher (1967).

*The unit 'kilogram' is used as a measure of force throughout the paper to be consistent with prior reports, and to avoid large numbers (i.e. a kilogram is assumed to be equivalent to both $9{\cdot}8 \times 10^5$ dyn., which would be consistent with the commonly used centimeter unit of length, or in the engineering system, to 2·21 lb).

This expression was found to have a correlation coefficient of 0·73. The error can be attributed to (1) not knowing the exact position of the trunk during each test, thus causing an unknown variation in both the assumed hip angles and torques, and (2) not being able to access the time rate of force application.

Recent experiments by Asmussen (1968) have disclosed that probably the method of lifting, (i.e. quick jerk or slow sustained pull) can significantly affect the abdominal pressure.

The amount of force created by the abdominal pressure was estimated by assuming the following three conditions which are similar to those proposed by Morris *et al.* (1961): (1) a diaphram area of 465 $cm^2$ and a pelvic area of 517 $cm^2$ upon which the abdominal pressure can act, (2) the line-of-action of the force acts parallel to the line-of-action of the normal compressing forces on the lower lumbar spine, (3) the force acts through finite moment arm distances from the centers of the discs. The moment arms have been assumed to vary as the sine of the angle at the hips, with the erect position having moment arms of 6·2 cm at the pelvis and 6·7 cm at the diapharm level, and the 90° hip angle position having 13·7 cm at the pelvis and 14·9 cm at the diaphram.

Additional compressive forces on the lumbar spine due to the abdominal muscles were assumed to be negligible, since Bartelink (1957) disclosed that the rectus abdominis, which could mechanically cause a spinal compressive force, was relatively inactive during lifting activities. The abdominal pressure was attributed, there-by, to the oblique and transverse muscles, which are not well positioned to directly assist or hinder in sagittal plane flexion or extension action of the trunk.

The line-of-action of the muscles of the lower lumbar back were assumed to act parallel to the normal compressing force on the vertebral/discs interface, and with a moment-arm of 5·0 cm.* The estimate of the magnitude of the muscle force required to maintain a particular trunk position against the gravitational forces that act on the body masses and any mass being held in the hands is accomplished by dividing the estimated torque at the center of the two discs by the 5 cm moment arm assumed for the back muscles. An example of the calculations is presented in the Appendix.

As mentioned earlier, it was believed necessary to include a 'muscle strength' evaluation in whole-body biomechanical models. The reason being that muscle strength varies greatly between and within a particular population, due to both anatomical variations as well as changes in specific types of daily physical activities. The method used to quantify the voluntary strength of various muscle groups was simply to place the subject in a specific 'test' position and ask him to 'Pull' or 'Push' with maximum efforts against various restraining straps attached to load cells.† By multiplying the forces measured via the load cells by the distances that the straps were placed from the various articulations, a set of maximum voluntary torque limits for the following muscle group actions was obtained for given angulations of the articulations:

Elbow flexion and extension,
Shoulder flexion and extension,
Hip extension,
Knee extension,
Ankle plantar flexion.

By comparing the articulation torques computed by the SSP model for a given person and task, with the maximum voluntary articulation torques obtained from the same person when performing the various strength

*This is an average moment arm, which is based on values published by Bartelink (1957), Munchinger (1962), Perey (1957) and Thieme (1950), as well as from cadaver measurements.

†This procedure is similar to that described by Clarke (1966).

tests, it is possible to gain an indication of the 'degree of loading' of a particular muscle group during the task. As an example of both the application of this procedure, and the spinal stress evaluation, the following investigation is presented.

### THE SSP MODEL ANALYSIS OF WEIGHT LIFTING

Much controversy presently exists regarding the recommended weight that can be safely handled by various industrial populations, or even what type of factors should be considered in arriving at a 'safe weight' limit. From the literature of others, two general factors were identified as possibly dictating what a reasonable limit for a single lifting effort might be.

Firstly, it was recognized that the lifting capacity of an individual would be partly a function of the person's muscle strength. It was believed that this statement could be easily tested by comparing the torques developed by a person performing a lifting task, with the maximum voluntary torques obtained from tests of specific muscle groups. Secondly, it was foreseen that the limit to how much a person could lift would be highly dependent on the strengths of various bone and ligament structures of particularly, the lumbar spine. If these latter structures were the cause of the performance limit, then the determination of lifting limits could not be based solely on *in vivo* tests, but instead would have to be statistically inferred from test results using cadaver spines, and clinical data. From the preceding discussion, the following investigation was structured.

A group of 13 volunteers (10 males and 3 females) who were not specifically oriented to weight lifting were employed in the following tests: (Their anthropometric data and lifting performances are presented in Table 1.)

(1) Procure anthropometric data, (i.e. age, weight, stature, and link lengths).

(2) Test each subject's voluntary strength for the seven isolated muscle group actions described earlier.*

(3) Have each subject perform a maximum lift with legs and back while holding handles attached to load cells secured

Table 1. Anthropometric data on industrial sample

| Subject number | Sex | Age (yr) | Weight (kg) | Link lengths (cm): Wrist-to-hand C.G. | Forearm | Upper arm | Shoulder-to-hip | Thigh | Shank | Height (cm) |
|---|---|---|---|---|---|---|---|---|---|---|
| 200 | M | 30 | 73 | 8·0 | 26·7 | 29·9 | 42·9 | 47·8 | 45·5 | 176 |
| 300 | M | 32 | 66 | 9·9 | 27·3 | 30·4 | 40·7 | 49·6 | 47·9 | 180 |
| 500 | M | 34 | 89 | 8·3 | 29·9 | 32·7 | 42·2 | 50·0 | 48·5 | 182 |
| 700 | M | 31 | 88 | 8·4 | 28·9 | 31·8 | 40·1 | 49·6 | 47·9 | 178 |
| 800 | M | 29 | 85 | 8·4 | 28·5 | 31·4 | 40·7 | 47·5 | 45·1 | 134 |
| 1000 | M | 24 | 68 | 7·6 | 27·4 | 30·5 | 42·2 | 47·0 | 44·5 | 173 |
| 1100 | M | 27 | 71 | 7·8 | 26·5 | 29·7 | 42·6 | 48·0 | 45·9 | 178 |
| 1400 | M | 27 | 75 | 7·9 | 31·4 | 34·1 | 47·7 | 52·1 | 51·3 | 191 |
| 1500 | M | 32 | 76 | 9·0 | 29·5 | 32·4 | 38·3 | 47·4 | 45·0 | 171 |
| 1800 | M | 28 | 58 | 7·1 | 25·1 | 28·3 | 48·5 | 42·9 | 39·1 | 167 |
| 1900 | F | 21 | 66 | 7·4 | 23·2 | 26·7 | 39·1 | 44·1 | 40·7 | 162 |
| 2000 | F | 43 | 50 | 6·5 | 23·7 | 27·1 | 34·9 | 43·3 | 39·6 | 153 |
| 2100 | F | 29 | 51 | 7·3 | 25·5 | 28·7 | 33·7 | 48·0 | 45·8 | 169 |

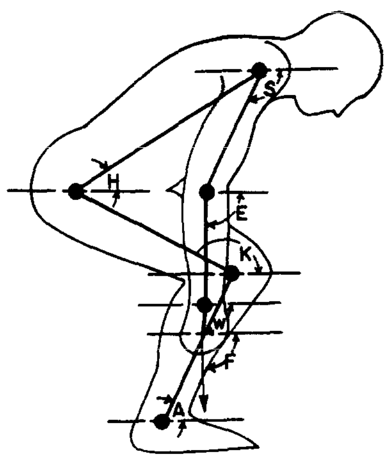
*The test of hip extension strength is performed with the subject's pelvis firmly supported by a rigid chair with a lap belt, thus eliminating strain on the lumbar spine during the test.

to the floor, and at the point of maximum effort take a lateral photograph to enable the measuring of the body position, as depicted in Fig. 4.

The data from the above procedure was used in the SSP model, which computed estimates of both the major articulation torques and the compressing and shearing forces at the L4/L5 and L5/S1 disc vertebral body end-plates during the lifting tasks.

When the articulation torques produced during the lifting task were compared to the maximum voluntary torques determined by the isolated muscle group strength tests, it was found that the predicted strength limits were not exceeded or equalled during the lifting task. In fact, the knee extension torque during the task came the closest, as it ranged from 73 to 95 per cent of the tested maximum voluntary knee extension torque. The hip extension torque during the lifting task was found to range from 18 to 75 per cent of the subject's maximum voluntary hip extension torque as determined by the prior strength tests. It was concluded from these results that for this particular lifting task the subjects were not limiting their performance due to the muscle group strengths that were tested.

An analysis of the compressing and shearing forces at the lower lumbar spine during the




| | Body angles (deg) | | | | | | Hand force | |
|---|---|---|---|---|---|---|---|---|
| Subject number | 'A' | 'K' | 'H' | 'S' | 'E' | 'W' | 'F' (deg) | Mag. (lb) |
| 200 | 58 | 142 | 44 | −111 | −96 | − 93 | −91 | 215 |
| 300 | 48 | 130 | 63 | −104 | −88 | − 93 | −93 | 240 |
| 500 | 49 | 127 | 55 | −107 | −99 | − 97 | −95 | 229 |
| 700 | 58 | 129 | 48 | −107 | −93 | − 95 | −88 | 288 |
| 800 | 50 | 129 | 66 | 100 | −92 | − 88 | −97 | 250 |
| 1000 | 53 | 144 | 48 | −105 | −94 | − 93 | −98 | 150 |
| 1100 | 57 | 140 | 40 | −117 | −97 | − 97 | −98 | 137 |
| 1400 | 50 | 138 | 57 | −102 | −94 | − 95 | −94 | 182 |
| 1500 | 51 | 128 | 73 | − 98 | −90 | − 94 | −94 | 300 |
| 1800 | 69 | 135 | 49 | −118 | −89 | − 83 | −91 | 193 |
| 1900 | 41 | 126 | 70 | − 96 | −79 | − 90 | −91 | 144 |
| 2000 | 73 | 130 | 48 | −103 | −94 | −100 | −91 | 110 |
| 2100 | 71 | 137 | 55 | − 96 | −96 | − 96 | −90 | 166 |

Fig. 4. Positions assumed during maximum lifting task.

lifting task, as predicted by the SSP model, disclosed that the subjects appeared to limit their compressive forces to a fairly constant magnitude, regardless of their potential hip and knee extension strengths. A plot of the compressive forces estimated to exist at the superior surface of the sacrum is displayed in Fig. 5. Similar results but slightly lower were found for the disc/L5 inferior surface, disc/L5 superior surface, and disc/L4 inferior surface.

The magnitudes of the shearing forces at these segmental levels were found to be relatively small, never reaching greater than 50 kg.

## DISCUSSION OF SSP MODEL AND LIFTING TASK

The use of the SSP model has disclosed that, in at least one particular type of weight lifting circumstance, the strengths of a group of major muscles (besides the back muscles) do not appear to dictate the whole-body lifting capacity. It is proposed that either the limited muscle strength of the back extensors, or the

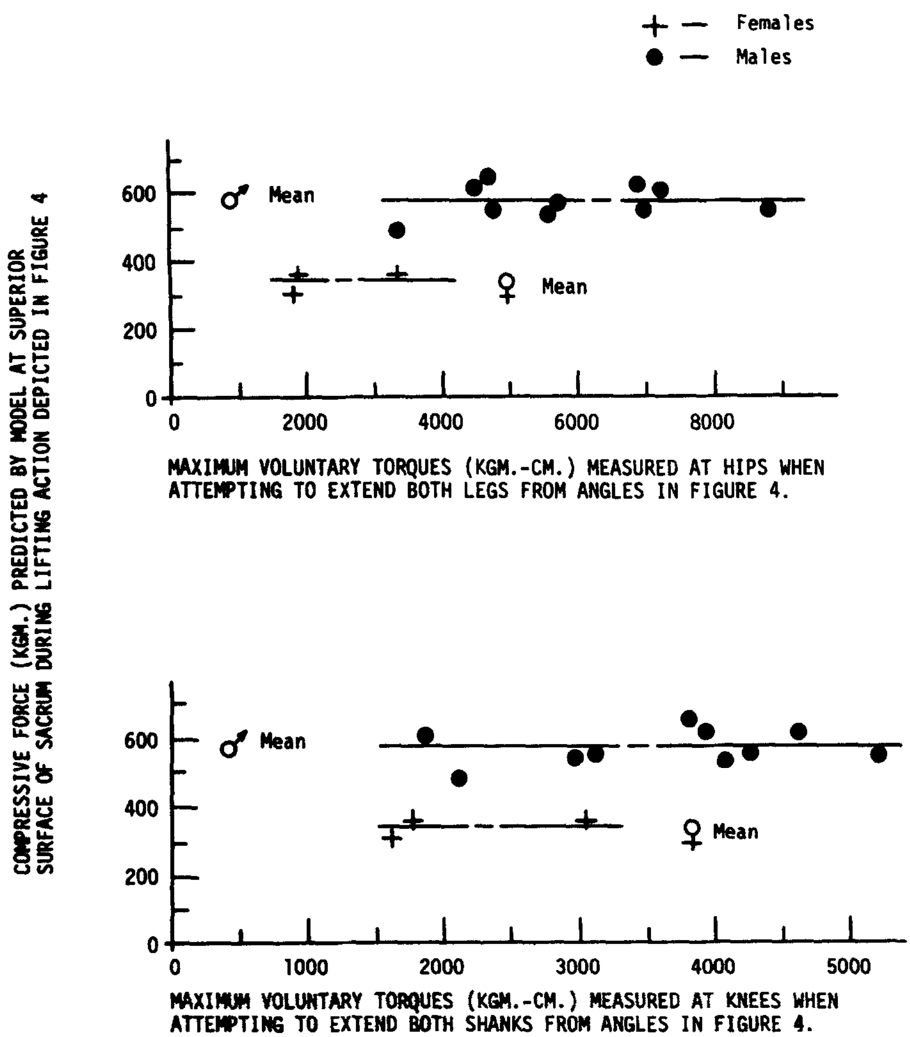



Fig. 5. Lumbar spine compression forces vs. knee and hip strengths.

sensing of possible injury to the vertebral column due to excessive compressing forces, inhibits the lifting action.

Further justification for the lifting performance being restricted by this latter mechanism, is given by the results of the cadaver spine testing of Sonada (1962), which disclosed that the young male lumbar spinal column can sustain compressive loads averaging 730 kg. However, both age and degeneration of the discs have been shown to reduce this value. In fact, one study by Evans (1959) produced a lumbar spine fracture with as low as 277 kg of compression loading. By testing maximum pull strengths, Troup (1968) estimated 527 kg as the average maximum compression force at the lumbosacral level for young, physically fit, male adults, assuming a 25 per cent abdominal pressure assistance. The SSP model has a pressure effect that averaged 11 per cent for the positions and torques achieved in the lifting task, thus accounting for the difference in these results. Sonada (1967)

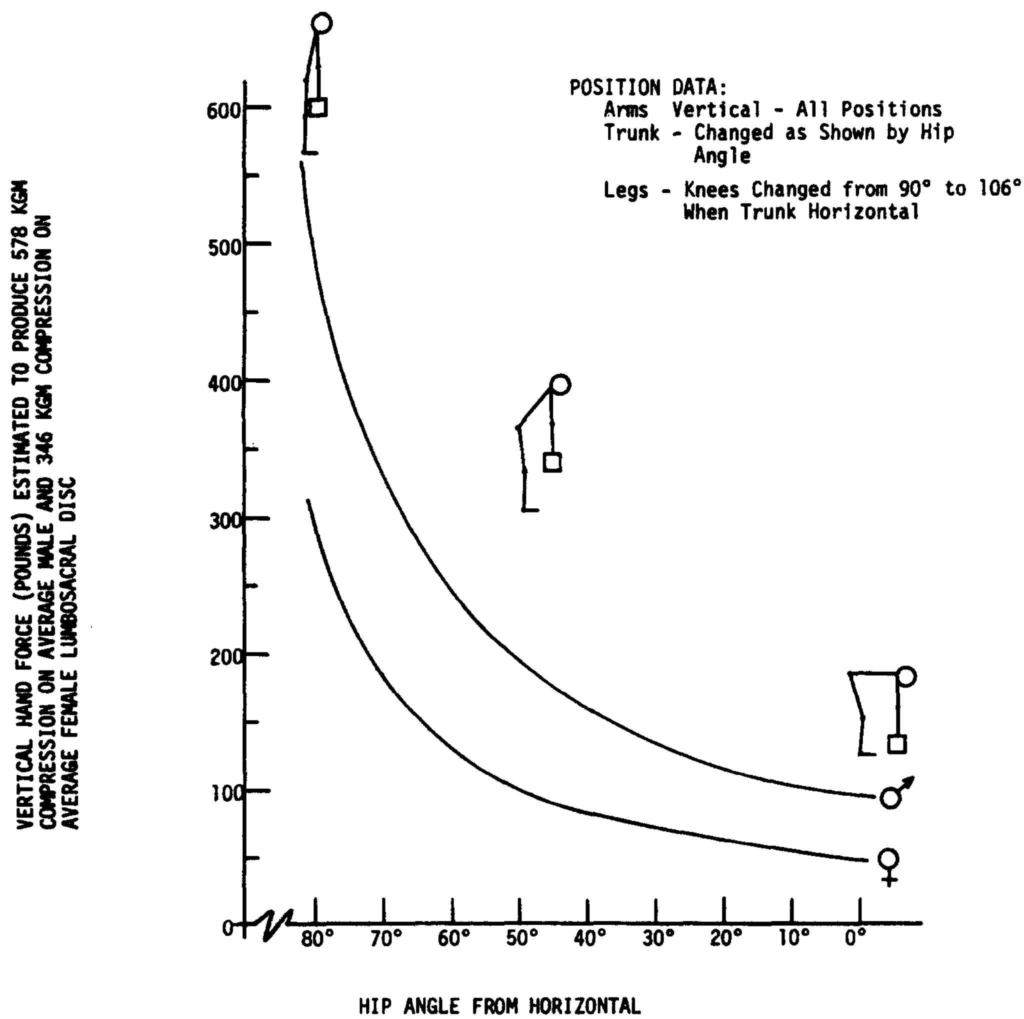



Fig. 6. Weight held in different body positions which produce constant compression forces on lumbosacral disc.

proposed that due to the smaller female vertebral body, a 'safe' maximum compressive force must also be smaller. This was also suggested by Chaffin (1968), from analysis of X-ray dimensions of low-back injuries.

Quite obviously, the average value of the spinal compressive forces estimated for the small sample of individuals employed in the lifting task described in this paper cannot be used at present as a recommended performance limit, though the compatibility of the results with others is encouraging. The fact that the individuals voluntarily chose widely varying body positions and produced a large range of lifting forces, but in so doing achieved relatively constant compressive force estimates on the lumbar spine and muscle tensions in the lumbar back muscles (regardless of leg strength) suggests that (1) the performance on the lifting task was dictated by the stresses on the lumbar back, (2) a healthy population which is not trained specifically for lifting is limited to a relatively constant lumbar stress capacity, and, (3) a female population may be limited to about 2/3 the stress withstood by the male lumbar back.

The use of the SSP model in analyzing different conditions has not been great enough to conclude that the preceding statements are absolutely correct, but it does provide a methodology for studying the effects on performance of such factors as body size, body position during materials handling tasks, sex, age, muscle group capacity and possibly back strength characteristics.

As an illustration of the effect of two of these factors on lifting performance, (i.e. sex and body position), Fig. 6 is presented. In developing this illustration, the hip and knee angles were discretely varied, as indicated, and at each chosen position the load on the hands was increased in one pound increments from zero until the compressive force limits of 578 kg and 346 kg for the male and female lumbosacral disc, respectively, were reached. The curves developed from this procedure clearly indicate the relative performance limitations that result from the suggested lumbar spinal column's capacity when one stoops-over to lift or carry an object.

It is proposed that only through the development of still more complex biomechanical models can the complex behavior of the musculoskeletal system be understood and used as a basis for equipment and methods design. The digital computer makes this a feasible alternative. As such, computer biomechanical models serve to give a total structure to the many variables which have and are being shown to affect the gross functions of the complex musculoskeletal system.

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## APPENDIX

Referring to the lifting activity depicted in Fig. 2, with average male dimensions, the torque resulting at the hip ($M_H$) is 2823 kg-cm, due to the weight of the trunk, neck, head, and arms, ($BW$) and the force on the hands ($FH$). The abdominal pressure ($P_{ABDOM}$) is predicted, based on this hip torque and the angle between the trunk and thighs, by the empirical relationship:

$$P_{ABDOM} = 10^{-4}[0{\cdot}6516 - 0{\cdot}005447\,(70°)]\,[2823]^{1{\cdot}8}$$

$$P_{ABDOM} = 44 \text{ mm Hg} = 0{\cdot}0598 \text{ kg/cm}^2$$

which gives:

$$F_{ABDOM} = (P_{ABDOM})(\text{Diaphram Area}) = (0{\cdot}0598 \text{ kg/cm}^2)(465 \text{ cm}^2)$$

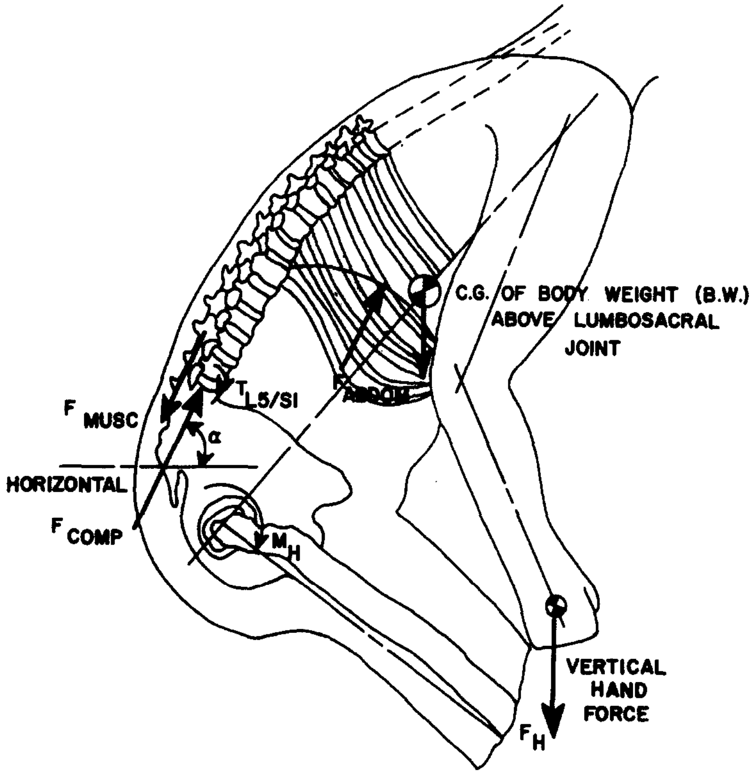
$$F_{ABDOM} = 27{\cdot}3 \text{ kg.}$$

For the position depicted in Fig. 2, it is estimated that the abdominal force ($F_{ABDOM}$) line of action is 9·1 cm anterior to the center of the lumbosacral disc. This, in turn, produces a torque of 248 kg-cm, which assists in extension of the trunk, thus reducing the compression forces on the spine that result from both the contraction of the erector spinae muscles and the combination of the body weight and 100-lb force acting on the hands.

The torque at the lumbosacral disc ($T_{L5/S1}$) due to the weight of the body above the disc and the 100-lb hand force, minus the torque created by the abdominal pressure, is estimated to be 2578 kg-cm. The component of this that is caused by the body weight and hand force is different than the hip torque because the disc location varies with the rotation characteristics of the pelvis and lumbar spine, as well as the trunk, neck, and head mass being less at the higher spinal levels.

The erector spinae muscle tension ($F_{MUSC}$) required to balance the torque at the lumbosacral disc is computed by dividing the torque $T_{L5/S1}$ by the assumed moment arm of the muscle group, (i.e. 5 cm) thus giving $F_{MUSC}$ = 516 kg.

The force of compression on the lumbosacral disc is found by balancing those forces at the disc that act perpendicular to the vertebral body end plates. Assuming the erect spinal column configuration depicted in Fig. 3,




and then rotating it and the pelvis as described earlier in the paper, it is found that the angle $\alpha$ at the superior surface of the sacrum is 66° for the task position indicated in Fig. 2. Thus, the force of compression ($F_{COMP}$) produced as a result of the combined forces acting at the lumbosacral disc is:

$$F_{COMP} = [(B.W. + V.H.F.) \div \sin\alpha] + F_{MUSC} - F_{ABDOM}$$

$$F_{COMP} = \left[\frac{(36+45)}{\text{Sin } 66^\circ}\right] + 516 - 27$$

$$F_{COMP} = 89 + 516 - 27 = 578 \text{ kg.}$$