Analysis and Measurement of Neck Loads

Sean P. Moroney, Albert B. Schultz, and James A. A. Miller

Department of Mechanical Engineering, University of Michigan, Ann Arbor, Michigan, U.S.A.

Summary: To examine the loads imposed on the structures of the neck by the performance of physical tasks, a biomechanical model of the neck was constructed. The model incorporated 14 bilateral pairs of muscle equivalents crossing the C4 level. A double linear programming optimization scheme that minimized maximum muscle contraction intensity and then vertebral compression force while equilibrating external loads was used to calculate the muscle contraction forces required and the motion segment reactions produced by task performance. To test model validity, 14 healthy adult subjects performed a series of isometric tasks requiring use of their neck muscles. These tasks included exertions in attempted flexion, extension, and left and right lateral bending and twisting. Subjects exerted maximum and submaximum voluntary efforts. During the performance, surface myoelectric activities were recorded at eight locations around the periphery of the neck at the C4 level. Calculated forces and measured myoelectric activities were then linearly correlated. Mean measured voluntary neck strengths in 10 male subjects were as large as 29.7 Nm. Four female subjects developed mean strengths that were approximately 60%–90% of those of the males. In both sexes, neck muscle strengths were approximately one order of magnitude lower than previously measured lumbar trunk strengths. Mean calculated neck muscle contraction forces ranged to 180 N. Mean calculated compression forces on the C4–5 motion segment ranged to 1164 N, lateral shear forces ranged to 125 N, and anteroposterior shear forces ranged to 135 N. Correlation coefficients between the calculated muscle forces and the measured myoelectric activities were as large as 0.85 in some muscles, but generally were smaller than this. **Key Words:** Neck biomechanics—Spine biomechanics—Biomechanical models—Myoelectric measurements.

The neck is susceptible to trauma and is frequently a site of pain (7,11,15). Episodes of neck pain have been estimated to affect at one time or another approximately 40% of the population (6,8,9). The pathogenesis of neck pain is often unknown, but evidence suggests that mechanical factors may sometimes be involved. For example, Kelsey et al. (10) found a marked increase in risk for cervical disc prolapse in association with frequent diving from a board (odds ratio of 2.3, for those diving 10 times or more in the past 2 years). Thus, the determination of the magnitudes of neck muscle contraction forces and of cervical spine compression and shear forces imposed by the performance of physical tasks is likely to have clinical relevance.

Loads internal to the lumbar trunk and spine have been studied (1,16–18). In those and similar studies, internal loads were calculated using biomechanical model analyses and model validity tested through measurement of trunk muscle myoelectric activities. Calculations and measurements generally were highly correlated.

The present study applied these techniques to the

cervical spine and its musculature. Fourteen subjects performed 30 isometric tasks calling for neck muscle contractions. These tasks included attempted neck flexion, extension, and left and right lateral bending and twisting. Biomechanical model analyses of these efforts were used to predict the required neck muscle contraction forces and measurements of the neck muscle myoelectric activity were used to test the validity of the predictions. A primary aim of this study was to examine quantitatively what loads arise internal to the neck when external loads are applied to the head. A secondary aim was to measure maximum voluntary strengths of the neck musculature.

Experimental Measurements

Fourteen adult volunteers, ten males and four females, participated in the experiments. Data on the subjects are presented in Table 1. None of the subjects reported a history of neck injury or notable neck pain.

Each subject performed 30 quasistatic tasks calling on his neck musculature. The subjects were asked to sit in a chair placed in a reference frame. Their shoulders were strapped to a board fixed behind the chair, and their hands placed in their laps. A 2.5 kg helment was worn by each subject. The helment was equipped with attachments to permit application of external loads. The helment could be custom-fitted by means of foam inserts. The subjects wore a bathing cap to maximize friction between head and helment. The C4 level of the neck was located by palpation of the vertebrae. Markers were placed on the side of the neck to indicate the

TABLE 1. *Subject anthropometric data*

Subject	Sex	Age (years)	Height (cm)	Weight (N)	Neck diameters A-P (cm)	Neck diameters Lat. (cm)
1	M	43	183	758	10.7	11.4
2	M	59	173	602	11.4	11.7
3	M	21	175	557	10.0	11.0
4	M	25	183	803	12.1	11.4
5	M	24	175	776	12.1	11.4
6	M	28	173	647	11.1	11.1
7	M	33	168	713	11.5	11.5
8	M	29	173	638	10.4	10.5
9	M	19	170	535	10.8	11.2
10	M	25	198	869	11.2	10.2
11	F	34	160	508	9.3	8.5
12	F	27	173	580	9.3	9.3
13	F	23	168	513	8.8	8.8
14	F	20	155	473	8.3	8.9
Means	M	30.6	177.0	689.8	11.1	11.1
SD	M	12.1	8.9	111.5	0.7	0.5
Means	F	26.0	163.8	518.6	8.9	8.9
SD	F	6.1	7.9	44.6	0.5	0.3

level of C4, and on the helment to indicate the approximate center of mass of the head plus helment. The coordinates of these points were recorded during testing by using a horizontal and a vertical scale along with a plumb bob and a level.

In one series of tasks, each subject was asked first to relax and then, with head and neck maintained in an upright posture, to exert for 5 s his maximum voluntary strength against a load cell in attempted flexion, extension, and left and right lateral bending and twisting (Fig. 1). Given feedback, he then exerted 25% and 50% of his maximum strength in a like manner. In another series of tests, the subjects were asked to resist applied moments in flexion, extension, and left and right lateral bending and twisting. These moments were pro-
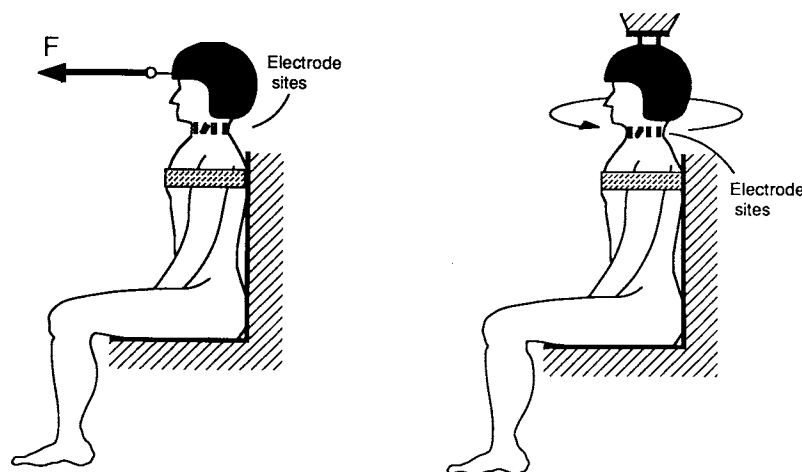

FIG. 1. Schematic of testing procedures. The seated subject has his trunk restrained and electrodes mounted on his neck at the C4 level. (Left) The subject resists the flexion moment created by an anteriorly directed force. (Right) The subject attempts to twist his neck to the left.

duced by free-hanging weights using cords attached to hooks on the helment and pulleys. The moments applied were approximately 10% and 30% of the maximum strengths. The maximum duration of the tasks was 10 s. The subjects rested for at least 2 min between tasks; a 10-min rest period was provided between sets of tasks (Table 2).

Eight bipolar electrodes were affixed to the surface of the neck at the C4 level at bilaterally symmetric locations. The electrode locations, denoted as (1) anterior, (2) anterolateral, (3) posterolateral, and (4) posterior, with approximate azimuth angles of 35°, 70°, 105°, and 150°, respectively, were determined by reference to topographical landmarks. These locations were (a) midway between the anterior midline and the anterior border of the sternocleidomastoid (SCM) muscle, (b) midway between the anterior and posterior borders of the SCM muscle, (c) midway between the posterior border of the SCM muscle and the anterior border of the upper trapezius muscle, and (d) midway between the anterior border of the upper trapezius muscle and the posterior midline. The myoelectric signals were preamplified, then further amplified, full-wave rectified, integrated with a reset time constant of 0.2 s (using Grass Instrument Company Model 7P3 myoelectric signal processors), and displayed on an eight-channel chart recorder. Mean signal levels were determined from these recordings.

BIOMECHANICAL MODEL ANALYSES

The contraction forces in the neck muscles and the compression and shear loads they imposed on the cervical spine were computed for each task and each subject by biomechanical analyses. Each analysis was based on the requirement that the structures superior to an imaginary cutting plane at the C4 level remain in equilibrium during the performance. The analysis was carried out in two steps. First, the net reaction needed to equilibrate the structures superior to the C4 level was computed. Then, a set of neck muscle contraction forces that could produce that net reaction and its resulting set of compression and shear loads on the C4–5 motion segment was calculated.

Knowledge of the external loads and of the weights of the subjects' heads and the helment was required for the computations. The external loads were the known cord-applied forces and moments. The weight of the head was assumed to be 7.3% of the subject's total body weight and to act midway between the ears, based on the data of Clauser et al. (3). The coordinates of the force application points were available for each task and for each subject from measured configuration data. These data were entered into the six equations expressing the requirements for equilibrium. These were that the applied and body segment weight forces had to be balanced by the forces transmitted across the C4 section along the left, posterior, and superior axes; and the moments of those forces had to be balanced by the three moments about those C4 section axes. These requirements enabled the six components of the net reaction transmitted across the C4 section to be computed. The contraction forces in 14 pairs of neck muscles and the cervical spine compression and shear loads were then calculated from the net reaction using a biomechanical model of the C4 cross section and solving the model equations by an optimization technique.

A scaled cross-sectional anatomy drawing of the

TABLE 2. *Sequence of tasks performed*

Attempted movements	Percent of max. vol. effort
Attempted movements	
1 Left twist	100
2 Left twist	25
3 Left twist	50
4 Right twist	100
5 Right twist	25
6 Right twist	50
Resisted loads	
7 Right twist	10
8 Right twist	30
9 Left twist	10
10 Left twist	30
Attempted movements	
11 Extension	100
12 Extension	25
13 Extension	50
14 Left bending	100
15 Left bending	25
16 Left bending	50
17 Flexion	100
18 Flexion	25
19 Flexion	50
20 Right bending	100
21 Right bending	25
22 Right bending	50
Resisted loads	
23 Flexion	10
24 Flexion	30
25 Right bending	10
26 Right bending	30
27 Extension	10
28 Extension	30
29 Left bending	10
30 Left bending	30

In the resist tasks, the loads applied were determined from the loads developed in the maximum effort tasks.

C4 level [the carefully detailed drawings of Eycle-shymer and Schoemaker (4) were used, although data from more modern texts could have served as well] provided the basic data on muscle centroid locations and cross-sectional areas (Fig. 2). A reproduction of the drawing was pasted onto a uniform sheet of cardboard. By weighing cutouts of each muscle cross section, and weighing the total rectangular cross section defined by the neck diameters, the relative area of the cross section of each muscle was determined. Cutout centroids were located by constructing cross plumb lines on the cutouts. Centroid locations relative to neck diameters were then determined by linear scaling. Muscle lines of action were determined similarly from scaled cross-sectional drawings at adjacent levels.

To construct the model, the origin of an orthogonal coordinate system was placed at the C4–5 disc center, with axes positive to the left, posterior, and superior. Muscle centroidal coordinates were expressed relative to the frontal and sagittal plane neck diameters; their areas were scaled relative to the product of the diameters (Table 3). The muscle forces were assumed to act at the muscle centroids.

In the model, the C4–5 motion segment was assumed to offer resistance to compression and shear forces, but not to bending or twisting moments.
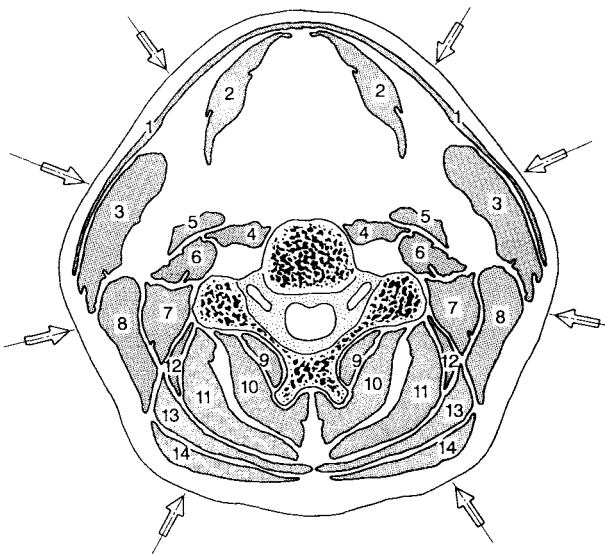

FIG. 2. A cross section of the neck at the level of the C4 illustrating the muscles modeled: (1) platysma, (2) infrahyoid, (3) sternocleidomastoid, (4) longus colli & cervicis, (5) scalene anterior, (6) scalene medius, (7) longissimus cervicis, (8) levator scapulae, (9) multifidus, (10) semispinalis cervicis, (11) semispinalis capitis, (12) spenius cervicis, (13) spenius capitis, and (14) trapezius. The arrows indicate the electrode locations.

This assumption, analogous to that used in the lumbar trunk model of Schultz et al. (17), was based on the data of Moroney et al. (14). They found cervical spine motion segments to have shear stiffnesses on the order of 100 N/mm and compression stiffnesses on the order of 1000 N/mm. Thus, sizable forces could be resisted by the cervical spine with only a few mm of vertebral relative displacement. On the other hand, they rotational stiffnesses to be on the order of 1 Nm/deg. In tasks analyzed here, the neck was maintained upright, so the motion segments probably developed rotational resistances of less than 1 Nm. The mean external moments in the tasks studied here were as large as 26 Nm. So, the assumption used seems reasonable.

The optimization scheme employed (2) first determined by linear programming the minimum muscle contraction intensity needed to equilibrate the applied load. Next, with this contraction intensity as an upper bound, the muscle equivalent forces that maintained equilibrium while minimizing the compression force on the spine were calculated, again by linear programming. This scheme called on nearly every muscle that could contribute to an activity to do so at nearly equal contraction intensities, while keeping antagonistic muscle activity to a minimum.

The predicted force in each muscle equivalent was paired with the myoelectric signal level in the electrode nearest to that equivalent. The 30 sets, one per task, of data pairs of mean measured myoelectric signal level and mean predicted contraction force were correlated by a linear least-squares regression and the correlation coefficients computed.

RESULTS

The mean voluntary moments developed by the subjects ranged to 25.9 Nm (Table 4).

In relaxed upright postures, mean myoelectric signals were approximately 10 μV (Table 4). The largest mean signals (approximately 208 μV) were recorded from the anterior electrodes during maximal flexion. The largest mean posterior electrode signals were approximately 50 μV and these occurred in maximal extension. The largest myoelectric signals during twisting or lateral bending were recorded from the anterolateral electrodes.

During attempted lateral bending, signal strengths were larger to the side of bending. In attempted twisting, the largest signals were recorded

TABLE 3. *Data incorporated into the C4 cross-section model*

Muscles modeled	Area	Centroid locations (left side)		Direction Angles of the line of action (left side)		
		Left	Posterior	Left	Posterior	Superior
Platysma	0.0040	0.208	− 0.353	115	105	30
Infrahyoid	0.0128	0.135	− 0.343	90	90	0
Sternocleidomastoid	0.0301	0.396	− 0.088	75	58	37
Longus colli and cerv.	0.0055	0.115	− 0.049	80	90	10
Scalene anterior	0.0075	0.228	− 0.049	105	90	15
Scalene medius	0.0079	0.240	0.010	105	103	20
Longissimus cerv.	0.0051	0.260	0.108	80	90	10
Levator scapulae	0.0228	0.323	0.147	110	90	20
Multifidus	0.0083	0.073	0.176	140	70	45
Semispinalis cerv.	0.0189	0.073	0.275	95	90	5
Semispinalis cap.	0.0248	0.188	0.284	90	90	0
Splenius cervicis	0.0030	0.260	0.225	80	100	15
Splenius capitis	0.0120	0.250	0.304	77	105	20
Trapezius	0.0144	0.188	0.373	120	90	30

Centroid locations are given in ratio to neck frontal plane and sagittal plane diameters. Areas are given in ratio to the product of these two diameters.

from the anterior and anterolateral electrodes on the side opposite the direction of twisting. Bilateral symmetry in response to the symmetric loads was evident in the myoelectric signal increases recorded by both the posterolateral and posterior electrode pairs. The measured myoelectric signals and the predicted muscle contraction forces showed coefficients of variation that ranged generally from 25% to 75%.

The largest predicted muscle contraction forces (Table 5) occurred in the SCM in maximum twisting attempts. The SCM was also the site of the largest predicted contraction forces in other maximum voluntary strength exercises, except in attempted extension. Generally, predicted muscle forces increased monotonically with increasing load. However, in a few cases, a decrease in predicted force with increasing load was observed. This occurred in those submaximal exercises for which the moment due to head and helmet weights equaled or exceeded the applied moment.

The percentage contributions of each muscle equivalent to an exertion varied. Some muscle equivalents, splenius capitis for example, contributed to all performances at nearly the same percent level. In contrast, others contributed significantly only to some types of exertions. For example. levator scapulae contributed strongly to lateral bending but negligibly to twisting. SCM was the largest single contributor to both attempted lateral bending and twisting. The infrahyoids made the largest contribution to attempted flexion. The

semispinales muscles accounted for over one-half the effort in attempted extension.

The mean predicted motion segment reactions for the maximum voluntary exertions are summarized in Table 6. The largest lateral shear (approximately 125 N) occurred during lateral bending. The largest anteroposterior shear (approximately 135 N) occurred in extension. The largest compression (approximately 1164 N) occurred during extension and the smallest (approximately 578 N) during flexion. For comparison, the estimated mean weight of the subjects' heads was 46.7 N. Thus, the calculated motion segment compression forces in the maximum exertions were 12.4 and 24.9 times head weight in attempted extension and flexion, respectively, and the shear forces were 2.7 and 2.9 times head weight in attempted lateral bending and extension, respectively.

The coefficients of linear correlation between predicted forces and measured myoelectric signal levels over all 30 tasks (Table 7) ranged from 0.29 to 0.85. Those for the anterior and anterolateral muscles were generally high, while those for the posterolateral and posterior muscles were more variable.

Complete tabulations of results are available from the authors.

DISCUSSION

More detailed information on the present study is provided by Moroney (13). The decision to model

TABLE 4. *Mean applied moments and mean myoelectric signal levels*

Task	Mean applied moment (Nm)	Mean left side myoelectric signal level (μV)			
		Anterior	Anterolateral	Posterolateral	Posterior
Relaxed upright		12 (8)	10 (3)	11 (2)	5 (3)
Resisted right twist, 30%	4.1 (2.8)	45 (49)	55 (64)	30 (22)	25 (21)
Attempted left twist, 50%	4.6 (1.5)	35 (18)	57 (73)	34 (29)	34 (27)
Attempted left twist, 100%	8.5 (3.1)	99 (58)	134 (102)	77 (58)	58 (39)
Resisted right bend, 30%	4.8 (2.9)	28 (19)	50 (42)	51 (44)	19 (19)
Attempted left bend, 50%	6.4 (3.9)	31 (24)	47 (33)	46 (33)	18 (13)
Attempted left bend, 100%	12.8 (7.3)	104 (60)	122 (70)	112 (68)	50 (46)
Resisted flexion, 30%	8.7 (4.6)	16 (13)	11 (2)	17 (11)	17 (16)
Attempted extension, 50%	12.6 (6.6)	12 (2)	11 (4)	19 (16)	19 (9)
Attempted extension, 100%	25.9 (13.4)	16 (6)	14 (3)	44 (55)	52 (28)
Resisted extension, 30%	3.6 (2.3)	86 (54)	90 (80)	35 (23)	14 (19)
Attempted flexion, 50%	5.3 (3.2)	99 (61)	76 (40)	35 (20)	9 (9)
Attempted flexion, 100%	10.4 (6.3)	207 (70)	165 (71)	103 (54)	36 (37)

Task	Mean right side myoelectric signal level (μV)			
	Anterior	Anterolateral	Posterolateral	Posterior
Relaxed upright	12 (2)	11 (1)	10 (2)	10 (1)
Resisted right twist, 30%	73 (64)	112 (58)	36 (18)	30 (39)
Attempted left twist, 50%	61 (44)	113 (58)	34 (14)	28 (18)
Attempted left twist, 100%	112 (48)	174 (80)	76 (35)	51 (25)
Resisted right bend, 30%	16 (6)	16 (16)	14 (13)	12 (4)
Attempted left bend, 50%	21 (15)	30 (36)	15 (13)	12 (4)
Attempted left bend, 100%	61 (37)	56 (55)	27 (44)	16 (14)
Resisted flexion, 30%	12 (4)	12 (1)	13 (3)	23 (8)
Attempted extension, 50%	13 (2)	13 (5)	14 (2)	22 (6)
Attempted extension, 100%	16 (3)	14 (2)	30 (14)	49 (21)
Resisted extension, 30%	94 (57)	83 (52)	41 (25)	13 (6)
Attempted flexion, 50%	97 (46)	86 (48)	42 (24)	13 (4)
Attempted flexion, 100%	209 (77)	188 (79)	99 (61)	23 (17)

Standard deviations in parentheses.

the neck muscles at the C4 level reflected both the convenience of making myoelectric measurements there and the assumption that knowledge of muscle contraction forces at that level would protray well typical magnitudes of neck muscle forces in general. Subjects were placed in a supported sitting posture for the experimental measurements so that they needed to pay little attention to maintaining their trunks upright in performing the various tasks asked of them. Internal loads in structures inferior to the neck probably have little influence on neck loads, so this choice of posture seemed reasonable. Measurement of myoelectric activities with fine wire intramuscular electrodes would have been preferable, in some ways, to measurement with surface electrodes. However, it seemed unwarranted to subject the volunteers to the discomfort and risk associated with use of intramuscular elec-

trodes without first carrying out a study using surface electrodes.

The maximum voluntary twisting strengths developed by the subjects may have been limited by inability of the head/helment interface to transmit large moments without slipping. For all six modes of exertion, mean female strengths tended to be 60% to 90% of mean male strengths. The mean product of the neck diameters in the males was approximately 55% greater than in the females, which probably accounts at least in part for this gender difference.

The neck muscle strengths in attempted flexion and extension found in the present study are roughly equivalent to those reported earlier by Foust et al. (5). If the neck muscle strengths measured in this study are compared with lumbar trunk muscle strengths in healthy males and females (12),

TABLE 5. *Calculated muscle contraction forces (N) for maximum voluntary efforts*

Muscle equivalent	Relaxed	Attempted left twist	Attempted left bend	Attempted extension	Attempted flexion
L Platysma		23 (8)	19 (11)		19 (14)
R Platysma					19 (14)
L Infrahyoid		72 (25)	60 (35)		62 (45)
R Infrahyoid		68 (30)	41 (45)		62 (45)
L SCM			140 (83)		145 (106)
R SCM		170 (58)	68 (47)		145 (106)
L Longi		31 (11)	26 (15)		26 (19)
R Longi					26 (19)
L Scal. ant.		42 (15)	35 (20)		36 (27)
R Scal. ant.					36 (27)
L Scal. med.	3 (1)	45 (15)	37 (22)	39 (17)	
R Scal. med.	3 (1)			39 (17)	
L Long. cerv.	2 (1)	29 (10)	24 (14)	26 (11)	
R Long. cerv.	2 (1)			26 (11)	
L Levator scap.	8 (4)	2 (3)	106 (62)	116 (51)	
R Levator scap.	8 (4)	14 (14)		116 (51)	
L Multifidus	3 (1)			42 (19)	
R Multifidus	3 (1)	47 (16)	39 (23)	42 (19)	
L Semispin. cerv.	7 (3)		2 (8)	96 (42)	
R Semispin. cerv.	7 (3)			96 (42)	
L Semispin. cap.	9 (4)	96 (22)	111 (57)	127 (56)	
R Semispin. cap.	9 (4)			127 (56)	
L Splenius cerv.	1 (1)	17 (6)	14 (8)	15 (7)	
R Splenius cerv.	1 (1)			15 (7)	
L Splenius cap.	4 (2)	68 (23)	56 (33)	61 (27)	
R Splenius cap.	4 (2)			61 (27)	
L Trapezius	5 (2)			73 (32)	
R Trapezius	5 (2)	81 (29)	5 (9)	73 (32)	

Standard deviation are in parentheses.
L, left; R, right.

the moments exerted about L5 are an order of magnitude larger than those exerted about C4. This must be due in part to the larger cross-sectional areas and moment arms of the muscles of the lumbar region. Mean intraindividual ratios of strengths in maximum attempted extension to maximum attempted flexion for the neck muscles in both males and females were approximately twice those found for the low back (12).

The coefficients of linear correlation between model predicted forces and measured myoelectric activities were weaker than those found earlier for the lumbar trunk. There are a number of reasons why this might have occurred. First, the assump-

TABLE 6. *Mean predicted motion segment reactions in maximal exertions*

Exercise	Lateral shear (N)	Anteroposterior shear (N)	Compression (N)
Relaxed	0 (0)	−2 (1)	122 (36)
Left twist	33 (8)	70 (24)	778 (228)
Extension	0 (0)	135 (69)	1164 (494)
Flexion	0 (0)	31 (63)	558 (375)
Left bending	125 (58)	93 (59)	758 (422)

Standard deviations in parentheses.

TABLE 7. *Coefficients of correlation of predicted muscle forces with measured myoelectric signals*

Muscle equivalent	Paired electrode	Left	Right
Platysma	Anterior	0.60	0.68
Infrahyoids	Anterior	0.83	0.83
SCM	Anterolateral	0.82	0.85
Scalene ant.	Posterolateral	0.77	0.73
Levator scap.	Posterolateral	0.33	0.29
Splenius cap.	Posterior	0.60	0.75
Trapezius	Posterior	0.49	0.41

tion in the regressions of linearity between contraction force magnitude and myoelectric signal level probably is valid only approximately. Nonlinear regressions were not examined since evidence from earlier studies of lumbar trunk muscles suggests that significant nonlinearity seldom arises in force/ activity relations of the kind studied here (17,18). Second, the model assumption that antagonistic co-contractions of muscles are minimal was clearly violated at least sometimes, which can be seen from the Table 4 myoelectric signal data. The assumption was used because it has proven reasonable in at least some similar circumstances (17,18), and validated models incorporating more realistic assumptions in this regard have yet to be established. Third, and probably most important, myoelectric activities were measured by surface electrodes, and so could be affected by crosstalk among signals from different muscles. This may explain in part why the correlation coefficients for the more anterior muscles were better than for the more posterior muscles; there are only a few anterior muscles, but many posterior ones. Fourth, while the biomechanical model makes no distinctions about the mechanical functions of different muscles, they actually may be functionally differentiated. The weakest correlations were for the levator scapulae and trapezius muscles. These muscles may have primary functions other than to development moments about the cervical vertebrae. The correlation coefficients for the other muscles ranged from 0.60 to 0.85, suggesting that the model used in these studies had a reasonable degree of validity. Further research can guide the development of cervical muscle action models with improved validity. Nevertheless, this first model of the quasistatic biomechanics of the neck muscles seems to be able to predict contraction forces adequately, at least in a semiquantitative sense.

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