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Mechanical Recruitment of Low-Back Muscles

Theoretical Predictions and Experimental Validation

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A biomechanical model for studying lumbar muscle load sharing for a class of physical tasks that involve gravitational loading (holding weights) of the upper body in an erect posture is presented. The model assumes that the lumbar muscles balance the externally applied flexion and lateral bending moments. The concept of a 'loading plane' whose axes are the two bending moments is introduced. Any point in the plane can be viewed as a 'loading-point' describing a combination of bending moments that are applied to the body. The study of lumbar-muscle load sharing revealed loading conditions that required activation or deactivation of a particular muscle. The loading plane thus could be divided into regions of activity and inactivity for each muscle, separated by a 'switching curve.' The concept of 'switching curves' proved very useful for examining previously described physiologic assumptions on the loading conditions of particular muscle groups, and for grouping the 22 muscles described in the model into ten functional units. Electromyographic validation studies were conducted and showed a high degree of correlation between the model predictions and actual measurements for the contralateral (with respect to the load) muscles and to a lesser degree of correlation for the ipsilateral muscles. [Key words: biomechanical model, lumbar muscle groups, EMG validation]

THE ABILITY of the human body to maintain normal posture and to balance external loads applied to the trunk and the upper limbs relies on the activity of a large number of muscles. The involved muscles balance the spinal column and maintain the overall mechanical integrity of the trunk. The movement of the upper trunk with respect to the lower trunk is comprised of three rotational movements: flexion, lateral bending, and rotation. The mechanical moments that will be responsible for such movements will correspondingly be: flexion moments, lateral bending moments, and torsional moments. Such moments will actively be produced by the muscles of the trunk; for example, muscles that cross the lumbar region.

The activation of the lumbar muscles is required not only to produce the aforementioned movements, but to sustain a given posture in the presence of external loading. Consider the loading condition represented by Figure 1. In this schematic representation, the subject holds a weight in one hand with the arm fully extended. The weight and the extended arm produce an external moment that is inclined to flex the upper trunk. Because the load is applied to the right of the center of the body, it also tries to rotate the upper trunk laterally to the right side. The two resulting bending moments (flexion and lateral bending) can be calculated by multiplying the loads (ie, the external weight and the weights of the arm and the forearm) by their respective moment arms. In order to prevent such a movement from taking place, opposing moments produced by lower back muscles must be generated. Hence, the lumbar muscles need to be activated not only to generate motion, but also to balance external moments while maintaining a given posture. Since the external moments are independent from one another, a minimum of three muscles will be required to generate balancing moments around the spine. However, a close examination of the lumbar region reveals that there are many more muscles available across any cross section in that region. For example, 22 muscles can act across the L3 level.[14] As there are many more muscles than are minimally needed to balance the external moments, a question arises as to how those muscles will "cooperate" in response to external loads. This question may be rephrased in mechanical terms: what is the distribution of forces in the individual muscles that is generated in response to given combinations of external moments? This paper proposes for the first time a mechanical recruitment scheme that suggests the order by which different muscles are recruited in response to external moments. It predicts the existence of "switching curves" for each muscle, ie, load combinations that will and will not activate a given muscle. The paper further describes the experimental results that were used to test the above hypothesis.

The experimental results show that the switching curves were able to predict the muscular response to external loads with a considerable degree of accuracy for muscles that are contralateral to the load, and with a lesser degree of accuracy for the muscles that are ipsilateral to the load.

METHODS

The process of calculating the muscle forces in the lower back in response to external loading can be described in the following manner: Consider an imaginary transverse cross section in the lumbar region (plane I in Figure 1). Such a cross section will transect horizontally through all the structures that cross that level, "exposing" the internal elements (muscles, ligaments, and bones) that produce force. The force in each individual element will be an unknown that will have to be resolved. Schultz et al[14] proposed a physiologic decoupling for external loads that involved the compression of the spinal column: the muscle forces balanced the external mo-

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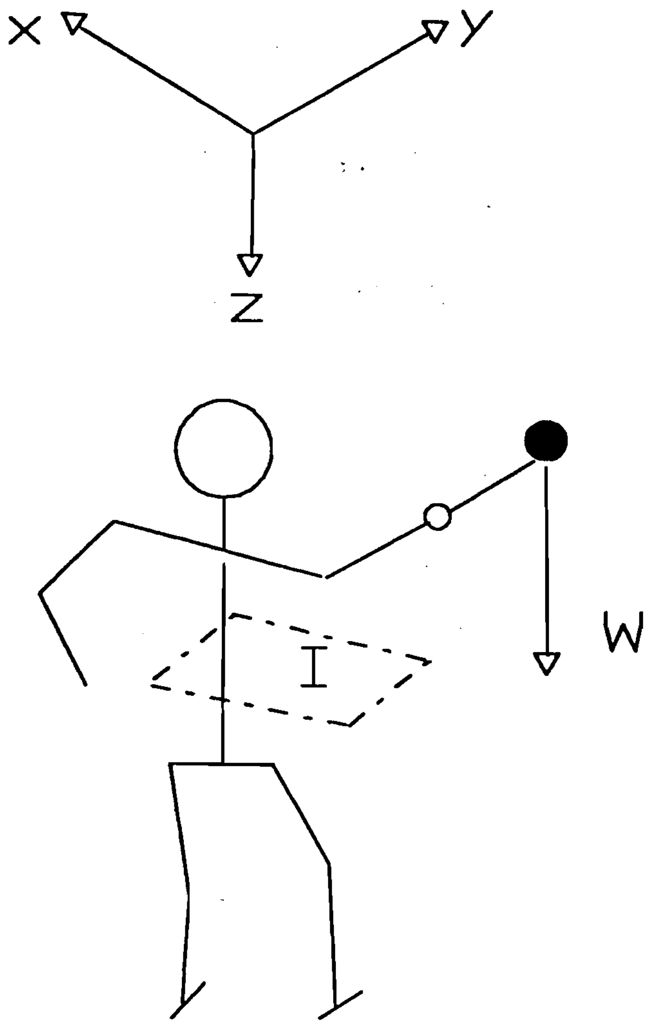

Fig 1. A 'free body' analysis at a lumbar transverse cross section.

ments, while the spinal column balanced the resulting compression and shear forces. As all the loads were transferred from the upper half of the body to the lower half of the body could be combined into six independent variables—three forces and three moments—six independent equations of static equilibrium could be written. Invoking the physiologic decoupling assumption, three equations of static equilibrium (the three moment equations) for the 22 unknown muscle forces could be written. Such a system is undetermined (many solutions that satisfy the above equations exist). Hence, we followed the same general approach suggested by Schultz et al,[14] which employs an optimization approach that converges on a single solution for the set of equations that describe the system.

A cost function that describes the spinal compression that arises exclusively from the muscle tensile forces was defined, and the following optimization problem was solved: find the individual muscle forces that will be required to satisfy the constraint equations (ie, the moment equilibrium equation) while minimizing the muscular spinal compression cost function, and at the same time not exceeding maximum predefined muscular stresses. The solution process involved specifying a low value of maximum stress, which was constrained to be identical in all muscles, and then gradually increasing that value until the above conditions were met. Linear programming algorithms were used for the solution and the anatomic data for the muscle moment arms and cross-sectional areas were based on cadaver studies[4] and ergonomics manuals.[3]

The class of isometric loading exercises described in this paper include gravitational loading with no external torsion performed in an erect posture. Physical exercises that involve holding weights in different positions of the upper limbs while standing erect may be included in this category. Any such exercise that involves a combination of weights and the arbitrary position of the upper limbs can be characterized by a single combination of the two bending moments. Since the bending moments uniquely define the distribution of the muscle forces in all the muscles, one can describe a two-dimensional plane, called the "loading plane," whose axes are the two bending moments. Figure 2 shows how three different tasks map onto the loading plane; hence, a physical activity that involves a slow movement of the upper limbs from one position to another can be described as a curve on the loading plane.

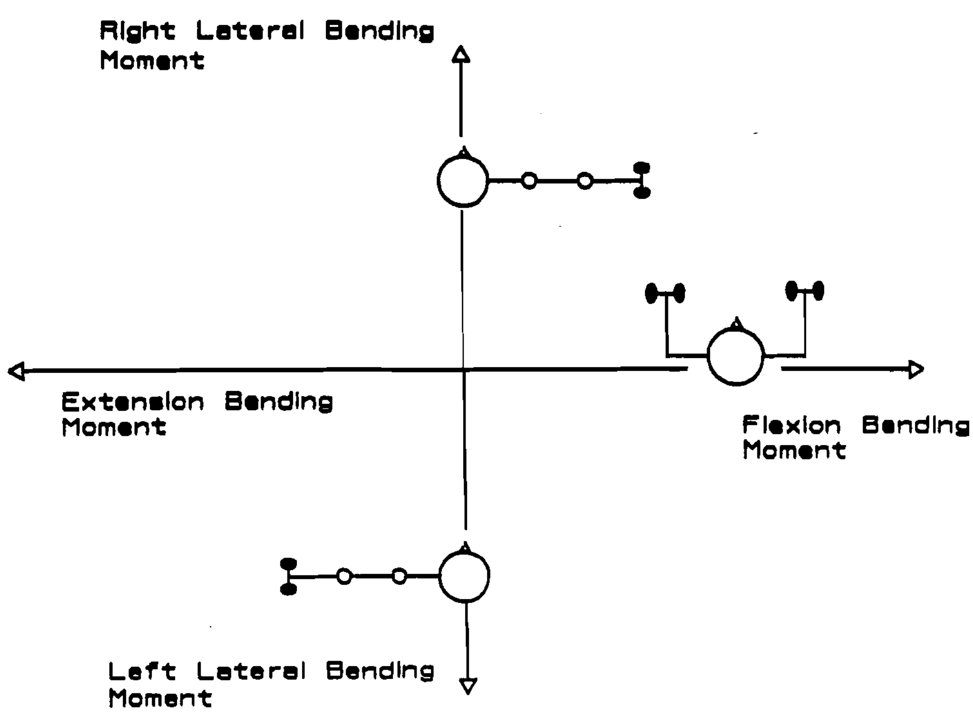

Fig 2. The 'loading plane' expressed by the two bending moments; characteristic tasks are illustrated on the axes.

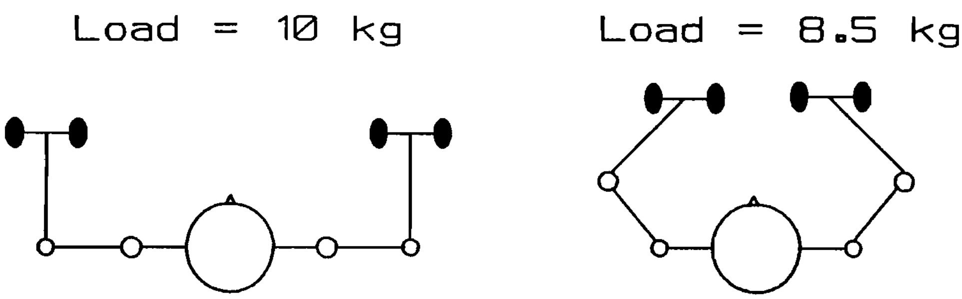

Fig 3. An example of two different physical tasks that give rise to the same bending moments at the lumbar region.

The top half of the loading plane corresponds to right lateral bending moments, and the right half of the loading plane corresponds to flexion bending moments. Note that there is a unique mapping of physical tasks to the loading plane, ie, a given combination of weights held in a particular position map into a single point on the loading plane; however, the inverse is not true; a single point on the loading plane can correspond to many physical tasks. Hence, the study of patterns of muscle activation in response to external loads is greatly facilitated by studying a portion of the loading plane and the corresponding distribution of muscle forces. This was achieved by specifying the values of the bending moments, solving the optimization problem described above, and storing the resulting distribution of the muscle forces. The process was repeated for a new set of values of the bending moments until the range of loading moments of interest was covered.

It is instructive to review the advantages of using the loading plane for load specification compared with the use of the upper limbs' position and the value of weight when studying low-back muscle force distribution. Figure 3 describes two different physical tasks that involve different weights that will generate the exact same bending moment loading on the lower back, and consequently an identical distribution of low-back muscle forces. Hence, any attempt to study the underlying pattern of muscle activation must be based on the loading moments and not on the details of the physical tasks that gave rise to those moments.

The bending moments in the theoretical study ranged from 0 to 100 N-m for the flexion moment and from -100 N-m to +100 N-m for the lateral bending moment (ie, from a left lateral bending moment of 100 N-m to a right lateral bending moment of the same value), in steps of 10 N-m. By specifying 11 discrete values on the flexion moment axis, and 21 discrete values on the lateral bending axis, a grid of 231 points was generated. Each point represents a different loading condition and the 22 muscles forces that correspond to that loading were stored.

RESULTS

The examination of the resultant muscle force distribution revealed that all of the muscles had a range of loading conditions where no activity was predicted, while different levels of activity are predicted for the rest of the loading plane. A convenient way to display such information is described in Figure 4, which presents the results for the left longissimus muscle. The curve shown in the figure separates the loading plane into two zones: an inactive zone that is generally below and to the left of the curve, and a zone of activity that is generally to the right and above the curve. Such a curve thus can be viewed as a switching curve: the external loading that determines the bending moment combination will "switch" the muscle on or off, depending on the location of the loading point with respect to the switching curves.

From considerations of symmetry, one expects the switching curve of the right longissimus muscle to be the mirror image of the left one with respect to the flexion moment axis. This is indeed the case, as shown in Figure 5. Note that the relative location of the activity/inactivity zones with respect to the switching curve also is reversed.

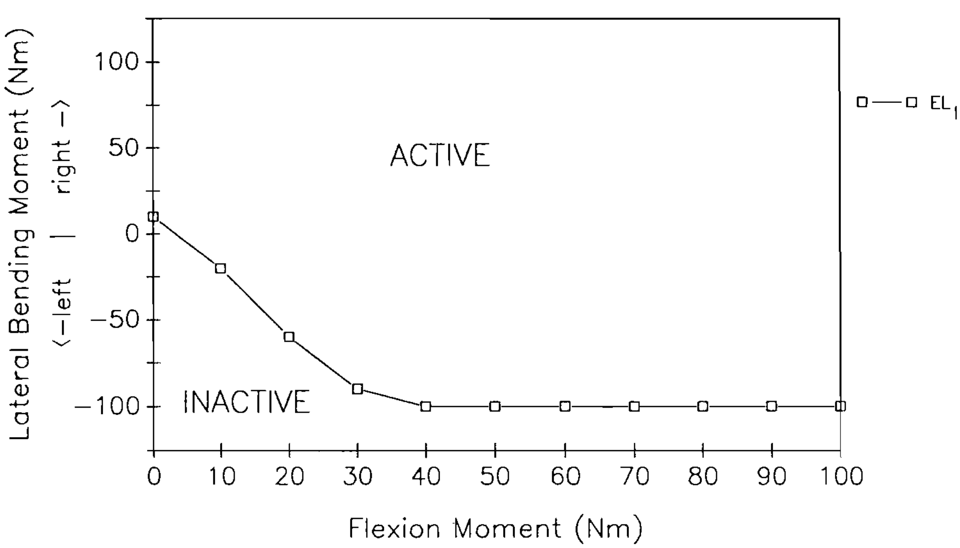

Fig 4. The switching curve of the left longissimus muscle.

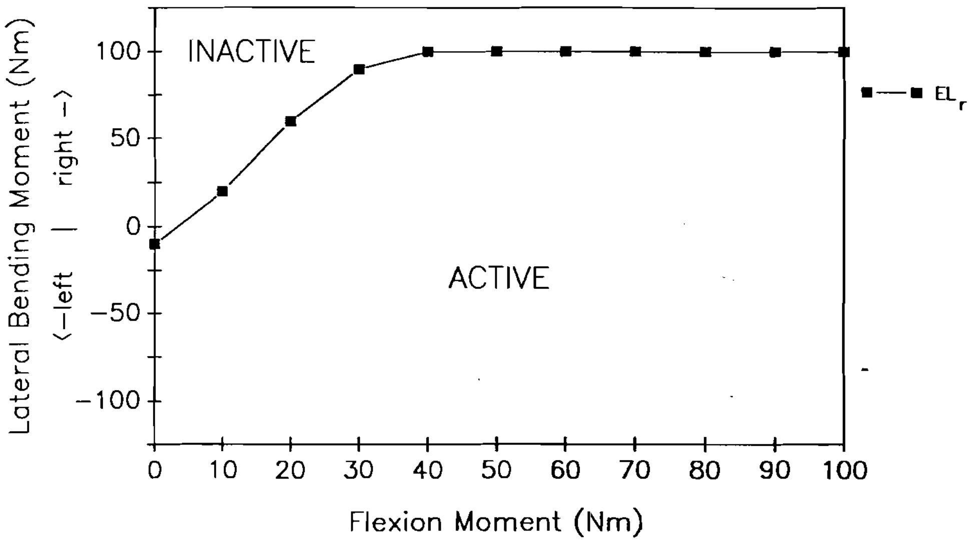

Fig 5. The switching curve of the right longissimus muscle.

Since the switching curves of all the muscles are displayed on the same loading plane, a number of curves may be overlaid on the same figure. Figure 6 displays the switching curves of five muscles on the left side of the body: the longissimus, the psoas, the lateral portion of the external oblique, the medial portion of the external oblique, and the rectus abdominus. Hence if one holds a weight in the left hand while the arm and forearm are extended laterally to the left, the resulting loading will have no flexion component and a significant left lateral bending. Such a loading condition will result in a predicted inactivity of all the muscles described in Figure 5. As the upper arm rotates forward, increasing the internal rotation angle, the flexion moment increases while the left lateral bending moment decreases. This will result in the activation of the left longissimus muscle first, followed by the activation of the psoas. With both hands holding the weight in front of the body, the loading consists of pure flexion moment, and among the five muscles only, the longissimus and the psoas are active. As the load is transferred to the right hand, which continues its rotation to the right, decreasing the internal rotation angle of the right upper arm, the flexion moment decreases and the right lateral bending moment increases.

Figure 6 predicts that the corresponding recruitment order will include the lateral portion of the external oblique, then the medial portion of the external oblique, and finally, the rectus abdominus. Hence, when the load is held laterally in the right hand, the model predicts that all five muscles on the left side will be active.

Electromyographic Validation

Valid tests of muscle force predictions generated by biomechanical models must involve physiologic measurements of muscular activity. Because it is practically impossible to insert a force transducer noninvasively into a muscle that is under study, researchers have resorted to measurements of the musclular electomyographic activity conducted by surface or wire electrodes.[7-11] The underlying assumption of all of these studies is the existence of a unique relationship between the electrical activity of the muscle and its mechanical force output. Such an approach would have been useful if such a relationship was unique and known for a given muscle,

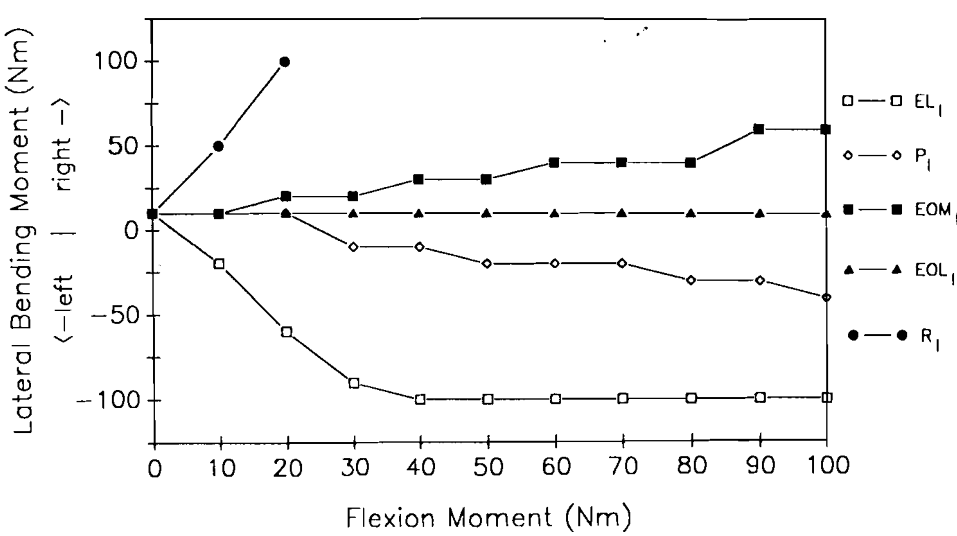

Fig 6. The switching curves of five muscles on the left side of the body: EL = longissimus; P = psoas; EOL = lateral portion of the external oblique; EOM = medial portion of the external oblique; R = rectus abdominus.

Table 1. Subject Population

Subject	Height (cm)	Weight (kg)	Age	Dominant hand	Arm length (cm)	Forearm length (cm)	Trunk width (cm)	Trunk depth (cm)
AR	193.04	68.2	23	R	38	31	30	16
NR	171.45	59.1	23	L	32	24	28	21
SG	167.64	71.8	25	R	31	25	34	22
DO	177.80	68.2	26	R	31	27	32	17
LB	181.00	76.1	20	R	31	30	29	17
ME	163.83	68.2	29	R	28	25	31	19
BK	179.07	75.0	21	R	34	29	32	17
TR	172.72	69.1	23	R	26	24	27	17

but as Basmajian and De Luca[2] point out, a wide range of linear and nonlinear force-EMG (electromyogram) relationships has been proposed in the literature.

The lack of a widely accepted function that relates the EMG measurement to the force output prompted our interest in finding a different approach for the validation of our theoretical predictions. Since the model predicted that the activation of any low-back muscle for this particular class of isometric exercises is based on the external loading, one can expect that the electrical activity will change markedly when the switching curve of that muscle is crossed. Hence, in order to test the validity of the switching-curve predictions, one can design an isometric physical exercise that will produce loading conditions on both sides of the switching curve of a particular muscle. The EMG activity level then should change, from the background level in the 'inactive region' to a markedly larger value in the active region. The advantage of this approach lies in its insensitivity to the actual relationship (linear or other) between the muscle force output and the EMG signal. The requirement to be able to separate between the 'activity' and 'inactivity' states lies in the determination of an EMG threshold level—EMG measurements that are smaller than the threshold are considered to express an 'inactive' state, whereas measurements that are larger than the threshold will determine an 'active' state.

Experimental Protocol

Eight male subjects ranging in age from 20 years to 29 years (23.75 ± 2.68) were tested. Their height ranged from 164 to 193 cm (175.8 ± 8.5) and their weight ranged from 59 to 76 kg (69.5 ± 4.9). All had no reported incidence of lower back pain. Other anatomic parameters (eg, forearm length, arm length, etc.) were measured and are summarized in Table 1.

The subjects were placed in a specialized testing apparatus designed to mechanically immobilize the subject at the L3 level. The details of this apparatus have been reported elsewhere.[12] Thus, this experimental setup closely simulates the anatomic configuration used in the biomechanical model. The EMG signal was detected with active surface electrodes that have been in regular use at the NeuroMuscular Research Center over the past 6 years. Their detection surfaces consist of two 1.0 cm long and 1.0 mm wide silver bars spaced 1.0 cm apart. The surface electrodes were placed on three bilateral muscle pairs: 1) rectus abdominus (RA), 2) medial oblique abdominals (OB) and 3) erector spinae (ES). The EMG signals were collected using the Muscle Fatigue Monitor (MFM®; a device developed at our Center), which provides the root-mean-square (RMS) value and median frequency of the EMG signal.[6,16] In addition, the raw EMG signals were recorded on FM tape. The subjects were asked to perform six tasks, and each task was repeated once; hence, a total of 12 experiments were performed on each subject. The six tasks were 0°, 45°, and 90° internal rotation of the right arm and the same tasks using the left arm. In all trials, the subject was required to maintain an erect posture and extend the upper limb horizontally. In each task, the subject was holding a 4.5-kg weight in the rotated arm. The EMG signal was collected for 10 seconds with a baseline activity measurement preceding each task (note that monitored muscles are postural muscles, thus they may display EMG activity even in the absence of external loading). The subjects were asked to relax during the unloaded/baseline measurements. The raw EMG signals were analyzed with the MFM to obtain the RMS values.

Figure 7 shows the EMG RMS traces of all the six monitored muscles, measured during the execution of tasks A, B, and C. The time the loads were applied and removed is indicated on the figures. Since our goal in this study was to identify the activity state of each one of the muscles, the only determination made in each experiment was whether a muscle was active or inactive. The activity state of each muscle in each experiment is indicated in the figure and was based on a step change in the muscle EMG trace. As may be seen in the figure, such a determination can be made by visually comparing the steady-state value of the EMG trace in the preloading state to the trace during the load application. The transients of the EMG traces that occurred as the load was applied always died out in less than 1 second; hence, the EMG level that occurred during the middle period of the load application, ie, the third to fifth second of load application, were used to make the determination. The EMG level after the removal of the load was used as a further indication that the muscle returned to its preloading state. It is instructive to note that the change in EMG level was so dramatic (always larger than twice the reference level), that even an untrained individual had no difficulty in making that determination. For the purposes of the statistical analysis presented later, a numerical score was given to the activity level: an active state was given a score of 1 and an inactive state was given a score of 0.

Data Processing and Statistical Analysis

The study described in this paper is aimed at testing the hypothesis that the activity state of the lumbar muscles under gravitational loading in erect posture, is determined by the combination of the external bending moments. Hence, a theoretical prediction on the activity state of the low-back muscles has to be made and compared with the experimentally measured EMG level.

Figure 8 shows the switching curves of the six muscles that were examined in this study and the loading combinations representing the different tasks (A-F). The loading combinations were calculated for an "average" person using anthropometric data from

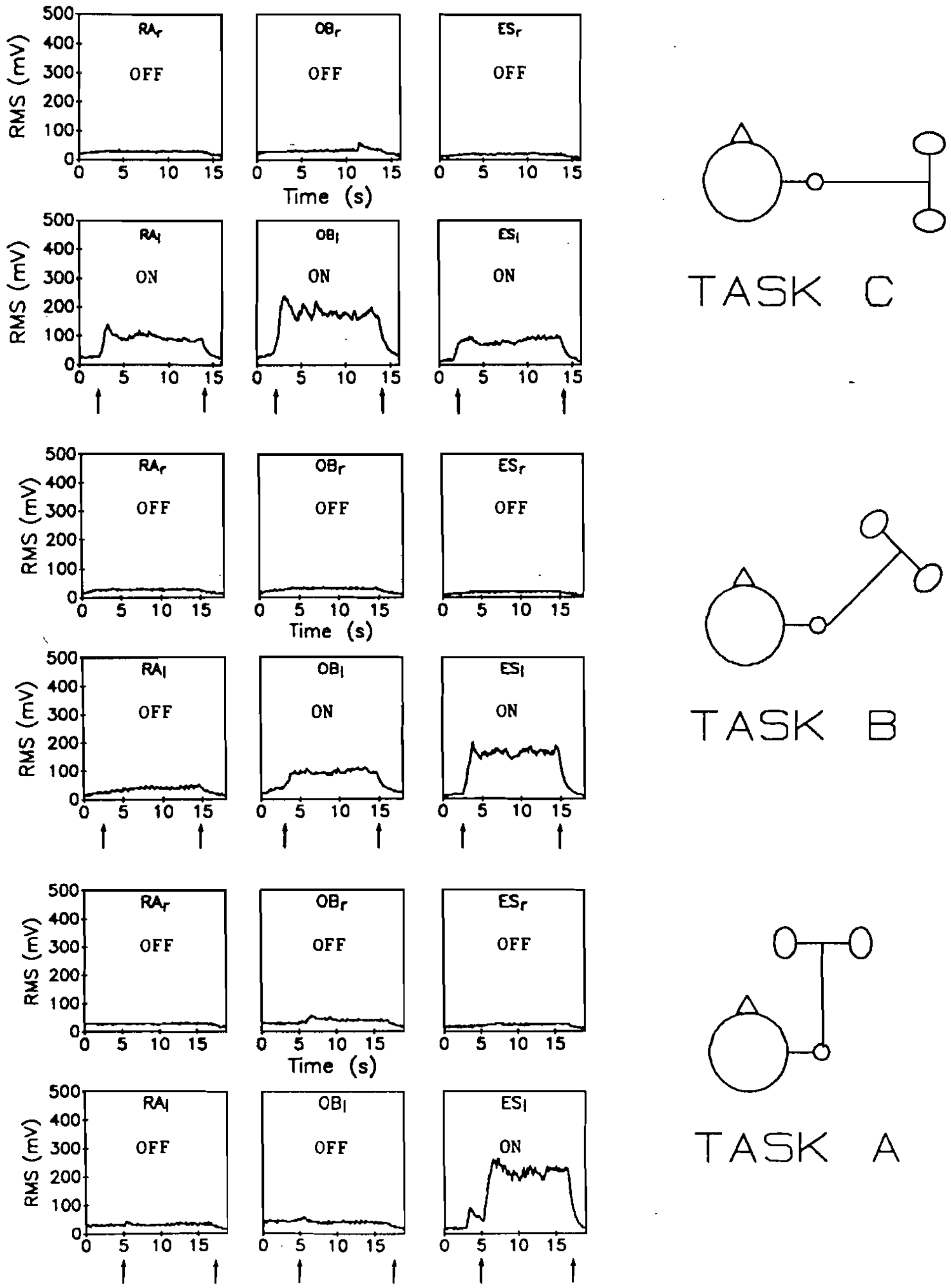

Fig 7. Electromyographic traces of all of the six monitored muscles in three different tasks. (*r* indicates the right muscles and *l* indicates the left muscles)

ergonomics manuals[3] to determine the moments generated by the applied loads. This was done to match the anatomic data for the lumbar musculature from the "average" person. A determination of the activity state of all the muscles under study can be made based on the location of each point representing a task on the loading plane. For example: the left ES muscle is predicted to be active for all the three tasks involving a right hand-held weight (A–C), and for tasks D and E involving the left hand-held weight, and inactive only in task F (left hand-held weight in 90° internal rotation). Similar determinations can be made for all the other muscles, and the results are presented in Table 2, which uses the symmetry property of the switching curves, which predicts that the activation pattern of all of the low-back muscles arising from a right hand load, will by symmetrical to the activation pattern that arises from a load held symmetrically in the left hand. Hence, the activation pattern of all the muscles can be studied in terms of an ipsilateral or contralateral loading, eliminating the need for continual referral to the right or left muscles and loads.

With the prediction of the muscle activity state in place, we can now move to the comparison between the experimental results and the theoretical prediction. The activity state of the left back muscles in tasks A through C for the first series of experiments compared with the prediction of the theoretical model is shown in Figure 9. As only the active muscles in each task are shown, theory predicts that the erector spinae (ES) is active in all three tasks, the oblique abdominal (OB) is active in tasks B and C, while the rectus abdominus (RA) is active only in task C. Seven out of the eight subjects show the predicted activation pattern, and even subject SG shows the predicted pattern in two out of the three tasks. The similar results that were obtained for tasks D through F

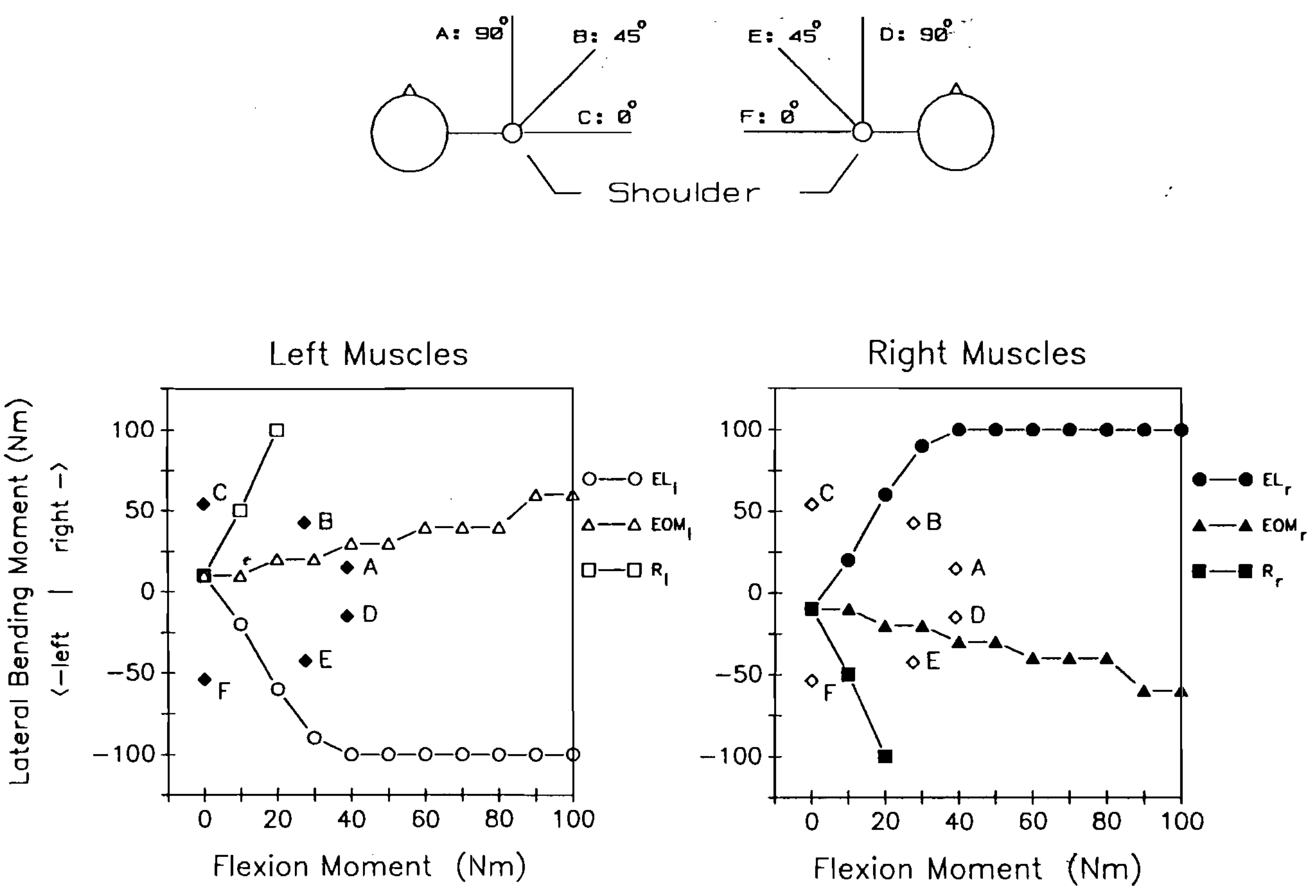

Fig 8. Switching curves for the left and the right muscles; A through F indicate the loading points (ie, the bending moment combinations) applied in this study. The stick figure illustrates schematically the different loading conditions that produced the above moment combinations.

suggested that the model predictions are strongly correlated with the experimental evidence.

The total experimental testing involved eight subjects, each tested twice in six different tasks. A total of 96 tests were conducted. The EMG data for each subject were analyzed for each one of the six muscles, and the results were summed based on the contralateral or ipsilateral location of the muscle with respect to the load. Table 3 shows the summary of the results for a given subject. The results are organized as a "truth table" for each muscle. For example, in the first series of experiments, the contralateral ES is predicted always to be on. This condition is indicative of having a score of 1 six times, once for each of the tasks A through F. This condition is expressed in Table 3 by the score of 6 at the top left cell and 0 everywhere else. Generally, the larger the numbers in the diagonal cells of the table, the better the correlation between theory and experiments.

The experimental results and the theoretical predictions were summed for each one of the six groups of muscles, ie, a summation across subjects. The results are presented in Table 4. The examination of the table for the contralateral ES in Experiment Series 1 reveals that it was predicted to be ON (ie, have score of 1) in all 48 experiments, and was found to be ON all 48 times. In comparison, the ipsilateral ES was predicted to be ON 32 times but was found to be ON only 13 times, always in tasks where it was predicted to be ON. It was also found to be OFF 35 times, 19 of which were cases where it was expected to be ON. As will be discussed later, the reason for this particular discrepency between the theoretical predictions and the experimental findings may be

Table 2. Summary of Predicted Activity

	Contralateral			Ipsilateral		
Trial	*RA*	*OB*	*ES*	*RA*	*OB*	*ES*
0° Internal rotation	1	1	1	1	0	0
45° Internal rotation	0	1	1	0	0	1
90° Internal rotation	0	0	1	0	0	1

RA = rectus abdominus; OB = oblique abdominal; ES = erector spinae.

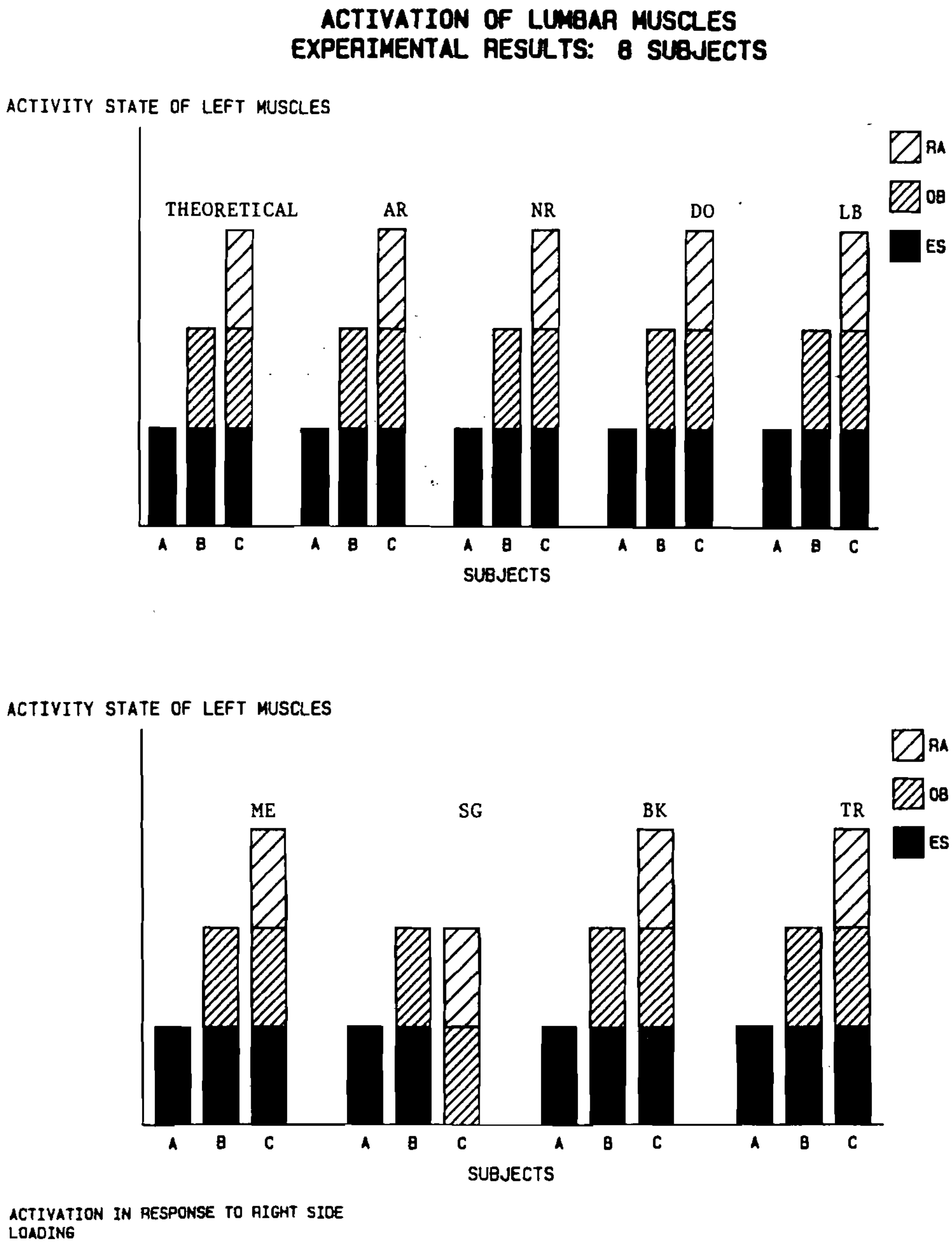

Fig 9. Activity states of the three left-side muscles in tasks A through C (right hand-held loads).

strongly related to the choice to the ES muscle as the example for comparison.

Chi square analysis was performed on the 2 × 2 tables for each muscle in Table 4. This analysis is generally used to determine how much of the observed activity is due to randomness and how much is due to the model predictions. For the one degree-of-freedom (*df*) statistical model, the chi square (χ^2)-statistic should be greater than or equal to 6.635 for $P < 0.01$ (ie, $\chi^2_{0.99} = 6.635$). The resultant χ^2 values are summarized in Table 5. Note that two sets of muscles, namely the ipsilateral obliques and the contralateral erector spinae, have no test statistic. This is because these muscles are always predicted ON (ES_c) or OFF (OB_i) and as there are zero values along one half of the table, a statistic cannot be calculated. These cases will be discussed individually.

From the results in Table 5, one can see that the statistic is significant for all trials but one for which a test statistic could be calculated. Thus, there is statistically strong correlation between the model predictions and the observed muscle activity. Furthermore, the model is extremely powerful for contralateral predictions with values greater than 30 and the model is somewhat less powerful in ipsilateral predictions. The results also show that there is no significant difference between the values calculated based on the first series of experiments and those based on the second series. The two cases for which the χ^2 statistic could not be calculated also provide acceptable correlations between the experimental and predicted observation. The contralateral erector spinae (ES_c) was predicted ON for all 48 trials for each set of data. For Series 1, it was experimentally ON 46 times. Thus, the model was correct 96% of the time. For Series 2 (repeated trials), the model was correct 100% of the time. The ipsilateral obliques did not provide as good results. The model was 71% correct in the predictions for Series 1 and 69% for Series 2.

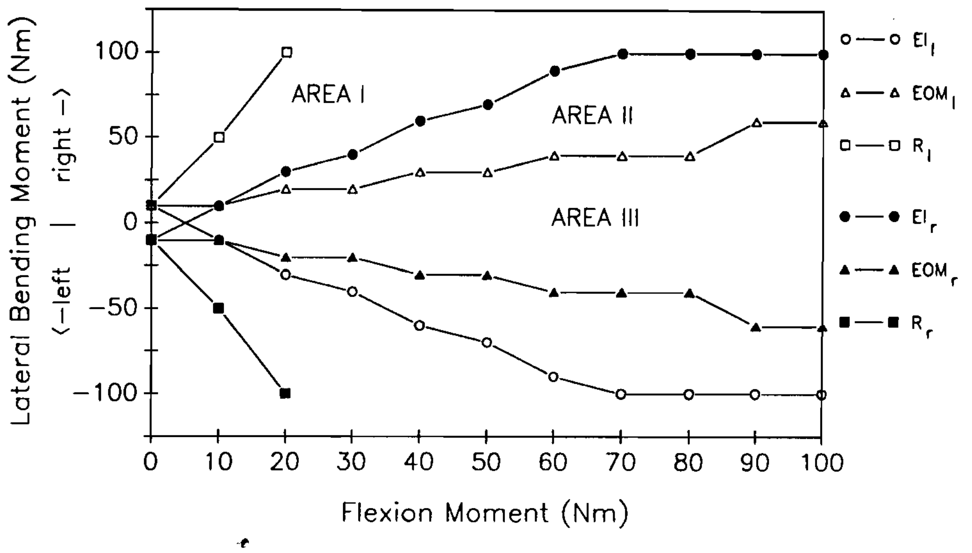

Fig 10. Switching curves of three bilateral pairs of muscles: ES = erector spinae represented by the longissimus; EOM = medial portion of the external oblique; RA = rectus abdominus (*l* indicates left and *r* indicates right).

DISCUSSION

The focus of this paper is the relationship between a general class of loading conditions applied to the upper body, and the resulting distribution of muscle forces in the lower back. Even by restricting the class of loading conditions to weight-holding tasks with no external torsion, there is still an infinite number of combinations of load and upper arm positions that can be generated. Recalling that 22 different muscles cross the lumbar area of interest (L3), it is easy to appreciate the difficulty in establishing a comprehensive framework for describing the relative role of the different muscles in opposing the external loads.

The basic concept of the analysis presented in this paper has its origin in the work of Andersson et al,[1] who suggested that the vertebral column and the muscular structures surrounding it can be functionally viewed as a mechanical joint. As Seroussi and Pope[15] point out, this amounts to a ball and socket joint (the spine) stabilized by a series of tension guy wires (the muscles). A system of this kind distributes the mechanical load such that the vertebral column balances the external forces, while the muscles balance the external moments. By limiting the physical tasks to weight-holding tasks with no external torsion, the load distribution among the different muscles thus will be determined by the combination of lateral and flexion bending moments only. Hence, the infinite number of tasks involving all the possible combinations of weights and upper limbs position maps to different points in a loading plane whose axes are given by the lateral and the anterior-posterior (flexion) bending moments.

It is important to stress that the actual solution to the mechanically redundant problem presented in this paper, namely, what is the load distribution among many muscles that cross the given level of the lumbar region in response to a set of external loads, relies on the optimization assumptions used in the mathematical solution. The particular solution presented in this paper uses a linear programming algorithm, which by its nature will tend to equili-

Table 3. Results from Subject AR for One Trial Series

	Experimental ON	*Experimental OFF*
Contralateral RA		
Predicted ON (2)	2	0
Predicted OFF (4)	1	3
Ipsilateral RA		
Predicted ON (2)	2	0
Predicted OFF (4)	0	4
Contralateral OB		
Predicted ON (4)	4	0
Predicted OFF (2)	0	2
Ipsilateral OB		
Predicted ON (0)	0	0
Predicted OFF (6)	0	6
Contralateral ES		
Predicted ON (6)	6	0
Predicted OFF (0)	0	0
Ipsilateral ES		
Predicted ON (4)	2	2
Predicted OFF (2)	0	2

RA = rectus abdominus; OB = oblique abdominal; ES = erector spinae.

Table 4. Results from All Subjects for One Trial Series

	Experimental ON	*Experimental OFF*
Contralateral RA		
Predicted ON (16)	14	2
Predicted OFF (32)	2	30
Ipsilateral RA		
Predicted ON (16)	4	12
Predicted OFF (32)	0	32
Contralateral OB		
Predicted ON (32)	32	0
Predicted OFF (16)	2	14
Ipsilateral OB		
Predicted ON (0)	0	0
Predicted OFF (48)	15	33
Contralateral ES		
Predicted ON (48)	48	0
Predicted OFF (0)	0	0
Ipsilateral ES		
Predicted ON (32)	13	19
Predicted OFF (16)	0	16

RA = rectus abdominus; OB = oblique abdominal; ES = erector spinae.

Table 5. Summary of Chi Square Analysis for Four Muscles

Muscle	Trial Series 1 chi square	Trial Series 2 chi square
Contralateral RA	39.42*	31.69*
Ipsilateral RA	3.41	8.73*
Contralateral OB	35.35*	39.53*
Ipsilateral ES	8.00*	8.91*

RA = rectus abdominus; OB = oblique abdominal; ES = erector spinae.
*Chi square statistic significant at $P < 0.01$.

brate the stress level across all the muscles that participate in the load sharing (where stress is the muscle force divided by the muscle's cross-sectional area). We do not have any mechanism for directly verifying the existence of such a criterion, and hence we do not mean to imply that this paper proves that the central nervous system tries to evenly divide the load among all the muscles. Still, our measurements and the comprehensive framework suggested in this paper for examining measurements of previous researchers suggest that such a model is a powerful tool in predicting the response of the lumbar musculature to a given set of external loads.

The identification of the loading plane as the determinant of the individual muscle forces enabled us to plot the switching curves for all the different muscles. The switching curve of an individual muscle divides the loading plane into two areas: one representing loading conditions that will cause activation of the muscle, and the other, conditions that will leave the muscle inactive. The switching curve can be viewed as the loading threshold required to activate a particular muscle. Seroussi and Pope[18] reported the existence of a threshold level of the lateral bending moment required to activate the centralized external oblique. They also reported that the threshold level is independent of the flexion moment. Both of these findings are corroborated in Figure 6, where the switching curve of the lateral portion of the external oblique is indeed parallel to the flexion moments axis, reflecting the independence of the flexion moment and the threshold level of the lateral bending moment.

The switching curves described in Figure 6 predict that a pure flexion load (represented by loading condition along the flexion moment axis) will not activate any of the abdominal muscles (note the location of the lateral portion of the external oblique (EOL), medial portion of the external oblique (EOM), and rectus abdominus [R] switching curves relative to the flexion moment axis), and indeed Schultz et al[13] reported that for such tasks the measured EMG from these muscles stayed at resting levels. The activation of the contralateral ES in response to a lateral bending moment is predicted by the switching curves and has been reported by many researchers, including Floyd and Silver,[5] Jonsson,[8] Andersson et al,[1] and Seroussi and Pope.[15] Jonsson[8] also points out that equal lateral loading of both hands showed "... no asymmetry of ES action, and usually very little increase in activity over the resting, unloaded condition . . ." Again, such a loading will bring the overall moment loading condition very close to zero (on the loading plane), which theoretically would not involve any activity of the muscles above the resting level.

The switching curves can serve as a useful tool in the examination of physiologic assumptions and approximation involving muscular activity of the lower back. In addressing the issue of mechanical redundancy in the low-back muscles, some investigators have proposed schemes of muscle activation or deactivation. In so doing, they effectively coupled the activity of different muscle groups. For example, Schultz et al[13] proposed that for pure flexion loading either ES or OB and RA (lumped together as a single equivalent muscle group) are active. Such an assumption is supported by the switching curves described in Figure 6. A 2 *df* model addressing the flexion and the lateral bending moment was described by Andersson et al,[1] who proposed to distribute the muscular load between the ipsilateral (with respect to the load) ES and the contralateral abdominal in the following manner:

1) $M_x > M_y$: Ipsilateral ES is active; contralateral abdominal is inactive.

2) $M_x < M_y$: Ipsilateral ES in inactive; contralateral abdominal is active.

where M_x is the flexion moment and M_y is the lateral bending moment. The ipsilateral abdominals are always considered to be inactive.

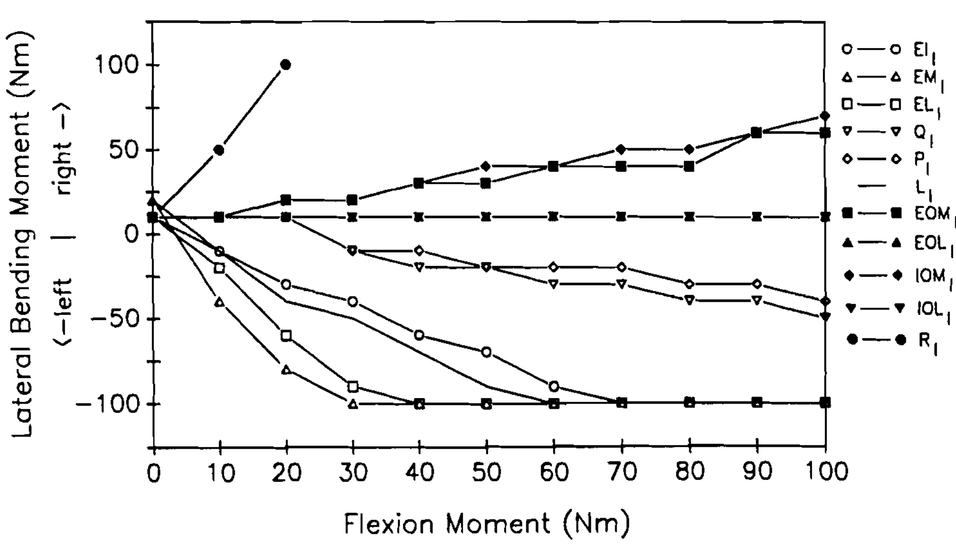

Fig 11. Switching curves of all the 11 muscles on the left side of the body: EI = iliocostalis; EM = multifidus; EL = longissimus; Q = quadratus; P = psoas; L = latissimus; EOM = medial portion of the external oblique; EOL = lateral portion of the external oblique; IOM = medial portion of the internal oblique; IOL = lateral portion of the internal oblique; R = rectus abdominus.

Figure 10 describes the switching curves of the bilateral abdominal (represented by EOM) and the bilateral erector spinae (represented by the ES). For small values of the loading moments, the above assumptions seem appropriate as both curves are fairly close to one another. But as the external loads increase, one can identify two areas where the above constraints are satisfied, areas I and III. However, since the ipsilateral muscles are activated in the clockwise direction, whereas the contralateral muscles are activated in the counterclockwise direction, there also is a broad range of loading conditions (area II) where both muscles are active. The range of lateral bending moments reported by Seroussi and Pope[15] in support of the above physiologic constraint was approximately 11 N-m. This relatively low loading moment may explain why the existence of area II was not reported earlier.

Finally, one can use the switching curves to study the appropriate way to group together the lumbar muscles into functional units. Based on the calculations of the switching curves of the 11 muscles on one side of the lumbar cross section (Figure 11), it appears that the muscles can be grouped into five functional units:

1) ES: consisting of the longissimus, multifidus, latissimus, and iliocostalis;

2) Paraspinal: consisting of the psoas and quadratus;

3) Medial abdominal: consisting of the medial portions of the external and internal obliques;

4) Lateral abdominal: consisting of the lateral portions of the internal and external oblique; and

5) Rectus abdominus.

This also will be the order of recruitment if the lateral bending moment is gradually increased for a given value of flexion moment, from negative values (representing ipsilateral bending) to large positive values (representing contralateral bending). The introduction of functional groups that are recruited under similar loading conditions suggests that the simplest biomechanical model that could capture the complexity of the lumbar region is composed of ten muscle groups (five functional groups on each side). Functional grouping may prove to be a more valid criterion to use when lumbar muscle models are studied than grouping based on geometric or anatomic considerations alone (Schultz et al[14]).

The EMG measurements were designed to test the concept of the switching curves, ie, does the external bending moment combination determine the activation of the lumbar muscles. Such an approach avoids the need to identify a force-EMG relationship, the nature of which varies considerably (see a summary of 22 different studies conducted between 1952 and 1978 in Basmajian and DeLuca[2]), and produces a true-false determination. The results suggest that the model predictions are highly correlated with the experimental results for the contralateral muscles (with respect to the load).

The poorest correlation was obtained for the ipsilateral muscles: the ipsilateral ES was judged correct only in 60% of the cases and the ipsilateral OB and RA were correct in 70% of the cases. The relatively high rate of failure for the ES may illustrate the caution that must be exercised when the load combination, represented by a point on the loading plane, is close to a switching curve of a given muscle. The prediction of whether that muscle is active or inactive depends on the relative location of the loading point with respect to the curve. As our determination of the loading point (calculated by loading the "average" person) and our determination of the switching curves were based on anatomic data from cadaver studies and not on the individual subjects' anatomic data, we would expect to have poorer predictions when the loading point and the switching curve are close. It is interesting to note that even though our predictions were based on "average" anatomic data, the results were still significant for the ipsilateral ES at the $P < 0.01$ level. If the loading point or the switching curve are slightly altered (for example, if the experimentally tested muscle was the lattissimus dorsi and not the longissimus), our prediction rate would have been 92% based on the 2×2 table calculated for the ipsilateral latissimus dorsi.

Our results support previous observations that simple mechanical models are usefully accurate in predicting the ES muscle activity (Andersson et al,[1] Schultz et al,[13] Seroussi and Pope[15]), and that the correlation is weaker for the abdominal muscles, in particular the obliques (Schultz et al[14]). One reason for the weaker correlation could be the involuntary introduction of twisting moments in the single-arm loading tasks, which would have affected the activation of the oblique muscles contrary to the model assumptions.

It is interesting to note that even though all of the anatomic data used in this model is based on an "average person", ie, the muscle cross sections and centroids were based on cadaver studies and ergonomics manuals, a high degree of correspondence was achieved between the model and the actual measurements. This coincidence suggests that the switching curves may be applicable to a broad range of anatomic values, without the need to correct for the size or the weight of the subject. Further studies on a larger population will have to be conducted in order to further study this property of the switching curves.

CONCLUSION

This study introduces the concept of mechanical recruitment of lumbar musculature. It suggests that for a class of physical tasks that involve holding weights in an erect posture and an arbitrary position of the upper arms, the lumbar muscle force distribution will be determined by the values of the overall flexion and lateral bending moments. The bending moments can be described as the "loading plane," where the physical task can be represented as a single point on that plane.

The theoretical calculations of the lumbar muscle load distribution were based on optimization criteria (ie, minimization of the spinal compression) and used linear programming techniques that produce a uniform distribution of muscle stresses. Studying the effect of different loading conditions on the muscle force distribution, it became apparent that all the muscles have regions of activity and inactivity separated by a "switching curve." Plotting the switching curve of all the muscles on one side of the body revealed that the muscles can be lumped into five functional groups. Those groups are activated sequentially as the lateral bending moment shifts from the ipsilateral to the contralateral side of the body. Hence, it may be impossible to activate all of the groups individually, and the activation order for the above exercise will be: ES, P + Q, lateral OB, medial OB, and RA. The only muscle that can thus be isolated is ES (provided that the flexion moment is larger than zero). Electromyographic validation of the switching curve concept was conducted and showed a high degree of correlation between the model predictions and the EMG measurements.

Our work suggests a comprehensive framework for studying the effect of mechanical loading on lumbar musculature recruitment. It further provides a functional grouping of the 22 muscles into five functional groups that are activated or deactivated depending on the combination of bending moments applied to the upper body. It should be stressed that this paper does not prove the existence of any optimization criteria or strategies in the central nervous system that would lead to the mechanical recruitment pattern in the lumbar muscles. It does suggest a comprehensive framework and a derived activation pattern that is correlated with experimental findings. This correlation, if it does not describe true behavior of the central nervous system in activating the lumbar musculature, at least points to a comparable model that produces outputs consistent with experimental evidence. As such, it points to the need for a more extensive study of the effects of mechanical loading on patterns of muscle activation in the lumbar region.

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