Discussion of the 3-D static strength Michigan model

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Chapter 1

Introduction

The purpose of the present computerized static strength Michigan model (Mm) is to predict the loads on several body segments and joints as functions of the body posture and forces applied on the hands. The calculated results are compared with muscle strength data in order to judge whether or not a static isometric occupational task is hazardous.

Our special interest is low back biomechanical modeling so only this part of the model is subjected to a closer examination. The model compares the calculated (predicted) disc compression force on the L5/S1 motion segment intervertebral disc with compression strength data. Compression of the lower lumbar intervertebral discs is considered to be the most important factor to judge whether a task is hazardous or not in view of low back overloading. Mechanical loading can be related to low back pain [12,31]. The present model incorporates in fact two types of low back models. One model is a free body cantilever model of the lumbo-sacral joint, involving the erector spinae muscle, abdominal pressure and disc compression and shear forces. The load situation is statically determined and can be analysed by means of force and moment equilibria. The second model is clearly an additional subroutine which can be used optionally. It involves five bilateral muscle pairs, disc compression and shear forces at the L4/L5 level. The load situation is statically indetermined and two optimization routines are used to distribute the necessary reaction loads over the five muscle pairs. These models will be discussed in section 2.5.

A first impression of the computerized model was obtained by means of some exercises with it and by reading the User's Manual [1]. The program is user friendly. Use of keyboard and mouse is explained very well and the screen functions as a good interface between user and program. However, it is unknown whether or not the calculations results are valid and how they are determined. In order to modify or extend the model it is important to get more insight into the model, especially its mechanical backgrounds. Since no complete description of the present model exists, a number of publications, reports and books are studied in order to get a better understanding of the mechanical backgrounds, the mathematical descriptions and validity of the Michigan model.
Chapter 2

The Mechanical Backgrounds of the Model

2.1 The link model of the body

The human body is represented by means of a multibody system consisting of rigid links which are coupled by joints. The joints are purely kinematic elements. This means that they restrict the relative motion of the links. Energy cannot be dissipated, accumulated or added to the system via the joints or the links. This is based on the fact that the model is used to analyse static isometric postures occurring in occupational activities. In figure A.1 the linkage system is depicted as published by Chaffin [13,14,24]. This figure gives not much information about the assumptions made to come to this rigid body representation of the human body, e.g. the characteristics of the joints, segment names and a coordinate system. In figure A.2 a more extensive picture is given of the rigid body model.

In the present Michigan model the ankle to ball of foot links are not considered as links which can be manipulated. The feet always remain on solid ground, with floor contact via the balls of the feet. A leg is represented by means of two links: the lower leg (ankle-to-knee) and the upper leg (knee-to-hip). The ankle, knee and hip joints are revolute joints. The legs can therefore only be positioned in the sagittal plane, but different floor levels are possible. The axis through both hips always remains horizontal and perpendicular to the sagittal plane. An arm is modelled by means of two links: the upper arm (shoulder-to-elbow) and the lower arm (elbow-to-centre-of-grip-of-hand). The elbows are represented by means of revolute joints and the shoulder joints are spherical joints. The arms can be positioned in three dimensional space.

The remaining joints, L5/S1, L2/L3 and L4/L5, are spherical joints. The presence of the L2/L3 and L4/L5 joints is only known from a closer examination of the computerized model and its User's Manual [1]. In Chaffin et al.[9,12,13,24] concerning biomechanical modelling only the L5/S1 joint is mentioned.

The head and the neck are not considered as links, but their weights are taken into account in the mass and the mass distribution of the L5/S1-to-shoulder link. The L2/L3 and L4/L5 are also considered in the model. Why is the L3/L4 joint not taken into account. What about the length, the mass and the mass distribution of the back links coupled by these joints?

The source of the antropometric data is based on the results of several investigations. The estimates of mass distribution and the link lenghts are based on the work of Dempster.
et al. [18,19]. The mass estimates are based on the work of Drillis and Contini [20]. No extra information concerning spread, sex differences etc. is mentioned by Chaffin. These references on the subject of anthropometric data were not studied by Vincent.

Chaffin and Erig [13] have carried out an analysis of the sensitivity of the Min to postural and antropometric data inaccuracies. The differences between the results of a load situation were compared with variations in the postural input and antropometry input. They conclude that the model is extremely sensitive to errors in postural input data and to a lesser extent to errors in anthropometric data. May be it is better to conclude that the model is sensitive for variation of postural input and less sensitive for variation of the antropometric data.

2.2 Postural input

The body postures are defined by means of angles and link lengths which can be entered into the computerized model. These lengths are available together with link weights from predefined antropometric data in terms of percentile antropometry. Also specific values can be input in order to obtain more flexibility concerning the input.

The angle definitions are such that a body posture can be defined on the basis of e.g. subject photographs from which the angles can be measured. Figure A.3 shows the kind of angles which can be input in the computerized model. It is important to mention that the angles are defined with respect to fixed spatial planes and regardless of each other. So in case an angle is adapted the other angles keep the same value. A more detailed description of the input angles is given below. It is based on the User's Manual [1].

The lower leg vertical angle is used to position a lower leg in the sagittal plane. It is the angle in the sagittal plane between the lower leg link and the positive Y-axis. The angle convention is the same for both legs.

The upper leg vertical angle is used to position the upper leg in the sagittal plane. It is the angle in the sagittal plane between the lower leg link and the positive Y-axis. The angle convention is the same for both legs.

The upper arm horizontal and vertical angles are used to position an upper arm in 3-D space. The vertical angle is the angle “between the upper arm and the horizontal plane”. It is not clear whether or not it is the angle in the frontal plane between the projection of the upper arm link in the frontal plane and the positive X-axis for the right arm and the negative X-axis for the left arm. The horizontal angle is the angle “formed by the upper arm with the X-axis when looking down at the figure”. Is this the angle in the horizontal plane between the projection of the upper arm in the horizontal plane and the frontal plane? Is a spherical coordinate description meant with the vertical and horizontal angles as angles and the link length as radius?

The forearm arm can be positioned in 3-D space by means of two angles: the forearm vertical and horizontal angle. The description of these is the same as used for the upper arm.

The trunk can be positioned in 3-D space by means of three angles. The trunk can be considered as the triangle formed by the centre of the axis through the hips and the shoulders. The flexion angle is the angle formed by the “trunk axis”, which is the centre of the axis through the hips to the centre of the axis through the shoulders, projected onto the sagittal plane and the positive Y-axis. In case the trunk is erect the angle is 90 degrees and when the trunk is aligned with the Y-axis the angle is zero. The lateral bending angle is the angle “formed by the trunk axis and the sagittal plane.” The trunk axial rotation angle is the rotation angle of the trunk about the axis, from L5/S1 to the centre of the line segment between
the shoulder joints. Why is suddenly the L5/S1 joint mentioned? Only the shoulders are used to
define the angle and both the other trunk angles are based on the trunk axis through the centre
of the line segment between the hip joints.

The angles are defined regardless of each other. Projections of body links in the various
reference planes are probably used and these projections depend indeed on the angles. Can a
body posture be defined uniquely this way?

The body postures can also be defined by means of mouse control in combination with
frontal, sagittal and horizontal stick figure (multi-body linkage system) views. These figures
use projections in the planes in which the stick figures are depicted. See figure A.4

In the computerized model some limitations of kinematic constraints are implemented,
however not consequently. The flexion angle can increase so an impossible posture can be
defined, whereas the positioning of the arms is limited on basis of a maximum and minimum
upper and lower arm included angle. However, the User's Manual supplies ranges of angles
which are realistic.

It is clear that the orientation and position of the back links, shoulders-to-L2/L3, L2/L3-to-
L4/L5, L4/L5-to-L5/S1 and L5/S1-to-hips, are not defined by means of the input angles. Their
orientation and position have to be coupled to the body posture via extra constraints or a kinematic
model. This matter is discussed in section 2.5.1.

2.3 Mathematical description of the kinematics

This section is based on Chaffin and Erg [13]. The first step is computation of the link
lengths, link centre of gravity distances to proximal link ends and the link weights based
on anthropometric data. Secondly, the joint location coordinates are calculated based on
postural input angles and anthropometric data derived from the previous step. Thirdly, the
orientation of the links is computed in terms of unit vectors which define the orientation of a
link. They are directed from the proximal link end towards its centre of gravity and its distal
end. Every vector is resolved with respect to a coordinate system originating from each joint
and having the same orientation as the reference coordinate system. Why is this? Finally the
proximal joint to centre of gravity vectors \( \vec{V}_n \) for each link are calculated by multiplying its
unit orientation vector with the distance between the proximal end and the centre of gravity.
The link vectors \( \vec{V} \) are computed in the same way. Figure A.7 shows a free-body diagram of
the right arm.

Thus e.g.:

\[
\vec{V}_n = (V_{nxn} \hat{i} + V_{ny} \hat{j} + V_{nz} \hat{k}) \cdot LL_n
\]

Where:

- \( n \) link number,
- \( \vec{V}_n \) link vector,
- \( \hat{i}, \hat{j}, \hat{k} \) unit vectors forming an orthonormal
  base at the joints,
- \( V_{nx}, V_{ny}, V_{nz} \) unit vector components with respect to orthonormal
  coordinate system at the joint,
- \( LL_n \) link length.
Figure A.5, adopted from Chaffin and Erig [13], shows the vector conventions used in the computerized Michigan model. In the figure also joint reaction loads are depicted. However, this is only meaningful in case a free body diagram representation is used such as figure A.7. It would be less confusing to use a figure with only kinematic quantities and one with solely load quantities. The numbering is not consistent, the vector $\vec{V}_6$ does not exist, the links or the joints are not numbered, no consistent relation exists between link numbering, kinematic vector numbering and load vector numbering.

### 2.4 Loads on the links

Next to the forces due to gravity, acting on the centres of gravity of the links, also loads due to external forces acting on the hand grips can be applied. By means of entering the force magnitude and two angles the latter forces can be specified. The gravity forces are automatically applied based on link weights and the centre of gravity vectors (See figure A.6 which is adopted from the User's Manual). The forces are probably expressed in terms of spherical coordinates.

In earlier 2-D sagittal plane models developed by Chaffin et al. the following static force and moment equilibria were used in a recursive algorithm form [11,12,15], see figure A.8:

\[ R_j = R_{j-1} + W_L \]  
\[ M_j = M_{j-1} + \left( \bar{J}CM_L \cos(\theta_j)W_L \right) + \left( \bar{J}j - 1 \cos(\theta_j)R_{j-1} \right) \]

- $R_j$: joint reactive force at joint $j$
- $M_j$: reactive moment at joint $j$
- $\bar{J}CM_L$: distance from joint $j$ to centre of mass of link $L$
- $\theta_j$: postural input angle
- $W_L$: weight of link $L$
- $\bar{J}j - 1$: link length measured from joint $j$ to joint $j-1$

Some unusual definitions are presented in the above algorithm. The reaction forces act at the joints. Normally they act at the links which are coupled by the joint. A free body diagram, force and moment sign conventions and static equilibrium constraints can lead to an algorithm as described above. These are not discussed by Chaffin. The centre of gravity to joint distances and the link lengths have confusing names. The index $j$ is not always used as an index. In this simple sagittal case joint numbering and link numbering could be defined by means of one and the same index. It seems that also only lifting activities can be analysed since only vertical forces are considered.

In Chaffin and Erig [13] equilibrium equations are mentioned for use in 3-D models, however only concerning the right arm model.

\[ \bar{M}_1 = \vec{V}_1 \cdot \vec{F}_{\text{RHAND}} + \vec{V}_{1cg} \cdot \vec{W}_1 \]  
\[ \bar{M}_2 = \bar{M}_1 + \vec{V}_2 \cdot -R_1 + \vec{V}_{2cg} \cdot \vec{W}_2 \]
Equation 2.3 is the moment equilibrium of the right lower arm. Equation 2.4 is the moment equilibrium of the right upper arm. $\mathbf{F}_{\text{HAND}}$ is the force vector acting on the right hand, $\mathbf{M}_1$ is the moment vector at the right elbow, $\mathbf{M}_2$ is the moment vector at the shoulder, $\mathbf{W}_1$ is the lower arm weight vector, $\mathbf{W}_2$ is the upper arm weight vector and $\mathbf{R}_1$ is the elbow reactive force. See figure A.7. Force equilibrium equations are not mentioned.

The equations 2.3 and 2.4 are wrong. Considering figure A.7 the following equations are valid:

\begin{align*}
\mathbf{M}_1 + \mathbf{V}_1 \cdot \mathbf{F}_{\text{HAND}} + \mathbf{V}_{1cg} \cdot \mathbf{W}_1 &= \mathbf{0} \\
\mathbf{M}_2 - \mathbf{M}_1 + \mathbf{V}_2 \cdot \mathbf{R}_1 + \mathbf{V}_{2cg} \cdot \mathbf{W}_2 &= \mathbf{0}
\end{align*} 

These equation express the moment equilibrium of the upper and lower arm link. The sum of all moments about the elbow and the shoulder joint equal zero.

In figure A.9 the forces and moments on the pelvic bone are depicted. The loads on the hips cannot be uniquely determined from the static equilibrium in case only the loads on the L5/S1 joint are known. It is also not clear whether or not extra constraints are defined since the hip joints are revolute joints. The reaction forces on the feet could be computed from the global static body equilibrium and the forces and moments on the leg links by means of equations such as equation 2.5 and 2.6. This is only possible if just one reaction force and no reaction moment acts on each foot, else the global problem is statically indetermined, unless extra constraints are defined. This problem is also valid with respect to a so called body balance criterion used to predict whether or not a person is capable of keeping his balance when holding weights in a certain posture.

In Garg and Chaffin [24] a body balance criterion is given (figure A.10). This criterion is expressed by means of two equations:

\begin{equation}
|RT_a| \leq \min(|W_a \cdot L_{ah}|, |W_a \cdot L_{ab}|)
\end{equation} 

\begin{equation}
W_a \geq \max R_h, R_b
\end{equation}

What are the backgrounds of these equations? It uses two reaction forces at a foot. When this is applied to both feet the global static body equilibrium cannot be solved uniquely unless extra constraints are defined.

In the present Mm body balance is defined by means of three states: Acceptable Balance: the subject is capable of maintaining the posture; Critical balance: the hand load and the body weight are distributed over the feet in such a way that the subject is having difficulty to keep the posture; Unacceptable balance: the hand load and the body weight are largely concentrated on one leg, the subject is supposed to be loosing the balance and falling forward, backward or sideways.

This body balance criterion seems to be based only on the distribution of the hand loads and body weight over the feet. E.g. in case of a sagittal symmetric load situation one foot supports 50% of the load. What kind of numerical criterion is used to define a specific status?
2.5 Low back modeling

This section is concerned with the low back biomechanics used in the present model. Some backgrounds of the low back models developed by Chaffin and Schultz are discussed.

2.5.1 Low back models of Chaffin

As mentioned before the position and orientation of the back links, see section 2.2, have to be adopted from the body posture. In the early models of Chaffin the centre of the axis through the hips to the L5/S1 joint and the L5/S1 joint to the centre of the shoulders were considered as back links. This was based on the idea that concerning the low back the L5/S1 joint is subjected to the highest loads and therefore is the most vulnerable location in the low back. [10, 16]

In the present 3-D strength model also some other low-back joints are incorporated: the L2/L3 and the L4/L5 joints. The User’s Manual mentions X-ray studies in literature which provide empirical relations between disc orientation and the three-dimensional body postures. We still could not trace this literature except for two studies which deal with this subject in sagittal plane activities (Chaffin [10], Anderson and Chaffin et al. [5]). Both studies will be discussed below. Note that in the present Michigan model a trunk flexion of zero degrees means an horizontal trunk position, whereas in the following studies it denotes an erect trunk.

In Chaffin [10] the following is reported. The geometry of a male average vertebral column was “developed” based on work of Fick [23] and Lanier [26]. The dimensions of this column were proportionally scaled, based on the hips to shoulders distance. The curvature change of the spinal column during flexion of the hips was deduced from the work of Dempster [18]. During the first 27° of trunk flexion the pelvis does not rotate. For each additional degree of trunk rotation two thirds of a degree is produced by the pelvis and one third by the lumbar spine. On the basis of data of Davis [17], Allbrock [3] and Rolander [28] it was assumed that 22 and 29 percent of the lumbar rotation occurs at L5/S1 and L4/L5 level respectively. These references are not studied by Vincent. No accuracy, spread, sex differences and experimental methods are mentioned by Chaffin. These could be important since the model results are sensitive to postural data inaccuracies as reported by Chaffin and Erig [13].

In Anderson, Chaffin et al. [5] only the L5/S1 joint is taken into account, the previous study [10] is not mentioned or referred to. It was assumed that the trunk and the knee angles were the most important parameters that have influence on the orientation of the sacrum (S1) and on the relative orientation of L5 with respect to S1. The results of the study are two quadratic regression equations based on extensive data collection on four(!) subjects by Anderson. These are depicted in figure A.11. Also in this study it was found that the pelvis does not rotate for about the first 27° of trunk rotation. No measurement methods, population information and sex differences are discussed. These are may be discussed in the PhD thesis of Anderson [4], which we do not possess.

In the early 2-D sagittal plane models erector spinae muscle force $F_{MUSC}$, L5/S1 disc compression force $F_{COMP}$, L5/S1 shear force $F_{SHEAR}$ and an intra-abdominal force $F_A$ were included. In figure A.12 adopted from [12] a Chaffin model is depicted. The abdominal force $F_A$ is the result of the intra-abdominal pressure which is assumed to be a function of the hip torque $M_H$ and the angles $\theta_H$ and $\theta_T$ depicted in figure A.12. It was assumed that an intra-abdominal force could relieve the disc compression. The moment arm $D$ of $F_A$ is a sine function of the hip angle $\theta_H$: increasing from about 7 cm in erect position to 15 cm in
horizontal trunk position.

The erector spinae which are modelled as one muscle resists the total moment induced by the body weight and the hand loads. The line of action of the erector spinae force and the disc compression force are parallel with a distance $E$ of 5 cm. According to Bartelink [7], antagonistic muscle forces of abdominal muscles are neglected since these are assumed to be inactive during lifting activities in the sagittal plane.

The model described above is statically determined involving three coplanar equilibrium equations and three unknowns: $F_M$, $F_{COMP}$ and $F_{SHEAR}$. The same model is still used in the present 3-D model although it is at most useful in 2-D sagittal load situations. The question therefore is: are the assumptions made in this simple model still useful in three dimensional loading situations? Are there some special reasons for using this 2-D model in 3-D situations?

### 2.5.2 Low back model of Schultz

In order to find a more comprehensive low-back description also a 3-D model is implemented based on work of Schultz et al. [29,31,32,33]. This model can be used optionally in the Michigan model. It is a model which distributes the reaction forces at the fourth lumbar motion segment over five bilateral muscle pairs. A description of this Schultz model is given below (see also figure A.13).

The reaction loads due to body weight and applied loads on the hands is calculated for the L3 cross sectional area, probably a cross section through the third lumbar disc. In the present model the L4/L5 cross section is used. Why this difference? Schultz assumed that the third lumbar motion segment resists compression force, lateral and anteroposterior shear forces. Resistance against bending and twisting was neglected, so the muscles resist all moments about the L4/L5 joint. This was based on experimental results: a 5 Nm moment on a cadaveric lumbar motion segment results in about 1 to 5 degrees of rotation in bending or twisting [34]. In physical activities moments of about 100 Nm occur and the motion segments rotate a few degrees at most [31,34].

Firstly intra abdominal pressure was taken into account in the model, but measurements showed that it had no significant or consistent influence on the load situation. Schultz could not find a good correlation between measured intra abdominal pressure and both the measured intradiscal pressure and predicted compression of the spine [33]. In the present Michigan model intra abdominal force is still used concerning the simple low back model as described before but not in the Schultz model. Schultz did not use it any longer in his models based on experimental experience.

Five bilateral single muscle equivalents were assumed to act across the L3 section. The equivalence means that all muscles are assumed to have the same maximum contraction intensity. Different cross sectional muscle surfaces lead to different muscle forces. The muscles are the rectus abdominis, the internal and external oblique abdominal muscles, the erector spinae muscles and the latissimus dorsi. The trunk cross sectional geometry was scaled for each subject from its measured trunk diameters.

The static indetermined problem was solved by assuming that the muscles only experience tensile forces, while a prescribed value of maximum contraction intensity was not exceeded and an objective function was minimized. The objective functions used were minimal disc compression and minimal maximum muscle intensity. In the present model these objective functions are also used. They are both used successively and it is not clear how their solutions are processed.
The model of Schultz was used to analyse several loading situations. Validation studies are discussed in chapter 3. In the model the relation between the cross-sectional surface and the body posture is not considered or discussed. The model assumes that the motion segment is cross-sectioned parallel to the disc or vertebra surface. Compression is defined perpendicular to the cross-sectional surface and shear is defined in the surface. It is therefore important to know the disc or vertebra orientation as a function of the body posture. Moreover, the cross-sectional muscle surface or the muscle moment arm can depend also on the body posture.

The assumption that the motion segment resists no bend or twist moments must be handled with care since facet joints, ligaments and the intervertebral disc can resist moments in reality. This remark shows probably the weakness of the Schultz model. It deals only with active muscle force prediction and disc compression and shear. Kinematics are not discussed and neither other passive structures besides the intervertebral disc. In e.g. extensive twist and flexion the contribution of the passive structures can be important and also their overloading can lead to damage.

2.5.3 Ligament strain in the low back

An output parameter of the present Mm is a certain ligament strain. A study of Anderson and Chaffin et al. [5] deals with the subject of estimating ligament strain in the posterior regions of the lumbo-sacral joint by means of the sagittal plane kinematic model mentioned in section 2.5.1. The interspinous, supraspinous, iliolumbar, sacrolumbar, articular, yellow ligaments are considered and also the lumbodorsal fascia. From the results of several investigations resting lengths were adopted [2,22,27] and constitutive models were developed in terms of regression equations [8,25,30,35].

What is the role of the above study in the present model? In the present Mm the estimated ligament strain is only a function of the trunk flexion angle. Constitutive models are not used, reference lengths are not discussed.
Chapter 3

The Validity of the Model

In this chapter the validity of the present static strength model developed by Chaffin and Schultz is discussed. Only the low back models which are used in the present Michigan model are considered. The disc compression prediction is based on the model of Chaffin which is at most valid in 2-D sagittal plane activities. The model of Schultz is in fact not used for judgement of the load situation in the sense of disc compression. What is the role of the Schultz model in the Michigan model?

3.1 Validity of the Chaffin model

From studies of the failure mechanism of cadaveric motion segments it is known that often disc end-plate fractures occur due to compressive forces [12]. Figure A.14 shows the results of these studies. On the basis of these data the National Institute of Occupational Safety and Health (NIOSH) has recommended that predicted L5/S1 compression loads above 3400 N must be considered potentially hazardous for some workers (Action Limit). Values greater than 6400 N are hazardous to most workers (Maximum Permissible Load). These criteria are used in the present computerized model.

Shear forces are also calculated but they are not compared to experimental values since these are hardly documented or reported in literature.

The model has never been validated in the sense of comparing measurements of erector spinae muscle activity or intra-discal pressure to model predictions. Figure A.15 adopted from Chaffin [12] shows how the simple low back model is often used to analyse several lifting activities. The model can certainly be used to show how loads on the low back can depend on the overall body posture, the body weight and the external hand loads in the sagittal plane.

3.2 Validity of the Schultz model

The distribution model present in the computerized Michigan model is based on the distribution model of Schultz. It is not utilized to judge a loading situation. Differences are discussed in section 2.5.2. Schultz et al. subjected their model to two more or less extensive validation studies. These will be discussed below. [32,33]
3.2.1 Validation involving standing and sitting positions

In [33] a validation study is described in which four volunteers performed twenty-five tasks each. In isometric sitting and standing positions various loads were applied. Figure A.16 gives a picture of the mean measured EMG amplitudes and the various tasks performed.

The lines of action of the external loads, the locations of the head, the upper limbs and the trunk centres of mass were determined. This information was used to determine the reaction loads at the third lumbar level. Myoelectric activity was measured by means of twelve bipolar surface electrodes. Intradiscal pressure in the third lumbar disc was measured by means of a pressure transducer built into the tip of a needle. Intra-abdominal pressure was measured by means of a swallowed pressure sensitive radio transducer.

The mean measured intradiscal pressure was correlated with the mean predicted compression on the third lumbar vertebra by means of a linear least squares regression analysis. This resulted in a correlation coefficient of 0.94.

Predicted muscle contraction forces were correlated with mean measured myoelectric-signal amplitudes. The electrodes were placed bilaterally at the lumbar back three centimeters from the midline at the T8, L1, L3 and the L5 levels; bilaterally over the rectus abdominus two centimeters from the midline at the level of the umblicus; bilaterally over the oblique abdominal muscles, three centimeters above and anterior to the iliac spines (the ilium?). The correlation was done for the rectus abdominis with the abdominal muscle measurements, the erector spinae with the lumbar measurements and the sum of the internal and external oblique muscles with the oblique abdominal measurements. The left and right erector spinae predictions correlated with coefficients of 0.94 and 0.91, respectively. The left and right rectus abdominus correlated with 0.56 and 0.55, respectively. The correlation coefficients of the left and right external and internal muscle predictions with the EMG measurements were 0.57 and 0.20, respectively.

As mentioned earlier no consistent relationship was found between the measured intra-abdominal pressure and loads imposed on the spine.

3.2.2 Validation involving bending and twisting

In [32] another validation study is discussed in which 20 tasks were performed by ten subjects. The tasks were of the weight-holding and force-resisting kind in upright, upright twisted, lateral bended and combinations of bended and twisted isometric postures. In this study only myoelectric measurements were done. The placement of the twelve bipolar electrodes was different in comparison to the previous experiment. Electrode pairs were placed bilaterally three centimeters from the back midline at the C4, T8 and L3 levels. Another electrode pair was placed bilaterally 6 cm from the back midline at the L3 level. A pair of electrodes was placed over the rectus abdominis at the level of the umblicus, bilaterally two centimeters from the midline. A pair of electrodes was placed bilaterally over the oblique abdominal muscles 3 cm from the midline superior to the anterior iliac spines. A ground electrode was placed on the right ankle. No reasons for the differences compared to the previous study especially concerning the back measurements were discussed.

For each of the twenty tasks the means over the ten subjects of the myoelectric signal amplitudes of each electrode were computed. Also the mean of the means over the two electrode pairs at the L3 level was computed for the left and the right side.

The mean of the predicted left and right erector spinae muscle force was correlated with
the measured lumbar myoelectric signal for each task. The rectus abdominis predictions were correlated with the corresponding EMG measurements. The oblique abdominal measurements were correlated with the sum of the external and internal force estimates means for each task.

In this study two object functions were used to solve the mechanical problem. These objective functions are described earlier in this report. The results of the calculations derived from both methods do not differ much, but the correlation between the muscle contraction forces and the measured EMG activity has a smaller spread in case the minimum contraction criterion is used. In figure A.17 the results are depicted.

The authors conclude that the results are "less impressive" compared to previous studies [6,29,33] but this is only true for the erector spinae muscles. In case the same model is used as discussed above, the correlation coefficients concerning the other muscles are even better. Comparison with other studies [6,29] is hardly sensible since other models were used and only the erector spinae muscle contraction estimates were correlated with EMG measurements. The experiments used to validate the more simple models were sitting in the same positions with different arm activities [6] and isometric sagittal slightly flexed and upright weight-holding tasks. In these activities the erector spinae muscles are primarily active and a very good correlation between predicted muscle activity and measured EMG activity was obtained.

The authors conclude that the model is valid in a "semi-quantitative" sense, which can be considered reasonable. In the various tasks, as seen also in the previous study, the model predicts a strong muscle contraction in case also a higher myoelectric activity was measured. Figure A.18 shows the calculation result for some tasks.

3.2.3 general remarks

The major conclusion from Schultz et al. [32] is loads imposed on the lumbar trunk involving bending and twisting can be predicted moderately well but not as well as loads imposed by sagittal symmetric activities.

Hardly discussed are the backgrounds of the way the myoelectric muscle activities are measured and processed. This could be a rather important factor in model validation. A study could be made involving the same tasks which are used in combination with different electrode placement and signal processing.

The model predicted large latissimus dorsi contraction in almost all tasks described in section 3.2.1. This is due to the fact that they are modelled synergists of the erector spinae with a larger moment arm. The anterior muscles were hardly recruited in the various tasks according to the model. Myoelectric activity however showed substantial muscle activity in these muscles. The correlations could therefore be rather bad for these muscles.

The latissimus forces were not taken into account in the correlation with the back EMG measurements. The sum of the predicted erector spinae and the predicted latissimus dorsi muscles could be correlated with the measurements. Maybe this leads to a better correlation.

According to the authors trunk flexion moments load the spine heavily, whereas twisting and bending load the spine much more less. Chaffin et al. [9,10,12] concluded also that flexion moments can be rather hazardous due to the large erector spinae muscle contraction forces since the line of action of these muscles is relatively close to the spinal column. This leads to high compression forces on the spine.

It is important to mention that Schultz achieves equal forces of the bilateral muscle groups in case of sagittal symmetric load situations. The distribution model implemented in the present Mm does compute non equal muscle forces in these cases. This seems invalid.
Note that the Schultz model only predicts active muscle forces. The effect of kinematics, stress and strain of passive structures as well as the effects of the loads on the facet joints can not be predicted. These items are important in e.g. extensive flexed, bended and twisted postures.
Chapter 4
Concluding Remarks and Questions

The general impression of the computerized Michigan model is that the program is user friendly. However, it is very difficult to get an idea of the mechanical backgrounds, even after studying several literature sources. The most important items not fully understood are summarized below.

- What is the exact mathematical description of the low back joints and links in relation to the overall body posture? [section 2.2 and 2.5.1]
- Can a unique body posture be defined by means of the input angles? [section 2.2]
- Some indistinctness is present regarding the description of the static force equilibria. Has this any influence on the results of the computerized model? [section 2.4]
- What is the role of the 2-D erector spinae model in the analysis of 3-D loading conditions? Are there any special reasons or assumptions to use the 2-D back model in the Michigan 3-D model? [section 2.5.1 and 3.1]
- What is the role of the estimated ligament strain? [section 2.5.3]
- What is the role of the Schultz distribution model in the Michigan model? It is not used to judge a loading condition in the sense of disc compression loading. [section 3.2]
- What about the implementation of the distribution model in the Michigan model? Why are there sagittal non-symmetric muscle forces in sagittal symmetric loading situations? Why is the L4/L5 disc level modeled whereas Schultz modeled the L3/L4 level. [section 3.2.3]

These items need further clarification in order to extend or adapt the model or to study its validity.
Bibliography


Appendix A

Figures

Figure A.1: segment representation of the body as used by Chaffin et al.
Figure A.2: alternative segment representation of the body
Figure A.3: input angles in the computerized model
Figure A.4: screen of the computerized Mm concerning postural input
Figure A.5: vector conventions in the link model
Figure A.6: conventions of the hand forces

Figure A.7: the right arm model
Figure A.8: 2-D sagittal symmetric strength model

When \( \vec{M}_s \) and \( \vec{R}_c \) and the geometry of the pelvis are known, \( \vec{R}_b, \vec{M}_b, \vec{R}_g \) and \( \vec{M}_g \) cannot be determined uniquely.

Figure A.9: static equilibrium of the pelvis
**Figure A.10:** body balance

\[ R_h + R_b = W_a \]

Max. of \( R_h \) or \( R_b \leq W_a \)

---

**Figure A.11:** sacral rotation and relative L5/S1 rotation vs torso rotation for some knee angles
Figure A.12: simple low back model developed by Chaffin
Fig. 1. Schematic of the model used for internal force estimation. The five bilateral pairs of single muscle equivalents represent the rectus abdominis, the internal and external oblique abdominal, the erector spinae, and those parts of the latissimus dorsi muscles which cut the trunk sectioning plane. Contraction forces in these model muscles are denoted R, L, S, E, and C, respectively, with the subscripts denoting left and right sides. Inclination angles $\beta, \alpha$ and $\gamma$ were all set to 45°. Motion segment compression force is denoted C. Anterior shear force $S_a$ and right lateral shear force $S_r$. Abdominal cavity pressure resultant $P$ here was set to zero. Muscle cross-sectional area per side were taken as, respectively for R, E, S, C, and L: 0.006, 0.0168, 0.0146, 0.0389 and 0.003 times the product of trunk cross section depth and width. Centroidal offsets in the anteroposterior direction were, in the same order, taken as 0.19, 0.19, 0.19, 0.22 and 0.28 times trunk depth. In the lateral direction, they were taken as 0.12, 0.45, 0.45, 0.18 and 0.21 times trunk width.

Figure A.13: the model of Schultz
Figure A.14: compressive failure values obtained from cadavers
Figure A.15: predicted L5/S1 disc compression forces for varying loads lifted in four different positions
MEAN MEASURED AMPLITUDES OF MYOELECTRIC SIGNALS (ROOT-MEAN-SQUARE VALUES AVERAGED OVER THE FIFTEEN-SECOND RECORDING PERIOD)

<table>
<thead>
<tr>
<th>Activity</th>
<th>Lumbar Back Electodes (μV)</th>
<th>Rectus Abdominis Electodes (μV)</th>
<th>Oblique Abdominal Electodes (μV)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Left</td>
<td>Right</td>
<td>Left</td>
</tr>
<tr>
<td>Standing</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Relaxed</td>
<td>5</td>
<td>16</td>
<td>16</td>
</tr>
<tr>
<td>Resisting horizontal forces</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Flexion</td>
<td>46</td>
<td>48</td>
<td>21</td>
</tr>
<tr>
<td>Extension</td>
<td>16</td>
<td>23</td>
<td>23</td>
</tr>
<tr>
<td>Right twisting</td>
<td>6</td>
<td>16</td>
<td>105</td>
</tr>
<tr>
<td>Upright, arms in, holding 8 kg</td>
<td>10</td>
<td>22</td>
<td>21</td>
</tr>
<tr>
<td>Upright, arms out</td>
<td>8</td>
<td>21</td>
<td>17</td>
</tr>
<tr>
<td>Upright, arms out, holding 8 kg</td>
<td>33</td>
<td>38</td>
<td>28</td>
</tr>
<tr>
<td>Flexed 30°, arms out</td>
<td>45</td>
<td>46</td>
<td>32</td>
</tr>
<tr>
<td>Flexed 30°, arms out, holding 8 kg</td>
<td>73</td>
<td>67</td>
<td>43</td>
</tr>
<tr>
<td>Sitting</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Relaxed</td>
<td>6</td>
<td>14</td>
<td>9</td>
</tr>
<tr>
<td>Resisting horizontal forces</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Flexion</td>
<td>39</td>
<td>32</td>
<td>19</td>
</tr>
<tr>
<td>Extension</td>
<td>7</td>
<td>11</td>
<td>36</td>
</tr>
<tr>
<td>Right twisting</td>
<td>20</td>
<td>22</td>
<td>15</td>
</tr>
<tr>
<td>Weight-holding with right hand</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Position A, holding 0 kg</td>
<td>5</td>
<td>17</td>
<td>13</td>
</tr>
<tr>
<td>Position A, holding 4 kg</td>
<td>15</td>
<td>26</td>
<td>21</td>
</tr>
<tr>
<td>Position B, holding 0 kg</td>
<td>7</td>
<td>13</td>
<td>12</td>
</tr>
<tr>
<td>Position B, holding 4 kg</td>
<td>19</td>
<td>20</td>
<td>17</td>
</tr>
<tr>
<td>Position C, holding 0 kg</td>
<td>9</td>
<td>15</td>
<td>13</td>
</tr>
<tr>
<td>Position C, holding 4 kg</td>
<td>26</td>
<td>25</td>
<td>25</td>
</tr>
<tr>
<td>Position C, holding 0 kg</td>
<td>10</td>
<td>11</td>
<td>12</td>
</tr>
<tr>
<td>Position C, holding 4 kg</td>
<td>29</td>
<td>13</td>
<td>23</td>
</tr>
<tr>
<td>Position I, holding 0 kg</td>
<td>6</td>
<td>11</td>
<td>14</td>
</tr>
<tr>
<td>Position I, holding 4 kg</td>
<td>16</td>
<td>11</td>
<td>31</td>
</tr>
</tbody>
</table>

Figure A.16: mean measured muscle activity in relation to the various tasks

31
<table>
<thead>
<tr>
<th>Muscle</th>
<th>Function used to predict muscle forces</th>
<th>Minimum spine compression</th>
<th>Minimum contraction intensity</th>
</tr>
</thead>
<tbody>
<tr>
<td>Erector spinae</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Left side</td>
<td>0.86</td>
<td>0.88</td>
<td></td>
</tr>
<tr>
<td>Right side</td>
<td>0.60</td>
<td>0.74</td>
<td></td>
</tr>
<tr>
<td>Rectus abdominis</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Left side</td>
<td>0.82</td>
<td>0.70</td>
<td></td>
</tr>
<tr>
<td>Right side</td>
<td>0.92</td>
<td>0.83</td>
<td></td>
</tr>
<tr>
<td>Sum of internal and external abdominal oblique forces</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Left side</td>
<td>0.55</td>
<td>0.77</td>
<td></td>
</tr>
<tr>
<td>Right side</td>
<td>0.34</td>
<td>0.67</td>
<td></td>
</tr>
</tbody>
</table>

Figure A.17: correlation coefficients between mean muscle contraction forces and mean measured myoelectric activities
Table 1: Mean predicted forces (N) on the L3 motion segment using the two different objective functions.

<table>
<thead>
<tr>
<th>Task performed</th>
<th>Minimum compression solution</th>
<th>Minimum intensity solution</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Compr</td>
<td>L4 shear</td>
</tr>
<tr>
<td>Standing without twist</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Stand relaxed</td>
<td>470</td>
<td>0</td>
</tr>
<tr>
<td>Resist flex. 15 kg</td>
<td>1390</td>
<td>0</td>
</tr>
<tr>
<td>ext. 15 kg</td>
<td>690</td>
<td>0</td>
</tr>
<tr>
<td>last bend. 10 kg</td>
<td>620</td>
<td>90</td>
</tr>
<tr>
<td>last bend. 20 kg</td>
<td>880</td>
<td>150</td>
</tr>
<tr>
<td>twist. 10 kg</td>
<td>730</td>
<td>150</td>
</tr>
<tr>
<td>twist. 20 kg</td>
<td>970</td>
<td>160</td>
</tr>
<tr>
<td>Bend right. hold 4 kg</td>
<td>840</td>
<td>110</td>
</tr>
<tr>
<td>Arms lat. hold 2 - 2 kg</td>
<td>500</td>
<td>0</td>
</tr>
<tr>
<td>Standing with 45° twist</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Stand relaxed</td>
<td>480</td>
<td>0</td>
</tr>
<tr>
<td>Arms lat. hold 2 - 2 kg</td>
<td>640</td>
<td>0</td>
</tr>
<tr>
<td>Hands at chest. 2 - 2 kg</td>
<td>780</td>
<td>20</td>
</tr>
<tr>
<td>Arms anterior. 2 - 2 kg</td>
<td>1190</td>
<td>40</td>
</tr>
<tr>
<td>Standing twisted and flexed 30°</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hands at chest</td>
<td>1700</td>
<td>0</td>
</tr>
<tr>
<td>Arms anterior. 2 - 2 kg</td>
<td>2290</td>
<td>0</td>
</tr>
</tbody>
</table>

*The mean moments imposed on L3 by these weights in the six resist tasks were 44.3, 40.3, 27.6, 55.8, 25.6 and 51.2 N m.

Table 2: Mean trunk muscle contraction forces (N) required for some of the tasks. These were calculated using the minimum intensity solutions. Rectus abdominis and latissimus dorsi contraction forces were at most 90° and 110° N. E = erector spinae. X = internal abdominal oblique. Y = external abdominal oblique. Subscripts indicate left or right side.

<table>
<thead>
<tr>
<th>Task performed</th>
<th>E₁</th>
<th>E₂</th>
<th>I₁</th>
<th>I₂</th>
<th>X₁</th>
<th>X₂</th>
<th>Required intensity (N/cm²)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Standing without twist</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Stand relaxed</td>
<td>60</td>
<td>60</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>6</td>
</tr>
<tr>
<td>Resist flex. 15 kg</td>
<td>510</td>
<td>510</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>23</td>
</tr>
<tr>
<td>ext. 15 kg</td>
<td>0</td>
<td>0</td>
<td>160</td>
<td>160</td>
<td>140</td>
<td>140</td>
<td>23</td>
</tr>
<tr>
<td>right bend. 10 kg</td>
<td>110</td>
<td>40</td>
<td>60</td>
<td>20</td>
<td>70</td>
<td>20</td>
<td>14</td>
</tr>
<tr>
<td>right bend. 20 kg</td>
<td>310</td>
<td>120</td>
<td>50</td>
<td>50</td>
<td>120</td>
<td>40</td>
<td>23</td>
</tr>
<tr>
<td>twist. 10 kg</td>
<td>60</td>
<td>210</td>
<td>170</td>
<td>0</td>
<td>0</td>
<td>110</td>
<td>19</td>
</tr>
<tr>
<td>twist. 20 kg</td>
<td>80</td>
<td>260</td>
<td>290</td>
<td>0</td>
<td>0</td>
<td>220</td>
<td>31</td>
</tr>
<tr>
<td>Bend right. hold 4 kg</td>
<td>340</td>
<td>30</td>
<td>10</td>
<td>10</td>
<td>90</td>
<td>0</td>
<td>15</td>
</tr>
<tr>
<td>Arms lat. hold 2 - 2 kg</td>
<td>50</td>
<td>50</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>0</td>
<td>6</td>
</tr>
<tr>
<td>Standing with 45° twist</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Stand relaxed</td>
<td>50</td>
<td>50</td>
<td>0</td>
<td>0</td>
<td>10</td>
<td>10</td>
<td>7</td>
</tr>
<tr>
<td>Arms lat. hold 2 - 2 kg</td>
<td>100</td>
<td>150</td>
<td>0</td>
<td>0</td>
<td>10</td>
<td>10</td>
<td>9</td>
</tr>
<tr>
<td>Hands at chest. 2 - 2 kg</td>
<td>250</td>
<td>140</td>
<td>10</td>
<td>0</td>
<td>10</td>
<td>0</td>
<td>11</td>
</tr>
<tr>
<td>Arms anterior. 2 - 2 kg</td>
<td>580</td>
<td>210</td>
<td>90</td>
<td>0</td>
<td>90</td>
<td>0</td>
<td>24</td>
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<tr>
<td>Standing twisted and flexed 30°</td>
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<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hands at chest</td>
<td>700</td>
<td>670</td>
<td>10</td>
<td>0</td>
<td>10</td>
<td>0</td>
<td>30</td>
</tr>
<tr>
<td>Arms anterior. 2 - 2 kg</td>
<td>1120</td>
<td>860</td>
<td>310</td>
<td>0</td>
<td>110</td>
<td>0</td>
<td>44</td>
</tr>
</tbody>
</table>

Figure A.18: model predictions in relation to various tasks