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ORIGINAL ARTICLE

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Electromyography of superficial cervical muscles with exertion in the sagittal, coronal and oblique planes

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Abstract The purpose of the current study was twofold: (1) to determine the isometric force and electromyographic (EMG) relationship of the sternocleidomastoid, splenii and trapezii muscles bilaterally in graded and maximal exertions in the sagittal, coronal and oblique planes, and (2) to develop regression equations to predict force based on the EMG scores. A newly designed and validated cervical isometric strength testing device was used to measure the cervical muscle isometric strength and force/EMG relationship in cervical flexion, extension, bilateral lateral flexion, bilateral anterolateral flexion, and bilateral posterolateral extension, all beginning with an upright seated neutral posture. A group of 40 healthy subjects were asked to exert their cervical motions in the directions of interest, while the force output and EMG from the sternocleidomastoids, splenii, and trapezii were sampled bilaterally at 1 kHz. ANOVA, correlation, and regression analyses were carried out. The force and EMG scores were significantly different between the directions of effort ($P<0.01$). All regressions were significant ($P<0.01$). All subjects registered the highest forces in pure extension and the lowest in pure flexion, showing a gradual decrease from the posterior to anterior direction. There was a modest correlation between EMG of the investigated muscles and force ($r=0.15–0.76$, $P<0.01$). EMG output was, for example, approximately 66% higher in flexion than in extension (while force output was roughly 30% less in flexion than extension) – thus relatively more muscle activity was required in flexion than extension to generate a given force. The intermediate positions (i.e. anterolateral flexion) revealed force/EMG ratio scores that were intermediate in relation to the force/EMG ratios for pure flexion and pure extension. The cervical muscle strength and cervical muscle EMG are therefore dependent on the direction of effort.

Keywords Neck muscles · Electromyography · Neck pain · Whiplash injuries · Torque

Introduction

Cervical pain and dysfunction have become a significant health and economic burden in our society [1, 10]. The cervical pain syndrome of the whiplash-associated disorders is one of the more frequently disputed conditions in this regard [3, 5, 6]. One of the primary reasons the medicolegal and social dilemma of whiplash persists is that the majority of whiplash victims are categorized by disorders that are largely subjective in nature, with no known diagnostic means available to prove that an acute injury has occurred, or in all but a small percentage to identify objectively an injury or cause for chronic pain. This brings to question the validity of whiplash as a chronic pain syndrome secondary to injury (thus the legal issue of causation). In low velocity impacts, in particular, the likelihood of trauma, let alone significant tissue damage to cause long-lasting symptoms, is highly questioned [4]. Studies of neck movements and tolerances in low velocity collisions are highly relevant to whiplash controversies given that as many as 65% of insurance claims result from low-velocity impacts [4]. For this reason, a number of human volunteer studies with experimental collisions have been conducted to evaluate an impact threshold (most often characterized by the change in velocity of the struck vehicle – delta V) below which acute symptoms (presumably from injury) are not expected and no (even transient) discomfort reported up to a change in velocity (delta V) of 8 km/h [2, 4, 15, 16, 21, 22, 23].

Another approach to the problem, however, and one that also allows for extrapolation to various head positions and directions of vehicle impact, lies in understanding the mechanical properties and behaviour of the tissues and tissue load-bearing capacity. This may be achieved, in part, and as a starting point, by understanding strength-producing capacity and associated electromyographic activity of the cervical muscles. These may ultimately serve as an index of force in experimental collisions, and assist in understanding the relationship between the collision and tissue load.

Ono and Kanno [17] found a sharp rise in neck bending moments at the beginning of the motion due to the inertial forces. In a study in which muscle moment and EMG of cervical erector spinae were investigated in women during submaximal and maximal isometric neck extension, Schuldt [19] found a nonlinear relationship. The ratio between the EMG and torque differed with the neck position ($P<0.01$). Queisser et al. [18] also explored this relationship and found it to be variable between subjects, which they could not reconcile by normalization of EMG. They reported that elimination of gravity, continuous monitoring of cervical motion and motor unit localization were essential for a reliable relationship. Nonlinear relationships in these muscles were also reported by Mayoux-Benhamou and Revel [14]. They reported peak EMG activity in cervical erector spinae in the slightly flexed position. Schuldt and Harms-Ringdahl [20] reported maximal activity in the cervical erector spinae during neck extension, and in the splenius and levator scapulae in lateral flexion.

When human subjects are subjected to anterior acceleration, the muscular strain as measured by integrated EMG (IEMG) increases sharply [9]. Under 7 *g* acceleration the EMG increases 5.9-fold compared to a 1 *g* exposure, and registers 37.9% of maximal voluntary contraction (MVC) EMG. In some individuals, the EMG score reaches 100% of MVC only under 4 *g* acceleration with simultaneous head motion and a twisted head position. In similar acceleration environments a difference in helmet weight also leads to a difference in EMG score, lighter helmets leading to a smaller EMG [8]. In another study eight healthy male subjects were accelerated from the rear [13]. EMG activities of cervical muscles were measured by a combination of inserted (semispinalis capitis, splenius capitis, levator scapulae) and surface electrodes (trapezius and sternocleidomastoids). The authors reported a reaction time of 13.2 ms from head acceleration and 65.6 ms from sled acceleration. From such observation the authors concluded that the muscles could affect the injury pattern. Thus the significance of the role of muscles in prevention/causation of cervical injuries is generally agreed, and further study is warranted.

A review of the literature reveals a lack of studies concerning cervical muscle strength and EMG activity despite the fact that perturbations of the neck in most automobile collisions are likely to be modulated by or injurious to cervical musculature. Information on muscle response to torque application is sparse, even in the sagittal plane, and non-existent in other planes. Therefore, a study was conducted to study the response of three superficial pairs of cervical muscles to an isometric torque in eight different directions.

Materials and methods

Subjects

Healthy adults with no recollection of musculoskeletal problems in the last year, or any known chronic musculoskeletal disorder of the neck were recruited locally by advertisement. Subjects were informed of the purpose and the protocol of the experiment, and signed an informed consent form. The study was approved by the Health Research Ethics Board.

Set-up and data acquisition

The experimental set-up consisted of an adjustable chair, sliding platform, and floor-mounted strength measuring device, as described in a previously [11]. The strength-measuring device consisted of a vertical telescopic 15-cm wide rectangular metal tube welded to a thick iron plate rigidly bolted to the floor (Fig. 1). The 12-cm wide inner tube could be raised or lowered and securely

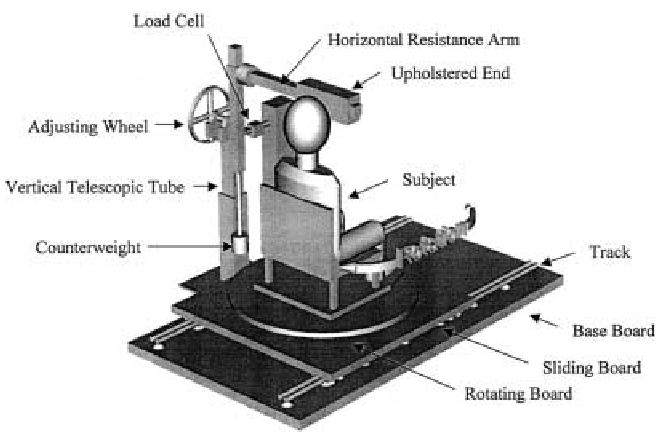

Fig. 1 The strength testing set-up

locked into place. On top of the inner tube a block bearing was mounted to which another hollow tube was attached allowing it to freely rotate. Attached perpendicularly to this tubing was an adjustable arm which was upholstered at the farther end for head contact and force exertion. At the lower end of the tubing a counter weight was attached with an adjustable length rod to compensate for the variable positioning of the horizontal resistance arm. A horizontal metal rod was built 14 cm below the pivot point at right angles to the tubing which could be attached to a fixed object with an intervening load cell (I-250). Thus, any force exerted on the upholstered horizontal arm was registered on the load cell, which was fixed in its mechanical path. The signals from the load cell were input into the force monitor for signal conditioning and processing.

The EMG system consisted of surface electrodes, electrode cables, preamplifiers and amplifiers. Silver-silver chloride surface electrodes of 1-cm diameter with recessed pre-gelled elements (HP 144445) were used with an inter-electrode distance of 2 cm. These electrodes were connected to fully isolated preamplifiers by means of short cables and tip plugs. The low-noise and low-nonlinearity preamplifiers were specially made and had a common mode rejection ratio of 130 dB and a wide bandwidth. These preamplifiers fed to low-power, high-accuracy instrumental amplifiers designed for signal conditioning and amplification. The amplifier system was run off an internal charged battery. The amplifier had AC coupled inputs with single pole RC Butterworth filter with a low cut-off frequency of 8 Hz. The RMS EMG signals were sampled at 1 kHz.

The force output monitor and EMG amplifiers were fed to a Metrabyte DAS 20 A-to-D board. The converted digital signals were stored on the hard disk of a 486 computer with a tape backup for archival purposes.

The strength was measured in units of force (newtons). This was obtained by dividing the product of the load cell reading and its moment arm with the moment arm of the force applied. The

Table 1 Demographic details of subjects

	Age (years)	Height (cm)	Weight (kg)
Male (*n*=21)			
Mean (SD)	24.4 (2.4)	177.8 (4.8)	78.0 (10.8)
Min	19.0	170.0	60.0
Max	29.0	185.0	100.0
Female (*n*=19)			
Mean (SD)	23.9 (2.8)	167.7 (6.1)	60.4 (6.7)
Min	20.0	157.0	52.0
Max	29.0	178.0	71.0

Table 2 Mean normalized peak (*P*) and mean average (*A*) forces (N) generated by male and female subjects in eight directions (*Rt ant lat* right anterolateral flexion, *Rt lat* right lateral flexion, *Rt post lat* right posterolateral extension, *Lt post lat* left posterolateral extension, *Lt lat* left lateral flexion, *Lt ant lat* left anterolateral flexion)

	Direction of exertion															
	Flexion		Rt ant lat		Rt lat		Rt post lat		Extension		Lt post lat		Lt lat		Lt ant lat	
	P	A	P	A	P	A	P	A	P	A	P	A	P	A	P	A
Male																
Mean	72	57	70	54	76	58	84	65	100	79	88	70	76	60	69	54
SD	18	14	23	16	26	19	25	18	28	21	29	22	23	17	22	16
Female																
Mean	41	30	43	32	52	39	64	49	72	56	64	50	54	42	42	32
SD	14	11	14	10	17	14	16	14	20	16	21	16	16	12	11	8

Table 3 Normalized peak EMG of the cervical muscles at maximal effort in eight different directions among male and female subjects (%MVC). Values are means (SD) (*L* left, *R* right, *scm* sternocleidomastoid, *spl* splenius, *trp* trapezius)

		Flexion	Right anterolateral flexion	Right lateral flexion	Right posterolateral extension	Extension	Left posterolateral extension	Left lateral flexion	Left anterolateral flexion
Male	L scm	100 (0)	90 (29)	45 (27)	26 (19)	24 (32)	47 (29)	83 (41)	96 (35)
	R scm	100 (0)	96 (32)	78 (34)	54 (34)	24 (26)	25 (18)	44 (26)	85 (25)
	L spl	58 (36)	36 (17)	36 (28)	48 (42)	76 (56)	90 (42)	100 (0)	84 (26)
	R spl	71 (40)	101 (55)	100 (0)	126 (101)	109 (96)	51 (49)	42 (35)	35 (17)
	L trp	36 (21)	36 (22)	61 (43)	83 (45)	100 (0)	100 (52)	80 (34)	66 (47)
	R trp	40 (26)	70 (72)	69 (34)	98 (53)	100 (0)	62 (33)	53 (39)	36 (30)
Female	L scm	100 (0)	90 (20)	31 (16)	11 (7)	11 (10)	38 (24)	62 (25)	92 (22)
	R scm	100 (0)	267 (770)	68 (20)	47 (22)	39 (98)	90 (335)	26 (24)	324 (1085)
	L spl	39 (15)	21 (7)	14 (10)	22 (11)	54 (27)	92 (24)	100 (0)	81 (41)
	R spl	50 (42)	86 (48)	100 (0)	114 (59)	71 (56)	27 (21)	17 (11)	29 (24)
	L trp	24 (14)	26 (9)	42 (26)	68 (19)	100 (0)	93 (28)	65 (29)	51 (20)
	R trp	27 (15)	53 (18)	64 (25)	89 (25)	100 (0)	70 (28)	40 (23)	28 (15)

Table 4 The normalized EMG area in percent-time product (%-s) of the cervical muscles in a progressive maximal effort in eight different directions among male and female subjects. Values are means (SD) (*L* left, *R* right, *scm* sternocleidomastoid, *spl* splenius, *trp* trapezius)

		Flexion	Right anterolateral flexion	Right lateral flexion	Right posterolateral extension	Extension	Left posterolateral extension	Left lateral flexion	Left anterolateral flexion
Male	L scm	100 (0)	90 (33)	35 (22)	17 (12)	18 (25)	38 (25)	75 (40)	91 (35)
	R scm	100 (0)	89 (26)	67 (27)	40 (25)	18 (20)	15 (10)	30 (17)	76 (26)
	L spl	58 (24)	39 (12)	30 (16)	39 (33)	73 (51)	88 (38)	100 (0)	84 (27)
	R spl	67 (31)	93 (32)	100 (0)	100 (40)	87 (67)	46 (44)	34 (24)	37 (17)
	L trp	40 (23)	41 (23)	52 (24)	70 (30)	100 (0)	94 (31)	72 (25)	59 (28)
	R trp	42 (20)	60 (32)	69 (24)	93 (34)	100 (0)	64 (28)	49 (26)	40 (24)
Female	L scm	100 (0)	82 (13)	21 (13)	9 (4)	9 (5)	29 (16)	53 (21)	78 (20)
	R scm	100 (0)	176 (383)	62 (23)	37 (19)	22 (45)	17 (31)	18 (22)	168 (436)
	L spl	41 (15)	24 (8)	17 (7)	21 (9)	52 (24)	95 (26)	100 (0)	75 (33)
	R spl	51 (29)	88 (40)	100 (0)	114 (58)	59 (31)	27 (16)	19 (10)	29 (21)
	L trp	30 (13)	31 (13)	37 (14)	60 (13)	100 (0)	100 (38)	63 (34)	45 (15)
	R trp	32 (13)	48 (14)	64 (24)	95 (34)	100 (0)	68 (27)	39 (23)	30 (14)

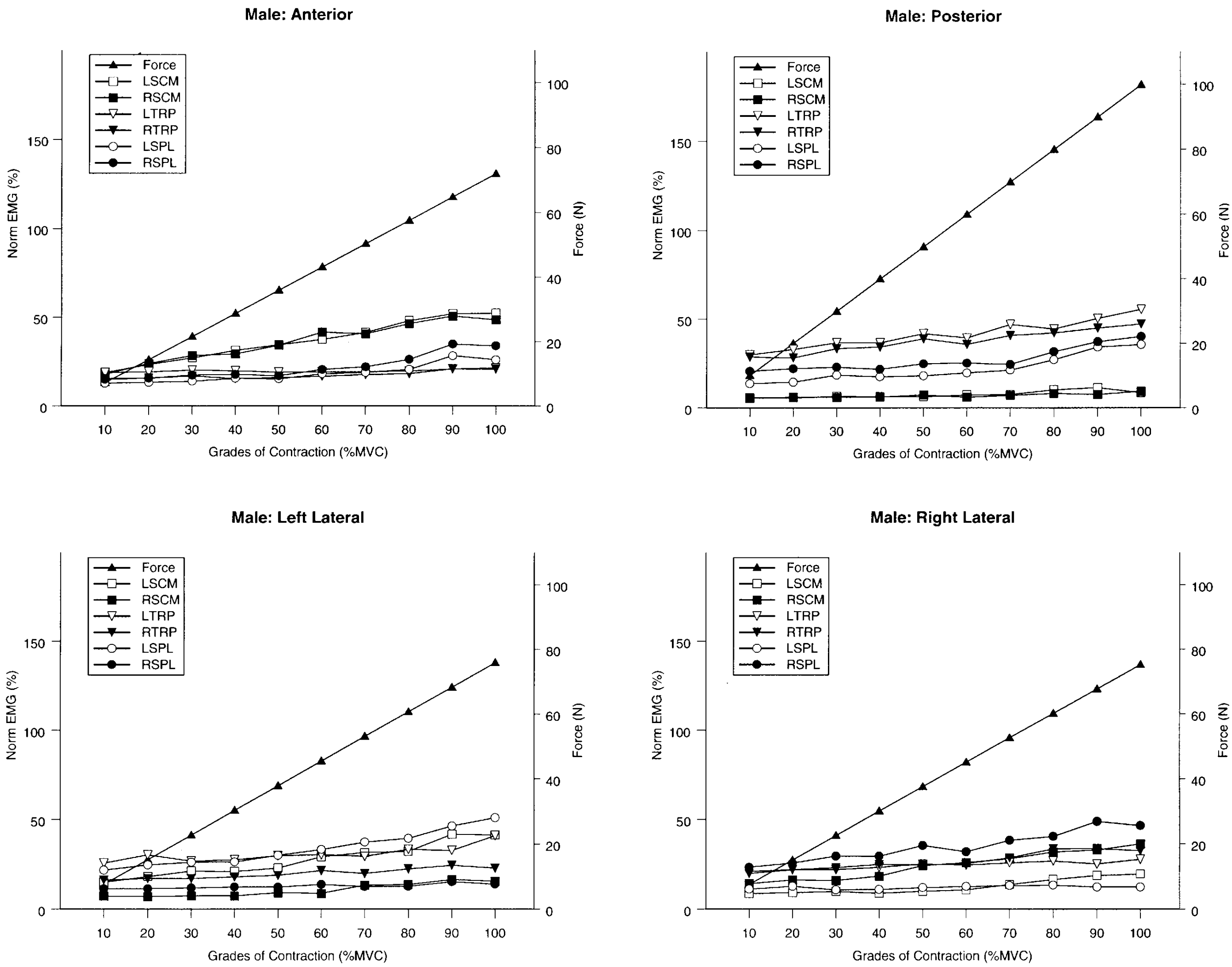

Fig. 2 Calibrated force and EMG output of all investigated muscles in flexion, extension and lateral flexion at 10% force levels in male sample

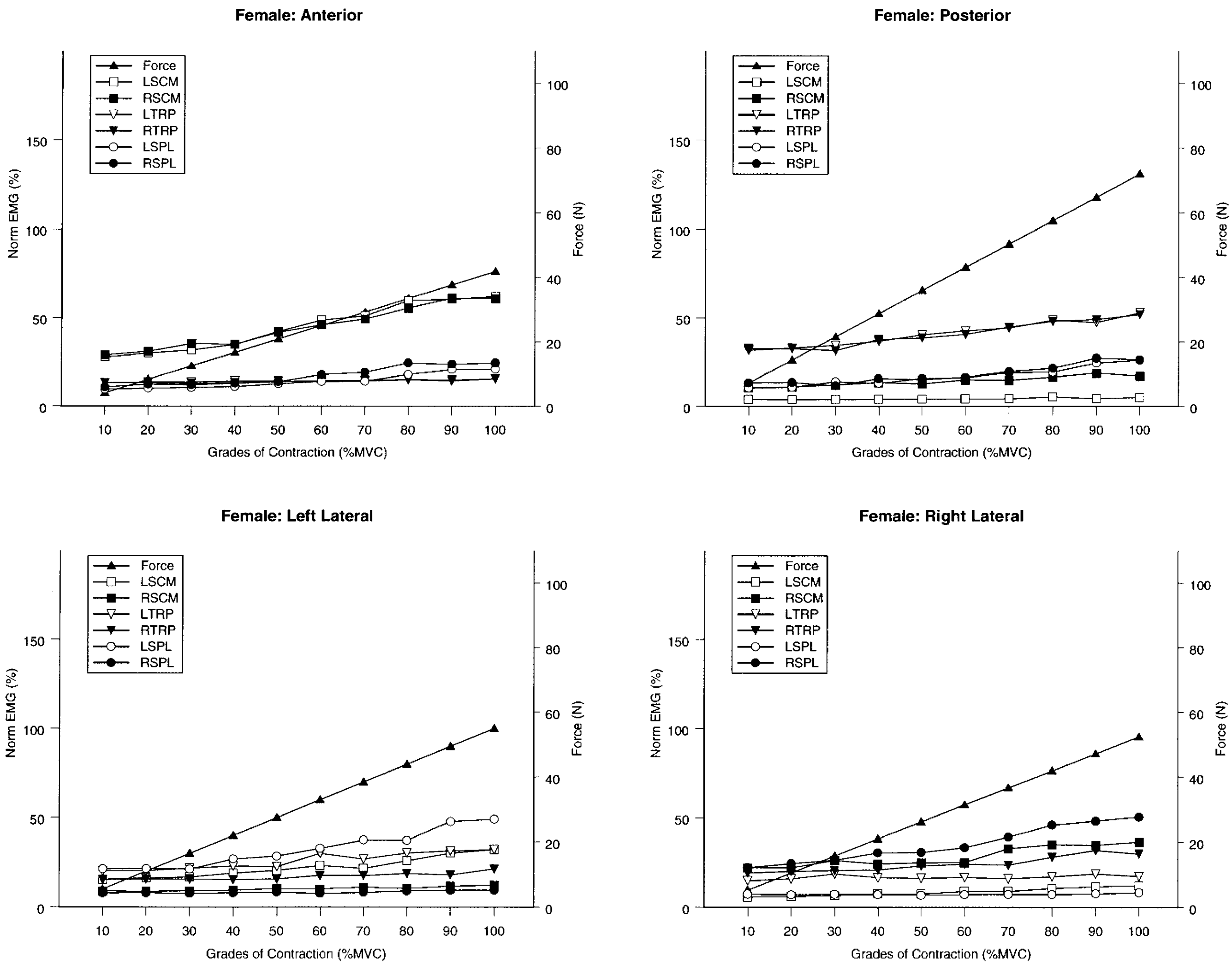

Fig. 3 Calibrated force and EMG output of all investigated muscles in flexion, extension and lateral flexion at 10% force levels in female sample

peak and average magnitude of the EMG were measured along with the EMG area. At the end of 5 s of exertion and recording the computer beeped, signalling the end of the recording period, and the subject then relaxed. All data channels were sampled at 1 kHz. Thus, collected data were converted to newtons and microvolts as appropriate. Peak and average strength and peak and average EMG along with EMG area were obtained from these sets and subjected to quantitative comparison and statistical analyses.

Test protocol

Age, weight and height were recorded for each subject, and each was then seated on the chair and stabilized in a neutral spinal posture. At this stage the height of the C7/T1 disc was measured from the floor. Using an adjusting wheel, a pivot point of the resistance arm was set at the same height. A horizontal upholstered bar was slid on the vertical portion of the resistance arm and adjusted to the subject's height. Subsequently the counterweight position was adjusted to balance the weight of the top portion of the resistance arm. On a day before the testing day, the protocol of MVC was explained to the subjects and they were given practice until they produced three consistent trials for MVC in flexion, extension, lateral flexion and intermediate planes. On the day of the experiment after placement on the experimental set-up each subject was given a warm-up exercise, which consisted of deep, but gentle cervical flexion, hyperextension, lateral flexion to the left and right, and finally right and left circumduction for a minimum period of 5 min. The exercises were repeated five times until the subject felt entirely comfortable with the task before the EMG and force recording began.

The order of directions measured for each subject was carried out in a computer-generated random sequence. After the upholstered resistance arm of the testing device had been placed in direct contact with the head of the subject in the relevant test position, subjects were asked to exert their maximum effort gradually building over a period of 5 s, terminating the effort at the auditory signal generated by the computer at the end of the sampling period. They were asked to concentrate on using their neck only, and were told to raise their feet from the footrest to minimize leverage from the lower extremities. The subjects were informed initially that in order to meet the objective of the project it was essential they reach their maximal effort unless any symptoms or pain arose, at which time they were instructed to stop immediately and inform the investigator. No encouragement was issued to them while they were exerting force.

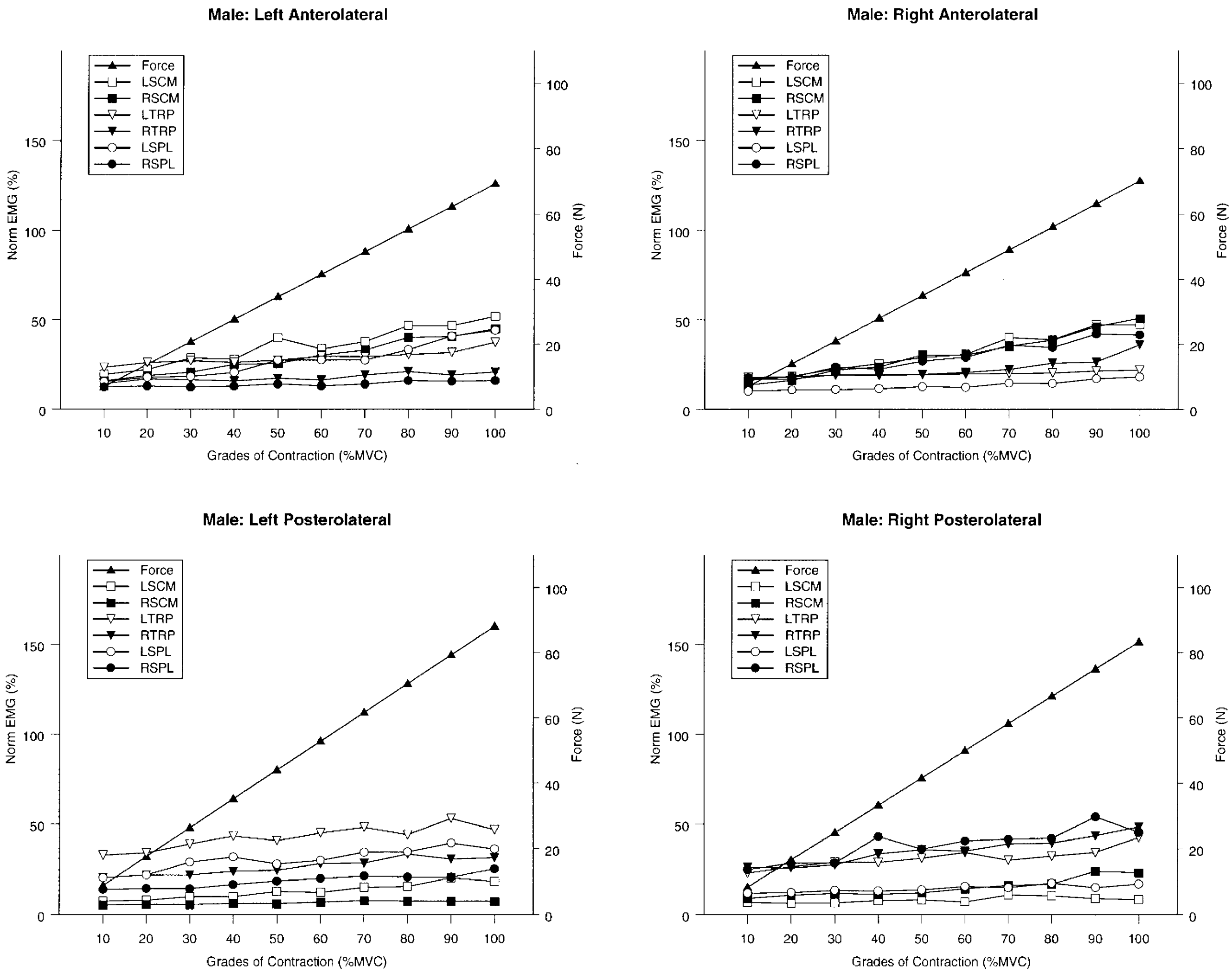

Fig. 4 Calibrated force and EMG output of all investigated muscle in right and left anterolateral flexion and posterolateral extension at 10% force levels in male sample

The cervical strength and the cervical muscle EMGs were measured in the sagittal, coronal, and intermediate planes resulting in eight test positions. The subjects were allowed a minimum of 2 min rest between any two tests. Only one trial in each position was done.

Data analysis

Data were loaded into the hard drive of the computer for windowing of the relevant section of the data stream. As soon as the torque began to rise from the baseline the computer software automatically drew a vertical marking the start of the activity. Taking the force value at maximal voluntary contraction as 100% the computer drew a vertical marker at every 10% of the torque interval. An 11-point smoothing routine repeated twice was used to smooth EMG signals for reliable interpretation of the data. Such smoothed EMG traces were subjected to normalization. The EMG score of the trapezius in extension, the sternocleidomastoids in flexion, and the splenii in their respective lateral flexion were chosen as 100% against which the score of corresponding muscles in eight directional efforts were appropriately normalized. For determination of EMG values corresponding to these 10% torque intervals, each EMG channel was plotted with the torque trace individually. The software performed the linear envelope detection of the RMS EMG (full-wave rectified) channel from which the segmental normalized EMG value as well as normalized peak and average EMG, and the EMG area were derived.

A statistical analysis of these torque and EMG data was carried out using SPSS to calculate descriptive statistics, ANOVA, and the correlation between torque and EMG. A regression analysis was also carried out to determine the predictability of the torque from EMG measurements.

Results

A total of 40 subjects (21 males and 19 females) were recruited. The demographic details of the subjects are provided in Table 1. The mean normalized peak and average forces in various directions of exertion are tabulated in Table 2. The mean peak EMG magnitudes and the mean EMG areas for the different muscles and each gender are

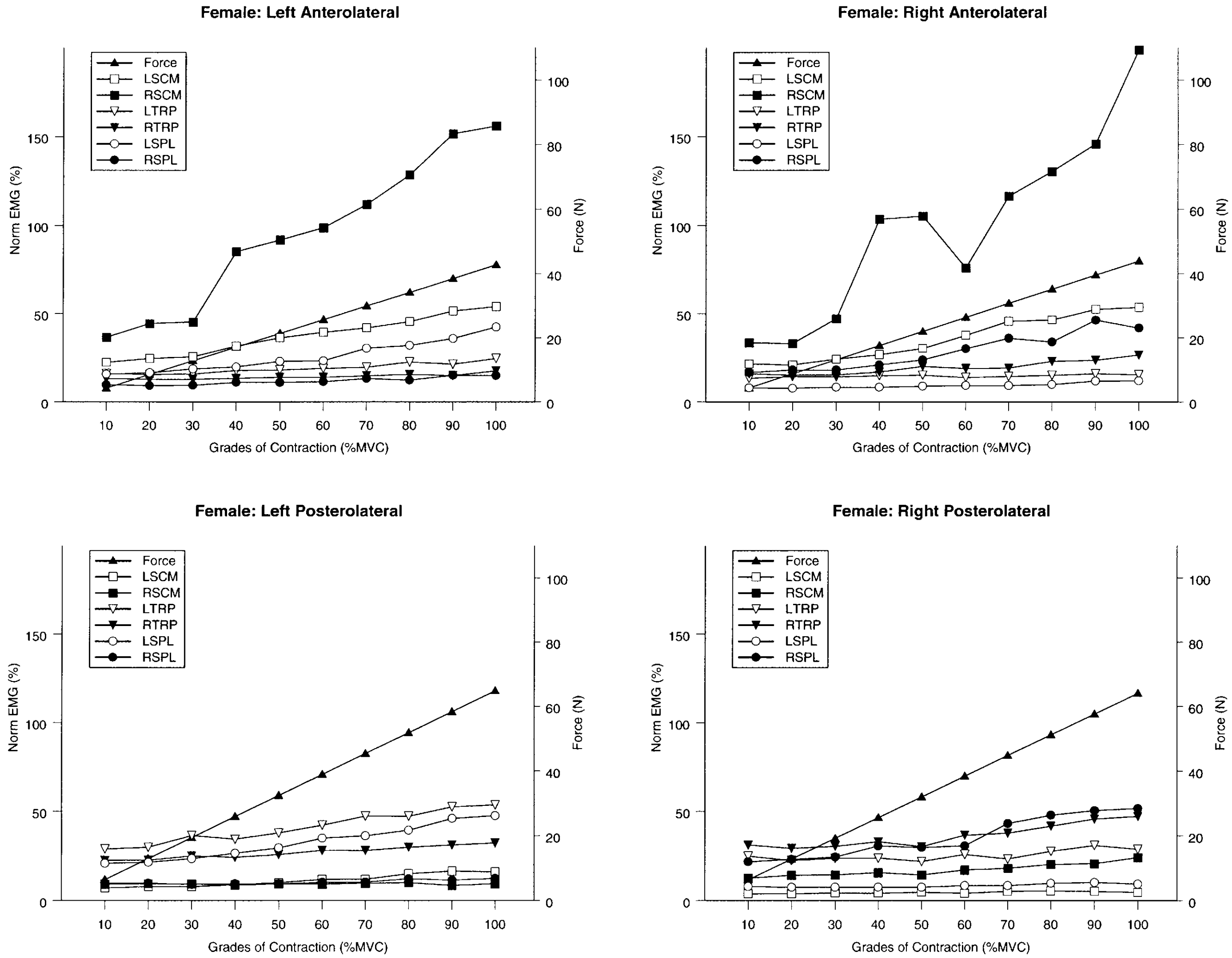

Fig. 5 Calibrated force and EMG output of all investigated muscle in right and left anterolateral flexion and posterolateral extension at 10% force levels in female sample

shown in Tables 3 and 4, respectively. Figures 2, 3, 4, and 5 show the calibrated force and EMG output of the investigated muscles in the various exertion directions for both genders, at 10% force levels.

Certain findings are noteworthy. In pure forward flexion, the maximal EMG output (for peak, average and area) was from the sternocleidomastoids, each muscle contributing equally to the effort. Similarly, though very small in magnitude, the EMG output of the left and right splenius and the trapezius were also symmetric. The average EMG output and the total area were similarly proportioned among these muscles. In the pure posterior extension, the peak and average forces generated by both males and females were greater than in the pure forward flexion effort.

The raw EMG scores of the right and left sternocleidomastoids, splenii and the trapezii were plotted at the ten percentile grade levels for males and females. In flexion there was a continuous rise in the magnitude of sternocleidomastoid score with increase in force. In extension the level of EMG remained low from beginning to end, rising very little. In lateral flexion the ipsilateral sternocleidomastoids and splenius scores rose in parallel. In anterolateral flexion, the force output of the sternocleidomastoids again rose steadily to a maximum level. In posterolateral extensions the ipsilateral splenius was clearly most active and contributed more than the others. The contributions of the ipsilateral sternocleidomastoids and the trapezii were greater than the contribution from their respective contralateral parts.

An analysis of variance of EMG parameters (peak and average EMG, and EMG area) revealed the same findings. All EMG parameters were affected by three main factors (Table 5). Peak EMG, for example, was significantly affected by gender of the subject, the muscle in question, and the direction of exertion. There were significant two-way interactions between gender and muscle and muscle and the direction of exertion. There was no

Table 5 ANOVA summary for peak EMG (*NS* not significant)

Source	df	*F*-value	*P*-value
Gender	1	77.379	0.001
Muscle	5	105.381	0.001
Direction	7	13.683	0.001
Gender×muscle	5	5.526	0.001
Gender×direction	7	0.510	NS
Muscle×direction	35	20.042	0.001
Gender×muscle×direction	35	0.125	NS

Table 6 Scheffé post-hoc multiple comparisons of EMG scores of different muscles (*a* peak, *b* average, *c* area; $P<0.05$) (*L* left, *R* right, *scm* sternocleidomastoid, *spl* splenius, *trp* trapezius)

	L scm	L spl	L trp	R scm	R spl	R trp
L scm	–	a,b,c	a,b,c	–	a,b,c	a,b,c
L spl	–	–	a,b,c	a,b,c	–	a,b,c
L trp	–	–	–	a,b,c	a,b,c	–
R scm	–	–	–	–	a,b,c	a,b,c
R spl	–	–	–	–	–	a,b,c
R trp	–	–	–	–	–	–

Table 7 Scheffé post-hoc multiple comparisons of EMG scores in different directions of activity (*a* peak, *b* average, *c* area; $P<0.05$) (*RAL* right anterolateral, *RL* right lateral, *RPL* right posterolateral, *LPL* left posterolateral, *LL* left lateral, *LAL* left anterolateral)

	Ant	RAL	RL	RPL	Post	LPL	LL	LAL
Ant	–	–	b,c	a,b,c	a,b,c	a,b,c	a,b,c	–
RAL	–	–	b,c	a,b,c	a,b,c	a,b,c	a,b,c	–
RL	–	–	–	–	–	–	–	–
RPL	–	–	–	–	–	–	–	a,b,c
Post	–	–	–	–	–	–	–	a,b,c
LPL	–	–	–	–	–	–	–	a,b,c
LL	–	–	–	–	–	–	–	b,c
LAL	–	–	–	–	–	–	–	–

significant interaction between gender and direction. The three-way interaction between gender, muscle and direction of exertion was also not significant. A Scheffé post-hoc multiple comparison of EMG scores (peak, average and EMG area) of different muscles and due to exertion in different directions revealed that all EMG parameters of all muscles were significantly different from the other muscles except the left sternocleidomastoids and splenius, which were not significantly different from their right counterparts (Table 6). The EMG parameters in different directional efforts in post-hoc comparison revealed that flexion and extension were not significantly different from posterolateral extensions. Furthermore, left and right lateral flexions were neither significantly different from each other nor were they significantly different from the posterolateral extensions (Table 7).

The correlation analysis (Table 8) showed a strong correlation between peak EMG and average EMG (r=0.95 to 0.99, P<0.01), peak EMG and EMG area (r=0.95 to 0.97, P<0.01), and average EMG and EMG area (r=0.99, P< 0.01). Similarly, peak and average force and force were significantly correlated with each other (r=0.94 to 0.98, P<0.01). EMG parameters of all muscles at their ten percentile contraction grade intervals were correlated with similar parameters of other muscles. These correlations ranged between a low of r=0.38 to a high of r=0.87 (P< 0.01). The EMG of each of the muscles separately demonstrated low to modest significant correlations between the two genders. A stepwise forward regression to predict force generated in different directions based on the EMG scores were all significant (P<0.01) but accounted for the variances ranging between a low of 22% to a modest value of 78% in males for peak values (Table 9). A similar result was obtained for the average force value with variances from 31% to 82% accounted for among males. In females, similar regressions were obtained with variances up to 84% explained.

Discussion

The exact mechanism of whiplash injury and the exact site affected has eluded research to date. Though some studies are reported to have identified the elusive whiplash injury, problems with patient type have made such conclusions as yet untenable [7, 12]. Given a lack of an objective test for the injury, experimental collision data are the most effective current method for assessing the likelihood of injury from a given collision. Delta V studies are a particular aid to the problem. Yet, this is only one objective data set. In an arena where much of the controversy is fed by the subjective nature of the patient's problem, further objectivity is greatly desired. If the human strength production capacities in various postural configurations in low-velocity impacts and the corresponding EMG could be determined, we may be able to extend the value for experimental collisions based on calibrated EMG scores by extrapolating the maximum force borne by the tissues before the possibility of damage.

Our study revealed that human cervical strength-producing capacity is directionally dependent. The human neck is weakest in flexion and strongest in extension. The strength-producing capability increased progressively as the deviation in direction of effort moved from anterior to the posterior direction on either side. However, the total EMG output demonstrated exactly the opposite pattern with maximal EMG in flexion, and least in extension, decrementally increasing with deviation from flexion to extension on either side. Using extension force in males as 100% of the force, the decline in force in flexion was 30%. The EMG output, on the other hand, increased by 66% from extension to flexion. Thus the muscle effort per

Table 8 Correlation between EMG and force parameters among male and female subjects (*L* left, *R* right, *scm* sternocleidomastoid, *spl* splenius, *trp* trapezius)

Muscle	EMG variable	Male			Female		
		Peak force	Average force	Force area	Peak force	Average force	Force area
L scm	Peak	0.153*	0.163*	–	–0.169*	–0.199*	–0.207*
	Average	–	–	–	–0.173*	–0.203*	–0.215**
	Area	–	–	–	–0.165*	–0.193*	–0.193*
R scm	Peak	0.226**	0.217**	0.172*	–	–	–
	Average	0.218**	0.213**	0.167*	–	0.170*	–0.178*
	Area	0.214**	0.210**	0.171*	–	–0.162*	0.160*
L spl	Peak	0.256**	0.286**	0.241**	0.350**	0.366**	0.345**
	Average	0.307**	0.343**	0.288**	0.344**	0.355**	0.332**
	Area	0.305**	0.343**	0.296**	0.363**	0.376**	0.363**
R spl	Peak	0.303**	0.299**	0.267**	0.358**	0.359**	0.361**
	Average	0.370**	0.369**	0.331**	0.375**	0.370**	0.372**
	Area	0.367**	0.368**	0.337**	0.378**	0.375**	0.383**
L trp	Peak	0.542**	0.544**	0.500**	0.748**	0.756**	0.737**
	Average	0.580**	0.568**	0.517**	0.762**	0.773**	0.755**
	Area	0.581**	0.575**	0.532**	0.765**	0.776**	0.771**
R trp	Peak	0.500**	0.516**	0.487**	0.701**	0.709**	0.696**
	Average	0.561**	0.563**	0.529**	0.716**	0.725**	0.709**
	Area	0.556**	0.562**	0.538**	0.719**	0.729**	0.724**

*P<0.05, **P<0.01

Table 9 Prediction of peak force using step-wise forward regression in different directions (model summary) (*Rt ant lat* right anterolateral flexion, *Rt lat* right lateral flexion, *Rt post lat* right posterolateral extension, *Lt post lat* left posterolateral extension, *Lt lat* left lateral flexion, *Lt ant lat* left anterolateral flexion) (*Predictors: a* constant+left trapezii×beta a_1, *b* constant+right sternocleidomastoids×beta b_1, *c* constant+right trapezii×beta c_1, *d* constant+right sternocleidomastoids×beta d_1+left splenii×beta d_2, *e* constant+left sternocleidomastoids×beta e_1, *f* constant+right sternocleidomastoids×beta f_1+right splenii×beta f_2, *g* constant+left trapezii×beta g_1+right sternocleidomastoids×beta g_2, *h* constant+left trapezii×beta h_1+right sternocleidomastoids×beta h_2+right trapezii×beta h_3)

	Direction	Predictors	df	*F*-value	*P*-value	R^2 (adjusted)	SE of estimate
Male	Flexion	a	1	6.675	0.018	0.221	16.5821
	Rt ant lat	b	1	8.968	0.007	0.285	20.1098
	Rt lat	c	1	10.666	0.004	0.326	21.978
	Rt post lat	d	2	15.526	0.001	0.592	16.5223
	Extension	a	1	7.887	0.001	0.256	24.7013
	Lt post lat	e	1	8.861	0.008	0.563	14.1764
	Lt lat	a	1	11.041	0.004	0.633	9.8628
	Lt ant lat	c	1	9.563	0.006	0.786	5.2394
Female	Flexion	a	1	12.915	0.002	0.398	10.8987
	Rt ant lat	c	1	97.237	0.001	0.842	5.893
	Rt lat	c	1	23.295	0.001	0.553	12.0093
	Rt post lat	f	1	19.791	0.001	0.676	9.4798
	Extension	a	2	28.459	0.001	0.604	12.9654
	Lt post lat	a	1	24.170	0.001	0.563	14.1764
	Lt lat	g	2	16.553	0.001	0.633	9.8628
	Lt ant lat	h	3	22.999	0.001	0.786	5.2394

newton force in flexion increased by 100%. The increase in muscle effort per newton progressively increased from extension to flexion on both right and left sides. It would appear that the force production capacity of the neck maybe modulated by the properties of the sternocleidomastoids. These muscles, though inserted anteriorly on the clavicle, arise from the posterior portion of the skull, which is located behind the centre mass of the head. Such an arrangement provides little to no mechanical advantage, reducing the strength-production capacity, and increasing the amount of muscle tension required to generate force. Perhaps it is due to this mechanical arrangement that in flexion the trapezii show little activity (5% to 6%) as they would easily stabilize the neck against the tension developed in the sternocleidomastoids and the splenii due to their size and mechanical advantage. Antagonism and stabilization is largely offered by the two splenii with an EMG output ranging between 20% and 25%.

From the results presented in this study it becomes feasible to speculate, for example, that rear-end collisions potentiate the injuries to the sternocleidomastoids and or supporting connective tissues. This is the mode of loading where the cervical region is most vulnerable due to the least strength-producing capacity. Experimental collision studies indeed reveal that the threshold (in terms of delta V) at which acute symptoms are reported is roughly half the delta V for acute symptoms in lateral or frontal collisions.

While a possible relationship between mechanical forces and injury precipitation has been argued, the correlation data presented show a low to moderate correlation between EMG and the forces produced. Furthermore, an inverse relationship between the force generated and EMG output make it difficult to point to a simple relationship. The step-wise forward regressions generally predicted between 20% and 25% variance explained. Only in a few cases were variances between 60% and 80% explained. This leaves between 20% and 80% of variances unexplained. Therefore, it may appear that there may be other variables that may modulate strength production. It is worth pointing out that the current study investigated only three pairs of superficial cervical muscles. Other superficial as well as deeper cervical muscles may also make significant contributions to neck strength generation, and neck stabilization. Due to the lack of this information, the current study presents only a trend of cervical muscle behaviour and possible mechanics rather than conclusive evidence.

In conclusion, the direction and muscle dependence in EMG output as shown in this study clearly supports the contention that the mode as well as site of injury in an automobile collision will be dependent on the direction of impact, if the force generated by the impact exceeds the threshold of the tissue tolerance. Thus it may be likely that exceeding the tissue tolerance (which is within a fairly narrow range) is essential for the injury precipitation. More study is needed.

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