INTERACTIVE BIOMECHANICAL COMPUTERIZED MODEL FOR PREDICTING HUMAN STATIC STRENGTHS IN THE SAGITTAL PLANE

THOMAS M. BAINES

December 1975

This report was partially sponsored by a Traineeship Grant from NIOSH under Contract No. OH-00161-03.

orign
UMRØ181

TABLE OF CONTENTS

LIST OF FIGURES

List of Figures (cont.)

I. INTRODUCTION

Because of an increasing awareness of the importance of human factors in the design of man-machine systems, the use of computerized human motion-strength models has become increasingly common. Many of these models have been developed in the Engineering Human Performance Research Laboratory at the University of Michigan, and include both sagittal plane and three-dimensional models for both sitting and standing postures (1, 2, 6, 17, 18, 21, 25).

In spite of the fact that there are already a number of these models developed, it was felt that there was a need for a basic, sagittal plane model that was easy to operate by a job designer, for example, by interaction with a computer teletype. Also needed was a program that provided flexibility in the input of the required variable of body link lengths and weights, articulation strengths and body position.

Therefore, in response to these needs, this new model was created. This model evaluates a given body position and hand held load against three criteria: 1) Can the body maintain its balance under that load and with that position? 2) Are any muscle strengths exceeded under that combination of load and position? and 3) What magnitude of lumbo-sacral inter-vertebral disk compressive loadings are produced by this combination of load and position? The body for which this evaluation is performed is described by length, weight and strength parameters. The input of these parameters can be by individual subject description or percentiles. Differences between male and female subjects are taken into account also. The model evaluates "static" (less than 4 seconds) situations such as when one is holding a weight, or pushing or pulling on a non-moving container. These situations should be considered to be infrequently occurring (less than 15 per hour) thereby eliminating metabolism limited work situations. The model is also restricted to symmetric sagittal plane activities.

Therefore, with the output of this model, the job designer can more accurately determine whether or not a given level of exertion can safely be performed by a given person. It also helps determine job work requirements as they relate to lower back problems (4, 7, 8). This is done by comparing the L_5/S_1 intervertebral disk compressive force that the model computes for a given load and body configuration with disk failure data obtained from cadaver experiments.

Figure 1 shows the major components of one of the body configurations that the model can evaluate. The figure shows a subject in the squat position, lifting a given load. Also shown are the body "links" which are used to describe the subject's body.

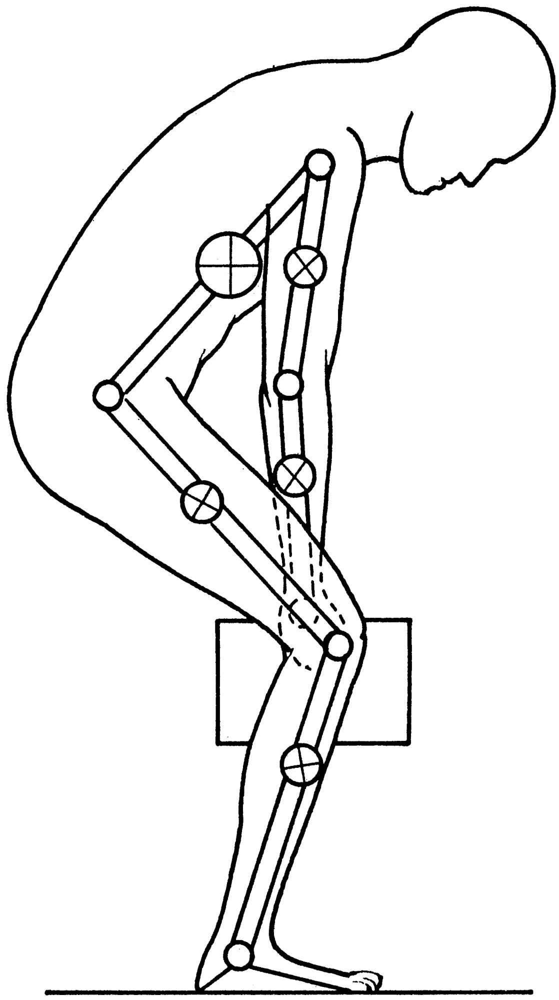


Figure 1: Diagram showing major components of sagittal plane model for squat mode, lift load.

II. PROGRAM DESCRIPTION

The macro logic of the computer program is illustrated in Figure 2, with the various input and stand modes listed in Figure 3. The program starts by requesting certain subject characteristics, depending on the input mode selected. The first of these characteristics deals with body dimensions and weight. They are used for computing link lengths, link weights and locations of link centers of gravity. The units used are centimeters and kilograms. A brief description of the link lengths and weights input mode is as follows:

1. Population percentiles. Individual link lengths are computed from population values for each individual link. Also computed from population data is subject stature and weight. The link weights are computed as a proportion of body weight, and centers of gravity are taken as proportion of link length. The specific population link length data and link weight and center of gravity data can be found in Appendix A.
2. Stature and weight. From the input values of subject stature and weight, the link lengths and weights and centers of gravity are computed. Link lengths and weights are computed as a proportion of stature and weight and centers of gravity are computed as a proportion of link length. The specific functions can be found in Appendix A.
3. Reference lengths, stature and weight. In this mode, the link lengths are computed as functions of three reference lengths: tibia length, radius length and stature. The link weights are proportions of body weight and centers of gravity are proportions of link lengths. See Appendix A for the specific functions.
4. Actual link lengths and body weight. This mode uses actual link lengths and computes link weights as a proportion of weight and centers of gravity as a proportion of link length.
5. Actual link lengths and weights. In this input mode, actual link lengths and weights are used. The link centers of gravity are computed as percents of link length.

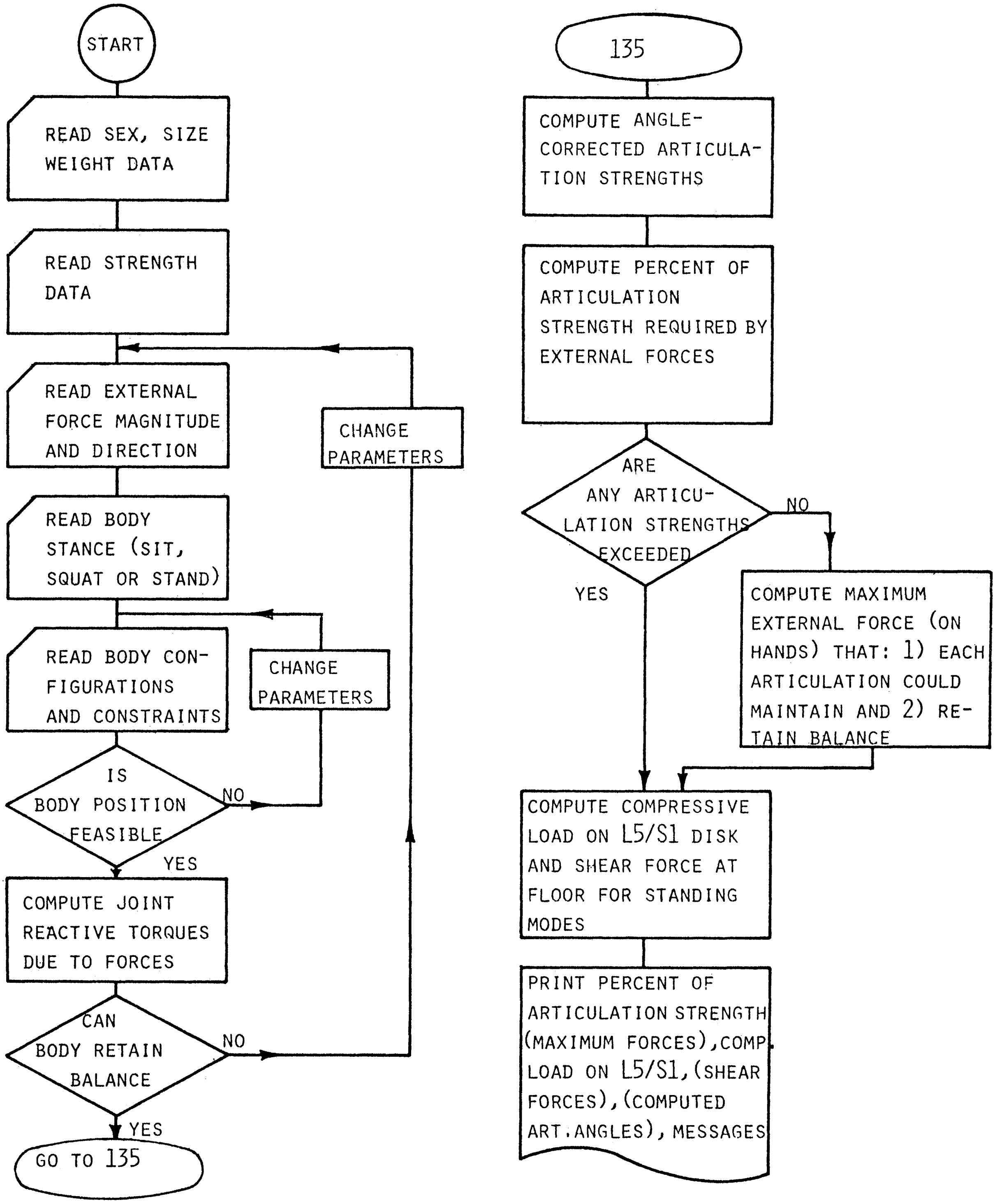


Figure 2: Macro Logic Flow Diagram.

I. BODY DIMENSIONS
   A. Link lengths and weights input modes
      1. Population percentile*
      2. Stature and weight
      3. Reference lengths, stature and weight
      4. Actual link lengths and body weight
      5. Actual link lengths and weights
   B. Articulation strengths input modes
      1. Population percentiles*
      2. Actual maximum voluntary torques at standard positions.

II. BODY STANCE MODES
   A. Sitting
   B. Squat
   C. Stand

III. BODY POSITION DESCRIPTION INPUT MODES
   A. Load position and hip angle
   B. All articulation angles

IV. L_5/S_1 DISK ANGLE INPUT MODES
   A. Actual angle
   B. Empirical equation calculation

* Male or Female Option

Figure 3: List of Program Input and Stance Modes

The next body characteristic that must be specified is articulation strength. These are stated as torques (kilogram-centimeters) at each articulation center. They represent maximum voluntary exertions calculated from data gathered under carefully controlled conditions and in standard positions. These maximum voluntary torques depend on a number of factors such as 1) body position, 2) individual characteristics such as health, training, technique, age, sex, etc., 3) motivation and other psychological factors and 4) fatigue level at time of exertion (17, 20).

The model uses body position and sex as variables, controls fatigue by limiting the applicability of the results to infrequent, short duration exertions and assumes the remainder of the strength factors as constants. The non-dominant arm has about 91% of the strength of the dominant arm (1). Therefore, the model assumes that each arm can exert 95.5% of the strength of the dominant arm. Also, the model assumes the arm strengths additive, which is valid if the trunk is adequately supported (1). Adequate support is tested in the model by a balance test, which will be discussed later. The two input modes for maximum voluntary torques are as follows:

1. Population percentiles. In this mode the maximum voluntary articulation torques for the subject can be determined from population data by specifying the percentile. The specific data is contained in Appendix A.
2. Actual maximum voluntary torques at standard positions. The program requires the input of actual strengths at each articulation for this mode. The standard positions are described in Section III.

Once the full description of body size, weight and strength characteristics has been done, the external forces must be entered. This involves stating whether the force that the subject is to exert is a push, pull (both horizontal) or lift (vertical). Also required is the force magnitude in kilograms. The model assumes that the load is balanced at the center of gravity of the hands and that there is no cantilever effect.

With the force magnitude specified, the model compares it with the subject's grip strength when lift and pull force directions are used.

This assumes that the subject is gripping the load with a firmly attached handle and that any force away from the hand that is greater than the grip strength would exceed the subject's capacity to grip the handle. This might approximate the real condition if the handle were able to revolve about its longitudinal axis. However, in fixed handle situations, the grip strength that results from muscle strength is augmented by friction, thereby increasing grip capacity. Therefore, when the theoretical grip is exceeded, the model provides two alternatives: 1) return and input a revised force magnitude, or 2) continue with the program thereby overriding the grip strength constraint.

The next variable to be specified is the body stance. This can be either sitting, squating or standing. The sitting push mode can be either with or without back support. If back support is specified, the model assumes an infinite ability to retain body balance (i.e., a high back to the seat) when pushing a load. This, in turn, means that the torso flexion strength would be infinite under this condition.

The seated mode assumed that the seat upon which the subject is sitting is slightly padded (and therefore the compression is negligible) and inclined at 8° from horizontal. However, thigh muscle compression on the seat results in a horizontal thigh bone. Also assumed is an included knee angle of 120° and a firm support for the foot to exert force on. Figure 4 shows this graphically.

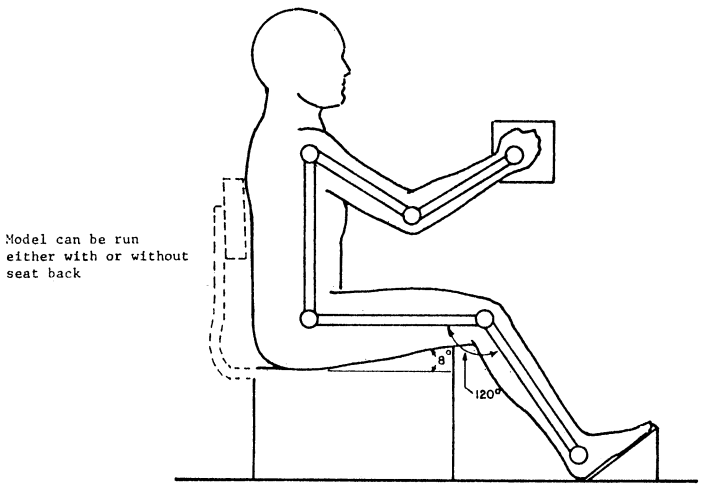


Figure 4: Graphical representation of seated configuration and assumptions

The squat position assumes a flat foot stance and an included knee angle of less than 180° as in Figure 5. A stability factor is permitted wherein body balance can be maintained over a wider range of forces and body configurations. The stability factor is represented as the distance between the ankles (talus) and is limited to one foot length. This is further explained in Appendix B-3. The standing position is a special case of the squat case wherein the included knee angle is set at 180°.

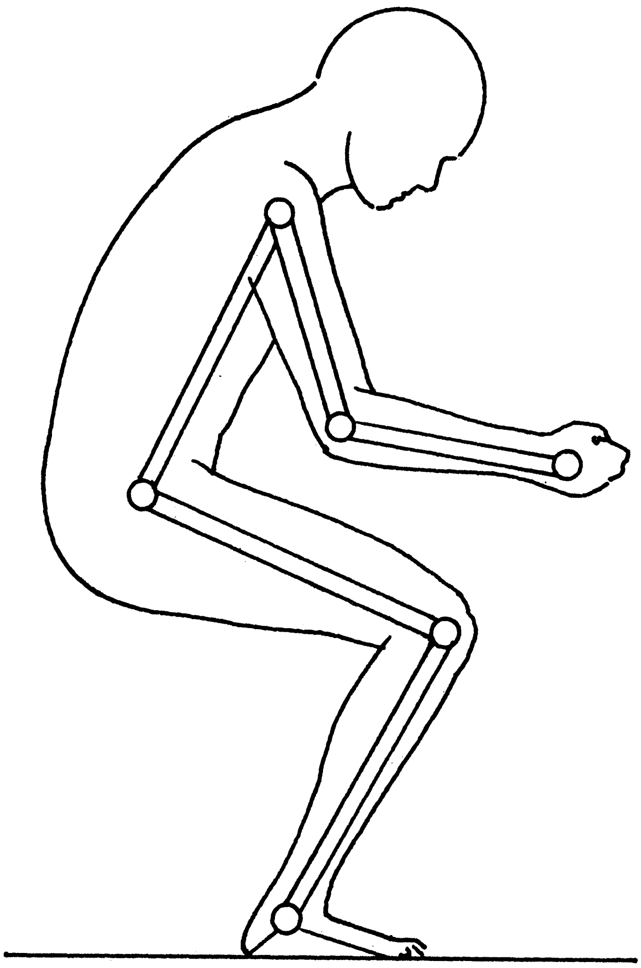


Figure 5: Graphical representation of squat configuration

Following the specification of the body stance, the body configuration must be identified. The two modes for this are as follows:

1. Load position and hip angle. In this mode the X and Z coordinates of the load must be read into the program in addition to the hip angle (see Figure 6). Then, on the basis of trigonometric relations the remainder of the angles are computed. A check is then made as to whether or not the load can actually be reached with the variables inputted. In the standing or squat modes, the ankle angle is computed from an empirical relationship. In the standing case, ankle angle is computed as a function of hip angle whereas for the squat case, it is computed as a function of squat height and load position (see Appendix B-1).

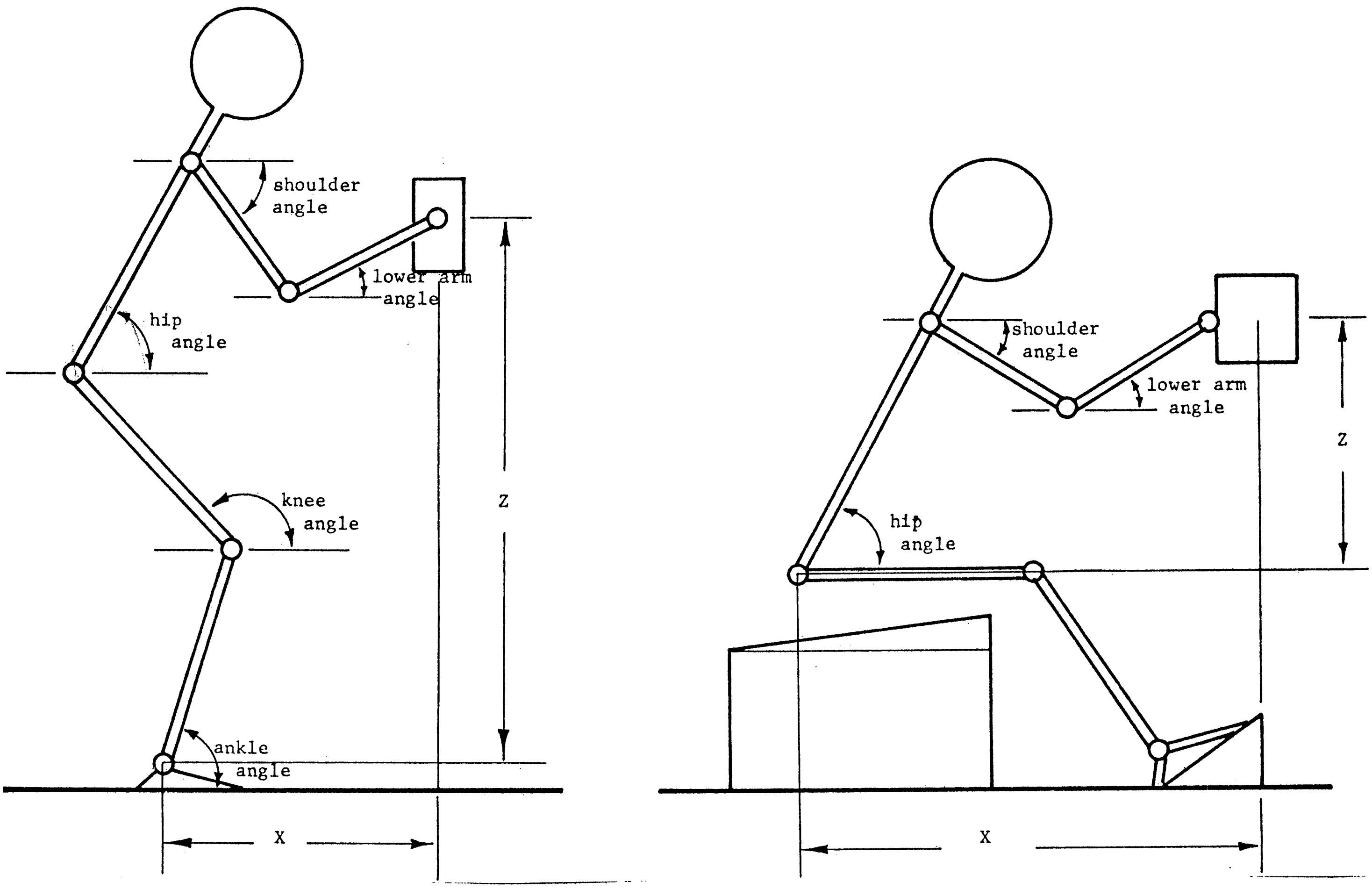


Figure 6: Body Configuration Definitions

2. <u>All articulation angles</u>. In order to fully specify body stance in this mode, all body angles are required as input. Each input angle is checked to be sure it is within the specified limits of motion. See Appendix B-6 for these limits.

The pelvic angle is defined as the angle that the pelvis makes with the horizontal (see Figure 7.) The angle is measured at the hip point and is measured from horizontal to a line extending from the hip to the center of the superior surface of the S1 vertebrae. When the load position input mode is used to describe body position, this angle is computed from an empirical relationship. However, when the articulation angles are used as input, the program user can choose between using the empirical relationship to compute the pelvic angle or input the angle directly.

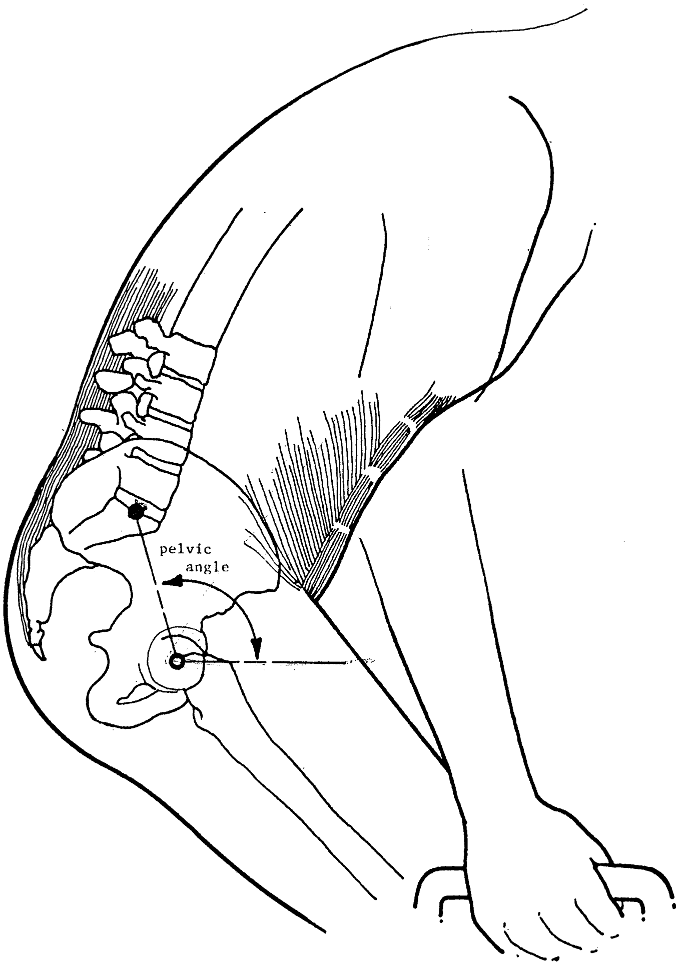


Figure 7: Pelvic (L_5/S_1 to hip) angle

With the body configuration defined, the model checks all of the included articulation angles to be sure that they are within the range of motion permitted of the joint. This then completes the body feasibility computation.

With the body fully dimensioned as to size, strength and position, the program proceeds to compute the reactive forces at each joint due to the forces acting on each link (the "articulation task torques"). These forces include gravity and the load. The exact procedure is outlined in Appendix B-2. Following these computations, the ability of the body to retain its balance under the given load and stance is calculated and forms the first criterion for the ability of the body to safely sustain a given load. This is done by comparing the reactive torques at the ankle and floor (squat and stand cases) or the hip and seat (sitting case). These computations are explained in depth in Appendix B-3.

Next, the model corrects the standard-position articulation strengths to take into account the effect of included angles that are other than the standard position. This is necessary because of the way muscles react to an externally applied force, i.e., by "pulling" across an articulation. The ability of the muscle to pull and thereby produce a torque varies with the included angle of the joint across which it is pulling. For example, when the upper arm is next to the torso, the lower arm is stronger in lifting when the included angle at the elbow is 90° than when it is fully extended (180°)(17). The model corrects the standard-position articulation strengths by a method which is fully explained in Appendix B-4. These corrected strengths are called maximum voluntary torques.

With these torques computed, the model compares the maximum voluntary torques with the articulation task torques, with the latter computed as a percent of the former. The results are then analyzed to see if any of the maximum voluntary torques are exceeded by articulation task torques expected of the joint by the external forces. This forms the second criterion for the ability of the body to sustain a given load.

If none of the maximum voluntary torques of the articulations are exceeded, then the model computes the maximum force that could be exerted by the hands and still not exceed the articulation's maximum voluntary torque. This is computed for each articulation. Since each articulation has a different maximum voluntary torque and task torque associated with it, these maximum hand forces will be different between the joints. Therefore, the body will be strength limited by the joint at which the lowest maximum hand held force occurs.

Also computed is the maximum hand held force that would still permit body balance to be retained. This must then be compared with the muscle strength limitation to determine what hand force magnitude the body is capable of for the given load and position. Details of these maximum hand force computations can be found in Appendix B-3.

Following this, the compressive load on the L_5/S_1 intervertebral disk is computed which forms the third criterion for hand held loads. If this and the other criterion are not exceeded, then the load and position specified would be acceptable for the subject of interest.

The disk compressive force is computed by the method described by Chaffin (1, 2, 5) with a few adjustments. These include the rearrangement of the abdominal pressure and abdominal force moment arm equations to reflect the model's use of radians and angular measure from horizontal. Also included in the model is the use of an adjustment of the back muscle moment arm. This is done by multiplying the assumed moment arm of 5 cm by the ratio of subject stature to a normalizing stature of 175 cm. This method of back muscle moment arm adjustment assumes that stature is a good predictor of back muscle depth by using the variation in population stature to correct for moment arm length. The underlying assumption is that the two are uniquely correlated, which is unproven as the author knows of no study verifying this. A comparison of the depth of the back muscle as determined from the data developed by Chaffin and Moulis (7) and the muscle depth as determined by the model is given in Table I.

Unfortunately, the criterion for maximum allowable disk compressive force is not as clearly defined as the balance or muscle force limitation criteria. Research indicates that much low back pain and injury is directly related to disk injury which could result from excessive disk

TABLE I: BACK MUSCLE DEPTHS (cm) AS DETERMINED FROM CHAFFIN AND MOULIS (7) AND AS DETERMINED BY MODEL.

	PERCENTILE		
	95th	50th	5th
MALE			
Actual muscle depth (7)	4.53	5.00	5.47
Model muscle depth	4.70	5.00	5.30
FEMALE			
Actual muscle depth (7)	4.24	4.80	5.36
Model muscle depth	4.35	5.00	4.91

compressive forces (8). However, the limits of compressive forces that the disks can withstand are very difficult to determine with accuracy.

When an intervertebral disk is subjected to a compressive load, the vertebral component that usually yields first is the end-plate (15, 24, 25). Several researchers have experimented with cadaver spines by loading them until they incur end-plate microfractures or other signs of disk-vertebrae breakdown. The results of work done by several researchers (14,15, 24, 26) are shown in the histogram in Figure 8. This histogram shows that an average of 592 kilograms of compressive force was required to produce recognizable end-plate fractures in test vertebrae. However, as the histogram also shows, there is a large variation in the data. This makes it hard to establish a fixed criteria for maximum disk compressive force.

Several things should be noted regarding the data presented in Figure 8. For example, all of the data presented is the result of testing various levels of lumbar disks, not just the L_5/S_1 disk that is of primary interest in the model. Inspection of the data indicates that there is no significant difference between the test data from any of the lumbar disk levels. Therefore, all of the lumbar data is presented in order to obtain a larger sample size. Also, it should be noted that all of the data is from male cadavers. Sonada (26) stated

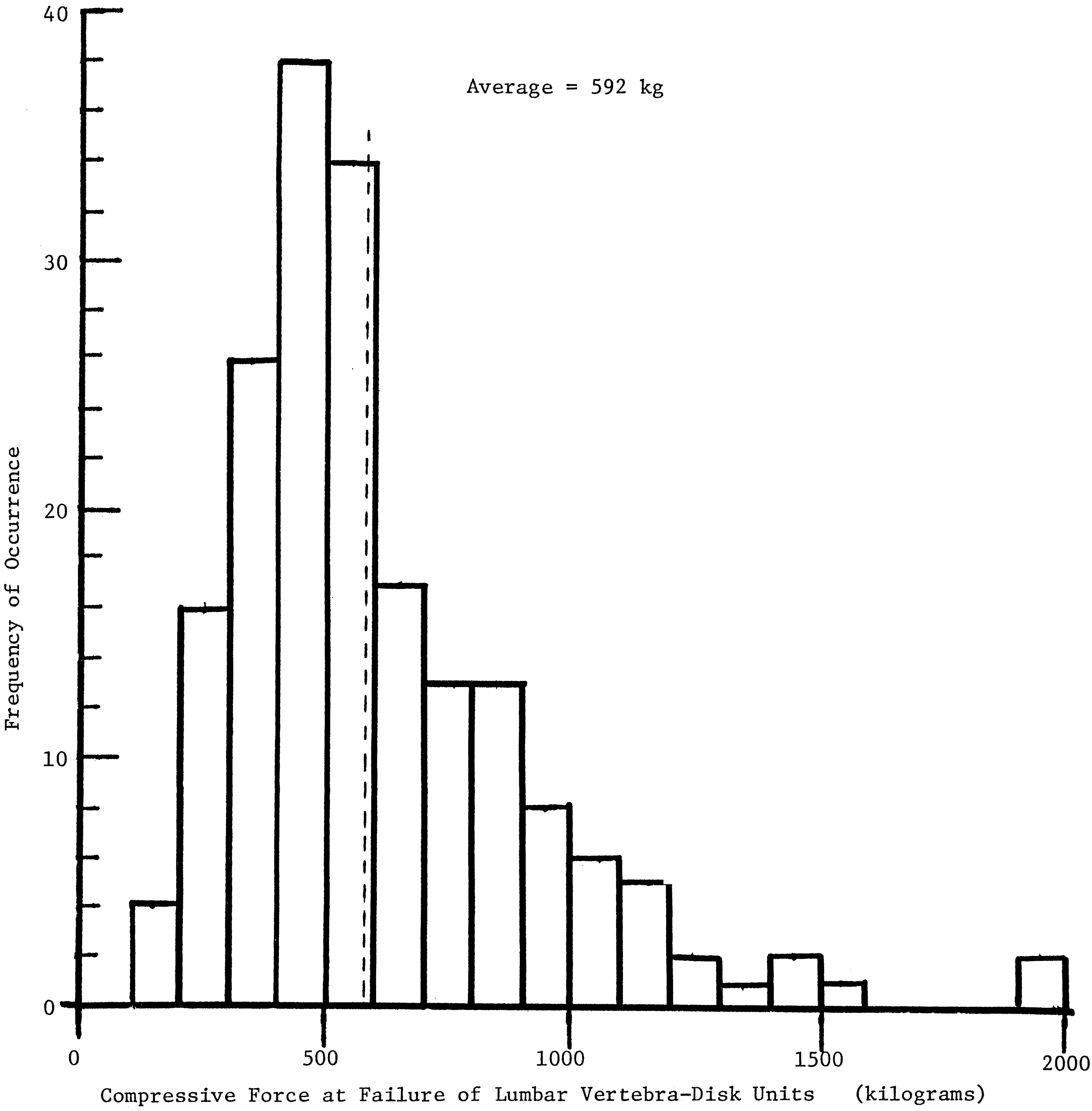


Figure 8: Histogram of Compressive Forces Required to Produce Failure in Lumbar Vertebra - Disk Units (14, 15, 24, 26)

that no significant difference in strength per unit area existed between women and men. This indicates that any difference in strength would be due to differences in surface areas of the vertebral bodies. For this reason, Sonada (25) estimates that the fracture load for women is about 5/6 of that for men.

Another source of variation that exists is an effect due to age. Figure 9 shows histograms for various age groups of test cadavers. This shows that there is a very definite trend with age, with the vertebral bodies from younger cadavers withstanding more compressive loads than those from older ones.

These and other sources of variation must be considered when analyzing these data. Also, the relationship between actual, live disks and cadaver disks should be analyzed as to the applicability of the test data. The end result is that the best judgment of the model user must be carfully employed in order to develop a criterion for the maximum compressive force which the lumbo-sacral region of the back can withstand. The model assists in this endeavor by computing the compressive force which would be generated by a certain load and body configuration.

Since the model does not compute maximum disk compressive force for a given load and body configuration, the model user must run the model at the maximum load in order to obtain the appropriate disk compressive force. For example, the user may input a given body description, body configuration and a hand held force of 5 kilograms. The model could respond by stating that the body is strength limited at the shoulder by 52 kilograms. In order to evaluate the disk compressive force, the user must re-run the model with the same input information but a load of 52 kilograms in order to compute the corresponding disk compressive force.

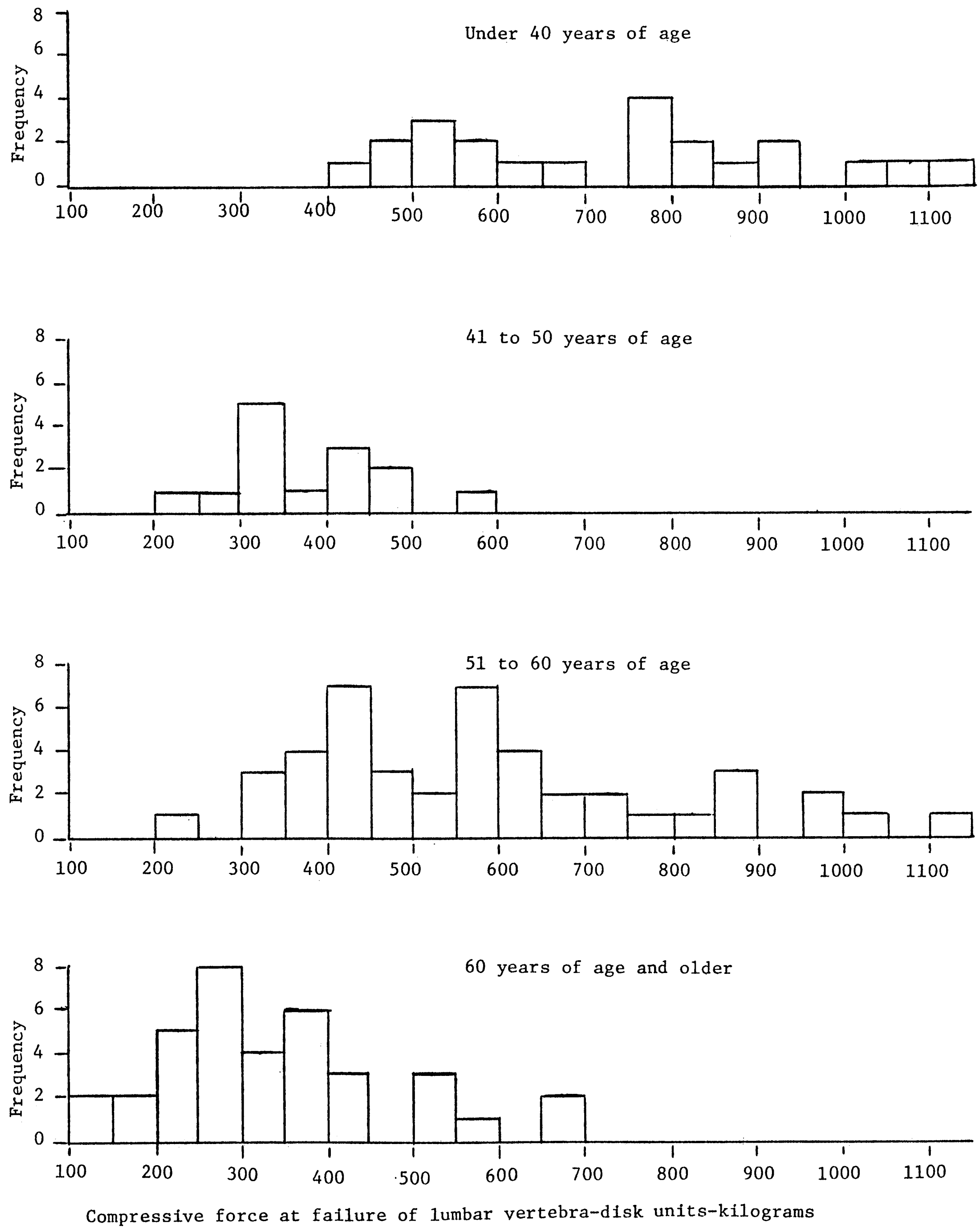


Figure 9: Histogram of Compressive Force Required to Produce Failure in Lumbar Vertebra - Disk Units Showing Age Dependency.

III. INPUT AND OUTPUT MODES

The information needed for input to the model is as follows:

I. Body Dimensions

A. Link Lengths and Weights Input Modes

1. Population Percentiles

The "Z" value corresponding to the percentile of the group of interest is needed for this input mode. Definitions of the percentiles and some "Z" values are as follows:

Percentile	Description	Z Value
95	Small and/or	1.65
90	weak	1.28
75		0.68
50	Average	0.00
25		-0.68
10	Large and/or	-1.28
5	strong	-1.65

2. Stature and Weight

Overall body stature in centimeters measured from the top of the head to the bottom of feet is needed as input for this mode in addition to total body weight in kilograms.

3. Reference Lengths, Stature and Weight

This in-mode requires stature and body weight as described above in addition to two reference lengths: the radius and tibia. The reference lengths are determined as follows:

RADIUS - Proximal end point is the palpable sulcus between the circumference of the radius and the capitulum of the humerous at the humeroradial articulation in the dimple posterolateral to the elbow. The distal end point is the palpable tip of the radial styloid process at the lateral border of the wrist.

TIBIA - The proximal end point is the palpable anterior border of the lateral condyle lateral to the patellar ligament; essentially a lateral displacement of the tibiale of the medial condyle. The distal end point is the palpable distal extremity of the medial malleolus.

These measurements should be in centimeters (13).

4. Actual link lengths and body weight

The body weight used as input is the same as defined above. The link lengths needed are wrist to hand center of gravity, lower arm, upper arm, trunk (shoulder to hip), thigh and shank. The measurements are taken over the body in the manner described by Dempster (11). The wrist to hand center of gravity is defined by Dempster (11) as follows:

WRIST AXIS LOCATION - On the palmar side of the hand, the distal wrist crease at the palmaris longus tendon, or the midpoint of a line between the radial styloid and the center of the pisiform bone; on the dorsal side of the hand, the palpable groove between the lunate and capitate bones, on a line with metacarpal bone III.

CENTER OF GRAVITY OF THE HAND - A point on the skin surface midway in the angle between the proximal transverse palmar crease and the radial longitudinal crease in line with the third digit; flattening or cupping the hand changes the relative location of this point very little, except to change the position normal to the skin surface.

All length measurements should be in centimeters.

5. Actual Link Lengths and Weights

This input mode uses the same actual link lengths as described above. The link weights are determined as described in Williams and Lissner (29) and should be in kilograms.

B. Articulation Strengths Input Modes

1. Population Percentiles

   The input for this mode is the "Z" value corresponding to the group of interest. The percentiles are explained in A-1 of this section which also includes "Z" values.

2. Actual Maximum Voluntary Torques at Standard Positions

   The strengths measurements used as input in this mode should be made on the subject in the standard positions pictured in Figure 10. The measurements should be carefully taken under well controlled conditions. Further details about strength testing can be found in Chaffin (3) and Martin and Chaffin (21).

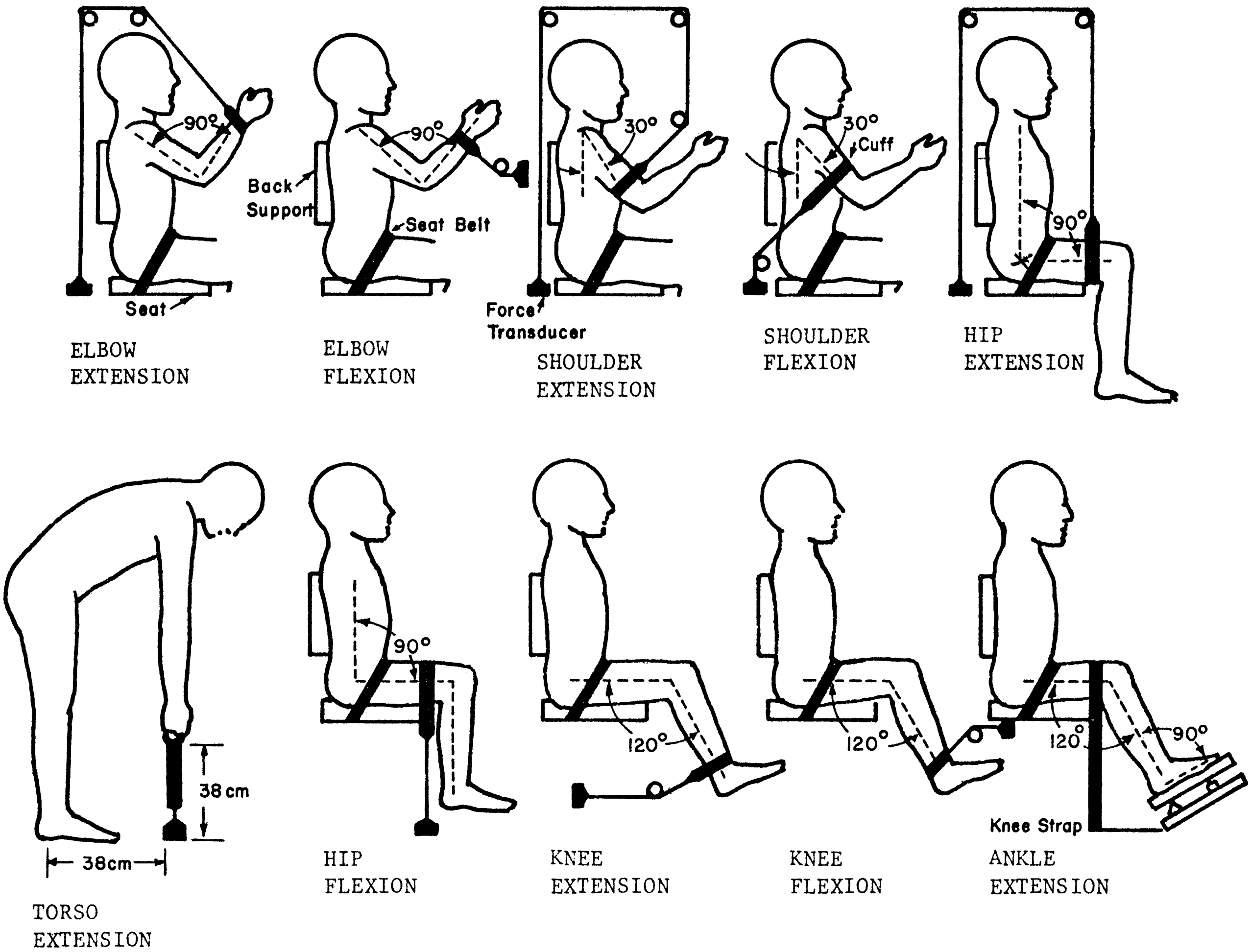


Figure 10: Standard Positions for Strength Measurement

II. Forces

A. Force Magnitude

The magnitude of the force that the subject is exerting must be specified in kilograms.

B. Force Direction

The direction of the force is requested as input at this point. The three force directions are push, pull and lift. They are restricted to horizontal and vertical forces, precluding exertions at intermediate angles.

III. Body Stance

A. Seated

The seated push position has two options available: with or without back support.

B. Squat

The squat position (knees bent) also has a stability factor which must be input. The stability factor is a number between 0 and 1 which represents the fraction of a foot length between the ankles as they are displaced in-front-off and behind the place where the ankles would be if the person were standing with the feet together. This is further explained in Appendix B-3 which deals with body balance.

C. Stand

The stand position (knees straight) has the same stability factor as described above (III-B).

IV. Body Position Description

A. Seated

1. Load Position

There are two basic data which must be input to describe the body position:

a. Load position - this is defined as: 1) X - horizontal distance from the hip to the load, and 2) Z - vertical distance from the hip to the load. The "Z" can be negative but when it is, the "X" must be longer than the thigh to prevent interference.

b. AHSHZ - angle at hips measured in degrees from the trunk to horizontal. This angle is limited to greater than -15° or less than 130°. Angles below horizontal are negative. In this mode and the others, a manikin can be of assistance in determining angles.

2. Articulation Angles

This input mode requires as input data the hip (torso to horizontal), shoulder (upper arm to horizontal) and elbow (lower arm to horizontal) angles in degrees. Angles below horizontal are negative. These angles are checked for range of motion (see Appendix B-6). Also available is the choice between the computation of pelvic angle from an empirical equation (Appendix B-1) or as direct data input.

B. Squat

1. Load Position

The data needed to describe the body position are as follows:

a. Load position - "X" is defined as the horizontal distance from the talus to the load. "Z" is the vertical distance from talus to load. Both "X" and "Z" are restricted to positive values.

b. AHSHZ - The hip angle (torso to horizontal) in degrees greater than -15° and less than 130°. Angles below horizontal are negative. The ankle angle is computed from an empirical relationship (Appendix B-1). The choice is given between an empirical calculation (Appendix B-1) for pelvic angle, or direct input of the angle.

c. Squat height - The distance, in centimeters, between the talus and hip.

2. Articulation Angles

The data required as input for this mode includes ankle angle (shank to horizontal), knee angle (thigh to horizontal), hip angle (torso to horizontal), shoulder angle (upper arm to horizontal) and elbow angle (lower arm to horizontal), all in degrees. Angles below horizontal are negative. All angles are checked for range of motion (Appendix B-1) and the choice is given between direct pelvic angle input or its computation from an empirical equation.

C. Standing

These input modes are just like the squat case above with the exception of no squat height or knee angle inputs.

The input formats are given in the printed messages that prompt the input data. These formats are Fortran IV designations and are for both integers (no decimal point to be used) and real numbers. It is very important to use the decimal point with real number inputs. Also important is the use of commas where called for.

OUTPUT

Figures 11 through 13 are example runs of the model. These figures show not only some of the prompt messages and input data but the output format. Several examples are given that reflect some of the model's capabilities.

The output of the model is for the most part self-explanatory. However, some details should be pointed out. For example, since trunk flexion strength data is not available, no comparison can be made between task torques and voluntary torque. However, trunk flexion task torque is printed out to provide the model user with its magnitude. The percent of maximum for the trunk flexion is printed as "-0.0" to draw attention to the fact that no comparison can be made. The only exception to this is when there is back support in the seated push case. Here "0.0" is assigned to the percent of maximum voluntary torque because the torso flexion strength in this case is assumed infinite.

#S16 P/U ~~RUN~~ 2RTA: BACK

p camp

```
#EXECUTION BEGINS
INPUT-FLEXABLE BIOMECHANICAL MODEL FOR TERMINAL USE/SITTING/SQUAT/ OR

STAND POSITIONS/PUSH,PULL OR LIFT LOADS

SELECT DATA INPUT MODE/PERCENTILES(ENTER "1"),STATURE AND WEIGHT

(ENTER"2"),WEIGHT AND REFERENCE LENGTHS(ENTER"3"), BODY WEIGHT AND

ACTUAL LINK LENGTHS(ENTER"4"), OR, ACTUAL LINK WEIGHTS AND LENGTHS

(ENTER"5"). USE I1 FORMAT
1

ENTER Z VALUE FOR STATURE(F5.2)
-1.28

ENTER Z VALUE FOR WEIGHT(F5.2)
-1.28

ENTER SEX OF SUBJECT (MALE="1",FEMALE="2")I1 FORMAT
1

SELECT ARTICULATION STRENGTH INPUT MODE. PERCENTILES(ENTER"1")
ACTUAL STRENGTHS(ENTER"2")FORMAT I1
1

ENTER "Z" VALUE FOR STRENGTH (F5.2)
-1.28

ENTER FORCE MAGNITUDE IN KILOS (F5.1)
45.

ENTER FORCE DIRECTION-PUSH("1"), PULL("2"), LIFT("3"). I1
1

ENTER BODY STANCE. SEATED("1"), SQUAT("2"), STAND("3") I1
1

WITH BACK SUPPORT("1") OR WITHOUT("2")? I1
1

SELECT INPUT MODE. LOAD POSITION("1") OR ARTICULATION ANGLES("2")-I1
1

ENTER X COORDINATE (CM)-POSITIVE VALUE ONLY-F4.1
25.

ENTER Z COORDINATE(CM)-F4.1
30.
ENTER ANGLE @ HIP  TRUNK/HORIZ (DEGREES) F5.0
90.

     BODY BALANCE AND MUSCLE STRENGTH NOT EXCEEDED

                   ART. TASK TORQUES(KG-CM) AND % OF ART. VOL. TORQUES
                         EL-FLX  EL-EXT  SH-FLX  SH-EXT  TR-FLX  TR-EXT
ART. TASK TORQUE          0.0    147.9   484.4    0.0    706.5    0.0
% ART.VOL.TORQUE          0.0     27.8    46.9    0.0     -0.0    0.0

MAXIMUM HAND FORCE MAGNITUDE FOR EACH ARTICULATION (KILOS)
    EL-FLX      EL-EXT      SH-FLX      SH-EXT      TR-FLX      TR-EXT
     0.0        135.7        96.9        0.0         -0.0        0.0
MAX HAND FORCE MAGNITUDE AND RETAIN BALANCE =999.9 KILOS
BODY LIMITED BY SMALLEST VALUE

FOR COMPARISON WITH HISTOGRAM(P.14)-COMP FORCE ON L5/S1 DISK FOR INPUT
HAND FORCE MAG=  0.0 KG      BACK MUSCLE FORCE=-135.0 KG

ARTICULATION ANGLES IN DEGREES
  SHOULDER ANG. (UA/HZ)=-107.5       ELBOW ANG.(LA/HZ)=  13.9

CONTINUE("1") OR STOP("2")?
```

Figure 11: Example Run of Model Using Population Data - Seated Stance, Push Force

```
SELECT DATA INPUT MODE/PERCENTILES(ENTER "1"),STATURE AND WEIGHT
(ENTER"2"),WEIGHT AND REFERENCE LENGTHS(ENTER"3"), BODY WEIGHT AND
ACTUAL LINK LENGTHS(ENTER"4"), OR, ACTUAL LINK WEIGHTS AND LENGTHS
(ENTER"5"). USE I1 FORMAT
2
ENTER SUBJECT STATURE IN CENTIMETERS (F5.1)
170.

ENTER SUBJECT WEIGHT IN KILOGRAMS (F5.1)
72.

SELECT ARTICULATION STRENGTH INPUT MODE. PERCENTILES(ENTER"1")
ACTUAL STRENGTHS(ENTER"2")FORMAT I1
2

INPUT FOLLOWING ACTUAL STRENGTHS IN KILOS-SEPARATE W/ COMMAS
- USE DECIMAL POINT - F6.1

ELBOW/FLEXION, EXTENSION
800.,520.

SHOULDER/FLEXION,EXTENSION
1200.,1100.

TRUNK EXTENSION
3900.

HIP/FLEXION,EXTENSION
1500.,2700.

KNEE/FLEXION,EXTENSION
800.,2200.

ANKLE EXTENSION STRENGTH
2300.

GRIP STRENGTH
52.5

ENTER FORCE MAGNITUDE IN KILOS (F5.1)
35.

ENTER FORCE DIRECTION-PUSH("1"), PULL("2"), LIFT("3"). I1
3

ENTER BODY STANCE. SEATED("1"), SQUAT("2"), STAND("3") I1
3
ENTER STABILITY FACTOR  0.0=<SF<=1.0 (F4.2)
0.5

SELECT INPUT MODE. LOAD POSITION("1") OR ARTICULATION ANGLES("2")-I1
1


ENTER X AND Z COORDINATES IN CM.-POSITIVE VALUES ONLY-
F4.1(SEPARATE W/COMMAS)
25.,70.
ENTER ANGLE @ HIP  TRUNK/HORIZ (DEGREES) F5.0
85.
ARMS DO NOT REACH LOAD
ENTER X AND Z COORDINATES IN CM.-POSITIVE VALUES ONLY-
F4.1(SEPARATE W/COMMAS)
25.,70.
ENTER ANGLE @ HIP  TRUNK/HORIZ (DEGREES) F5.0
45.

     BODY BALANCE AND MUSCLE STRENGTH NOT EXCEEDED
  A=ART. TASK TORQUES(KG-CM)    B=% OF ART. VOL. TORQUES
     EL-FX    EL-EX    SH-FX    SH-EX    TR-FX    TR-EX
A=   420.3      0.0     14.6      0.0      0.0   2096.9
B=    54.6      0.0      1.2      0.0      0.0     70.8

     HP-FX    HP-EX    KN-FX    KN-EX    AK-EX
A=     0.0   2397.6    943.2      0.0    646.3
B=     0.0     99.7     95.3      0.0     28.1

MAXIMUM HAND FORCE MAGNITUDE FOR EACH ARTICULATION (KILOS)
     EL-FX    EL-EX    SH-FX    SH-EX    TR-FX    TR-EX
     65.7      0.0    899.1      0.0      0.0     59.7

     HP-FX    HP-EX    KN-FX    KN-EX    AK-EX
      0.0     35.2     38.0      0.0    167.4
MAX HAND FORCE MAGNITUDE AND RETAIN BALANCE =150.0 KILOS
BODY LIMITED BY SMALLEST VALUE

FOR COMPARISON WITH HISTOGRAM(P.14)-COMP FORCE ON L5/S1 DISK FOR INPUT
HAND FORCE MAG= 392.0 KG      BACK MUSCLE FORCE= 35 .7 KG
SHEAR FORCE AT FLOOR=    0.0 KILOS
ARTICULATION ANGLES(DEGREES) AT JOINTS - LIMB TO HORIZ

               ELBOW     SHLDR       HIP      KNEE     ANKLE    PELVIS
               -47.2    -132.6      45.0      97.0      97.0     107.9

CONTINUE("1") OR STOP("2")?
```

Figure 12: Example Run of Model Using Stature, Weight and Actual Strengths as Input Data - Standing Body Stance, Load Position Defined.

```
SELECT DATA INPUT MODE/PERCENTILES(ENTER "1"),STATURE AND WEIGHT
(ENTER"2"),WEIGHT AND REFERENCE LENGTHS(ENTER"3"), BODY WEIGHT AND
ACTUAL LINK LENGTHS(ENTER"4"), OR, ACTUAL LINK WEIGHTS AND LENGTHS
(ENTER"5"). USE I1 FORMAT
4
ENTER SUBJECT WEIGHT IN KILOGRAMS (F5.1)
60.1        66
ENTER SUBJECT STATURE IN CM. (F5.1)
160.  161
ENTER FOLLOWING LINK LENGTHS IN CENTIMETERS(F4.1)
WRIST TO HAND CG
6.          7
LOWER ARM
23.         26.5
UPPER ARM
27.1        29
TRUNK(HIP TO SHOLDER)   50
45.7
THIGH
40.1        43
SHANK
36.7        42
FOOT
23.2        25
SELECT ARTICULATION STRENGTH INPUT MODE. PERCENTILES(ENTER"1")
ACTUAL STRENGTHS(ENTER"2")FORMAT I1
1
ENTER "Z" VALUE FOR STRENGTH (F5.2)
0.0   -0.68
ENTER SEX OF SUBJECT(MALE="1", FEMALE="2") I1 FORMAT
2
ENTER FORCE MAGNITUDE IN KILOS (F5.1)
55.     3.6, 11.4
ENTER FORCE DIRECTION-PUSH("1"), PULL("2"), LIFT("3"). I1
3
LOAD EXCEEDS GRIP STRENGTH.PROCEED?(ENTER"1")OR,REVISE
LOAD INPUT?(ENTER"2").I1

1

ENTER BODY STANCE. SEATED("1"), SQUAT("2"), STAND("3") I1
2
ENTER STABILITY FACTOR  0.0=<SF<=1.0 (F4.2)
0.5
SELECT INPUT MODE. LOAD POSITION("1") OR ARTICULATION ANGLES("2")-I1
2
INPUT ANGLE @ KNEE  THIGH TO HORIZ (DEGREES) F5.0
135.     140, 125, 105          100, 95, 90
INPUT ANGLE @ ANKLE  SHANK TO HORIZ (DEGREES) F5.0
65.        90, 90, 90
ENTER ANGLE @ HIP, TRUNK/HORIZ (DEGREES) F5.0
60.
ENTER SHOULDER ANGLE(UPPER ARM/HORIZ)DEGREES-F5.0
-110.
ENTER ELBOW ANGLE(LOWER ARM/HORIZ)DEG. F5.0
10.       80, 80, 60               80, 75, 70
PELVIC ANGLE INPUT MODE: MEASURED ANGLE("1")OR EQUATION
CALCULATION("2")I1 FORMAT
1
ENTER PELVIC ANGLE(L5/S1-HIP TO HORIZ)DEG. F5.0
106.       135
```

```
   BODY BALANCE RETAINED BUT MUSCLE FORCE EXCEEDED
  A=ART. TASK TORQUES(KG-CM)    B=% OF ART. VOL. TORQUES
```

	EL-FX	EL-EX	SH-FX	SH-EX	TR-FX	TR-EX
A=	804.3	0.0	532.4	0.0	0.0	2918.9
B=	356.2	0.0	174.1	0.0	0.0	175.4

	HP-FX	HP-EX	KN-FX	KN-EX	AK-EX
A=	0.0	2913.2	48.2	0.0	907.7
B=	0.0	391.4	24.7	0.0	81.1

```
FOR COMPARISON WITH HISTOGRAM(P.14)-COMP FORCE ON L5/S1 DISK FOR INPUT
HAND FORCE MAG= 598.2 KG      BACK MUSCLE FORCE= 549.6 KG
SHEAR FORCE AT FLOOR=    0.0 KILOS

CONTINUE("1") OR STOP("2")?
```

_ 6 2 . 1

Over Est. for 3

see RUM p 59

p 9

Figure 13: Example Run of Model Using Actual Link Lengths and Population Strengths--Squat Body Stance, Articulation Angles Define Body Configuration.

Frequently, during standing lifts, the knee flexion capability is rapidly exceeded. This is because the model does not include "locked" articulation strengths, but only the muscle strengths. Since the "locked" flexion strength of the knee is higher than the muscle strength at 180° included angle, the model errs on the conservative side. The same thing applies to the locked elbow case. After careful consideration the model user may want to ignore the results of locked articulations or change the body configuration so that the articulations would not be locked.

Another detail that should be noted is the disk compressive force output. The model is capable of computing negative compressive forces, but since these have no meaning in the anatomical sense, these negatives are reduced to zero compressive force.

In the computation of maximum hand force that will not destroy balance, several limiting cases are built in. In the sitting push case with back support, since balance cannot be lost, a value of 999.9 is assigned. Also, in cases where the hand force reaches 150 kilograms, the model stops the iteration. It is judged that other factors such as muscle strengths or excess disk compressive forces would limit the ability of the body to sustain loads greater than this so no further iteration is necessary.

Ankle flexions are another special case. Since the model is designed for situations under which the ankle flexion strength is not relevant, this strength is not used for maximum strength comparisons. The only time ankle flexion strength is a factor is when there is a negative torque at the ankle. Since under this condition backward balance is lost long before ankle flexion strength is exceeded, the flexion strength criteria is not used. Therefore, when there is a negative ankle torque, this torque is printed out under "AK-EX" and is negative. The percent of articulation voluntary torque is printed out as "0.0" and the maximum hand force magnitude is assigned the value "-999.9".

IV. MODEL OUTPUT EXAMPLES

Figures 14 through 18 show the results of several simulations using the biomechanical model described. Figure 14 shows the magnitude of force that the seated subject is capable of exerting with the hands at various horizontal positions during a lift, pull and push with and without back support. Also shown is what limited the subject from exerting more force at that position. For example, the push without back support is balance limited but when the balance constraint is altered by adding back support, the forces that the subject can exert are increased by about 2.5 and the subject becomes limited by shoulder flexion and elbow extension strengths. The lifts are elbow flexion strength limited up to the maximum force that the subject can exert and then become shoulder flexion limited. The pulls show a rapid increase in maximum force capability. This is because at the longer distances from the body, the arms are straighter and the torso can exert a larger proportion of the total force required rather than only the shoulder extension muscles.

The data in Figure 15 show the forces that a standing 5th percentile male can exert. The lift forces rapidly decrease with horizontal distance because of the increasing load moment arm relative to the shoulder flexion muscle moment arm. The push and pull modes are all balance limited and reflect only link weight moment arm changes. By increasing the stability factor to 1.0 in the push case, the maximum forces the subject is capable of exerting are increased by a factor of about 2.6.

Figure 16 shows essentially the same information that Figure 15 shows except that the subject is female. As can be seen in these figures, the lifting capacity of the female subject is about one-third that of the same percentile male. The pulling and pushing capabilities are about the same as the male with all of them being limited by balance constraints.

Figures 17 and 18 show other male-female comparisons for equal percentile subjects. The lifting curves in Figure 17 are quite similar in shape with the female curve being about one-half the male curve which shows the difference in strengths. The push mode is balance limited and therefore similar. Figure 18 shows pulling maximums that are quite divergent. This is because of the male case being balance limited and

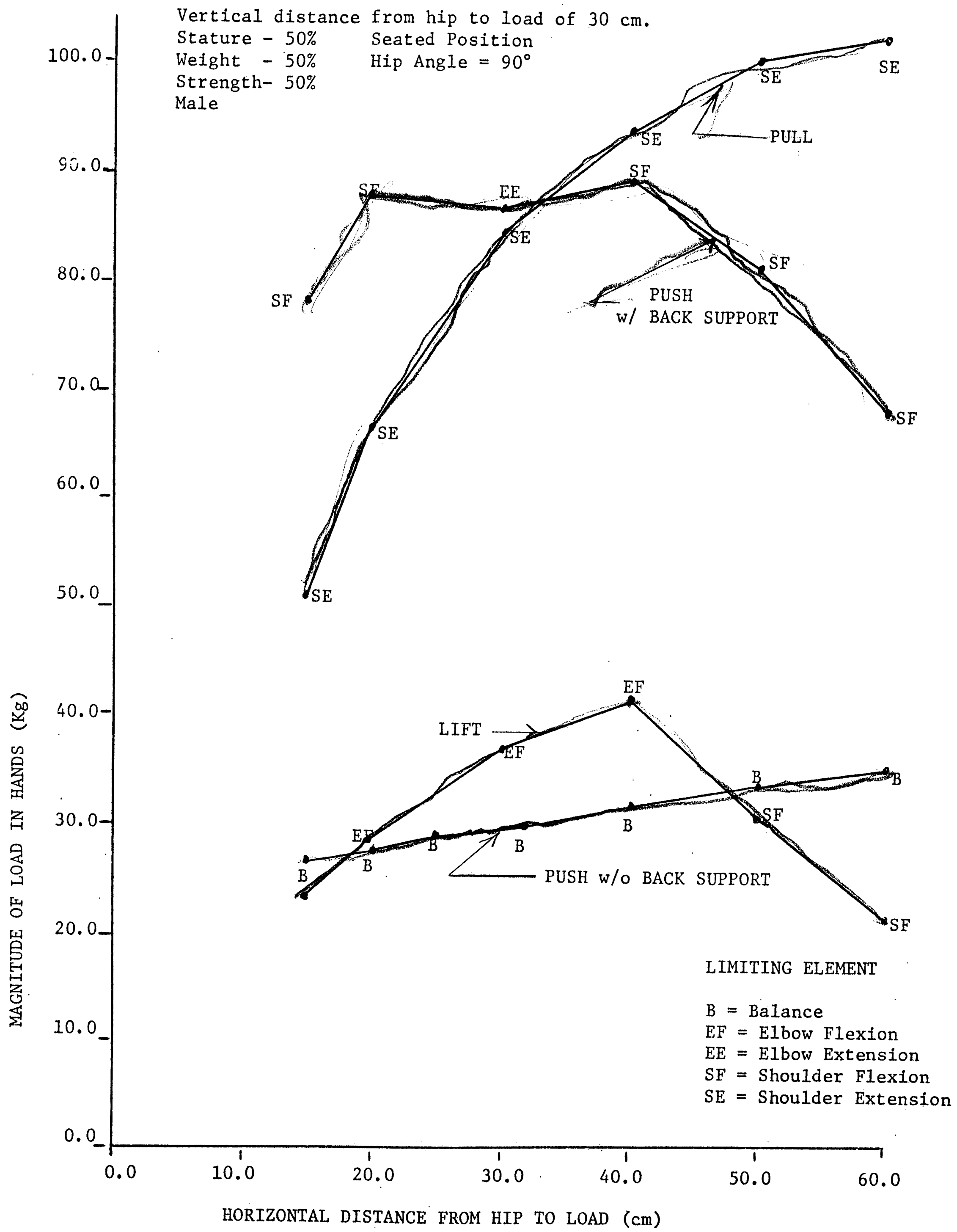


Figure 14: Plots of Hand Force vs. Horizontal Load Position for Various Force Directions.

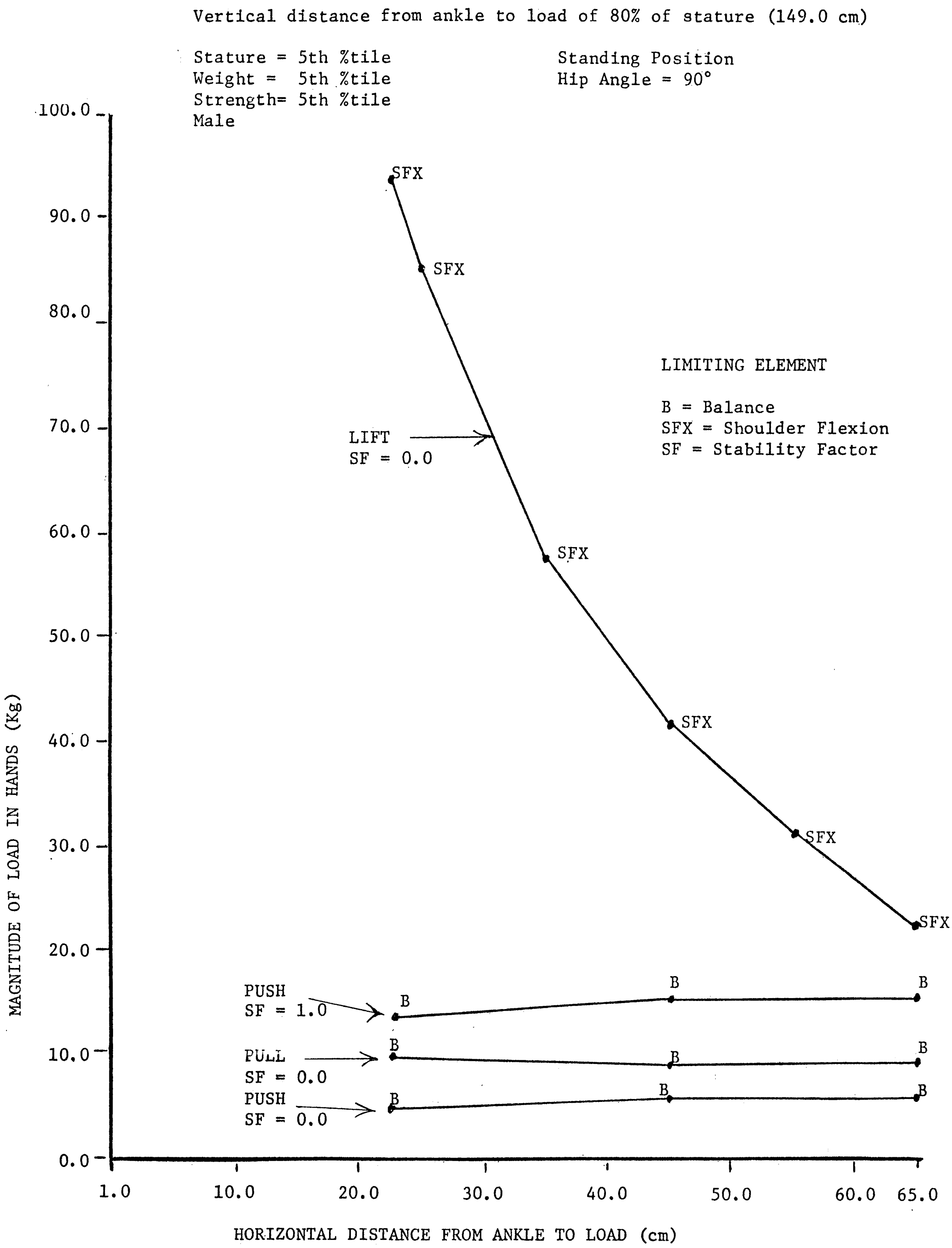


Figure 15: Plots of Hand Force vs. Horizontal Load Position for Various Force Directions.

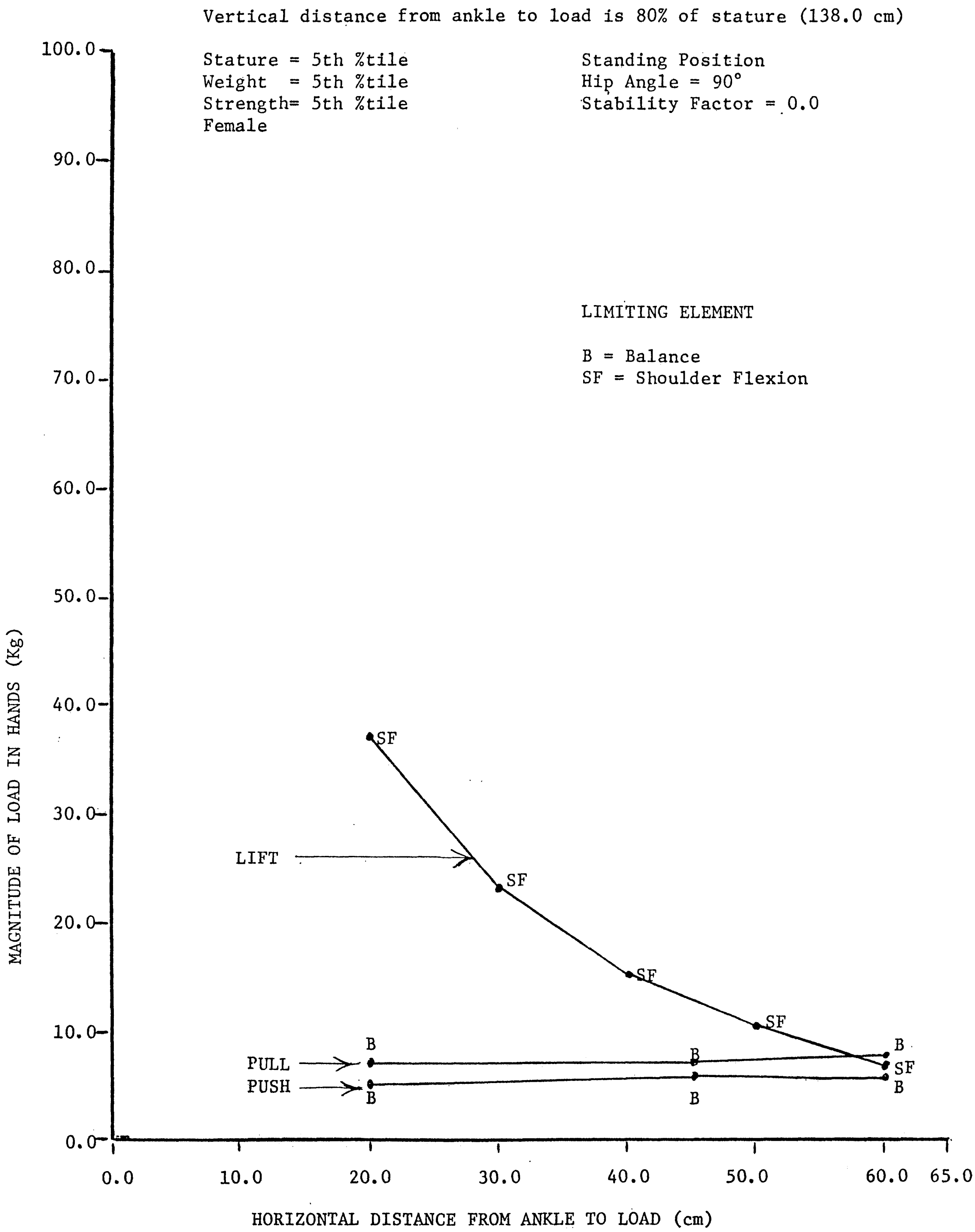


Figure 16: Plots of Hand Force vs. Horizontal Load Position for Various Force Directions.

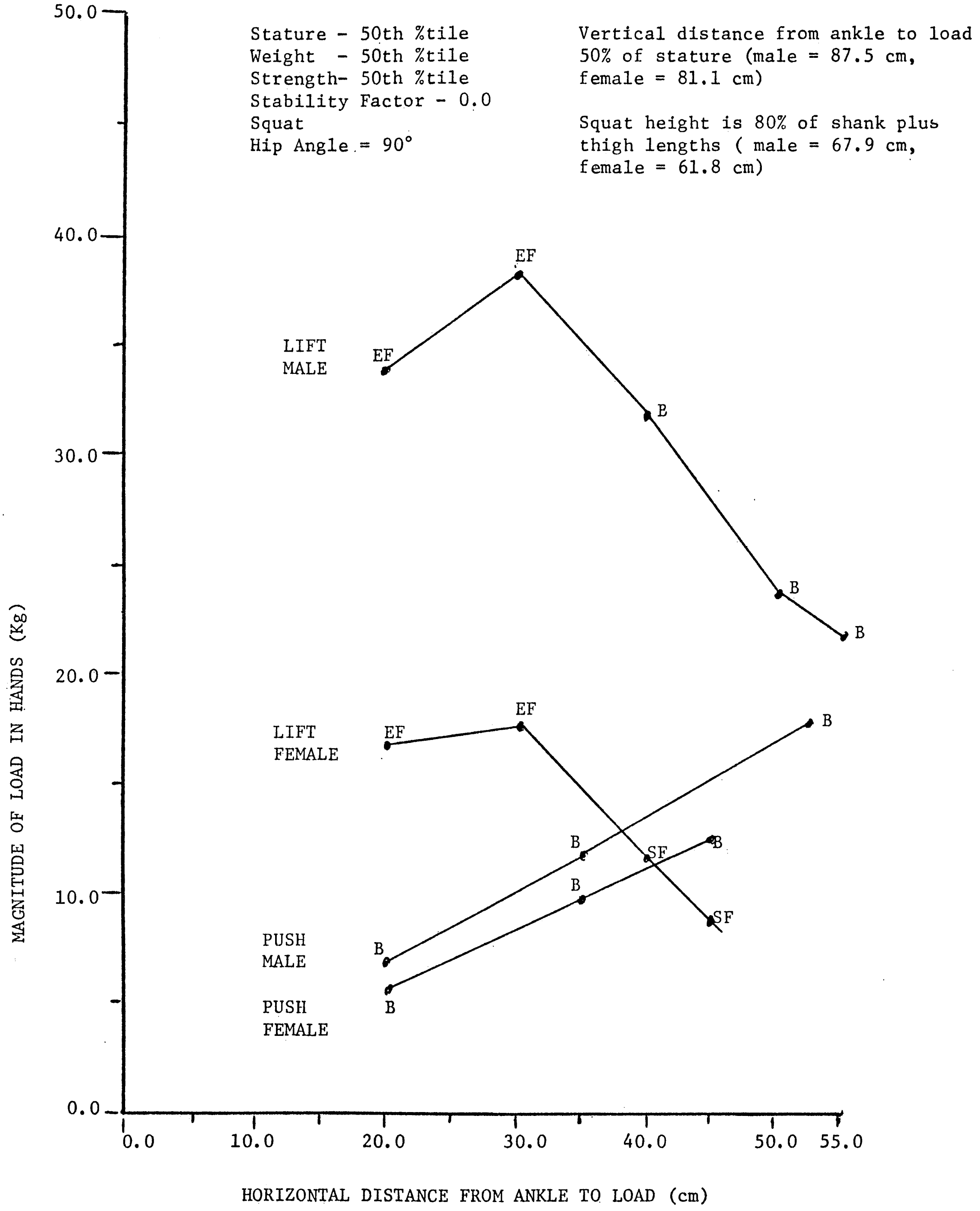


Figure 17: Plots of Hand Force vs. Horizontal Load Position for Various Force Directions - Male and Female.

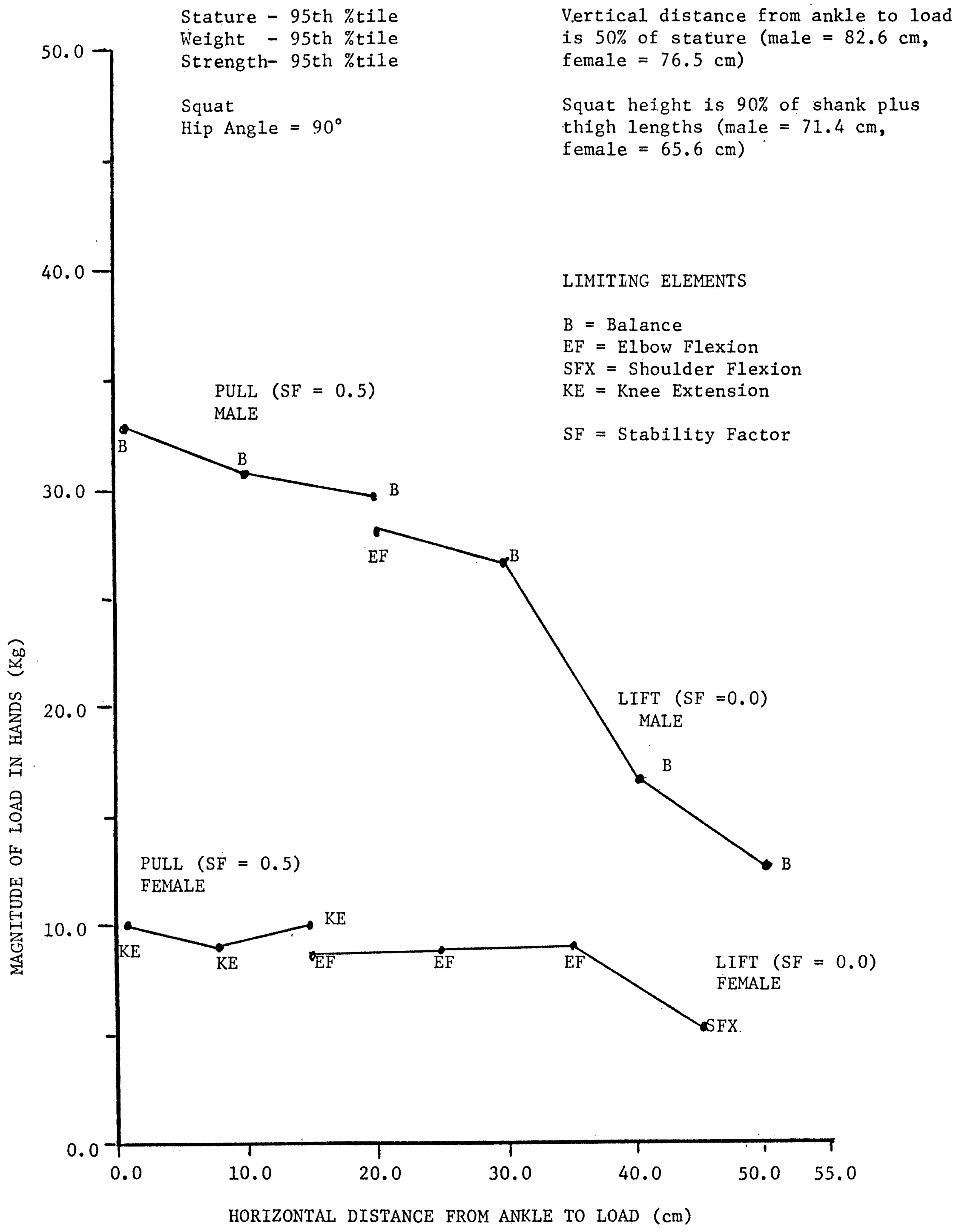


Figure 18: Plots of Hand Force vs. Horizontal Load Position for Various Force Directions - Male and Female.

the female case being limited by rather low knee extension strengths. These pulling cases have the horizontal coordinate of the load almost at the ankle point. This is because the ankle equation used in the load position body configuration input mode predicts an ankle angle of about 90° which results in a hip position well behind the ankles.

The disk compressive force ranged in value from negligible up to a maximum of about 500 kg. In all cases the compressive forces were judged to be within acceptable limits. In several of the lifting cases, the knee flexion strengths were theoretically the limiting factor. But, since these limits were close to the ones reported and because of the "locked knee" effect described earlier, these knee limits were ignored.

V. SUMMARY

A model has been created that evaluates a fully described subject in a specified body configuration exerting a push, pull or lift force with the hands. The criteria for evaluation are 1) can the subject retain balance, 2) are any maximum voluntary torques exceeded, and 3) are the compressive forces on the L_5/S_1 disk exceeded. If body balance is retained and no articulation torques are exceeded, the model determines the maximum hand held force that could be held and still not exceed the balance or articulation torque criterion.

Whereas there already exist several models similar to the basic model used here, this model is unique because of the flexibility that the user has for inputting the body dimensions (link lengths and weights), and body configuration. Also, the model is designed for terminal interaction use and is explicit in its prompt messages which makes it easy for people who are unfamiliar with such models to rapidly become proficient in its use.

The computer program is written in FORTRAN IV and has a storage requirement of 26,000 bytes. The central processor time (CPU) required for one full iteration of the program ranges between about 40 and 90 microseconds on the Amdahl 470V/6. The CPU time required depends on the number of iterations the model must make to find the maximum hand held load without the subject losing balance. The average time for a group of 40 iterations was 60 microseconds each.

Although this model evaluates only static capabilities, the model can be applied to slow, well-controlled force exertions, where the effects of acceleration and momentum are negligible. Therefore, with these capabilities and those described above, this model can effectively simulate many of the working conditions encountered by a wide range of workers. It is believed that effective use of information thus derived can greatly improve the interface between the worker and his physical work environment.

REFERENCES

1. "A Biomechanical Analysis of Materials Handling Activities," Department of Industrial Engineering, The University of Michigan, Ann Arbor, 1967.

2. Chaffin D.B. "A Computerized Biomechanical Model-Development of and Use in Styding Gross Body Actions," The American Society of Mechanical Enginners Publication Number 69-BHF-5, New York, N.Y., 1969.

3. Chaffin, D.B. "Ergonomics Guide for the Assessment of Human Static Strength," American Industrial Hygiene Association Journal, pp. 505-511, July, 1975.

4. Chaffin, D.B. "Human Strength Capability and Low-Back Pain," Journal of Occupational Medicine, 16(4):248-254, April 1974.

5. Chaffin, D.B. "Some Advantages and Limitations of Macro Static Models for Determining Spinal Injury Potential During Weight Lifting," The Department of Industrial and Operations Engineering, The University of Michigan, Ann Arbor, April, 1975.

6. Chaffin, D.B. "Some Effects of Physical Exertion - A Research Report," The Department of Industrial and Operations Engineering, The University of Michigan, July, 1972.

7. Chaffin, D.B. and Moulis, E.J. "An Empirical Investigation of Low-Back Strains and Vertebrae Geometry," Journal of Biomechanics, 2:89-96, 1969.

8. Chaffin, D.B. and Park, K.S. "A Longitudinal Study Low-Back Pain as Associated with Occupational Weight Lifting Factors," American Industrial Hygiene Association Journal, pp.513-125, December 1973.

9. Churchill, E. and Bernhardi, B. "WAF Training Body Dimensions," WADC Technical Report 57-197, Wright Field, Ohio, 1957.

10. Davis, P.R. and Troup, J.D.G. "Pressures in the Trunk Cavities when Pulling, Pushing and Lifting," Ergonomics, 7(4):465-474, October 1964.

11. Dempster, W.T. "Space Requirements of the Seated Operator," WADC Technical Report 55-159, Wright Field, Ohio, 1955.

12. Dempster, W.T. and Gaughran, G.R.L. "Properties of Body Segments Based on Size and Weight," Am. J. of Anat., 120:33-54.

13. Dempster, W.T.; Sherr, L.A. and Priest, J.G. "Conversion Scales for Estimating Humeral and Femoral Lengths and the Lengths of Functional Segments in the Limbs of American Caucasoid Males," Human Biology, 36(3):246-262, September 1964.

14. Evans, F.G. and Lissner, H.R. "Biomechanical Studies on the Lumbar Spine and Pelvis," Journal of Bone and Joint Surgery, 41A:278-290, 1959.

15. Evans, E.G. and Lissner, H.R. "Studies on the Energy Absorbing Capacity of Human Lumbar Intervertebral Disks," The Seventh Stapp Car Crash Conference Proceedings, Springfield, Illinois, 1965.

16. Fitzhugh, F.E. "Grip Strength Performance in Dynamic Gripping Tasks," Department of Industrial and Operations Engineering, The University of Michigan, Ann Arbor, December 1973.

17. Garg, A. and Chaffin, D.B. "A Biomechanical Computerized Simulation of Human Strength," AIIE Transactions, 7(1):1-13, March 1975.

18. Garg, A. and Chaffin, D.B. "A Biomechanical Computerized Simulation of Human Strength," Department of Industrial and Operations Engineering, The University of Michigan, June 1974.

19. Graphical Predictions of Human Strengths for Two Handed IV/EVAL - Phase One Report, Industrial Engineering Human Performance Group, The University of Michigan, Ann Arbor, April 1971.

20. Kroemer, K.H.E. "Muscle Strength as a Criterion in Control Design for Diverse Populations," Chap. 5 in Ethnic Variables in Human Factors Engineering, ed. A. Chapanis, (Baltimore: The Johns Hopkins Press), 1975.

21. Martin, J.B. and Chaffin, D.B. "Biomechanical Computerized Simulation of Human Strength in Sagittal-Plane Activities," AIIE Transactions, pp. 19-28, 1972.

22. Morgan, C.T.; Cook, J.S.; Chapanis, A. and Lund, M. Human Engineering Guide to Equipment Design, (New York: McGraw-Hill), 1963.

23. Park, K.S. and Chaffin D.B. "A Biomechanical Evaluation of Two Methods of Manual Load Lifting," AIIE Transactions, 105-113, June 1974.

24. Perey, O. "Fracture of the Vertebral End-Plate in the Lumbar Spine, An Experimental Biomechanical Investigation," Acta Orthopaedia Scandenavia, Supplement 25, 1957.

25. Schanne, F.J., Jr. "A Three-Dimensional Hand Force Capability Model for a Seated Person," Ph.S. Thesis, The University of Michigan, Ann Arbor, Michigan 1972.

26. Sonoda, T. "Studies on the Strength for Compression, Tension and Torsion of the Human Vertebral Column," Journal of Kyoto Prefect Medical University, 71:659-702, 1962.

27. Troup, J.D.G.; Chapman, A.E. "The Strength of the Flexor and Extensor Muscles of the Trunk," Journal of Biomechanics, 2:49-62, 1969.

28. Weight, Height and Selected Body Dimensions of Adults - United States, 1960-1962, National Center for Health Statistics, Series 11, Number 8, U.S. Department of Health, Education and Welfare, Public Health Service, Washington, D.C., June 1965.

29. Williams, M. and Lissner, H.R. Biomechanics of Human Motion, (Philadelphia: W.B. Saunders, Co.), 1962.

APPENDIX A

This appendix contains the data base used for the length, weight and center of gravity description of the body links. It also contains the population maximum voluntary strength data. All reference numbers refer to the main reference list.

Table A-1

Population Body-link Lengths (centimeters)

Percent of Population Larger	Male					Female				
	σ	95%	50%	5%	Ref.	σ	95%	50%	5%	Ref.
Hand (wrist to hand c. of g.)	0.2	6.6	6.9	7.4	11	0.2	6.1	6.4	6.6	9
Lower Arm	0.8	25.7	27.2	28.4	11	0.8	23.1	24.1	25.7	9
Upper Arm	0.8	29.0	30.0	31.5	11	0.6	26.7	27.7	28.7	9&11
Trunk (shoulder to hip)	1.8	45.5	48.8	51.6	9	1.8	42.2	45.2	48.3	22
Thigh (upper leg)	1.4	41.4	43.7	46.0	11	1.1	39.1	40.9	42.7	11&22
Shank (lower leg)	1.8	37.1	40.9	42.9	11	1.5	34.0	36.3	38.9	22
Foot (overall)	1.2	24.4	26.2	28.2	11	1.3	22.1	23.9	25.9	22
Stature	6.4	165.1	175.0	186.2	22	6.0	152.9	162.1	172.7	9

Note: The above dimensions were obtained by correcting "over-the-body" dimensions, as described in Dempster (11).

Table A-2

Population Dimensions to Segments Centers of Gravity (centimeters)(11)

Dimension	Percent link length
Elbow to lower arm c.g.	43.0% of lower arm
Shoulder to upper arm c.g.	43.6% of upper arm
Hip to trunk, neck, head c.g.	39.6% of (trunk length + 20.33% of Stature*)
Knee-to-thigh c.g.	56.7% of thigh
Ankle to shank c.g.	56.7% of shank

*22.33% of stature represents the distance from the shoulder to top of head and must be summed to trunk length to find the link c.g.

Center of gravity locations based on percent of link lengths, as proposed by Dempster (11).

Table A-3

Population Weight Distributions
(kilograms)(1)

Link	Percent of Total Body Weight
Hand	0.75%
Lower arm	1.80%
Upper arm	3.10%
Head, neck and trunk	55.40%
Thigh	10.45%
Shank	4.60%
Foot	1.60%

Table A-4

Total Body Weight Percentiles (kilos)(28)

Percent of Pop. Heavier	σ	95th	50th	5th
Male	12.51	57.15	75.30	98.43
Female	*	47.17	62.13	90.25

*Because of the skewedness of this distribution, the σ's were back calculated from the data in the reference and lumped into groups judged to be coherent from inspection of the data. The results are as follows:

Percentile group	σ
greater than or equal to 50%	9.56
less than 50%, greater than 15%	12.97
less than 15%, greater than 2.5%	16.51
less than 2.5%	19.52

Table A-5

Population Body - Link Lengths (centimeters)
As Calculated from Reference Lengths

Link	Relationship	Reference
Hand	(0.0395)(Stature)	11
Lower Arm	(1.0709)(Radius Length)	13
Lower Arm	(0.1572)(Stature)	12
Upper Arm	(5.80752)+(0.9646)(Radius Length)	13
Upper Arm	(0.1735)(Stature)	12
Trunk	(0.2798)(Stature)	12
Thigh	(13.28253)+(0.8172)(Tibia Length)	13
Thigh	(0.2399)(Stature)	12
Shank	(1.0776)(Tibia Length)	13
Shank	(0.2505)(Stature)	12
Foot	(0.1484)(Stature)	11

Table A-6

Distribution of Trunk Lengths and Mass (1)

Trunk Link	Length	Mass (% of trunk mass)
L_5/S_1 to shoulder	.805 * LTK*	65.5%
Hip to L_5/S_1	.195 * LTK	34.5%

* LTK = length of trunk (cm)

Table A-7

Population Strengths Articulation Torques (cm-kg)

Articulation and Direction	Male			Female		
	Mean	σ	Ref.	Mean	σ	Ref.
Elbow Flexion	708.4	134.2	6	289.7	87.7	6
Elbow Extension	436.4	73.1	6	152.1	34.6	6
Shoulder Flexion	855.3	168.5	6	305.4	81.8	6
Shoulder Extension	848.9	162.1	6	301.2	78.4	6
Hip Flexion	1562.4	245.8	6	480.6	107.2	6
Hip Extension	3437.0	996.7	6	1055.8	361.4	6
Torso Flexion	---**	---**		---**	---**	
Torso Extension	3583.0	1270.0	*	2109.0	623.0	*
Knee Flexion	524.1	75.2	6	225.2	65.1	6
Knee Extension	1856.4	443.7	6	810.3	301.2	6
Ankle Extension	2265.3	635.1	6	1119.5	396.3	6
Grip Strength	52.61	11.07	16	26.89	7.62	16

*These strengths are the result of the PEST (Pre-Employment Strength Test) Program which is currently under way at the University of Michigan. The way these numbers were obtained is fully explained in Appendix B-4.

** Data not available. See Appendix B-4 for explanation.

APPENDIX B

This Appendix contains a description of some of the mathematical formulations used in the model.

Appendix B-1

Ankle Angle and Pelvic Angle Relationships

There are two input modes that use empirical angle relationships rather than actual angle inputs. One of these is the ankle angle (angle at ankle, measured from shank to horizontal) for the load position input mode in the standing and squat positions. The other is the pelvic angle (angle at hip measured from a line connecting the center of the L_5/S_1 disk to the hip, to horizontal). This angle is automatically calculated in the load position input mode and is optional in the angle input mode.

In order to develop the empirical angular relationships, an experiment was performed wherein a subject was photographed from the side while performing a series of lifting, pushing and pulling tasks. The positions of the "load" on which the subject had to exert force (actually a load cell) was varied throughout a matrix. The exertions were made at 100% of capacity, 66% of capacity and 33% of capacity. The exertion levels (with the exception of 100%), the load positions and the directions of exertion were randomly assigned. Each combination was then photographed. The photographs were then analyzed and relevant data variables extracted. These variables included load position (X and Z), distance from hip to ankle ("D" length), ankle angle, hip angle, knee angle and pelvic angle. The resulting data was then reduced by multiple linear least squares regression techniques. The resulting equations are listed in Figure B-1-1.

Since the "D" length which was used in these equations is specific to the subject, each D was replaced with (D)(178.0)/STATURE. This then adjusts the "D" length to the stature of any model input subject.

The pelvic angle for the sitting case is a special case of the standing case wherein the knee angle is 180°. This works well for the push and pull case but leads to a pelvic angle of 130° when the lift case is used in this manner. Therefore, the sitting lift case uses the general pull case, a decision that was made by best judgment.

The ankle angle prediction equation yielded rather unsatisfactory results when applied to the standing posture. For this reason, a different set of data was regressed to obtain the equation used. The data was gathered by observing a subject bending through a series of hip angles with the knees straight. The hip and ankle angles were measured with a protractor and the resulting data reduced by the multiple linear least squares regression technique. A plot of the data used is shown in Figure B-1-2.

Figure B-1-1

Table of Equations Used in Predicting Ankle and Pelvis Angles

Equation	Stance	Load Exertion	R^2
AASHZ[1] = 1.136-(0.00243)(X)-(0.002276)(Z) + 0.00912(D)	squat	pull	0.68
AASHZ = 1.386-(0.00609)(X)+(0.00166)(Z) - (0.0187)(D)-(0.00017)$(D)^2$	squat	push	0.83
AASHZ = -1.935-0.001825(X)-0.00267(Z) + (0.1642)(D)-0.00299$(D)^2$ + (0.000018)$(D)^3$	squat	lift	0.87
AASHZ = 1.7599-(0.01127)(AHSHZ) - (0.06956)$(AHSHZ)^2$	stand[3]	all	0.99
AL5S1 = AHSHZ + 1.555-(0.7788)(AHSHZ) + (0.05212)(AKTHZ)	all (sit)[2]	pull (lift)[2]	0.73
AL5S1 = AHSHZ + 2.1762-(1.0243)(AHSHZ) - (0.05402)(AKTHZ)	all	push	0.99
AL5S1 = AHSHZ + 1.4143-(1.0364)(AHSHZ) + (0.291)(AKTHZ)	stand	lift	0.98

1. Variables defined as follows: AASHZ - angle @ ankle, shank to horiz.; AL5S1 = angle @ hip, L5S1 to hip line/horiz.; AHSHZ = angle @ hip, shoulder/horiz.; Z = vertical distance, talus to load, D = length, talus to hip.
2. Judged a special case - see text.
3. This equation from a different source, see text.

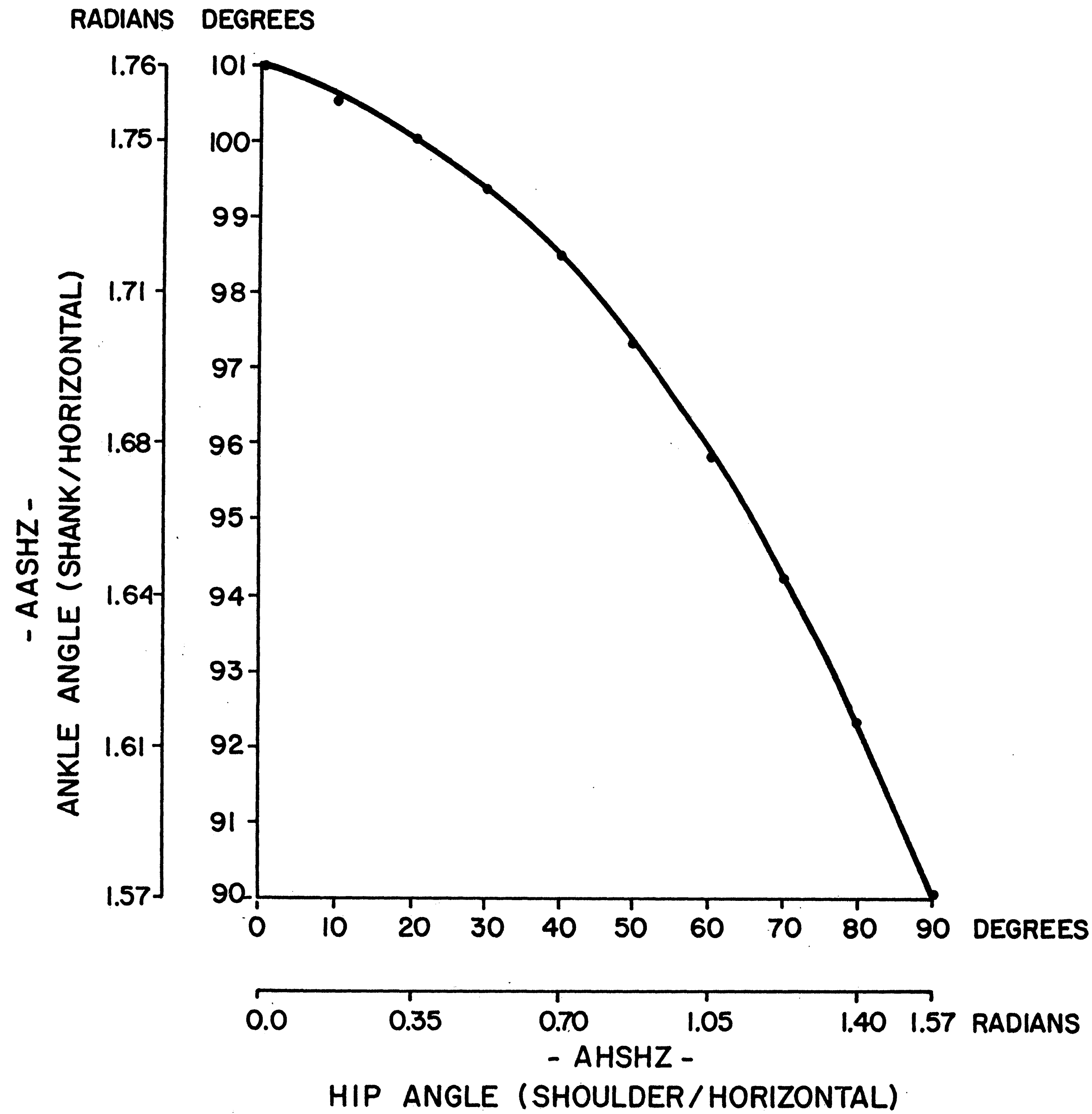


Figure B-1-2: Plot of Ankle Angle vs. Hip Angle Relationship Used For Standing Case.

Appendix B-2

Articulation Torques

The torques at each articulation that are the result of external forces (gravity and load) are computed by taking the product of the appropriate moment arms and forces. The elbow is computed first. The torques that make up the elbow torque are the lower arm weight/lower arm center of gravity, length, hand weight/distance to hand center of gravity, and hand held load/distance to hand center of gravity. This is shown in Figure B-2-1 and in the following equation:

$$\begin{aligned} TE = {} & COS(AELAHZ)[(CGELA)(WLA)+(LLA+LHD)(WHD)+(LLA+LHD)(\tfrac{FMD}{2})] \\ & + SIN(AELAHZ)[(LLA+LHD)(\tfrac{FMH}{2})] \end{aligned}$$

where:

TE = torque at elbow
AELAHZ = angle at elbow, measured from the lower arm to horizontal
CGELA = center of gravity of lower arm (measured from elbow)
WLA = weight of lower arm
LLA = length of lower arm
LHD = length of hand (wrist to c.g. of hand)
WHD = weight of hand
FMD = vertical force
FDH = horizontal force

With the elbow torque computed, the model proceeds to the shoulder to calculate its torque. The torques making up the total shoulder torque consist of the torque due to upper arm weight/upper arm c. of g., the elbow torque and the total force vector at the elbow. This total force vector is carried throughout the computations as horizontal and vertical forces. The shoulder torque is computed as follows:

$$\begin{aligned} TS = {} & COS(ASUAHZ)[(CGSUA)(WUA)+(LUA)(WED)] + TE \\ & + SIN(ASUAHZ)[(LUA)(FMH/2)] \end{aligned}$$

where:

ASUAHZ = angle at shoulder, measured from the upper arm to horizontal

WUA = weight of upper arm

LUA = length of upper arm

WED = vertical force at elbow

$= WLA + WHD + \frac{FMD}{2}$

The model then proceeds to compute the successive articulation torques, in order, away from the load. The model computes the torque that each single link would experience rather than treating "double links" (arms or legs) as one link. Therefore, the load is split between the two arms and then "recombined" at the shoulder. As throughout the model, all units are centimeters and kilograms.

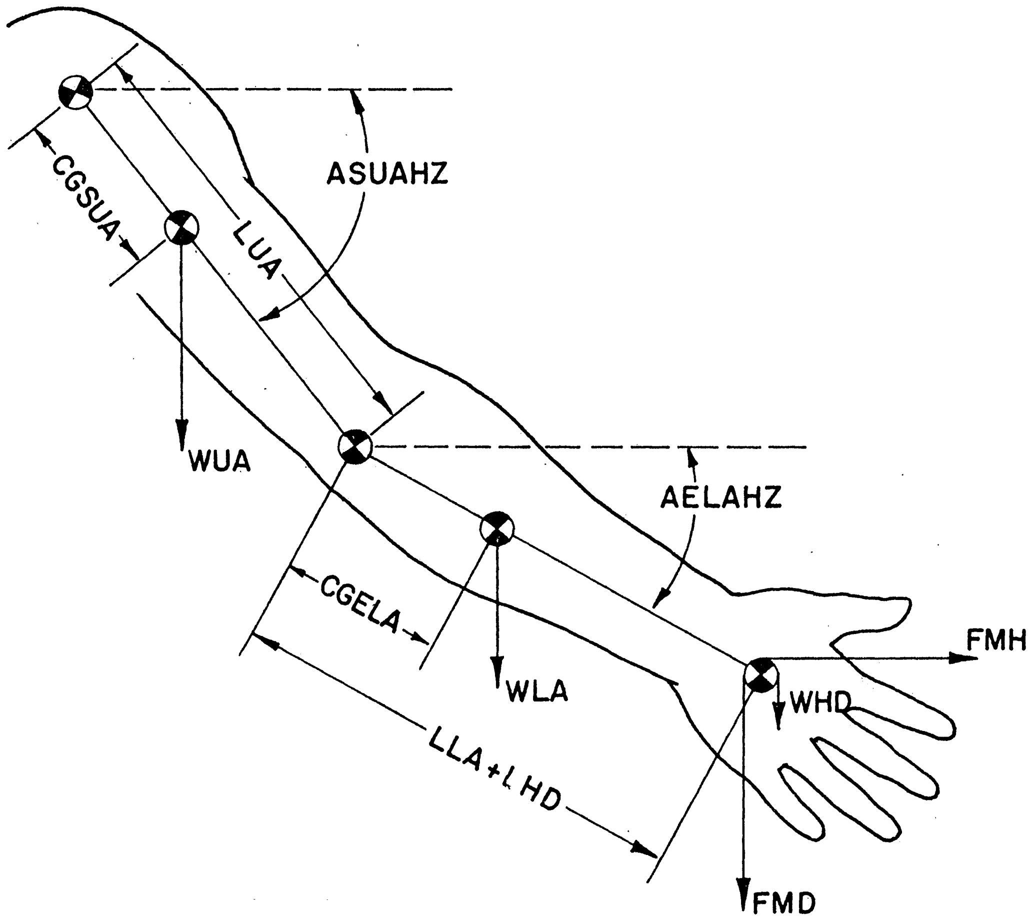


Figure B-2-1: Diagram of Link Length and Weight Relations used in Computing Articulation Torques.

Appendix B-3

Body Balance Computation

The standing person maintains body balance by positioning the body in a manner such that the resultant rotational torques caused by body weight or any other external forces acting on the body are counteracted by the equal in magnitude but opposite in direction reactive rotational torques exerted by the body support medium. This support medium would be the floor in the standing case or a chair and floor in the seated case. The force would be exerted at the heel or ball of the foot (standing case) or on the feet or posterior aspect of the buttocks (sitting case).

For the standing case, the torque at the ankle is a result of body weight and hand load and is computed as explained in Appendix B-2. If this torque is positive, it is compared to ankle extension strength to be sure it is not exceeded. If it is negative, it is not compared to ankle flexion because it is assumed that backward body balance would be lost long before any ankle flexion strength would be exceeded. If there is sufficient strength, then the support force is computed as the sum of the body weight and the downward vertical component of any other force acting on the hands. The torque produced by this force is the product of the support force and the horizontal distance between the ankle and either: (1) the posterior part of the foot (heel) when tending to fall over backwards; or (2) the ball of the foot when tending to fall forward.

If there is a stability factor specified in the input, it too is taken into account. The stability factor is the total distance that the ankles of the two feet are displaced ahead and behind where the ankles would be if the person were standing with the feet together. It is expressed as a fraction of a foot length and is limited to one foot length. Figure B-3-1 shows this relationship.

If the body tends to fall backward, the reaction torque is calculated as follows:

$$RH = [(5)(LFT/26.2)+(SF)(LFT)/2][(WT+FMD)/2]$$

where:

RH = reaction torque at heel

5 = assumed horizontal distance (cm) from ankle to force point of heel

LFT = length of foot

26.2 = assumed length (cm) of normalizing foot

SF = stability factor

WT = body weight

FMD = force magnitude in hands, down

Since the 5 cm is assumed for a "normal" foot, it is adjusted by the ratio of actual foot length to 26.2. Also, the horizontal forces are divided in half so that the comparison can be made to the forces that are calculated for the torque on one ankle.

If the body tends to fall forward, the reaction torque is computed as follows:

$$\mathrm{RB} = \left\{[\mathrm{LFT}-(5+6)(\mathrm{LFT}/26.2)]+(\mathrm{SF})(\mathrm{LFT})/2\right\} \left\{(\mathrm{WT}+\mathrm{FMD})/2\right\}$$

where:

RB = Reaction torque at ball

6 = assumed horizontal distance (cm) from end of foot to force point of ball

All other variables and constants are as defined above. Figure B-3-2 shows the foot length relationships graphically.

For the sitting case, it is assumed that the subject is sitting on a slightly padded (essentially incompressible) seat the front of which is inclined upward from horizontal by 8°. However, the compression of the thigh tissue results in a horizontal thigh bone. Also assumed is an included knee angle of 120°, a firm support for the foot to exert force on and a rigid ankle. Figure B-3-3 shows this graphically. Dempster (11) reports a mid-talus to floor length of 8.2 centimeters for a 175 centimeter male. This represents 4.69% of the stature and is the relationship used in the model.

In the sitting case, the torques are computed around the hip. The torque due to body weight and other external forces is compared to the

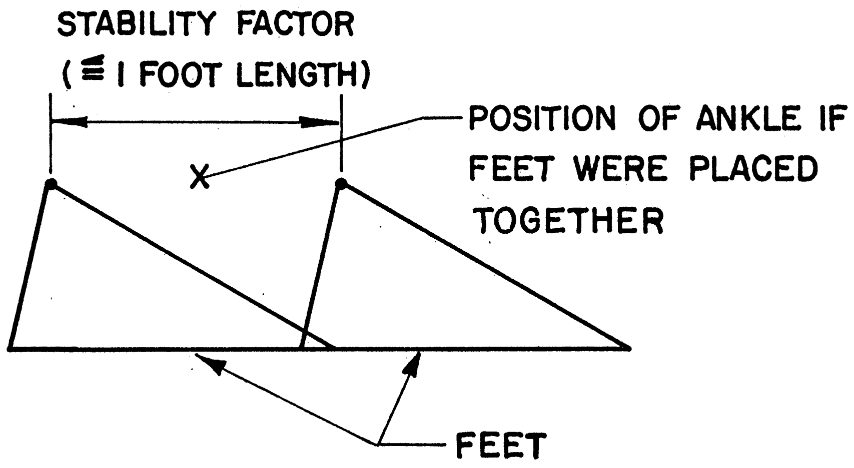


Figure B-3-1: Schematic Representation of Feet Showing Stability Factor.

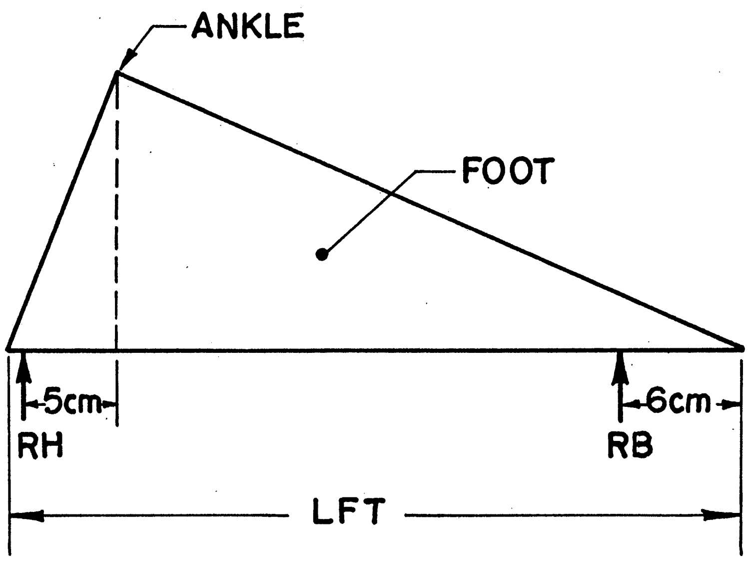


Figure B-3-2: Schematic Representation of Foot Showing Reaction Force Points.

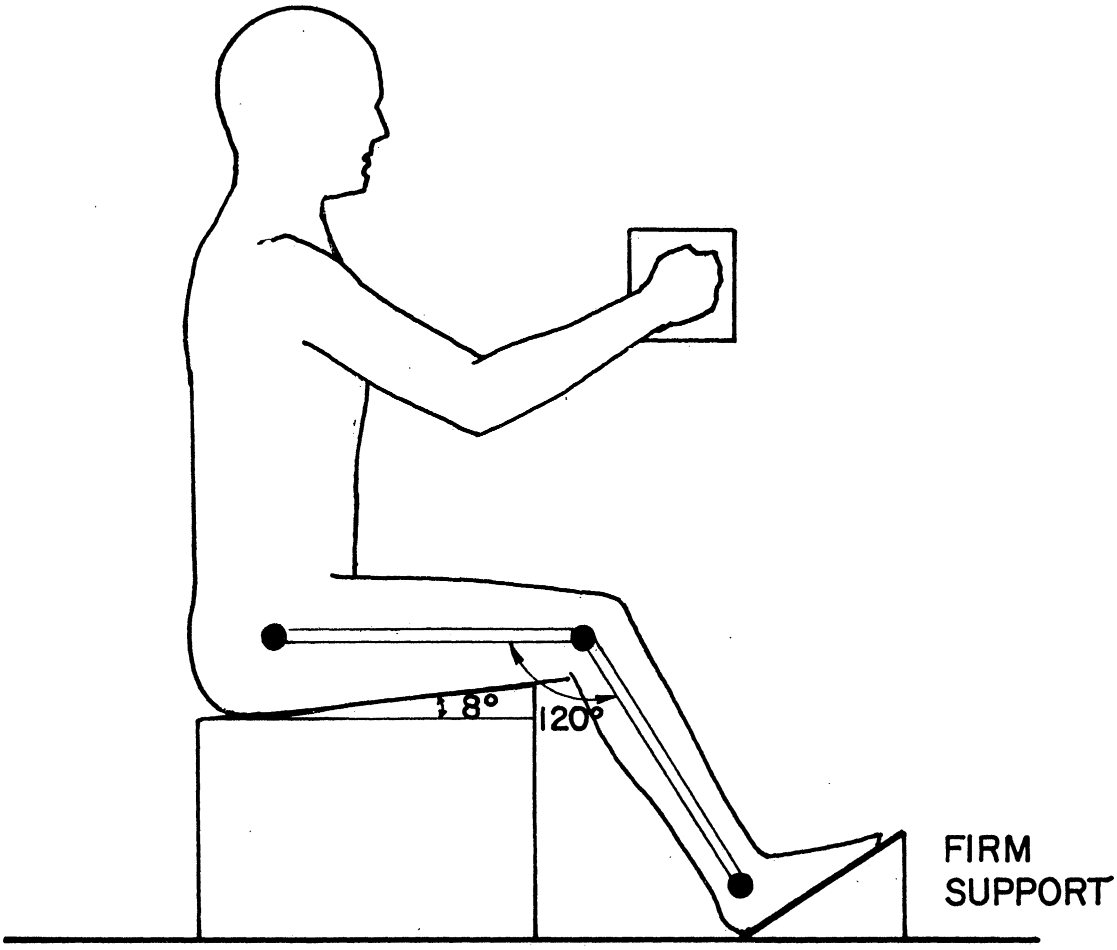


Figure B-3-3: Body Position Assumptions for Sitting Case support medium reaction forces. If the hip torque is positive (i.e., the body wants to fall forward) then the hip torque is compared to that torque produced by the vertical forces down with the moment arm extending from the middle of the foot to the hip. Also compared is the knee torque against the knee strength to be sure that the strength is not exceeded. The reaction torque equation is as follows:

$$G = [LSK + (0.0469)(ST)]/2$$

$$TF = (.5)[(0.0469)(ST) + CGASK)]\ (2)(WSK) + (G + CGKTH)(2)(WTH) + (G+LTH)(WPD)$$

where:

G = common length factor

TF = reaction torque at foot

LSK = length of shank

ST = stature

CGASK = center of gravity, from ankle to mid-shank

WSK = weight of shank

CGKTH = center of gravity, from knee to mid-thigh

WTH = weight of thigh

LTH = length of thigh

WPD = weight at pelvis, vertical, down

If the torque at the hip due to body weight and other outside forces is negative then there are two cases considered. If there is a back support, then the balance constraint is ignored under the assumption that backward balance cannot be lost or that the body will be limited by strength considerations first. If there is no back support then the only torque that can be generated to counteract the hip torque is the weight of the legs and feet with the appropriate moment arms. The equation is as follows:

$$THP = (LTH-CGKTH)(2)(WTH) + [LTH + (0.5)(LSK-CGASK)](2)(WSK) + [(0.5)(LSK+(0.0469)(ST))+LTH](2)(WFT)$$

where:

THP = torque at hips in a positive direction

WFT = weight of foot

In order to compute the maximum hand held force that could be maintained and still retain balance, the model simply increments the input force magnitude by one kilogram (providing the input force does not destroy balance) and iterates through the torque computations and balance criterion. This is done until (1) balance is exceeded, or (2) a limit of 150 kg. is reached. When balance is exceeded, one kilogram is subtracted from the resulting force magnitude and that force is then printed out as the maximum. The 150 kg. limit was included to eliminate excessive iterations and was judged to be slightly above the upper strength limit.

Appendix B-4

Adjustment of Maximum Voluntary Torque to Account for Included Angles

The ability of a muscle to exert a torque at an articulation varies with the included angle of the articulation across which the muscle pulls. Since the muscle strengths that are used as input for the model are all at a standard position, they must be corrected when the included articulation angle is other than standard. This is done by the following relationship (21):

$$ST_{\alpha} = PT_{\alpha} \frac{ST_{\alpha 1}}{PT_{\alpha 1}}$$

where:

ST_{α} = subject torque at angle of interest

PT_{α} = population curve value at angle of interest

$ST_{\alpha 1}$ = subject strength value at standard position angle

$PT_{\alpha 1}$ = population curve value at standard position angle

α = included articulation angle of interest

$\alpha 1$ = included articulation angle at standard position angle

The population curves are the results of careful testing of a large number of subjects and represent the variation in articulation strength with included articulation angle. The data thus collected was reduced by polynomial least squares regressions techniques. The resulting curve equations are listed in Table B-4-1.

Sufficient applicable data exists for all strengths with the exception of torso strength. Schanne (25) reports a population curve type relationship that, although it appears to be a reasonable representation of the way torso extension and flexion vary with angle, the values are judged to be significantly lower than what can reasonably be expected of a population. Troup (27) reports some data which appears to be comprehensive, however, the subjects were all college physical education majors or physical education teachers. The resulting data was therefore judged to be significantly higher than the worker population to which the model is to apply. Both relationships are shown in Figure B-4-1.

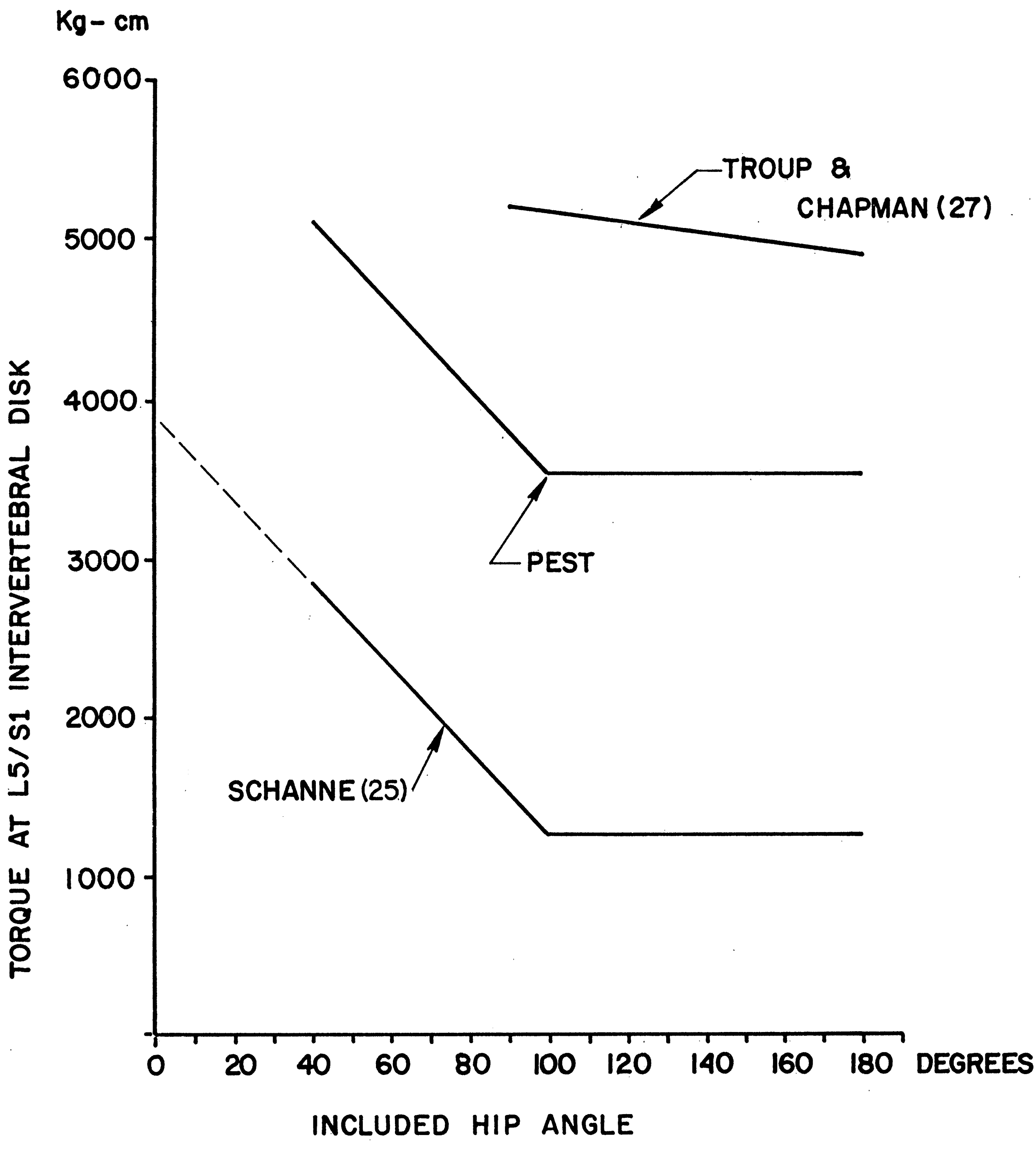


Figure B-4-1: Plot of Torso Strengths vs. Included Hip Angle from Two Sources and the Resulting Relationship used in The Model.

The third source of data on torso strength comes from the PEST (Pre-employment Strength Testing) project being conducted at the time of this writing at the University of Michigan. One of the strength modes tested was torso extension. A large group of male and female subjects from a worker population were tested. The subjects were instructed to perform a straight legged lift against a strain gage which was placed at a location 38.1 cm horizontal and 38.1 cm vertical, relative to the ankle. The result was data in the form of the force that the subject could exert. The data was described by the mean and standard deviation of the male and female subjects.

In order to get the one data point shown in Figure B-4-1, the model was run with average male body descriptors (stature, height and strength) and the appropriate load position. The hip angle was adjusted so that the arms were as straight as possible. This resulted in a 100° included hip angle and a torque of 3584 kg-cm. This point was then placed on the graph and Schanne's relationship paralleled to give the final torso extension relationship. The resulting data points were then regressed using the usual polynomial least squares regression techniques.

The population strengths for torso extension (male and female), were obtained by using the average and one standard deviation forces as input to the model. The model was then run with the appropriate body descriptors. For example, if the force was for a female subject and was the average minus one standard deviation, the body descriptors were the same. The data that resulted from operating the model in this manner was then used for the population strengths which are listed in Appendix A.

Table B-4-1: Table of Curve Equations Representing Articulation Torque as a Function of Articulation Included Angle for a Broad Population.

Articulation and Direction of Exertion	Equation Resulting from Polynomial Least Squares Regression[2]	R^2	Standard Position (Included Angle, Degrees)	Population Curve Value at Standard Position (cm-kg)[4]	Ref.[1]
Elbow Flexion	PFSE[3] = -469.83 + (1861.5)(AELAUA) - (793.9)(AELAUA)^2 + (99.716)(AELAUA)^3	.997	90°	881.8	
Elbow Extension	PESE = 1335.4 - (1616.7)(AELAUA) + (1240.2)(AELAUA)^2 - (403.94)(AELAUA)^3 +(45.485)(AELAUA)^4	.999	90°	567.3	
Shoulder Flexion	PFSS = 1239.3 + (32.391)(ASHUA) - (375.11)(ASHUA)^2 + (184.38)(ASHUA)^3 - (31.832)(ASHUA)^4		30°	1177.5	
Shoulder Extension	PESS = 1017.0 + (292.75)(ASHUA) - (279.1)(ASHUA)^2 + (127.52)(ASHUA)^3 - (42.373)(ASHUA)^4 + (5.269)(ASHUA)^5		30°	1104.0	
Hip Flexion	PFSH = 3729.7 - (10546.0)(AHTTK) + (12913.0)(AHTTK)^2 - (6707.2)(AHTTK)^3 + (1640.7)(AHTTK)^4 - (157.1)(AHTTK)^5		90°	1516.3	
Hip Extension	PESH = 1044.2 + (5491.3)(AHTTK)-(5157.7)(AHTTK)^2 + (1820.9)(AHTTK)^3 - (221.3)(AHTTK)^4	.997	90°	2653.9	
Torso Flexion	No data available				
Torso Extension	PEST = 7075.3 - (3526.2)(AHTTK) + (1154.4)(AHTTK)^2 - (123.82)(AHTTK)^3	.998	90°	3904.8	
Knee Flexion	PFSK = -342.82 + (1321.7)(AKAH) - (1232.1)(AKAH)^2 + (634.34)(AKAH)^3 - (105.89)(AKAH)^4	.999	120°	811.0	
Knee Extension	PESK = -2802.1 + (14144.0)(AKAH) - (17866.0)(AKAH)^2 + (11118.0)(AKAH)^3 - (3253.9)(AKAH)^4 + (353.6)(AKAH)^5	.998	120°	2234.0	
Ankle Extension Grip Strength	Strength Assumed Independent of Angle				

1. References numbers refer to main reference list.
2. All angles are in radians, strengths in cm-kg.
3. For variable definitions, see Appendix C.
4. As determined from equation.

Appendix B-5

Maximum Muscle Strengths

The maximum load that can be held in the hands (either push, pull or lift) and still not exceed the strength of the articulation is computed by the following relationship:

$$MVT = N.L.T. + \left(\frac{T_{1kg} - N.L.T.}{1_{kg}}\right)(ML_J)$$

therefore:

$$ML_J = \frac{MVT - N.L.T.}{T_{1kg} - N.L.T.}$$

where:

ML_J = maximum load that can be held in hand - for J articulation

MVT = maximum voluntary torque

N.L.T. = no load torque at articulation (no load in hands)

T_{1kg} = torque at articulation resulting from 1 kg load on hands.

This relationship is shown graphically in Figure B-5-1.

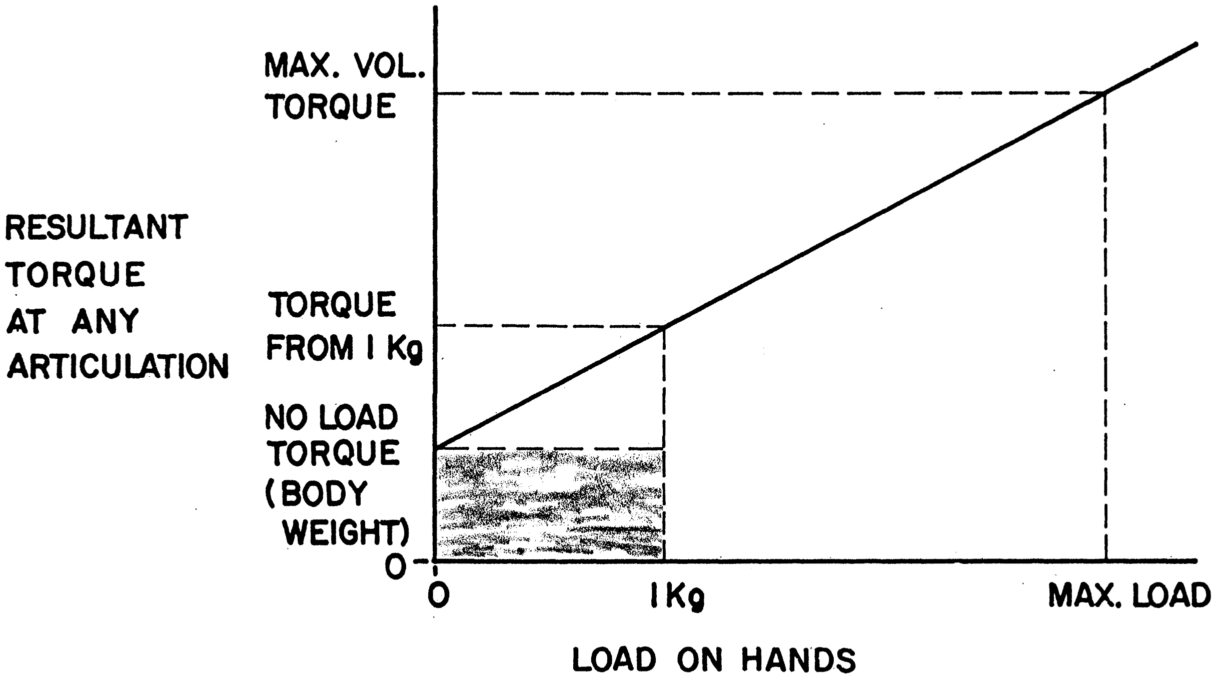


Figure B-5-1: Graphical Representation of Maximum Articulation Muscle Strength Algorithm.

Appendix B-6

Articulation Range of Motions

The model uses the following articulation ranges of motion:

elbow - 40° to 180° included angle
shoulder - -40° to 180° included angle*
knee - 50° to 180° included angle
ankle - 60° to 110° from horizontal

* upper arm to torso - negative angle means upper arm behind torso.

The basic hip range of motion is from 40° to 180° (included angle). However, since it is observed that many people can not flex the torso to an included angle of 40° with the knees locked straight (180°), an algorithm was devised to take this into account. A subject was observed flexing the torso to the maximum with the knees straight and then with successively bent knees. The angles were measured with a protractor and the results plotted. Figure B-6-1 shows these results. These data points were then fitted with a curve by a polynomial least squares regression technique. The resulting equation is then used in the model to define the hip range of motion for various included knee angles.

$$\text{AHMAX} = -2.62 + 4.736(\text{AKAH}) - 2.313(\text{AKAH})^2 + 0.386(\text{AKAH})^3$$

$$R^2 = .999$$

AHMAX = Maximum that hip can flex for given included knee angle

AKAH = included knee angle

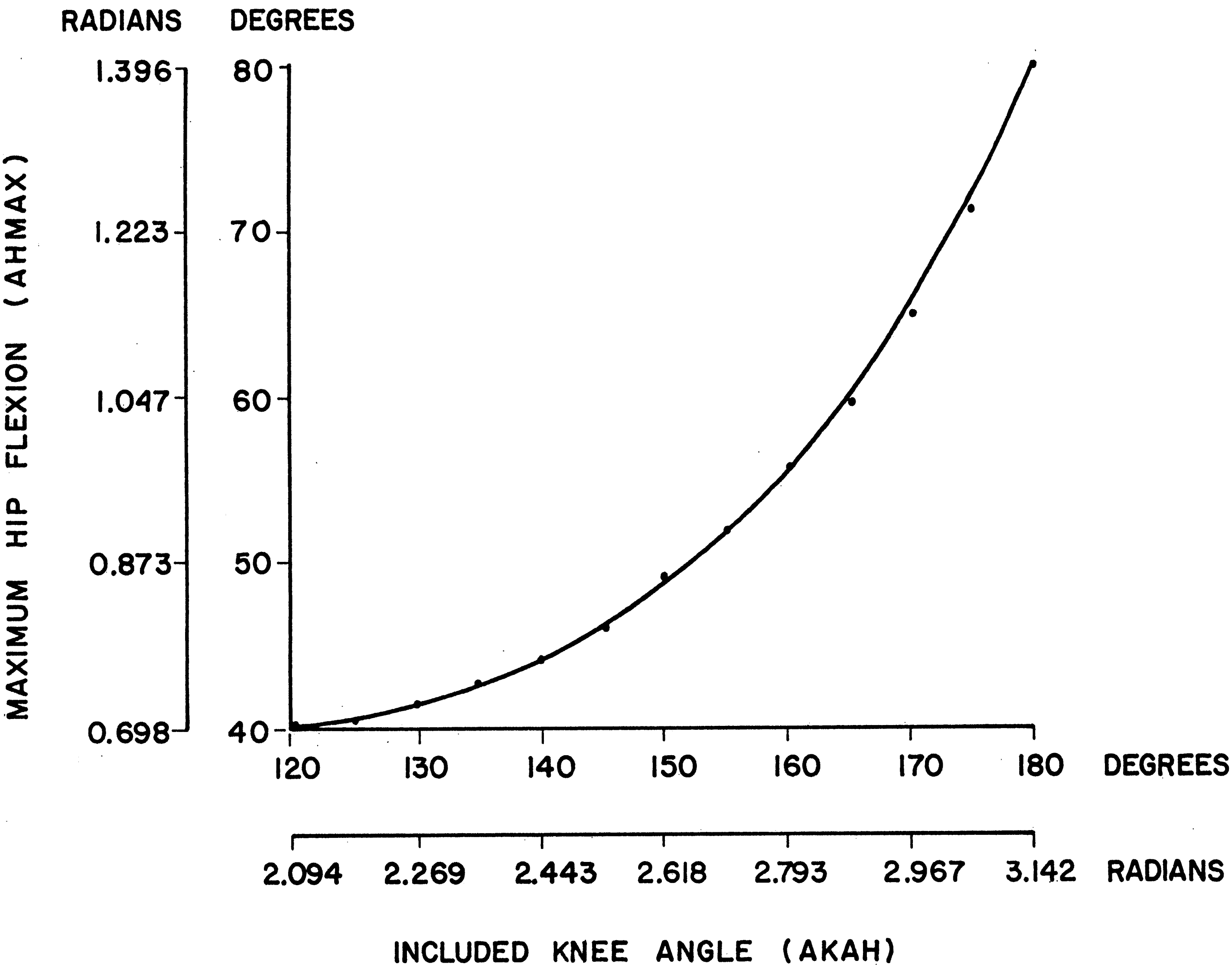


Figure B-6-1: Plot of Maximum Hip Flexion vs. Included Knee Angle.

APPENDIX C

LIST OF VARIABLES AND PROGRAM LISTING
(For Program Listing, see accompanying document.)

A. BODY DIMENSION VARIABLES

Character	Definition
LHD	Length of hand (wrist to hand c.g.)
LLA	Length of lower arm
LUA	Length of upper arm
LTK	Length of trunk
LTH	Length of thigh
LSK	Length of shank
LFT	Length of foot
WT	Body weight
ST	Body stature
WHD	Weight of hand
WLA	Weight of lower arm
WUA	Weight of upper arm
WTK	Weight of trunk (head, neck, and trunk)
WTH	Weight of thigh
WSK	Weight of shank
WFT	Weight of foot
CGELA	C. of g., from elbow to <u>lower arm</u>
CGSUA	C. of g., from shoulder to <u>upper arm</u>
CGHTK	C. of g., from hip to <u>trunk</u>
CGKTH	C. of g., from knee to <u>thigh</u>
CGASK	C. of g., from ankle to <u>shank</u>
RLLA	Reference length, lower arm (radius)
RLSK	Reference length, shank (tibia)
ZST	"Z" value for stature
ZWT	"Z" value for weight
ZSTR	"Z" value for strength

B. ANGLE VARIABLES

Character	Angle At	Sides From	To
AKTHZ	Knee	Thigh	Horizontal
AHSHZ	Hip	Shoulder	Horizontal
AHHH	Hip	Hand	Horizontal
AHSH	Hip	Shoulder	Hand
ASUAH	Shoulder	Upper Arm	Hand
ASHH	Shoulder	Hip	Hand
ASHHZ	Shoulder	Hand	Horizontal
ASUAHZ	Shoulder	Upper Arm	Horizontal
AELAUA	Elbow	Lower Arm	Upper Arm
AELAHZ	Elbow	Lower Arm	Horizontal
ASHUA	Shoulder	Hip	Upper Arm
AL5S1	Hip	L5/S1	Horizontal
ASHS1	Shoulder	Hip	S1
AS1SHZ	S1 Disc Surface	Shoulder	Horizontal
AASHZ	Ankle	Shank	Horizontal
AHTTK	Hip	Thigh	Trunk
AKAH	Knee	Thigh	Shank
ASSHN	Superior Surface of S1	Normal to S1	Horizontal
AASH	Ankle	Shank	Hip
AAHHZ	Ankle	Hip	Horizontal
AHHS	Hip	Hand	Shoulder

OTHER ANGLES

AHMAX	Maximum included Hip Angle (AHTTK) calculated as a function of included knee angle (AKAH)
BSUAHZ	BSUAHZ converted to Degrees
BELAHZ	AELAHZ converted to Degrees
AE	AELAHZ converted to Degrees
AS	ASUAHZ converted to Degrees
AH	AHSHZ converted to Degrees
AK	AKTHZ converted to Degrees
AA	AASHZ converted to Degrees
A5	A15S1 converted to Degrees

C. STRENGTH AND TORQUE VARIABLES

Articulation and Direction of Torque	Individual Subject Strength	Population Strength Value	Subject Strength/Angle Corrected	Percent of Maximum Value Torque	Task Torque Output Value	Maximum Hand Force/ Each Artic.
Elbow Flexion	FSE	PFSE	AFSE	PTEF	ELFX	MTEF
Elbow Extension	ESE	PESE	AESE	PTEE	ELEX	MTEE
Shoulder Flexion	FSS	PFSS	AFSS	PTSF	SHFX	MTSF
Shoulder Extension	ESS	PESS	AESS	PTSE	SHEX	MTSE
Hip Flexion	FSH	PFSH	AFSH	PTHF	HFX	MTHF
Hip Extension	ESH	PESH	AESH	PTHE	HEX	MTHE
Torso Flexion				PTTF	TRFX	MTTF
Torso Extension	EST	PEST	AEST	PTTE	TREX	MTTE
Knee Flexion	FSK	PFSK	AFSK	PTKF	KFX	MTKF
Knee Extension	ESK	PESK	AESK	PTKE	KEX	MTKE
Ankle Extension	ESA		AESA	PTAE		MTAE
Grip Strength	SGP					

D. ARTICULATION TASK TORQUE VARIABLES

Articulation	Original Torque Result of First Torque Computation	Storage Variable for Original Torque	Storage Variable for the No-Load Torque (FM = 0.0)
Elbow	TE	TEO	TEN
Shoulder	TS	TSO	TSN
Hip	TH	THO	THN
L5/S1	TS1	TS1O	TS1N
Knee	TK	TKO	TKN
Ankle	TA	TAO	TAN

Since the model must find articulation no-load torque and 1Kg torque, the original articulation torques are stored in storage variables, the force set to 0.0 and the no-load torque computed and stored in another storage variable. Then the force is set to 1.0 and the process repeated. Then, what used to be the "original" torque variables (e.g., TE, TS, TH, etc.) then become 1 kg torque variables for the maximum hand held force computations.

E. FORCE AND TORQUE VARIABLES

FM	Force magnitude
FMO	Storage variable for original force magnitude
FMD	Vertical force magnitude
FMH	Horizontal force magnitude
FMS	Storage variable for original horizontal force magnitude
WED	Weight at elbow - vertical
WSD	Weight at shoulder - vertical
WSDO	Storage variable for original weight at shoulder
WPD	Weight at hip - vertical
WKD	Weight at knee - vertical
PR	Abdominal pressure
FAB	Force due to abdominal pressure
FMU	Back muscle force
FCOMP	Compressive force on L5/S1 disk
RB	Torque due to reaction force at ball of foot
RH	Torque due to reaction force at heel of foot
TF	Torque about feet (sitting balance check)
THP	Torque at hip (sitting balance check)

F. DISTANCES

X	Horizontal distance from talus to load
Z	Vertical distance from talus to load
LSH	Shoulder to hand
A	A check to be sure LSH is not longer than the length of upper and lower arms plus hand.
G	A length factor that is common in the "TF" computation.
MA	Moment arm of abdominal pressure
D	Talus to hip
K	Horizontal distance from talus to hip
N	Vertical distance from talus to hip
LHH	Hip to hand

G. OTHER VARIABLES

The following are variables which store the results of input mode choices:

IMSQ	Body configuration input mode choice - squat
IML5	Pelvic angle choice - input or computation
IMST	Body configuration input mode choice - seated
KBS	Body stance
IFD	Force direction
IMS	Strength input mode
SEX	Sex of subject
IM	Stature and weight input mode
BSPT	With or without back support - seated push case
IGP	Grip strength choice to proceed or revise
CS	Continue or stop choice

The following variables are used for program flow control:

IB	IE	IQ
IC	IF	IP
ID	IG	IZ

Other Variables

SF	Stability factor for standing and squat cases
BMHP	Output form for maximum hand held load and retain balance - sitting case
BMAK	Same as above but for stand and squat case
RD	Constant to convert angles from radians to degrees

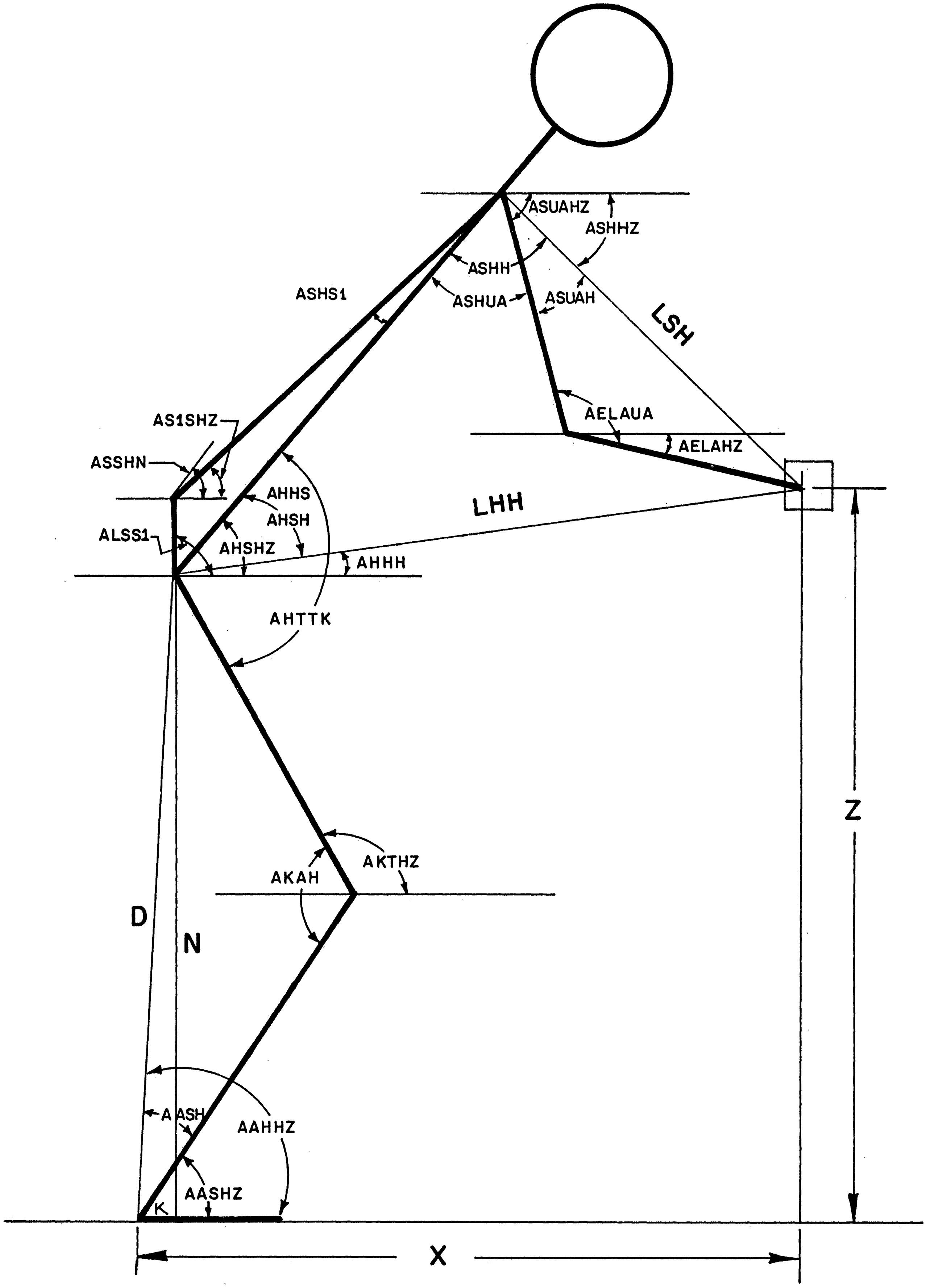
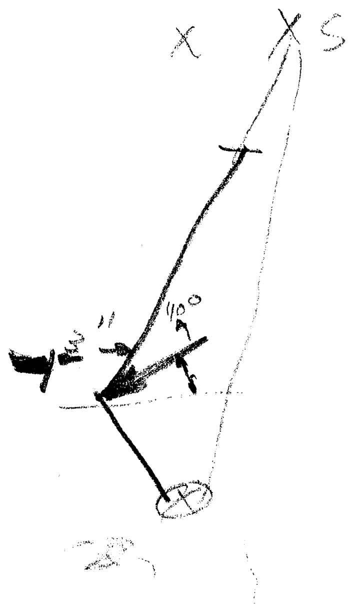


Figure C-1: Graphical Representation of Body Angles and Distances.



40°