International Journal of Industrial Ergonomics, 8 (1991) 261–277
Elsevier

Lumbar load during one-handed bricklaying

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(Received May 14, 1990; accepted in revised form May 6, 1991)

Abstract

Lumbar load, as indicated by the moment of force and the force at the lumbosacral disc, was determined for one-handed bricklaying tasks using a dynamic 3-D model, 'The Dortmunder'. The grasp height differed (90, 50, 10 cm). By contrast, the final postures were assumed almost upright in all cases. This resulted in considerable variance in the postures during the computer-simulated movements. The task duration varied (2.0, 1.5, 1.0 s). The lower the grasp height and the shorter the time, the higher the lumbar load (moment of force at L5-S1 up to 140 Nm, compressive force up to 6 kN), and the larger the differences between dynamic and static calculations. Increasing brick mass (0, 5, 10 kg) leads to an upward shift in the moment and compression curves (20 Nm or 1 kN per 5 kg). For the assessment of lumbar load during the analysed bricklaying tasks, the lumbosacral moment of force was first classified according to Tichauer (1978). Bricklaying involving a 50 cm grasp height requires 'selection of labor, careful training and rest pauses'. Lower grasp heights or bricks of 10 kg should not occur throughout 'the entire working day'. Lumbosacral force was then compared with lumbar strength values provided in the literature. These vary within a wide range (0.8–13 kN). Strength mean $\pm$ *s.d.* amounts to 5.0 ± 2.2 kN for the total sample ($n = 507$), to 5.8 ± 2.6 kN for males ($n = 174$) and to 4.0 ± 1.5 kN for females ($n = 132$). Strength depends primarily on age. Assuming linear regression models, strength (in kN) is 10.53–0.97/decade for males ($r^2 = 0.39$) and 7.03–0.59/decade for females ($r^2 = 0.35$). A strength prediction model considering 3 additional factors was developed ($r^2 = 0.62$) in order to explain most of the remaining variance. Since average values may overestimate an individual's strength, the mean or regression model value should be reduced by the *s.d.* of the respective sample. This would result in a lumbar load limit of 5.5 kN for 25-year-old men and 2.6 kN for 55-year-old men. Corresponding values for women are 4.1 and 2.3 kN. If the brick-supply stack is 90 cm high, the lumbar load limits will not be exceeded for any person in these age groups. By contrast, all limits would be exceeded for a 10-kg 1-s brick transfer from a grasp height of 10 cm. In conclusion, to ensure that the predicted lumbar load during bricklaying remains below the limits, the brick-supply stack should be above 50 cm.

Relevance to industry

This paper provides a biomechanical determination and assessment of the load on the lumbar spine during bricklaying tasks with different grasp heights. Bricks should be pre-positioned to permit grasping in erect postures.

Keywords

Bricklaying, body postures, biomechanical analysis, lumbar load, movement simulation, compressive strength.

Introduction

Construction workers frequently manipulate loads. They also complain of low back pain and are often absent from work (Hettinger, 1985; Zuidema, 1985). An analysis by David (1985) of the UK 'Health and Safety Statistics' for the years 1977/78 revealed that the incidence of back injuries at work amounted to 7.7 per 1000 employed in the construction industry compared with 5.8 for all industries. David ascribed 20–30% of 'over 3 days injuries' to manual materials handling. The incidence of spinal diseases is above average within the construction industry. Bricklayers are particularly affected (Klotzbücher, 1987). They are consequently classed as a 'population at risk with regard to lumbar injury' (Zuidema, 1985).

The aim of this study is the biomechanical determination and assessment of lumbar load during bricklaying. Bricklaying involves trunk inclination, lateral turning, sagittal flexion, and probably also lateral bending and twisting. Asymmetrical loads, such as found in one-handed lifting tasks, require three-dimensional analyses. Most of these have been static (e.g., Schultz et al., 1982). Spatial dynamic simulation calculations are relatively rare (e.g., Jäger, 1986; Chen and Ayoub, 1988). In the context of assessing lumbar load, values for lumbar compressive strength are collated and the main factors indicated.

2. Methods

2.1. Biomechanical model

Lumbar load during bricklaying is determined using an extended version of a biomechanical human model developed by Jäger (1986). The model 'The Dortmunder' permits the calculation of time curves for mechanical quantities such as pressure, force, and moments of force at the lumbar discs for predefined movements. The movable segments assumed in the model are the feet, shanks, thighs, hands, forearms, upper arms, and shoulders as well as a trunk with fifteen subdivisions and the segment 'head', including the neck as a single rigid body. The provision of jointed sections within the trunk permits simulation of sagittal and lateral bending as well as twisting within the trunk. However, the analyses in this paper only refer to sagittal flexion of the spine. A more detailed description of the skeletal structure of the model is contained in appendix A.

One-handed brick-turnover tasks are asymmetrical to the median sagittal plane. An adequate analysis model requires a more comprehensive simulation of the musculature than for symmetrical activities where 'one-back-muscle-models' predominate (e.g., Morris et al., 1961; Chaffin, 1969). The three-dimensional model here considers a total of 3 abdominal muscles and 1 back muscle on each side of the body. In the back region these are the erector spinae muscle group and, at the front

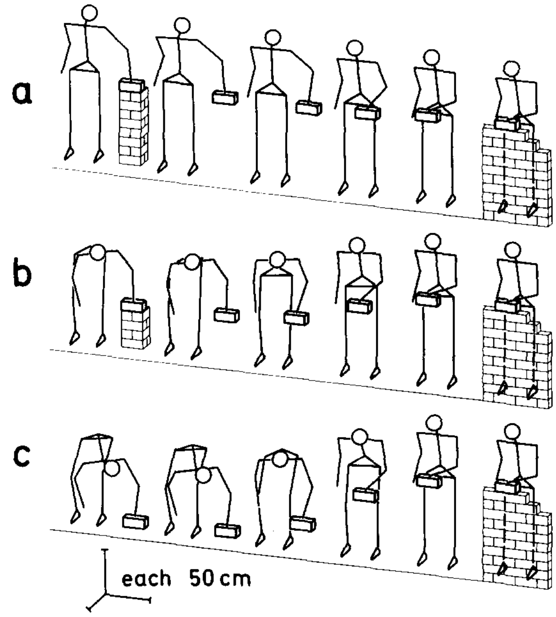

Fig. 1. Motions during bricklaying, biomechanically analysed according to different grasp heights: (a) 90 cm, (b) 50 cm, (c) 10 cm.

of the body, the oblique abdominal muscles, both the internal and external parts, as well as the rectus abdominal muscle. Appendix B contains a more detailed description of the muscular structure considered in the biomechanical model.

Although discussion about the support of the spine provided by the intra-abdominal pressure is contradictory (McGill and Norman, 1987a), it is ascribed a minor supportive function in the biomechanical model in this paper. A detailed description is contained in appendix C.

2.2. Task simulation

2.2.1. Body postures

One-handed turnover of bricks, the task chosen for the biomechanical analysis of lumbar load, is typical in bricklaying. A schematic representation of three movements during the turnover of bricks is provided in figure 1. The assumption of three grasp heights (90, 50, 10 cm) is based on height differences in the brick-supply stacks (see a, b, c).

The lateral distance (30 cm from the shoulder joint) is identical for all three grasp positions. During bricklaying (final posture in a, b, c), the same positions for the brick relative to the body were assumed, i.e., 110 cm above the ground and approx. 20 cm in front of the body for an assumed body depth of approx. 20 cm. After establishing the heights of the supply stack and the wall, the corresponding initial and final postures were modelled in a laboratory for a subject 175.6 cm tall. The three grasp postures and the bricklaying posture were each adopted five times. A tape measure was used to measure the positions of the relevant body joints relative to a coordinate system constructed of rods. Trunk, arm, and leg angles relative to the horizontal and vertical axes of the coordinate system were determined using a goniometer. The measured distance and angle values were averaged, and related to a body height of 173.3 cm (50th percentile of the German male population; Jürgens, 1981).

As figure 1a shows, the brick is grasped in the upper position through lateral abduction of the upper arm. The forearms hang down almost vertically. The trunk approximates to an upright position (7.5% sagittally inclined). A spine with normal lumbar lordosis and normal thoracic kyphosis was assumed here. By contrast, the lower and middle grasp positions require forward inclination and turning of the trunk to the left (see b and c). A fully flexed spine in the sagittal plane, as assumed for the lower height, corresponds to an inclination of the upper spinal segments of approx. 36° (table A.1, appendix A) in addition to the assumed initial pelvic inclination of 80°. For the medium grasp height, 50% of the total sagittal flexion mobility within the trunk as well as a 60° pelvic inclination were assumed. The displacement of the trunk to the left in the middle and low grasp positions is due to a 30° rotation of the hip. An additional 10° rotation of the shoulders to the left means that the 'shoulder axis' is not parallel to the 'hip axis' (see b, c), which, in turn, is not parallel to the 'foot axis'. Shoulder, hip and foot axes are understood here as the link lines between the shoulder, hip or ankle joints, respectively.

The final postures for all three turnover movements were assumed identical (figure 1): the legs vertical, the trunk 7.5° sagittally inclined, the spine with 'normal curvature', the right arm hanging down with a slight elbow flexion, the left upper arm hanging vertically downwards, and the left forearm held nearly horizontally and turned approx. 40° inwards. The position of the hands was assumed to be equivalent to the forearms throughout. During the movement, the flexed spine assumed in the medium and low grasp postures gradually adopts the normal curvature found in the final posture. The assumption that the load always remains in the field of vision permitted the determination of the head's position (see a–c).

2.2.2. Movement simulation

The trajectories of both the body segments and the brick were not measured using a motion analysis system. Instead, they were computer-simulated according to Jäger and Luttmann (1987) on the following assumptions: the left foot remains at rest during the movement, the left heel being the origin of an inertial coordinate system. Rotation of the shank is described as a rotation around the ankle, and movement of the thigh as superposition of the rotation of the thigh around the moving knee joint. Accordingly, the trajectories of the 'upper' body segments and the load can be interpreted in terms of a superposition of the rotations in the 'lower' joints. It should be noted, however, that the rotational axes are usually not parallel.

In figure 1 the brick is seen to describe curved lines in space as a result of the rotations in the arms, trunk, and legs. Since the brick itself is not rotated, the path of the brick only represents a translational movement. The above rules were applied to determine the trajectories of the joints and the centers of gravity of body segments and load. The following assumptions were made: initially and finally all the body segments and the load are at rest. They are therefore accelerated at the beginning and retarded at the end. Acceleration and retardation phases are assumed equally long and amount to half the total duration. The angular acceleration is sinusoidal, positive during acceleration, negative during retardation (Slote and Stone, 1963). The angular velocity follows a square sine curve, increasing during acceleration and decreasing during retardation. The time curve of the angle traversed during the rotation of a body segment can be determined through integration of the velocity curve.

3. Results

A height of 173.3 cm, corresponding to the 50th percentile of the German male population (Jürgens, 1981), was assumed for the following calculations. A body mass of 69 kg was assigned (Jelliffe, 1966). Clothing weight was disregarded, the upward vertical height shift due to footwear was assumed to be 3 cm.

3.1. *Several indicators of lumbar load during bricklaying*

Figure 2 shows the time curves of several indicators of lumbar load and the kinematics for the one-handed turnover of bricks of differing mass (0, 5, 10 kg). In the upper eight diagrams (Ia–d, IIa–d) the continuous lines represent the results of dynamic model calculations and the broken lines the results of static analyses. The assumed movement corresponds to the posture sequence in the middle of figure 1. The brick is grasped laterally at a height of 50 cm and laid medially in front of the body at a height of 110 cm. The diagrams in row III indicate the coordinates of three important points during the movement: the brick in the left hand ('load'), the lumbosacral disc ('L5-S1'), and the center of gravity of the body segments above L5-S1 ('body'). The assumed turnover duration of 1.5 s represents a 'medium' value for whole-body movements.

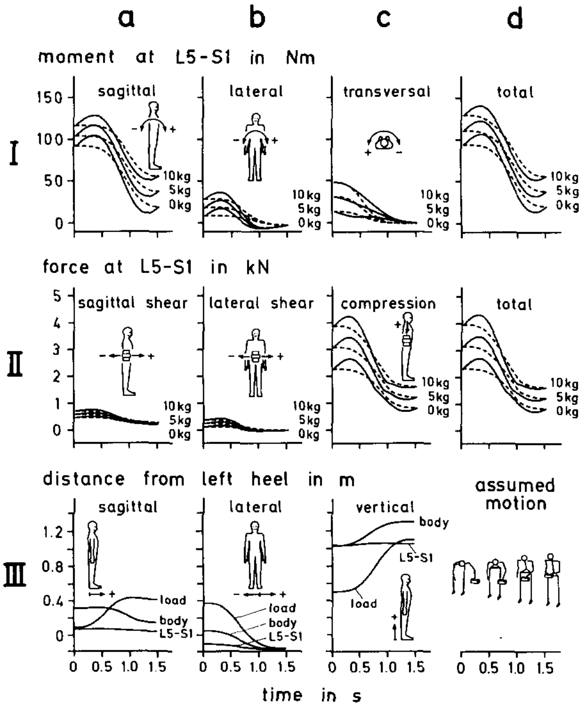

Fig. 2. Various indicators of lumbar load (moments of force and forces at the lumbosacral disc) and kinematics during bricklaying, assumed grasp height 50 cm. Key: ——— dynamic analysis - - - - - - static analysis

The time curves for the lumbosacral moment of force and its components in a trunk-related coordinate system are represented in the upper 4 diagrams Ia–d). Diagram Ia shows the 'sagittal lifting moments' which would also occur during bilateral symmetrical activities. The 'lateral bending moments' (Ib) describe the asymmetry of the load handling relative to the median plane. The 'transversal torsion moments' (Ic) take effect in the cross-sectional plane of the trunk at the level of the L5-S1 disc and describe the effect of the forces perpendicular to the longitudinal trunk axis. Diagram Id shows the curves for the total moment of force, this resulting from the vectorial addition of the three components of moment. Comparison of the three moment diagrams reveals that the sagittal lifting moments (Ia) assume higher values than the lateral bending moments (Ib) and the transversal torsion moments (Ic). Consequently, the curves for the total moment (Id) essentially reflect the curves for the sagittal lifting moment.

The second row of diagrams (IIa–d) shows the time curves for the forces transmitted via the lumbosacral joint. The components of the shear force are represented in IIa–b, and the compressive force in IIc. The total force indicated in diagram IId is derived by vectorial addition of these three orthogonal forces. Comparison of the magnitude of the force values reveals that the shear forces are considerably smaller than the compressive forces. Consequently, the latter represent the main component of the total force. The relatively large compressive forces result from the large sagittal moments (Ia) which are compensated for by the back muscles. Since, in the approximation in the biomechanical model, the force vector of these muscles has a line of action parallel to the spine (app. A) and has short lever arms, back muscle forces result in compression of the L5-S1 disc.

3.2. Influence of grasp height, working speed and brick weight on lumbar load

Whereas in figure 2 lumbar load was described using several indicators for a single activity with constant parameters, figure 3 represents the time curves of a single indicator for different grasp heights and working speeds. The compressive force on L5-S1 was calculated for the turnover of bricks (0, 5, 10 kg) for differing grasp heights (90, 50, 10 cm) with a single laying location and differing task-execution values (2.0, 1.5, 1.0 s). The continuous lines represent the results for dynamic, and the broken lines the results for static model calculations. The assumed movements of the body segments and the load are indicated in figure 1.

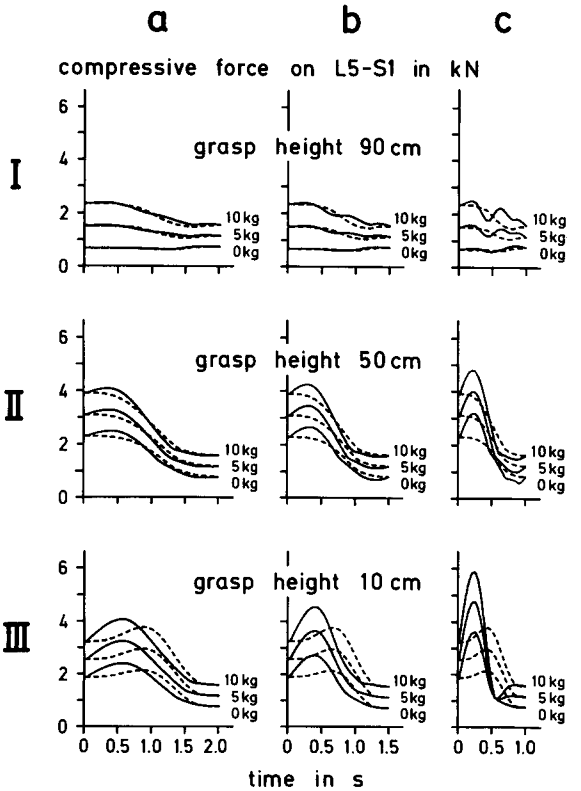

Fig. 3. Lumbar load during bricklaying for various grasp heights, time requirements, and load masses; the assumed motions are shown in figure 1: figure 1a corresponds to the diagrams Ia–c from figure 3, figure 1b corresponds to the diagrams IIa–c from figure 3, figure 1c corresponds to the diagrams IIIa–c from figure 3. Key: ——— dynamic analysis - - - - - - static analysis

Comparison of the compressive-force curves reveals that, as the grasp height decreases (90, 50, 10 cm), an increasing number of curves with a pronounced maximum (IIa, IIIa) develop from almost horizontal force curves (Ia). The same effect is apparent in the comparison of the diagrams Ib–IIIb or Ic–IIIc. This is mainly due to the difference in trunk inclinations with the attendant difference in the lever arms of the upper body and load. At the beginning of the movement the trunk is erect for a high grasp height (figure 1a) and almost horizontal for a medium grasp height (figure 1b). In the case of the extremely low grasp height of 10 cm, the trunk is inclined so far forwards and sagittally flexed that the shoulders are lower than the hips. The lever arms of the upper body and brick for the medium grasp height reach their maximum value at the beginning of the movement. However, at the low grasp height the maximum is only reached towards the middle of the movement. This leads to maximum values in the static compressive force curves in diagram IIa at the beginning of the movement and in IIIa towards the middle.

The influences of inertia are indicated by the differences between static and dynamic analyses. During a turnover act with upright trunk (Ia–c) the dynamic effects are relatively small since the heavy upper body is moved only slightly. The grasp height of 50 cm (IIa–c) is associated with a steeply inclined and flexed trunk. Consequently, large acceleration and inertia forces arise during erecting and straightening. The extremely low grasp height of 10 cm (IIIa–c) produces the highest values for lumbar load. As the paths taken by the body segments and the load are the longest here, the magnitude of the requisite acceleration and inertia forces is the greatest.

4. Discussion

4.1. Assessment of analysed bricklaying tasks

The lumbar load during a typical bricklaying task is described in section 3.1 on the basis of force and moments of force acting at the lumbosacral disc. Force and moments are vectorial quantities. It is customary in ergonomics and occupational health to use the total moment and the compressive force at the lumbosacral disc in the assessment of activities involving lumbar load. However, since the NIOSH guidelines (1981) only relate to lifting activities symmetrical in the sagittal plane, they cannot be applied to the asymmetrical bricklaying tasks analysed here.

4.1.1. Lumbar moment of force

Tichauer (1978) provides a classification of professional load-manipulation activities based on total values for moments of force. Four moment categories with relevance to the selection and training of working persons or to the adherence of rest breaks are identified. The limit values given by Tichauer are 40, 85 and 135 Nm (350, 750, 1200 in.lb). When this moment classification is applied to the 1.5-s bricklaying task, all curves in diagram Id in figure 2 exceed the 85-Nm limit. Even maintaining the inclined upper body posture (i.e., static without any external load) results in a lumbar load which, according to Tichauer, demands a 'selection of labor, careful training and rest pauses'. In diagram Id, the limit of 135 Nm is only exceeded in the dynamic curve for a 10-kg brick. However, the results in figure 3 must be additionally considered when determining the moments of force during bricklaying. Even though another indicator of lumbar load is involved, figure 3 still reveals that the lower the grasp height or quicker the task execution, the higher the lumbar load. Consequently, a lumbosacral moment greater than 135 Nm can occur for low loads when the grasp height is lower or the execution quicker than assumed in figure 2 (50 cm, 1.5 s). For example, the peak total moment at L5-S1 amounts to 153 Nm for a grasp height of 10 cm, 1s duration and 0 kg load and to 145 Nm for 10 cm, 1.5 s and a 5 kg load. Tichauer states that, where moment values exceed 135 Nm, the turnover activities should not continue for 'the entire working day'. There is also need for 'great care in recruitment and training' in such cases.

4.1.2. Lumbar compression

4.1.2.1. Collation of strength data. Compressive strength values can be used in the assessment of compressive forces. Table 1 contains 13 literature references (1st column) which indicate the static compressive strength of lumbar segments of various length, ranging from total lumbar spines to isolated vertebrae. Lumbosacral and thoracolumbar joints are included. There are between 5 and 145 segments per reference ('*N*', see 2nd column). Not all segments were included in the present study ('*n*', see 3rd column). Segments were rejected on account of the donor being too young, dynamic testing, and the use, inter alia, of thoracic or 'incomplete' segments. The latter are specimens

Table 1
Investigations into static lumbar compressive strength.

Reference	*N*	*n*	Strength in kN	
			Mean	*s.d.*
Wyss and Ulrich, 1954	8	8	5.89	2.24
Brown et al., 1957	5	5	5.20	0.54
Perey, 1957	145	142	5.15	2.10
Decoulx and Rienau, 1958	9	9	4.41	1.14
Evans and Lissner, 1959	11	11	3.51	1.22
Roaf, 1960	18	3	4.83	2.06
Eie, 1966	25	16	3.70	1.60
Farfan, 1973	65	39	3.84	1.22
Hutton et al., 1979	58	23	5.35	2.67
Hansson et al., 1980	109	109	3.85	1.71
Hutton and Adams, 1982	33	33	7.83	2.87
Brinckmann and Horst, 1983	22	22	6.42	2.00
Brinckmann et al., 1989	98	87	5.35	1.76
Female	–	132	3.97	1.50
Male	–	174	5.81	2.58
Total	606	507	4.96	2.20

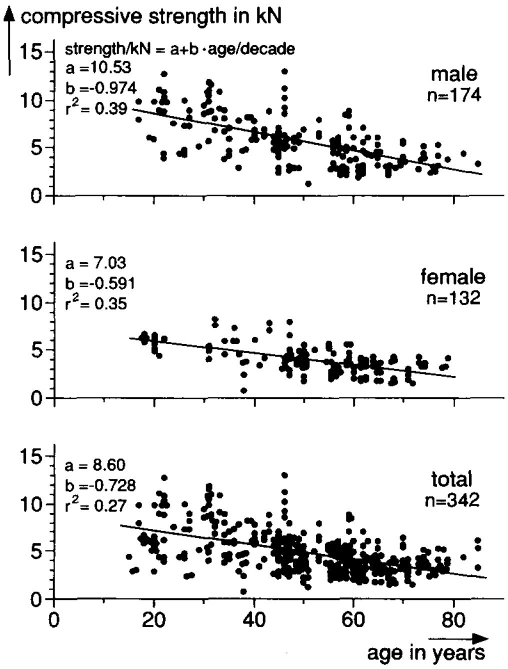

Fig. 4. Strength values from investigations into static lumbar compression, taken from the literature, according to age and gender.

consisting of less than a complete vertebra. Mean and standard deviation values (*s.d.*) for the compressive strength are listed in the right-hand columns. The lower part of table 1 shows the values for the male and female samples, as well as for the sum of all segments, including the specimens from donors of undocumented gender. Owing to the high standard deviation values, correlation analyses were performed on strength and various influencing factors.

The dependence of compressive strength on age, the factor with the greatest influence (Jäger and Luttmann, 1991), is demonstrated in figure 4. The upper diagram provides the strength test results for segments from male donors, the middle diagram the data for female donors, and the lower diagram refers to all segments. Figure 4 indicates in particular the age range of the donors. In spite of the wide range, more specimens were available from older donors than from young donors. Where the assumed age is > 65 years, no strength values exceed approx. 6 kN in any of the diagrams. For an age < 30, there are no specimens of known gender with a strength value under approx. 4 kN. For the total sample, the lowest strength values are around 3 kN (5 specimens). The regression lines and linear regression functions indicated in figure 4 demonstrate the decrease in compressive strength with age. Although the strength is greater for males than for females, in the case of men it decreases more rapidly with age.

The wide scattering which remains in the age and gender-related analyses must be due to influences other than age or gender. The result of additional variance analyses is shown in the following equation:

$$\begin{aligned}&\text{compressive strength/kN}\\&\quad = (7.26 + 1.88 \cdot G) - (0.494 + 0.468 \cdot G)\\&\qquad \cdot A/\text{decade} + (0.042 + 0.106 \cdot G) \cdot C/\text{cm}^2\\&\qquad - 0.145 \cdot L/\text{unit} - 0.749 \cdot S\end{aligned}$$

The square regression coefficient is 0.62. The variance analysis is based on 262 values taken from 120 persons for whom the 5 factors with the greatest influence are documented: age 'A', gender 'G' (female coded with '0', male '1'), the cross-sectional area of the specimen 'C', the so-called lumbar level 'L' ('0' for L5-S1 discs, '1' for L5 vertebrae, etc. and, finally, '10' for T12-L1 discs). The fifth factor, 'structure S' refers to whether the strength is determined for a disc ('0') or a vertebra ('1'). The intercept difference between male and female elements is almost 2 kN. The decrease in strength with age is almost two times greater for males. In contrast to the male sample, the dependence on the cross-sectional area was relatively small for the female sample. The decrease in strength with lumbar level is almost 0.15 kN per unit of lumbar level, i.e., between a disc and the adjacent vertebra. The strength of vertebrae is almost 0.8 kN lower than that of intervertebral discs.

4.1.2.2. Comparison of the calculated force with strength values. During one-handed bricklaying tasks (see section 3) there is wide variance in the compressive force on the lumbosacral disc dependent on working speed, grasp height, and brick mass (approx. 0.7–6 kN, cf. figure 3). The data are taken from males of 'average height and dimensions' (173.3 cm, 69 kg). The measured values for the compressive strength of lumbar segments from male donors vary similarly (approx. 1.3–13 kN). Consequently, some bricklaying tasks lead to a lumbar load which exceeds at least the lower strength values for the lumbar spine. The results in figure 3 show that a compressive force value of 5.8 kN is not exceeded by any of the curves for grasp heights of 90 or 50 cm. A value of 5.8 kN also represents the average strength for lumbar spine elements from male donors (table 1). However, higher force values were calculated for a grasp height of 10 cm, a 1-s duration, and a 10-kg brick (IIIc in figure 3). In this case, the compression curve reaches a maximum of nearly 6 kN.

Ideally, activities should be planned to exclude every possible health risk for the working persons. However, given a minimum value of about 1 kN in the compressive strength data collated in figure 4, such demands are clearly unrealistic. On the other hand, assessments of load manipulation based on the average value may overestimate an individual's strength. Average strength implies that the strength values are lower for a large percentage of the persons (50% for normal distribution). Regression functions also represent a form of average value, for example, for a given age. Taking the mean strength minus the single standard deviation (*s.d.*) is one possible compromise solution in the determination of the upper limit values for lumbar load. A limit based on 'mean minus *s.d.*' results in a value of 5.5 kN for 25-year-old men, 4.5 kN for 35-year-old, 3.6 for 45-year-old, and 2.6 kN for 55-year-old men. Corresponding values for women are 4.1, 3.5, 2.9, and 2.3 kN. The 'mean values' were taken from the regression lines in figure 4 and the values for the *s.d.* are valid for the male and female samples in table 1.

4.2. Criticism of the method

4.2.1. Assumptions and limitations of the biomechanical model

Owing to the structural diversity of the human skeletal and muscular systems and the complexity of movements during manual materials handling, not all factors can be considered in a single biomechanical model. Neither can a single paper consider, or even quantify, all of the variables which might influence load on the lumbar spine. Description of the fundamental relationships necessitates assumptions about, and simplification of, the real situation. Any interpretation of model-dependent calculations should not ignore the effect of these demands on the results. For example, the biomechanical model was validated by comparing disc compression predictions with disc pressure measurements from Andersson et al. (1977). Although the correspondence was fairly good (Jäger and Luttmann, 1989), Andersson et al. determined the intradiscal pressure during symmetrical load-holding tasks, whereas this paper provides compressive force values for asymmetrical load manipulation. Consequently, the model is only partially validated, i.e., for static or slow symmetrical movements. It is not validated for shear or highly dynamic activities.

Many assumptions in the biomechanical model concern the simulation of the skeletal structure. For example, punctiform joints were assumed although, in reality, a translation is usually superimposed upon the rotation of a body segment. This applies particularly to the joints in the lumbar spine. The position of their pivots relative to the disc shifts laterally, ventrally or dorsally according to the flexion (White and Panjabi, 1978). On account of the assumption of punctiform joints in the spine, the model fails to consider two further phenomena, motion coupling (Schultz et al., 1979) and the effect of the resistance of spine passive tissues. At large angles of trunk flexions, such as in bricklaying in low grasp positions, spine passive tissues can resist considerable external loads (Miller et al., 1986). Inclusion of the restorative moments of the discs, ligaments, and the muscles in the model would result in a reduction in the moments of force produced by the muscle equivalents and, consequently, in the calculated compressive and shear forces at the discs. In reality, spine passive tissues can be expected to reduce muscle loading.

In addition, different body weights, dimensions or proportions are not considered in the analysis of different grasp heights. The movement and the resultant time curves for lumbar load, which form the basis for the calculations, depend in particular on the postures at the beginning and end of the movement. These are determined for a person of average height. The corresponding postures for particularly tall or short people should be measured where required and further model calculations carried out for the determination of lumbar load.

The level of the calculated forces which are transmitted by the spine essentially depends on the moment arms of the muscle forces. The force-producing moment arms of the muscle equivalents are assumed constant in the model although the moment arm of the muscle force can vary as a result of the innervation of different muscle parts. This applies in particular to the fan-like, oblique abdominal musculature.

A dorsal moment arm of 5 cm was assumed in this model for the back muscles in accordance with the assumptions of Morris et al. (1961), Chaffin (1969) or Schultz and Andersson (1981). Larger dorsal arms, between 5.4 and 6.0 cm, were recently determined using computer tomography measurements (Kumar, 1988; Mc Gill et al., 1988; Tracy et al., 1989; Chaffin et al., 1990). Even greater moment arms, up to 6.8 cm, are provided by Reid and Costigan (1985) and Nemeth and Ohlsen (1986). The longest moment arm of 7.4 cm (Bean et al., 1988) contrasts sharply with the results of the anatomical measurements provided by Langenberg (1970), who determined the center of gravity of the erector spinae muscle group at approx. 4 cm dorsally from the disc center. The assumption of a 7.4 cm dorsal moment arm for the muscle force of the erector spinae would reduce, for example, the maximum in the dynamic 5-kg curve in figure 3 diagram IIb by approx. 30 per cent as compared with a 5 cm moment arm. This variance is within the same range as McGill and Norman (1987b) or Chaffin et al. (1990) provided for simplifying assumptions about muscle moment arms.

The description of the effects of the muscle system in the lumbar trunk region is realised in the biomechanical model in this paper using relatively few muscle forces (appendix B). For example, only the medial muscle parts of the fan-like, external and internal oblique abdominal muscles are considered, the parts responsible for the oblique binding of the abdominal wall (Lindh, 1980). The erector spinae muscle group was only described by one force-producing muscle equivalent for each side of the spine. This represents a considerable simplification of the actual muscle structure. Macintosh and Bogduk (1987) subdivide the erector spinae muscle in the lumbar region into four functional parts on the basis of the different attachments and the direction of the fibers. Similar to the model provided by Bloswick and Chaffin (1990), both of the cords of the rectus abdominal muscle were reduced to one medial force. Finer differentiation of the muscular and ligamentous systems was achieved in the models conceived by Schultz et al. (1983) and McGill and Norman (1986) with more than 20 muscle equivalents. Such detailed descriptions of the muscular structure require further assumptions or considerable extension of the methodology for activity analysis. For example, Schultz et al. apply the criteria that the compression of the L3 motion segment is minimized, that the maximum contraction intensity is limited to 100 N/cm^2 in any muscle, or that the largest muscle contraction intensity is minimized. In the model by McGill and Norman (1986), film analysis of the movement, electromyographical measurements, and biomechanical model calculations were employed in parallel.

After determination or measurement of the initial and final postures, the model in this paper first carries out a movement simulation and then the calculation of the lumbar load. The movement simulation assumes sinusoidal profiles for the angular acceleration during the flexion of all segments (section 2.2.2) although Slote and Stone (1963) have only shown this behaviour for 'discrete voluntary ballistic forearm flexion in the sagittal plane'. In this paper it is additionally assumed that the rotations of all segments begin and end simultaneously. Accordingly, interindividual and intra-individual variations in segment acceleration are not considered although different lifting techniques can greatly influence the spinal moments and forces (e.g., Leskinen et al., 1983; Hart et al., 1987; Bush-Joseph et al., 1988; Jäger and Luttmann, 1989). The calculated dynamic curves for lumbar load therefore represent approximations since, in contrast to measurements of dynamic kinematics or of a 'direct indicator of lumbar load' such as intradiscal pressure (Nachemson and Morris, 1964), they are based on a larger number of assumptions. In spite of this, the curves for simulated movements and the resultant lumbar load curves demonstrate the importance of considering inertial factors in the analysis of materials handling tasks, such as bricklaying, in addition to the static postural factors.

4.2.2. Analysis and application of the compressive strength data

The lumbar load during bricklaying was assessed in section 4.1.2.2 using the static compressive strength data provided in the literature. The experiments on dynamic strength could not be considered because they varied so widely both in terms of methodology and aims. However, once a sufficient number of samples is available it will probably be possible to draw conclusions for everyday slow dynamic activities from the investigations on fatigue fracture (e.g., Carter and Caler, 1983; Liu et al., 1983; Brinckmann et al., 1988; Porter et al., 1989).

The lumbar compressive strength values were taken from the literature without any reference to the donors' occupations. A higher bone mineral content (BMC) is revealed for physically active persons than for inactive persons (Dahlén and Olsson, 1974). BMC correlates with the ultimate compressive strength of lumbar vertebrae (Hansson et al., 1980). For example, the results of the measurements on power lifters (Granhed et al., 1987) demonstrate an increase in BMC related to the annually lifted load. It may therefore be the case that trained bricklayers have a higher spinal strength than the sample analysed in section 4.1.2.2. It is also conceivable that longterm employment in physically active professions may counteract the strength decrease with age indicated in figure 4.

The spine comprises osseous vertebrae and more or less visco-elastic intervertebral discs. Neither the vertebrae, consisting of spongiosa and cortex, nor the discs, with their fibrous annulus and gelatinous nucleus, are homogeneous. Consequently, the mechanical strength is dependent both on the direction of the force and on the rate of the load. Strength values should therefore be differentiated in terms of compression, tension, shear, torsion, flexion, and extension. To date, a sufficient number of investigations has only been conducted into static lumbar compressive strength. Examinations of the other types of load mainly referred to the mechanical behavior rather than the maximum strength (Schultz et al., 1973 and 1979; Berkson et al., 1979). Whole body movements, such as in bricklaying, comprise compression, lateral and sagittal shear, flexional, and torsional components etc. (figure 2). Further research is required in order to derive corresponding values for such superpositions.

Characteristics other than the 5 analysed factors, for example, body weight, professional activity, diet, and illnesses, also influence the compressive strength. Factors such as differences in the test equipment and in the preparation of the specimens used in the tests (e.g., surrounding the specimens in metal, plaster, etc.; the use of fresh, embalmed, frozen, or defrosted specimens, etc.) should also be considered. These influences on the compressive strength are currently being investigated.

It is assumed in most statistical tests that the elements in a sample must be independent of one another (Sachs, 1984). However, this statistical requirement cannot be fully met when several lumbar segments are taken from a single donor and all segments are considered in the statistical tests ('segment-related analyses'). Since, though, in lumbar strength investigations only 1, 2 or 3 segments are taken from a single person (average in this paper's sample 1.9), the number of specimens from one donor remains small when compared with the total number in the sample. Significant differences in the strength data analyses cannot be shown if only one randomly chosen value from each person is used in 'person-related' analyses. The results of the regression analyses for person and segment-related samples are very similar and cannot be differentiated statistically.

5. Conclusions

Lumbar load during one-handed bricklaying was analysed using biomechanical model calculations. The movements of body and brick were not measured using a motion analysis system but, instead, computer-simulated. To illustrate the asymmetry of such tasks, lumbar load was characterized here by the lumbosacral moment of force with its components in the sagittal, frontal, and transversal plane as well as by lumbosacral force with its compression, and lateral and sagittal shear components. The lumbar load during bricklaying increases (a) with decreasing grasp height (90, 50, 10 cm) as a result of more inclined postures, (b) with decreasing execution time due to the stronger influence of inertia and (c) with the increasing weight of the brick. Some moment and force values resulting from dynamic calculations deviate considerably from static calculations. According to the classification for the lumbosacral moment of force (Tichauer, 1978), bricklaying involving a 50 cm grasp height requires 'selection of labor, careful training and rest pauses'. Lower grasp heights or bricks of 10 kg should not occur throughout 'the entire working day'.

Cadaveric compressive strength data provided in the literature were collated. Lumbar compressive forces during bricklaying and lumbar compressive strength are of the same order of magnitude. The calculated force values vary between 0.7 and 6 kN, the strength ranges between 0.8 and 13 kN. In order to explain the variance, the factors with the greatest influence on strength, i.e., age ($r^2 = 0.27$) and gender ($r^2 = 0.14$), were quantified. Assuming linear regression models, strength (in kN) is $10.53 - 0.97$/decade for males ($r^2 = 0.39$) and $7.03 - 0.59$/decade for females ($r^2 = 0.35$). A further regression model for the prediction of lumbar compressive strength, in which three additional influencing factors were considered, was developed in order to explain the remaining variance ($r^2 = 0.62$).

Average strength values (males 5.8 kN, females 4.0 kN) should not be used to determine limits for the highest permissible load on the spine since this might overestimate an individual's load-bearing capacity. One suggestion for guaranteeing a better safety margin is to reduce the mean or regression model value by the standard deviation of the respective sample (males 2.6 kN, females 1.5 kN). This would result in a lumbar load limit of 5.5 kN for 25-year-old men and 2.6 kN for 55-year-old men. Corresponding values for women are 4.1 and 2.3 kN. However, even then, health risks cannot be fully excluded from bricklaying, nor is everyone capable of performing all types of bricklaying tasks. Age and gender should be considered when comparing the above lumbar load limits with the predicted lumbar compression during bricklaying. If the brick-supply stack is 90 cm high, the lumbar load limits will not be exceeded for any person in these age groups. By contrast, all limits would be exceeded for a 10-kg 1-s brick transfer from a grasp height of 10 cm. In conclusion, to ensure that the predicted lumbar load during bricklaying remains below the limits, the brick-supply stack should be above 50 cm.

Appendix A

The skeletal structure of the biomechanical model 'The Dortmunder' is presented in figure A.1. It comprises 30 rigid segments and 27 punctiform joints. The segments are assumed to be cylindrical. Figure A.1 contains numerical values for the following descriptive variables: segment length and radius, the location of the segment's center of gravity and the segment weights. The somatometrical data were assumed to be identical for body segments on both sides of the body. However, for the sake of clarity, they were not indicated in the stick model in figure A.1 for both sides. In addition, figure A.1 contains the values for the hip and shoulder width, the trunk length, as well as the length and weight of the

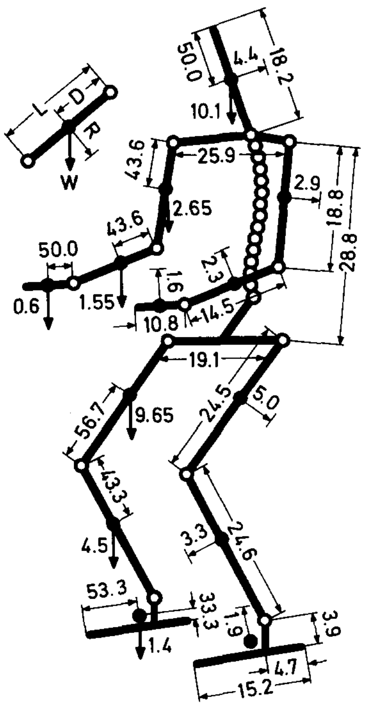

Fig. A.1. Skeletal structure of the biomechanical model. Key: L segment length as percentage of the total body height; R segment radius as percentage of the total body height; D distance of a segment's center of gravity as percentage of the total segment length; W segment weight as percentage of the total body weight.

Table A.1
Coordinates and sagittal inclination of the discs and disc mobility (BH: body height).

Joint	Coordinates in % BH			Inclination in degree	Mobility in degree			
	x	y	z		Ventral	Dorsal	Lateral	Torsional
left heel	0	0	0	–	–	–	–	–
left hip	4.70	0	53.00	–	–	–	–	–
L5-S1	1.46	−9.55	58.62	−30.48	2.2	16.4	1.5	2.5
L4-L5	2.02	−9.55	60.82	−6.70	3.7	10.2	3.0	1.0
L3-L4	2.15	−9.55	63.16	1.52	3.0	9.0	4.0	1.0
L2-L3	1.99	−9.55	65.41	9.18	3.0	8.0	3.0	1.0
L1-L2	1.53	−9.55	67.51	16.50	2.0	6.6	3.0	1.0
T12-L1	0.85	−9.55	69.47	20.55	1.3	3.0	4.0	1.0
T11-T12	0.16	−9.55	71.22	19.14	1.8	1.8	4.5	1.0
T10-T11	−0.31	−9.55	72.77	17.41	1.8	1.0	3.5	1.0
T9-T10	−0.78	−9.55	74.20	17.97	2.4	1.0	3.0	2.0
T8-T9	−1.23	−9.55	75.57	17.49	2.4	0.8	3.0	3.5
T7-T8	−1.49	−9.55	76.85	9.22	2.8	0.8	3.0	4.0
T6-T7	−1.66	−9.55	78.11	5.06	2.8	0.6	3.0	4.0
T5-T6	−1.73	−9.55	79.37	−3.50	2.8	0.3	3.0	4.0
T4-T5	−1.52	−9.55	80.69	−10.97	2.8	0.5	3.0	4.0
T3-T4	−1.28	−9.55	81.80	−13.42	2.4	0.1	3.0	4.0
Top	4.70	−9.55	100.00	–	–	–	–	–

head and neck segment. The division of the trunk into 15 segments will be explained in the paragraph after next.

Most of the somatometrical data were taken from the literature. The length of the body segments were mainly taken from Drillis and Contini (1966). The hand length refers to an outstretched hand. Since the hand is bent and, consequently, shortened when grasping objects, the distance between the wrist and the grasp point was assumed to be half of the total hand length. The location of the segments' centers of gravity relative to the adjacent joints and the weight of the segments are mainly adopted from Dempster (1955). The values for the neck-head segment were calculated from information provided by Plagenhoef (1971). Both the location of the center of gravity and the sagittal position of the ankle joint relative to the heel were calculated from details provided by Fritz (1979) and Lanz and Wachsmuth (1972). All of the values for the radius of the segments were determined in laboratory measurements for 5 males. In this, the circumference of the particular segment was measured at three points using a measuring tape, and averaged. These values were then related to the particular body height and converted into radius values.

The trunk comprises 15 segments including the segment of the pelvis between the hip joints and the lumbosacral joint, five lumbar and 9 thoracic segments which result from parallel cuts of the torso at the level of the discs L5-S1, L4-L5,...,T3-T4. Assuming an upright posture and hanging arms, the disc T3-T4 is at approximately the same height as the shoulder joints. Table A.1 contains the coordinates, the angles of sagittal inclination for 'normal curvature', and the spinal mobility angles of the discs which are considered in the biomechanical model. The coordinates are indicated as percentages of the total body height (BH) relative to a right-handed orthogonal inertial system whose origin is located in the left heel. In this system the x-axis points forwards, the y-axis to the left, and the z-axis upwards. The x-coordinate of the left hip joint (4.70% BH) and the y-coordinates of the discs (−9.55% BH) in table A.1 result from the corresponding values in figure A.1. The vertical position of the lumbosacral joint, the lowest disc, was calculated in accordance with Lanier (1939): 19.5% of the vertical distance between the hip and shoulder axes. The dorsal position of L5-S1 was determined by evaluating lateral X-ray photographs and diagrams from Sobotta (1967): in the sagittal plane, the link line between the middle of L5-S1 and the middle of the hip joint axis is at an angle of 30° from the vertical.

Starting from the L5-S1 coordinate, the coordinates for the centers of the next highest discs were determined. Orne and Liu (1971) provide coordinates for the centers of the lower endplates of the vertebrae, as well as the height of the vertebrae and discs. These data were used as a basis for the calculation of the centers in the discs (see x and z coordinates in table A.1). Parallel endplates were assumed for each vertebra. The 'sagittal curve' of the spine and tangents at the level of the centers of the discs were derived from the calculated x and z coordinates. The disc planes were set perpendicular to the tangents. The resultant angles of inclination are contained in the 5th column of table A.1. They are positive for dorsal inclination and negative for ventral inclination. The inclination difference between the L5-S1 and L3-L4 is 32° (Farfan, 1973).

The values for the sagittal mobility of the lumbar discs L5-S1...L1-L2 were adopted from Bakke (1931) without modification. The thoracic mobility values (T12-L1...T3-T4) were reduced by 1/3 since Bakke conducted measurements on isolated spinal sections. Lower flexibility was assumed due to the relatively rigid thorax. The sum of the ventral mobility angles, i.e. a fully flexed spine (100%), corresponds to a 36° inclination of the T3-T4 midplane compared with an unflexed trunk. The torsional and the lateral mobility of the spine were adopted unmodified from White and Panjabi (1978).

The locations of the centers of gravity within the trunk segments are listed in table A.2. The aforementioned x-y-z system and an upright posture were assumed. The coordinates in the sagittal plane (x) were all set to 4.7% BH, i.e., viewed laterally, they lie on the link line between the hip and shoulder joints. In the frontal plane all centers of gravity were assumed to be medial ($y = -9.55\%$ BH). The vertical locations were determined such that each center of gravity is at 45% of the total segment height as measured from the lower limit. A total trunk weight of 49.2% (Fritz, 1979) and a pelvic weight of 11.6% body weight (BW) were assumed. The trunk weight above L5-S1 (37.6% BW) was divided in proportion to the vertical segment height. The radius values for the trunk segments were determined by measuring, in the described laboratory investigations, the circumference at 15 suitable points along the trunk.

Appendix B

The lumbar musculature in the biomechanical model 'The Dortmunder' is illustrated for a posterior, lateral, and cranial view (figure B.1). The erector spinae muscle group is simulated by the

Table A.2
Center of gravity coordinates, weight, and radius of the segments of the trunk (BH: body height; BW: body weight).

Segment	Coordinates in % BH			Weight in % BW	Radius in % BH
	x	y	z		
Hips...L5-S1	4.70	−9.55	55.53	11.59	9.3
L5-S1...L4-L5	4.70	−9.55	59.61	3.56	9.0
L4-L5...L3-L4	4.70	−9.55	61.87	3.80	8.7
L3-L4...L2-L3	4.70	−9.55	64.17	3.66	8.3
L2-L3...L1-L2	4.70	−9.55	66.35	3.40	8.3
Ll-L2...T12-L1	4.70	−9.55	68.39	3.18	8.2
T12-L1...T11-T12	4.70	−9.55	70.26	2.85	8.2
T11-T12...T10-T11	4.70	−9.55	71.92	2.51	8.0
T10-T11...T9-T10	4.70	−9.55	73.41	2.32	8.0
T9-T10...T8-T9	4.70	−9.55	74.82	2.22	8.2
T8-T9...T7-T8	4.70	−9.55	76.15	2.08	8.4
T7-T8...T6-T7	4.70	−9.55	77.42	2.05	8.4
T6-T7...T5-T6	4.70	−9.55	78.68	2.05	8.5
T5-T6...T4-T5	4.70	−9.55	79.96	2.14	8.6
T4-T5...T3-T4	4.70	−9.55	81.19	1.79	8.7

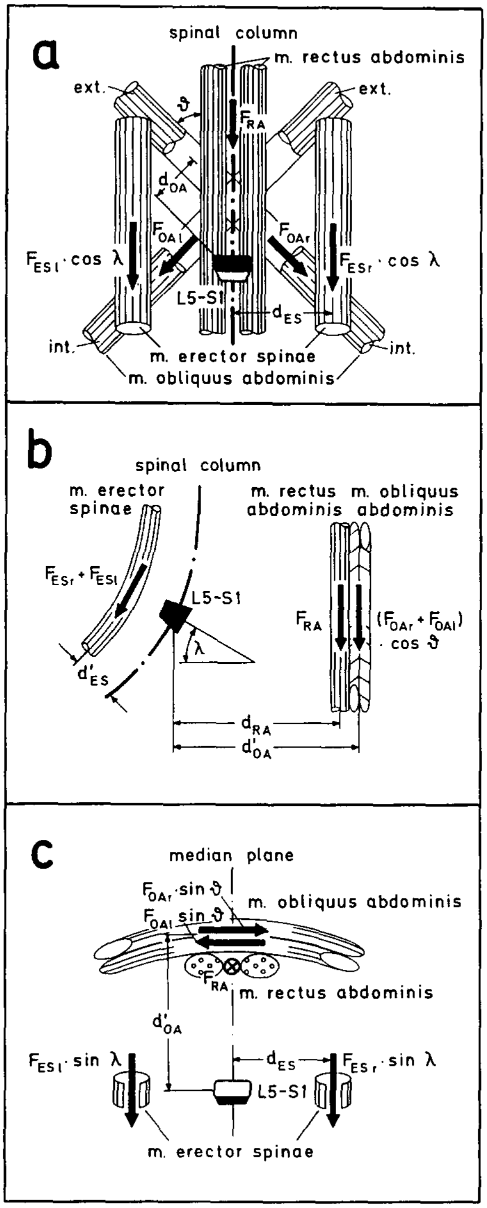

Fig. B.1. Muscular structure of the lumbar region of the biomechanical model; see text for the abbreviations and values for the variables. (a) posterior view; (b) lateral view; (c) cranial view.

bilateral resultant forces F_{ES}. The subscripts 'r' and 'l' refer to the right and left sides of the body respectively. The dorsal moment arm in the sagittal plane d'_{ES} (figure B.1b) was set to 5 cm (adapted from Morris et al., 1961; Chaffin, 1969; Schultz and Andersson, 1981). The lateral moment arm $d_{ES} = 5$ cm (figure B.1a, c) is based both on measurements for the determination of the center of gravity within the cross-sections of the erector spinae muscles (Langenberg, 1970) and electromyographical measurements (Jäger et al., 1983). In the latter, bipolar surface EMGs were derived in the lower back region. A mapping technique with 60 electrode configurations was used to determine the location of the maximum myoelectrical activity from the midline. The rectus abdominal muscle runs immediately below the aponeurosis of the oblique abdominal muscles. The sagittal moment arm d_{RA} (figure B.1b) of the rectus abdominis was estimated from anatomy textbooks (Tittel, 1978; Benninghoff and Goerttler, 1980) as 10% of the total body height. The two muscle cords of the rectus abdominis run closely parallel to each other. Their effect is represented by a resultant force F_{RA} operating along the midline of the body since, essentially, the rectus abdominis operates antagonistically to the back musculature and accordingly exerts only small lateral bending moments (Tittel, 1978).

The medial parts of the oblique external and oblique internal abdominal muscles are included in the model for the description of the oblique abdominal musculature. The external muscle from one side of the body combines with the internal muscle from the opposite side, and vice versa, to form, via a tendinous network, the diagonal binding of the abdominal wall (figure B.1a). Two resultant forces for the oblique abdominal musculature are assumed (F_{OAr}, F_{OAl}). The muscle moment arm d_{OA} in the posterior view (figure B.1a) was set to 1% of the total body height. The moment arm d'_{OA} in the lateral view (figure B.1b) was assumed to be 10% of BH. The direction of the force F_{ES} in the lateral view (figure B.1b) was assumed vertical to the median plane of the disc L5-S1 ($\lambda = 30.5°$, see appendix A). The angle ϑ of the diagonal abdominal muscles was set at 45° (figure B.1a–c).

Appendix C

According to Chaffin (1969), the intra-abdominal pressure is dependent on posture (angle α between the trunk and thighs) and on the load, expressed as the moment of force at the hip joints in the sagittal plane (M_H). The pressure (p_{Abd}) is converted to a force (F_{Abd}) via the effective area of the diaphragm and to an abdominal moment of force (M_{Abd}) via the moment arm at the lumbosacral disc.

$$p_{Abd} = 4.74 \cdot 10^{-3} \cdot (119.6 - \alpha/\text{degree}) \cdot (M_H/Nm)^{1.8} \text{ Pascal} + p_0$$

$$F_{Abd} = p_{Abd} \cdot 465 \text{ cm}^2 \qquad M_{Abd} = F_{Abd} \cdot 9.1 \text{ cm}$$

In the measurements by Morris et al. (1961), upon which Chaffin's model is based, the 'base pressure' p_0 in upright postures ($\alpha = 180°$) amounts to approx. 10 mm Hg (1.3 kPa). This phenomenon is not considered by Chaffin. In this paper p_0 was set at approx. 2.0 kPa. Consequently, the moment of force due to the intra-abdominal pressure is equal to the opposing moment of force due to the weight of the body segments above L5-S1 in an upright posture. The 'base pressure' is always selected so that an equilibrium in the moments for upright posture is maintained. This assumption is confirmed by the fact that the back musculature is inactive during an upright, relaxed, unloaded posture (Floyd and Silver, 1955).

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