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COMPENSATORY MUSCLE ACTIVITY FOR SITTING POSTURE DURING UPPER EXTREMITY TASK PERFORMANCE IN PARAPLEGIC PERSONS

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ABSTRACT. Compensation for the loss of postural activity of the erector spinae (ES) muscle in spinal cord injured (SCI) subjects was investigated. All SCI subjects had clinically complete lesions below the T3 level. Body disbalance was invoked by requiring sitting subjects to execute reaching movements over individually predetermined distances in a horizontal plane. Myoelectric activity of the latissimus dorsi (LD) muscle and the trapezius pars ascendens (TPA) muscle both in the SCI subjects and in controls was recorded. The body disbalance was measured in terms of changes in the position of the body centre of gravity. The effects of anticipation for body displacement were examined by cuing the direction of the reaching movement. Our results indicate that paraplegic subjects use both LD and TPA to stabilize their sitting posture, in contrast to non-disabled persons. Secondly, the movement anticipation is in general slower in persons with paraplegia. Furthermore, the paraplegic subjects showed considerable impairments concerning the processing of precued information prior to a goal-directed upper extremity movement.

Key words: electromyography, motor preparation, paraplegia, posture, reaction time, sitting.

Biomechanical and (neuro-)physiological phenomena concerning postural abilities of the non-motor impaired have been studied extensively and systematically in recent years (1, 2, 3, 4, 5, 6, 7, 8, 9, 12, 14, 16, 17, 19, 26, 27, 28, 29, 34). Knowledge on impairments in sitting posture and other sensori-motor functions, however, is largely dominated by the acquired clinical expertise of clinicians and paramedical staff in medical rehabilitation institutes. A functional approach to sitting, relating a specific sitting posture to an actual task or task performance, has been formulated by Corlett et al. (10). Dynamic control of sitting posture prerequisites a time-dependent sensory adaptation based on exteroceptive, proprioceptive and enteroceptive information and the capacity to implement this information into the motor system in order to cope with variable task-related needs and demands. Complex sensory and motor impairments, as seen in, for example, paraplegia, limit optimal sensori-motor control and task performance. This may consequently evoke compensatory mechanisms of motor control.

In the present study mechanisms of compensatory postural motor control during task performance in spinal cord injured (SCI) persons were investigated. Motor behaviour and task performance of these SCI persons can be described either in terms of impairments and disabilities, or, alternatively, in terms of compensation and residual functions. Paraplegia concerns a relatively well definable and more or less stable pathology. Hence straightforward inferences can be made concerning the loss of function due to total or partial impaired spinal cord efference and afference.

The aim of the present study is twofold. The first major question concerning sitting paraplegic persons is how the partial loss of postural activity of the erector spinae (ES) muscle is compensated. In medical rehabilitation practice it is observed that paraplegic subjects with complete spinal cord lesions located below segment T2 partially compensate the loss of postural ES activity by increased activity of the latissimus dorsi (LD) muscle and the trapezius pars ascendens (TPA) muscle. Both the cervical innervation and the anatomical location of the LD and the TPA are favourable in support of these clinical observations. In non-paraplegic subjects neither the LD nor the TPA is thought to participate in stabilising the spine while maintaining an upright sitting posture.

The second major question concerns the extent to which motor programming in medically rehabilitated paraplegic persons still is impaired. Movement anticipation capacities of SCI persons in relation to postur-

Table I. *The levels of complete spinal cord lesions*

	Level							
	T4	T5	T6	T7	T8	T10	T11	T12
Number of subjects	2	1	1	1	1	1	1	7

al muscle activity are not well known. In non-motor impaired subjects the movement precuing paradigm has often been used to study movement anticipation (11, 22, 23, 25). By using precued information relevant to a subsequent motor action a subject may reduce motor preparation time (i.e. reaction time) in an attempt to increase performance. It is hypothesized that paraplegic persons, although extensively retrained, may still suffer from impairments concerning motor preparation prior to selective, goal-directed upper extremity movements. This may consequently lead to decreased task performance, i.e longer reaction times, in comparison to non-disabled persons.

In this study a reaching task similar to the one described by Chari & Kirby (8) was used in a precue choice reaction time task paradigm (23). Precuing a movement was done by presenting visual information which was linked to the possible location of a motor response, i.e. a reaching movement in one out of six predetermined directions. Restoring and keeping sitting balance, a major function of the ES, and task performance were the prime aspects of interest.

METHODS

Subjects

Eleven male and four female SCI subjects whose age ranged from 14 to 57 years (mean = 34.3; SD = 11.8) participated in this experiment. All subjects disengaged from an active medical rehabilitation process at least 10 months earlier (10 months–15 years). The levels of spinal cord injury are given in Table I. Neither secondary pathology (e.g. pressure ulcers) nor upper extremity impairments were present. All participants were thought to have an individual consistent sitting pattern, acquired both during and after the medical rehabilitation period. All but two participants were right-handed.

A second group consisted of 15 healthy right-handed non-disabled subjects (3 males, 12 females) 19–36 years of age (mean = 22.3; SD = 4.9).

Task and apparatus

The subjects were seated in a multi-adaptable chair, either exactly simulating their own wheelchair (as regards the SCI persons) or simulating a normal office chair (5 deg. tilted, 7 deg. reclined). Arm rests were either very low or removed. On a table in front of them one central push button (*X*) and 6 peripheral push buttons (*Y*) in a semi-circle were mounted at either 30%, 60% or 90% of their individual maximal reaching distance, as shown in Fig. 1. Table-leaf was at elbow height. Button diameter was 38 mm. Positive contact force was about 2 N. Information relevant to the task was presented centrally on a standard IBM monochrome monitor (diameter 33 cm) at a distance of approximately 1.2 m.

Six squares that spatially corresponded to the six *Y*-buttons were displayed. In the precue condition three "—" signs appeared in either the three right or the three left squares during the warning signal (WS), pre-informing the subject on the location of the imperative stimulus (ISt). In the non-precue condition "—" signs appeared in all six squares. The WS was displayed for 1 s. The ISt (i.e. one "■" sign displayed in one of the 6 squares) immediately followed the WS. The WS was one of three alternatives as shown in Fig. 2 *a–c*.

The subject had to start each trial by pressing (and remain pressing) the centrally placed button (*X*) with both hands. After 3 s the WS was displayed. The ISt marked the point in time at which the subject had to release the centrally placed button (*X*) and move either his left or his right hand as fast as possible to the *Y*-button corresponding to the location of the ISt on the monitor. Subsequently he had to move back again as fast as possible in order to press button *X* again with both hands. Subjects were instructed not to lean on the push buttons while operating them. Intertrial interval (ITI) was at least 3 s and could be continued indefinitely by not pressing the *X*-button after having reached towards a *Y*-button.

Three series, named A, B and C, in which reaching distance was either 30%, 60% or 90% of the subjects individual maximal reach in all six directions (determined beforehand), were presented in random order. Each series consisted of 120 trials randomly divided between two precue conditions (information during the WS or not) and six reaching directions, giving 12 possible reaching conditions to be executed 10 times each. Performance errors were detected and an "ERROR" message was displayed for 1 s. Written instructions as to the task and the reaching movement were given at the start of the experiment. Subjects were given 12 training trials in advance in order to become familiarized with the task. Each task series took about 16 min. Time between task series was at least 4 min.

Bipolar EMG lead-off positions on the LD and the TPA were determined by using low frequency electro-stimulation (0.5 Hz, 2 ms block-impulse) of minimal intensity (3–5 mA). After light abrasion and cleaning with alcohol Ag–AgCl surface electrodes (2 mm diameter) (Beckmann) were attached. Electrode distance was 30 mm. A reference electrode was attached at the lower anterior part of the tibia. EMG signals were amplified, filtered (time constant 65 ms), negatively rectified and subsequently A/D converted and stored on hard disc using an AT&T 6300-AT (AT&T, Massachusetts) and a LABMASTER data acquisition board (Scientific Solutions Inc. Solon, Ohio). Sample frequency was 200 Hz. Sample time was 5 s, starting simultaneously with the WS. Signals were analysed off-line. After each series an EMG-calibration signal (20 V sinusoidal, 40 Hz) was recorded using a WAVETEK 187 generator (Wavetek, San Diego Inc, San Diego, California). Additionally the projection of the centre of gravity on the seat was recorded in order to quantify dynamic

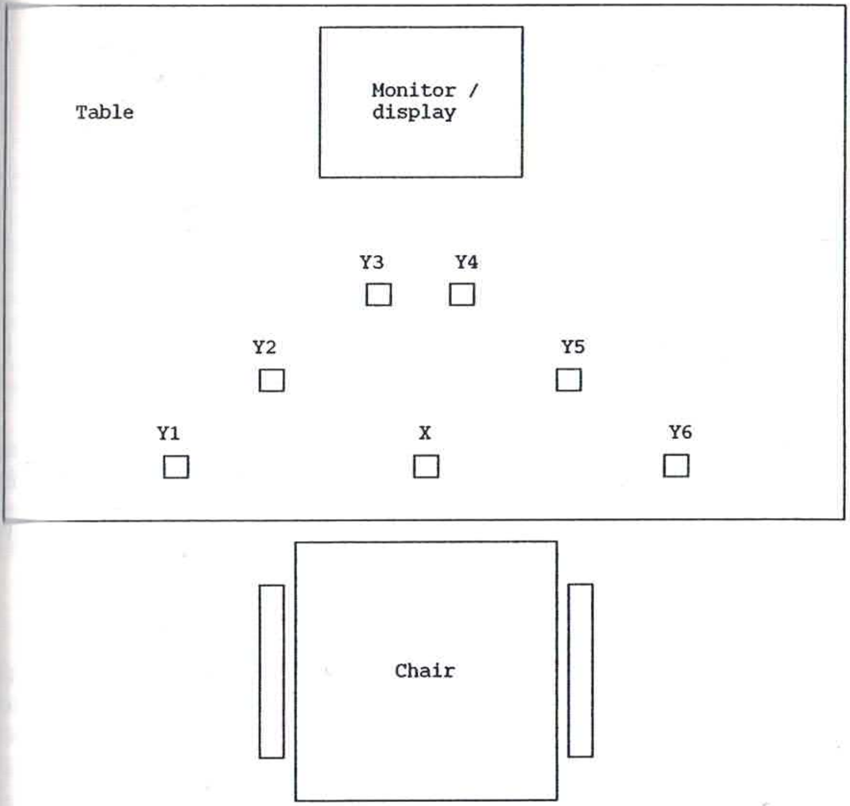

Fig. 1. Arrangement of the experiment. X=central button. Y=peripheral button.

sitting-(dis)balance. Four dynamic force transducers were mounted at four corners underneath the seat. Signals from these force transducers were also A/D converted and stored on disc similar to the EMG-signals. Two-dimensional pelvic movement relatively to the seat was recorded using a GRAFBAR GP-7 sonic digitizer (Science Accessories Co., Southport, Connecticut) placed behind the subject. A sonic pulse generator was attached to the sacrum using adhesive tape. Task presentation as well as the recording of reaction time (RT) (i.e. time between the onset of the ISt and the release of button X), movement time (MT) (i.e. time to reach button Y), task condition, possible task errors, pelvic coordinates (PC) and trial number were executed on a second AT&T 6300-AT computer using a separate timer board and a 8255 VIA interfacing with other peripheral devices.

Data analysis

All individual trials were selected according to (*a*) subject group, (*b*) reaching distance, (*c*) reaching direction and (*d*) cue information. Trials in which task errors (i.e. wrong button press) or signal registration errors had occurred were removed from further analysis as were trials with RT <125 ms or MT <20 ms. Resultant EMG and force transducer signals, individually offset corrected with respect to the initial 200 ms epoch of the WS, were computed per subject within a time-range of (MT$-$600 ms) to (MT$+$1 000 ms). This was the time-range in which a Y-button was pressed. EMG signals were converted to µV. Force transducer signals were converted to a so-called arbitrary force unit (AFU) (AFU$=$F$\times$C). Similar overall group signals were computed per group.

Per condition mean changes in EMG-values within a time-range of (MT$+$0 ms) to (MT$+$200 ms) were computed, i.e. mean EMG increment per second. In the above mentioned time-range maximal disbalance was expected due to the reaching movement and the subsequent return movement. This was consistent with the changes in force transducer signals recorded simultaneously as can be seen in the two examples given in Figs. 3 and 4.

Per reaching distance two-way analyses of variance (ANOVAs) with unequal cell numbers and with group and reaching direction as factors were conducted on these mean changes in EMG values. RT data were selected according to reaching series, group and cue information. Two-way ANOVAs with unequal cell numbers (factors: group and cue information) were used per task series.

RESULTS

EMG and force transducer data of 4 SCI and 2 non-SCI subjects were discarded due to errors in recording the calibration signals. Results of a task-error analysis are given in Table II.

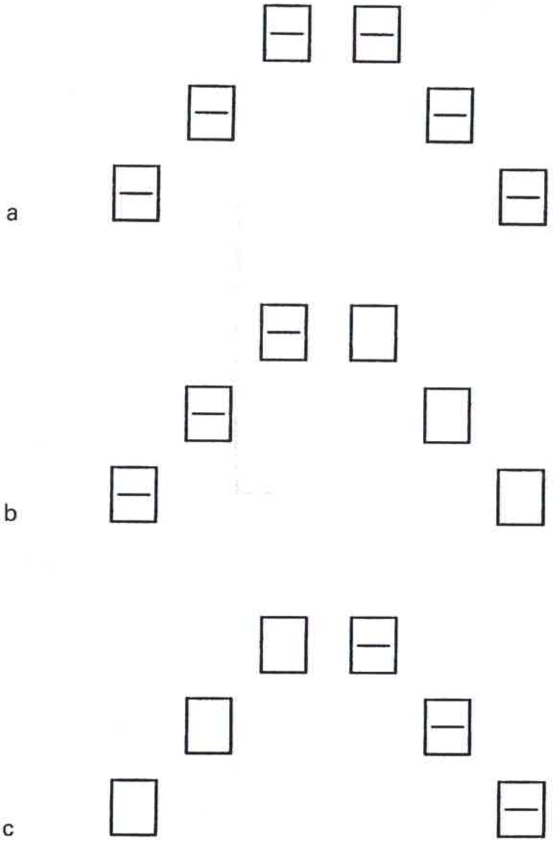

Fig. 2. Warning signal display. (*a*) Warning signal display with no precue as to the movement alternatives. (*b*) Warning signal display precuing the left movement alternatives. (*c*) Warning signal display precuing the right movement alternatives.

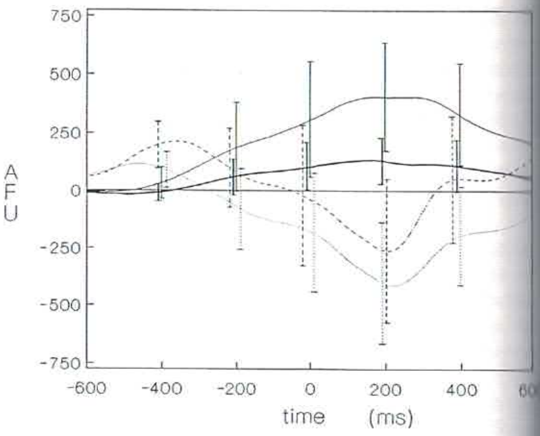

Fig. 3. Force transducer signals of the SCI subjects. 95% confidence intervals of the group mean are given. Button Y5 is pressed at $t=0$ ms. AFU = arbitrary force unit (AFU = F×C). —, front/right; **—**, front/left; ···, back/left; ----, back/right.

In both groups no significant differences in the RT data between left precue and right precue conditions were found. Consequently the RT data for these conditions were pooled per group. Subjects from both groups showed faster RTs in conditions containing relevant information. Non-disabled persons showed faster RTs in all cue conditions compared to SCI persons. These results were observed in all series, as can be seen in Fig. 5 and Table III. In series B and C significant interaction was found between the factors group and information. In series A group versus cue information interaction just failed to attain significance ($p=0.06$).

Due to the method used to compute resultant EMG and force transducer signals with MT as a reference point in time, possible RT- and MT-dependent changes in individual signal recordings were eliminated. No significant differences in EMG signals due to the factor cue information were found within the time-range of (MT+0 ms) to (MT+200 ms). Therefore EMG data were pooled for the factor information. Mean changes in EMG activity and results from statistical analyses (two-way ANOVAs with factors group and reaching direction) are presented in Tables IV and V.

Significantly larger increases in EMG activity in the above mentioned time-range were observed in SCI subjects in all series with respect to both LDs and both TPAs, except for the left LD in series A and the left TPA in series C. Significant effects for reaching direction were found in the right LD in series B and in the left TPA in series B and C. Significant group versus reaching direction interaction was only found in the right LD in series B and C, reflecting larger augmentations in EMG activity of SCI subjects while reaching to the three right hand *Y*-buttons.

DISCUSSION

One aim of this study was to investigate how paraplegics compensate for the loss of postural activity of the ES. A significant increase in EMG activity of the LD and the TPA in SCI persons coincided with a time-range in which increased body disbalance was registered by means of seat force transducers. Such strong EMG phenomena were not found in non-disabled persons. An alternative explanation for these EMG

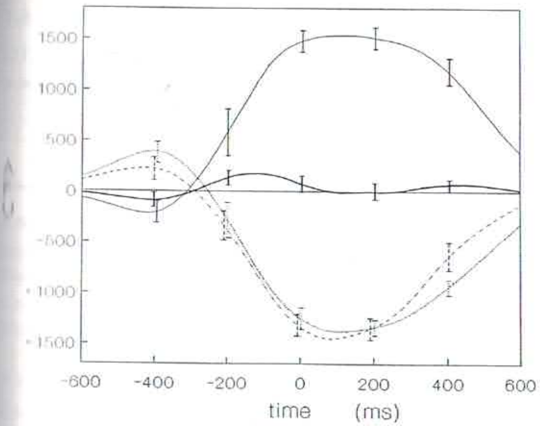

Fig. 4. Force transducer signals of the non-SCI subjects. 95% confidence intervals of the group mean are given. Button Y5 is pressed at $t=0$ ms. AFU = arbitrary force unit (AFU = F×C). ——, front/right; ——, front/left; · · ·, back/left; - - - -, back/right.

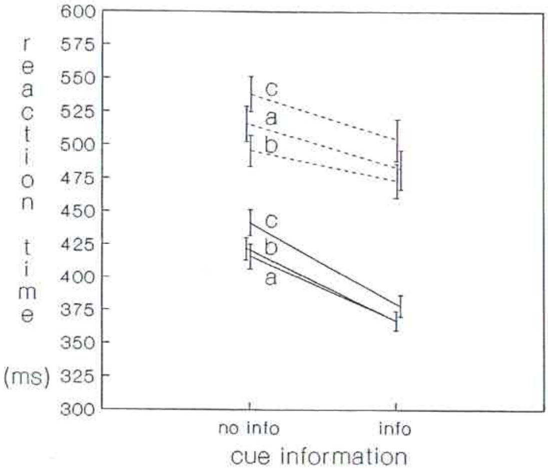

Fig. 5. Mean reaction time during all task series. 99% confidence intervals of the group mean are given. ——, non-SCI subjects; - - - -, SCI subjects. a = A series, b = B series, c = C series.

increases is hand support while pressing a button. However, this explanation can be rejected since increased EMG activity in paraplegic persons was simultaneously recorded in contralateral LDs and TPAs. Excessive use of muscular force necessary to press a button is also rejected as a possible explanation for a number of reasons. Firstly no evidence for such actions, i.e. no significant increase in EMG activity of the LDs and TPAs due to button press force, was found in non-disabled persons. Secondly positive button contact force was approximately 2 N which is relatively low. Finally subjects were explicitely instructed not to lean on the buttons during task performance. Sex differences regarding body tissue composition, e.g. subcutaneous fat, may have influenced between group EMG differences. In SCI persons, on the other hand, an increase in the amount of subcuta-

Table II. *Task error analysis (% of the total number of trials per series)*

RT = reaction time

	RT <125 ms (%)	Wrong button press (%)
A series		
SCI subjects	2.66	2.77
Non-SCI subjects	2.97	0.35
B series		
SCI subjects	1.11	1.66
Non-SCI subjects	2.02	0.47
C series		
SCI subjects	1.66	1.11
Non-SCI subjects	2.26	0.47

Table III. *Reaction time ANOVA statistics of A–C series*

DF = degrees of freedom, F = ANOVA F-value

	DF	*F*	*p*
A series			
Group	1	555.514	0.000
Info	1	79.091	0.000
Group×info	1	3.548	0.060
Residual	3 275		
B series			
Group	1	504.359	0.000
Info	1	86.513	0.000
Group×info	1	15.263	0.000
Residual	3 256		
C series			
Group	1	537.270	0.000
Info	1	97.534	0.000
Group×info	1	8.588	0.003
Residual	3 208		

Table IV. *Mean EMG increase (in µV/s) in A–C series*

LD = latissimus dorsi, TPA = trapezius pars ascendens, L = left, R = right

	SCI subjects ($n = 11$)				Non-SCI subjects ($n = 13$)			
	LD L	LD R	TPA L	TPA R	LD L	LD R	TPA L	TPA R
A series								
Direction 1	+4.0	−14.6	−85.2	−211.4	−5.0	−1.2	+1.2	−32.6
Direction 2	−13.2	−9.2	−101.2	−149.8	−4.0	−0.2	−26.6	−28.8
Direction 3	+4.8	−7.6	−127.0	−257.4	−5.0	+1.4	−37.8	−26.2
Direction 4	−11.2	−25.6	−70.4	−200.2	−1.8	−0.2	−2.8	−29.4
Direction 5	−12.8	−22.0	−42.2	−217.8	−3.2	+0.0	−6.6	−28.4
Direction 6	+7.6	−34.2	−26.8	−228.2	−4.2	−4.6	−4.4	−14.0
Mean	−3.4	−18.8	−75.4	−210.8	−3.8	−0.8	−12.8	−26.6
B series								
Direction 1	−17.2	−38.8	−26.6	−277.8	−8.6	−1.8	−3.8	−37.6
Direction 2	−14.8	−14.8	−62.0	−196.0	−6.2	−1.0	−51.2	−29.4
Direction 3	−14.2	−9.4	−65.8	−316.6	−4.2	+0.2	−71.2	+2.8
Direction 4	−21.6	−24.4	−20.0	−178.4	−2.6	+1.4	+16.4	−42.2
Direction 5	−3.0	−33.8	−34.6	−263.2	−5.4	+1.2	+8.6	−20.4
Direction 6	−6.8	−53.6	−57.0	−253.8	−7.0	−2.0	−7.0	−32.2
Mean	−13.0	−29.2	−44.4	−247.6	−5.6	−0.4	−18.0	−31.6
C series								
Direction 1	−11.8	−2.6	−25.6	−179.6	+3.6	−7.8	−21.4	−3.8
Direction 2	−21.2	−7.8	−42.8	−113.0	−3.6	−8.6	−93.4	−27.8
Direction 3	−9.8	−14.2	−53.0	−274.6	+3.2	−3.4	−70.6	−4.0
Direction 4	−10.8	−47.8	+39.8	−237.6	−4.2	+4.0	+12.2	−5.8
Direction 5	−6.0	−26.4	+0.8	−182.8	−11.0	+2.6	−18.2	−18.8
Direction 6	−12.0	−44.6	−25.8	−264.2	−9.0	+2.8	−30.2	−21.4
Mean	−12.0	−24.0	−17.8	−208.6	−3.6	−1.8	−37.0	−13.6

neous fat has also been observed. We argue that the increased activity of the LD and the TPA in sitting SCI subjects signifies compensatory postural adjustment rather than activity related solely to upper extremity movements. Our findings agree with clinical reports on increased LD and TPA activity in sitting SCI persons. No conclusive evidence as to the direction specificity of the mechanisms of compensatory postural adjustment in SCI persons can be drawn from the data concerning reaching direction and group versus reaching direction interaction.

The second goal of this study was to investigate whether motor response programming in SCI persons is also impaired. During informative cues reaching direction parameters were open for preprogramming. RT results indicate that motor preparation occurs both in SCI and non-SCI persons. These findings are consistent with earlier reports (13, 18, 22, 23, 25, 31). The SCI subjects however reacted slower in all cue conditions prior to a required selective, goal-directed movement. Furthermore they seem not to be able to prepare a required selective, goal-directed movement as much as non-SCI persons. This is reflected in the significant group versus cue information interaction. The relatively small reaching distances used in the A series and the associated small level of sitting disbalance may be the cause of the lack of interaction in the A series, although a tendency towards interaction can be observed. Sex differences between groups, although present, cannot be used as an explanation of RT differences. Noble et al. (21) and Welford (33) for example state that in choice RT tasks women tend to have longer RTs than men of the same age. This is in total opposition to our data concerning between group RT differences. Age differences between groups may have contributed to RT differences. However, in an other experiment by Vuurman & Seelen (32), consisting of a visual vigilance task, performed by the

Table V. *EMG ANOVA statistics of A–C series*

LD = latissimus dorsi, TPA = trapezius pars ascendens, L = left, R = right, DF = degrees of freedom, F = ANOVA F-value

		LD L		LD R		TPA L		TPA R	
	DF	F	p	F	p	F	p	F	p
A series									
Group	1	0.008	0.927	34.772	0.000	13.035	0.000	46.746	0.000
Direction	5	0.646	0.665	2.240	0.051	1.290	0.268	0.223	0.953
Group × direction	5	1.084	0.369	1.390	0.228	0.420	0.835	0.337	0.890
Residual	276								
B series									
Group	1	7.477	0.007	52.992	0.000	6.674	0.010	38.185	0.000
Direction	5	0.863	0.506	2.657	0.023	4.743	0.000	0.226	0.951
Group × direction	5	1.379	0.232	2.595	0.026	0.708	0.618	0.548	0.740
Residual	276								
C series									
Group	1	5.900	0.016	19.698	0.000	1.001	0.318	38.148	0.000
Direction	5	0.737	0.596	0.970	0.437	3.574	0.004	0.301	0.912
Group × direction	5	1.010	0.412	3.951	0.002	0.440	0.821	0.882	0.494
Residual	276								

same members of both groups, between group differences in mean RT amounted to 34 ms. Similar small RTs in choice RT tasks between persons in their twenties and thirties were found by Morikiyo et al. (20) and Salthouse (24). Even RT differences of –9 ms between 18–24 year olds and 25–34 year olds are mentioned by Welford (33). Between group RT differences in our study range from 75 ms to 125 ms. We argue that these differences cannot solely be attributed to age differences. Relating our RT results to the two-stage model of programming, as proposed by Haagh et al. and Spijkers et al. (15, 30) we suggest that both motor programming and program loading concerning upper extremity reaching movements may be impaired in paraplegic subjects. An increase in RT in paraplegics may be due to possible discrepancies between the required (compensatory) motor control parameters and the impaired postural motor system.

The regaining of postural stability, as a prerequisition to optimal upper extremity ADL, in sitting paraplegics may be improved both in time and in quality during medical rehabilitation if specific sensorimotor training of the LD and the TPA is carried out additionally to existing training programs. The recording of movement preparation may be a sensitive and clinically useful method to assess the degree to which SCI subjects can compensate or already have compensated for the loss of postural motor control.

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