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Trunk use and co-contraction in cerebral palsy as regulatory mechanisms for accuracy control

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Abstract

In the present study, we examined whether individuals with cerebral palsy (CP) systematically vary motion of the trunk and co-contraction in the upper limb as a function of accuracy demands. Four participants with spastic tetraparesis, four with spastic hemiparesis, and four healthy controls were asked to repeatedly move a spoon back-and-forth between two target locations. The task was externally paced. In half the trials the accuracy demands were increased by filling the spoon with water. In addition, a condition in which the trunk was fixated was examined. When the movements were controlled for speed, trunk motion hardly varied as a function of accuracy. Co-contraction in the shoulder, however, was systematically higher under high-accuracy demands. Trunk fixation yielded differential group effects on the co-contraction of the shoulder muscles. It increased in control participants, tended to decrease in hemiparetic participants, and was unaffected in tetraparetic participants. Collectively, the present findings show that the increased trunk involvement and high co-contraction levels in CP should not exclusively be regarded as disorder-related phenomena. Regulation of co-contraction in the shoulder is a general strategy to cope with variations in movement-accuracy constraints, while increased trunk involvement proves a secondary reaction to these constraints.

Keywords: CP; Spasticity; EMG; Spoon use; Accuracy demands; Multi-joint coordination

1. Introduction

Cerebral palsy (CP) is a condition caused by chronic, non-progressive brain damage in young children due to, e.g., an intra-uterine, peri- or postnatal oxygen shortage, infection, intoxication or cerebro-vascular accident (CVA), a cerebral contusion, prematurity, or a brain tumour at a very young age. Usually CP is associated with various motor deficits (i.e., spasticity, dystonia, ataxia, athetosis or hypotonia; Leland Albright, 1996), sensory deficits (i.e., impaired proprioception and stereognosis; Cooper, Majnemer, Rosenblatt, & Birnbaum, 1995), but also with seizures and behavioural and cognitive problems. Approximately 60% of children with CP experience serious forms of spasticity (Sugden & Keogh, 1990). Spasticity is a disorder characterized by hypertonia as reflected by resistance to an externally imposed movement, particularly when the speed of such an imposed movement increases beyond a certain threshold (Sanger, Delgado, Gaebler-Spira, Hallett, & Mink, 2003). Spasticity is generally accompanied by decreased dexterity, disordered coordination of synergistic muscles, increased co-contraction of antagonistic muscles, and stereotyped movement synergies, such as an increased involvement of the trunk (Barnes, McLellan, & Sutton, 1994; Filloux, 1996; Lance, 1980). The present study examines whether the last two phenomena can be employed as regulatory mechanisms for accuracy control in people with cerebral palsy.

Excessive trunk involvement is characteristic for upper limb motion in CP (Steenbergen & Meulenbroek, 2003; Steenbergen, Van Thiel, Hulstijn, & Meulenbroek, 2000; Van Roon, Steenbergen, & Meulenbroek, 2004; Van Roon, Van der Kamp, & Steenbergen, 2003; Van Thiel & Steenbergen, 2001). Originally, the increased trunk involvement was interpreted as a pathological movement synergy that serves to counteract the reduced functional range of motion of the

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shoulder and elbow joints. Recently, however, we found evidence that this increased involvement of the trunk might also be used as an adaptive mechanism to improve the accuracy of reaching movements instead of being a primary symptom of the cerebral disorder (Van Roon et al., 2004). Participants with CP displaced their trunk more with increases of accuracy demands of a task, similar to control participants (cf. Steenbergen, Marteniuk, & Kalbfleisch, 1995; Van der Kamp & Steenbergen, 1999).

However, these findings may have been confounded by movement speed since this factor has also been shown to affect the degree of trunk displacement during upper limb motion (Rosenbaum, Loukopoulos, Meulenbroek, Vaughan, & Engelbrecht, 1995; Rosenbaum, Slotta, Vaughan, & Plamondon, 1991; Van Roon et al., 2004; Wang & Stelmach, 2001). Greater-massed body segments such as the trunk (with a larger inertia) contribute less to the end effector displacement when movements are performed faster. In the present study, we therefore tried to keep the movement speed across participants and across conditions as constant as possible by externally pacing the movements via a metronome. We reasoned that if trunk displacement is a mechanism for accuracy control, the trunk should be recruited more extensively in tasks with high accuracy demands, even in movement speed controlled conditions.

Another mechanism that is known to increase movement accuracy is the regulation of muscular co-contraction. In healthy participants, increasing co-contraction levels of antagonist muscles around the elbow and shoulder has recently been proposed as a primary means by which the nervous system may stabilize the position of the limb to improve movement accuracy (Gribble, Mullin, Cothros, & Mattar, 2003; Van Galen & Schomaker, 1992; Van Galen & Van Huygevoort, 2000; Van Gemmert & Van Galen, 1997).

Our second aim of the present study was therefore to examine whether a similar regulation of co-contraction for the control of movement accuracy would be employed by individuals with CP. There is ample evidence for a decreased inhibition of antagonists during the contraction of agonist muscles in people with spastic CP (Barnes et al., 1994; Brouwer & Ashby, 1991; Filloux, 1996; Lance, 1980; Milner-Brown & Penn, 1979; O'Sullivan et al., 1998). This leads to increased levels of co-contraction, commonly regarded as a characteristic symptom of the disorder. The question pursued here is whether these participants are still able to regulate co-contraction for the sake of accuracy control. Since group differences with regard to co-contraction levels were not our assessment goal, we applied an individual-based EMG data normalization technique that allowed us to isolate variations in co-contraction as a function of accuracy demands per individual.

To test our predictions regarding increased trunk involvement and accuracy demand depending variations in co-contraction, we designed a combined EMG-motion recording experiment. Adolescents with a spastic paresis due to CP and healthy control participants were asked to move a spoon, either empty or filled with water, between two targets positioned within reach in the midsagittal plane. In half the trials, trunk motion was made impossible. Based on our assumption that both the involvement of the trunk and the co-contraction of arm muscles could be regulated for accuracy control, we expected co-contraction to increase when the trunk was fixed, as a compensation for having lost the supposed trunk-involvement strategy.

Finally, an explorative research question was whether there are differential effects of different types of motor deficit (hemiparesis versus tetraparesis). Therefore, both hemiparetic and tetraparetic adolescents participated. They had to perform the task with their least affected hand. The reason for that was that this hand is used in daily life for spoon-use tasks. Furthermore, the task was quite complex, to the extent that it was impossible for the spastic participants to perform it successfully with their most impaired hand. In addition, the ipsilesional side in hemiparesis, while often denoted the unimpaired side, shows subtle deficits as well, such as impaired dexterity and sensory deficits (Brown et al., 1989; Cooper et al., 1995; Duque et al., 2003; Mercuri et al., 1999). Still, the amount of functional loss of the ipsilesional upper limb is not well known (Jung, Yoon, & Park, 2002).

In sum, our two main research questions were: (1) can people with CP regulate the amount of trunk involvement for accuracy control, and (2) can they, for the same purpose, regulate the amount of co-contraction of antagonistic muscles at the elbow and shoulder?

2. Method

2.1. Participants

Four individuals being diagnosed with having spastic tetraparesis as a consequence of CP (mean age 17; 5 years, S.D. 2; 0 years, range 15; 4–19; 4 years), four individuals being diagnosed with having spastic hemiparesis as a consequence of CP (mean age 17; 6 years, S.D. 0; 10 years, range 16; 4–18; 4 years), and four healthy control participants with no known history of neurological disorders (mean age 22; 5 years, S.D. 3; 0 years, range 19; 6–26; 6 years) voluntarily participated in the experiment. At the time of testing, all participants with CP were students at the Werkenrode Institute (Groesbeek, The Netherlands), where they followed an adapted educational program. They all had sufficient physical and cognitive abilities to perform the task under study, had normal or corrected to normal vision, and used no medication to alleviate spasticity. None of the participants suffered from apraxia or neglect. Three tetraparetic participants sat unstrapped in a wheelchair from which they performed their daily activities. All other participants were able to walk independently. Dexterity of the participants was evaluated with the Purdue Pegboard test (fine dexterity; Tiffin, 1968) and the Box-and-Block test (gross dexterity; Mathiowetz, Volland,

Table 1
Participant information

	Sex	Age (year; month)	Position target 2 (cm)[a]	Hand-preference[b]	PP[c]	BB[d]		Diagnosis
Tetraparetic participants						PH	NPH	
1	F	18; 11	40	R	32	52	38	CP, spastic tetraparesis
2	F	16; 2	42	L	33	48	39	CP, spastic tetraparesis
3	M	19; 4	57	R	28	38	28	CP, spastic tetraparesis
4	M	15; 4	52	R	25	42	8	CP, spastic tetraparesis
Hemiparetic participants								
1	M	17; 11	47	L	19	24	7	CP, spastic hemiparesis, psychomotor retardation
2	F	18; 4	43	L	40	55	19	CP, spastic hemiparesis, epileptic
3	F	17; 4	45	L	40	43	19	CP, spastic hemiparesis
4	M	16; 4	52	R	42	50	8	CP, spastic hemiparesis, epileptic
Control participants								
1	M	22; 8	56	R	57	83	81	
2	F	19; 6	49	L	48	63	56	
3	F	20; 11	45	R	49	74	73	
4	M	26; 6	57	R	47	69	59	

[a] Distance between the front edge of the table and the midpoint of the target that was farthest away from the participant. See Section 2.2 for a detailed description of the method to determine this distance.
[b] As indicated by the participant.
[c] PP: purdue pegboard-score = the total number of pins placed into the holes in three 30 s periods with the preferred hand.
[d] BB: box and block test-score = total number of blocks transported in 1 min, PH: preferred hand, NPH: non-preferred hand.

Kashman, & Weber, 1985) according to the instructions in the test manuals. One-tailed *t*-tests showed that the differences between both CP groups and the control group were statistically significant for all dexterity scores: the Purdue Pegboard-score for the preferred hand and the Box-and-Block test-score for both the preferred hand and the non-preferred hand ($P < 0.05$ for all *t*-tests). No significant differences were found between the hemiparetic group and the tetraparetic groups as regards the dexterity measures. Additional participant information is provided in Table 1. Participants were naive as to the purpose of the task and gave signed consent prior to data collection. This study was approved by the local ethics committee and performed in accordance with the ethical standards laid down in the 1964 Declaration of Helsinki.

2.2. Experimental setup

Two targets were placed on the table and within reaching distance of the participant. A target was made of a translucent plexiglass cylinder of 4 cm in diameter and 4 cm in height. The top of the cylinders was completely covered by a circular metal plate. Determination of the location of the targets in the functional workspace was as follows. Participants were instructed to hold a spoon using a power grip and reach forward with it as far as possible by extending their arm *without moving their trunk* and to touch the table surface with the spoon. One target was placed at that position on the table. The other target was positioned 30 cm closer to the participant in the midsagittal plane. The spoon that was used to perform the experimental task, had a handle that measured 12 cm in length and 1 cm in diameter, and had an aluminium round bowl (diameter of 4.5 cm and depth of 1.2 cm). The spoon's weight was 80 g.

Two 3D motion-tracking devices (Optotrak 3020, Northern Digital) were used to record the positions of six infrared light emitting diodes (IREDs), placed on the wrist, elbow, both shoulders, and on each side of the sternum on the chest. The sampling rate was set at 100 Hz and the spatial accuracy was better than 0.2 mm in the *x*-, *y*-, and *z*-dimension.

Electromyographic activity of the following muscles was recorded using surface electrodes: brachioradialis (elbow flexor), triceps lateral head (elbow extensor), deltoid anterior (shoulder flexor), and deltoid posterior (shoulder extensor). We also recorded the activity of the biceps long head (bi-articular flexor acting primarily at the elbow). However, the results found for the brachioradialis and biceps long head were comparable. Therefore, we only report the results for the brachioradialis, since this muscle acts only at the elbow, and not at the shoulder.

EMG activity was sampled at 2 kHz (common mode rejection ratio 90 dB, high-pass 20 Hz, low-pass 500 Hz). These signals were subsequently amplified by means of an EMG-interface module consisting of a custom-made, front-end physiological amplifier. EMG signals were digitally converted at 1024 Hz by an ODAU II system (Northern Digital, Waterloo, Canada), enabling 16-bit synchronized collection of analogue and digital data with the Optotrak IRED-displacement data. Adhesive, disposable pre-gelled Ag/AgCl surface EMG disc electrodes (diameter 9 mm, inter-electrode distance 2 cm) were placed in a bi-polar derivation, parallel to the fibers at the bellies of the muscles under study (for electrode placements, see Delagi & Perotto, 1981). The reference electrode was placed on the acromial end of the clavicle on top of the contra-lateral shoulder. The electrode locations were prepared by cleaning and rubbing the skin with alcohol and gel until skin resistance was below 10 kΩ. For verification purposes, we also videotaped the experimental sessions.

2.3. Procedure

Participants were seated at a table on a wooden chair without armrests and with a high back. The table could be adjusted in height such that the following criteria were met. When the forearms were placed on the table, the elbows were flexed at 90°. Additionally, the feet were flat on the ground (or placed on a foot rest) and the knees were flexed at 90°.

Next, the location of the targets in the workspace was determined (see above) and each experimental condition was practiced two times. The control participants used the hand that they indicated to use for everyday unimanual tasks. The tetraparetic and hemiparetic participants performed the task with their least affected hand, as indicated by them and confirmed by the dexterity assessment.

At the start of each trial participants were asked to sit upright, with their back against the back of the chair, their preferred hand holding the spoon with a power grip, and the other hand resting on the tabletop. The convex side of the bowl of the spoon had to touch the top of the cylinder nearest to the participant before each trial. We standardised movement speed across participants as good as possible by external pacing. Movements had to be made at a predefined pace indicated by a computer-generated acoustic signal that was presented every two seconds and lasted 50 ms. Participants were instructed to start the movement at the sound of a click, arrive at the other target at the sound of the next click (two seconds later), start the return movement on the next click (again two seconds later), arrive at the first target on the next click (again two seconds later) and so forth. Participants were instructed to make these discrete back-and-forth movements as fluently as possible, and to touch the targets with the convex side of the bowl of the spoon. One trial consisted of transporting the spoon eight times between the two targets, so that four movements were made in the forward direction and four movements in the backward direction. In this paper, we will only report the results for the forward movements as analyses showed that results were largely the same for both movement directions.

In half of the trials, the trunk was attached to the back of the chair at the height of the armpits by means of a non-elastic strap (1.5 cm in width), so that forward and lateral trunk displacements and trunk rotations were minimized. Scapular movements remained possible. In the other half of the trials the trunk was left free to move. In addition, in half of the trials the bowl of the spoon was filled with approximately 6 g of water, while it was empty in the other half of the trials (manipulation of accuracy demands during the movement). When any water was spilled during movement the trial was immediately repeated. However, spilling was hardly observed.

The factor trunk (fixed versus non-fixed) was counterbalanced using an ABBA-design. Half the participants in each group started with the trunk attached to the back of the chair and the other half started without trunk restraint. Within each trunk block, five trials were performed with a filled spoon and five trials with an empty spoon. Half of the participants in each group started with five high-accuracy trials followed by five low-accuracy trials in the first two trunk blocks (ABAB). In the last two trunk blocks the order of presentation of the accuracy conditions was reversed (BABA). The other half of the participants started with the low-accuracy trials in the first two trunk blocks.

The total experiment took approximately three to three and a half hours. However, only 50–60 min were spent on the actual performance of the 40 experimental trials. The remaining time was spent on preparations (instructions, placement of the electrodes and IREDs, determination of the location of the targets). Furthermore, short breaks were scheduled between the four trunk blocks, and participants could always indicate when they needed extra rest.

2.4. Data analysis

The positional data of the IREDs were filtered using a zero phase lag, second-order Butterworth filter with a cut-off frequency of 10 Hz and then differentiated to calculate movement velocity and acceleration. Elbow angle data were derived by calculating per recorded sample the enclosed angle (in 3D space) between the upper arm (vector joining the IRED on the ipsi-lateral shoulder and the IRED on the elbow) and the forearm (vector joining the IRED on the elbow and the IRED on the wrist). Shoulder angle data were derived by calculating per recorded sample the angle in the horizontal plane between the vector joining the IRED on the elbow and the IRED on the ipsi-lateral shoulder and the vector joining the IRED on the ipsi-lateral shoulder and the IRED on the contra-lateral shoulder. The angle data were subsequently preprocessed in a similar fashion as the IRED-position data.

Preprocessing of the raw EMG data consisted of applying a root mean square filter with a time constant of $t = 0.02$ s, which resulted in a rectified, filtered surface-EMG signal for each of the four muscles. To synchronize the EMG signals with the IRED-position recordings, a constant time shift of 50 ms was applied to compensate for the plant delay. Then, a third-order, zero phase-lag, low-pass Butterworth filter with a cut-off frequency of 10 Hz was applied. Finally, the EMG data were normalized for each muscle and for each participant separately by dividing the EMG values by the median EMG value for that muscle across all movements. In that way, we were able to examine the effects of accuracy demands and trunk restraint on the muscular co-contraction. Overall between-group differences in co-contraction levels could not be determined, but determining these differences was not

Table 2
Median, minimum and maximum MT, and the number of movements before and after the selection of movements that fell within a window of 300 ms around the individual median (see Section 2.4)

	All movements (forward and backward)			Selected forward movements			
	Median MT (ms)	Minimum-maximum MT (ms)	N^{a}	Median MT (ms)	Minimum–maximum MT (ms)	N	N as% of total N of movements
Tetraparetic participants							
1	1770	1100–2920	320	1775	1620–1920	78	24
2	1710	820–3660	274	1770	1600–1860	35	13
3	1780	940–2590	318	1790	1630–1930	78	25
4	2530	1460–3450	308	2525	2390–2680	54	18
Hemiparetic participants							
1	1790	960–4070	313	1790	1640–1930	41	13
2	1640	860–2830	320	1640	1490–1790	53	17
3	1892	970–4110	312	1880	1750–2040	43	14
4	1590	720–2910	305	1600	1440–1740	35	11
Control participants							
1	1960	1190–2590	318	1920	1810–2110	73	23
2	1890	1290–2580	314	1890	1740–2040	71	23
3	1760	1050–2450	320	1725	1610–1910	76	24
4	1850	1190–2520	318	1880	1700–1990	96	63

The selected forward movements were used in the subsequent analyses.
[a] The maximum number of movements is 40 trials × 8 movements = 320 movements.

our goal. We used this normalization technique, since MVCs could not be reliably determined in the CP groups because of the weakness of the paretic muscles (see also Damiano, Martellotta, Sullivan, Granata, & Abel, 2000). The median instead of the mean values of the EMG signals were used, because the median is less sensitive to outliers, i.e., extreme EMG data spikes.

Semi-automatic segmentation routines were used to define the start and end of each movement. Specifically, the moment at which the tangential velocity of the wrist rose above and fell below 5% of peak wrist velocity defined the start and end of a movement, respectively.

Although externally paced by an audio signal, movement times varied, in particular in the experimental groups (see Table 2). To investigate whether the experimental factors, in isolation or in combination, affected the level of muscle activation and trunk recruitment apart from their effects on movement speed, we selected, for each participant separately, a subset of the experimental data for which all movements did not significantly differ as regards mean wrist velocity. We first calculated the median movement time across all movements for each participant. Subsequently, we selected those movements of which the movement time fell within a 300 ms time-window around the individual median (see Table 2), since any significant effects of the experimental factors on the mean wrist velocity were then eliminated ($P > 0.05$).

Trunk involvement was calculated by taking the average displacement in the forward direction of the two IREDs placed on the chest to the left and right of the sternum.

We used two different measures of muscular co-contraction to be able to determine the effects of accuracy demands and trunk restraint. For the first measure we determined the percentage of the movement time during which the normalized EMG-activity for *both* muscles of an antagonistic muscle pair was larger than 100%, that is, higher than the median phasic EMG-value for that muscle (Lamontagne, Richards, & Malouin, 2000). This measure provided an indication of the duration of phasic co-contraction during a movement.

For the second measure of co-contraction we determined the time-course of the EMG-activity of the brachioradialis and the deltoid posterior during forward movements, when these muscles act as antagonists at the elbow and shoulder, respectively. EMG-activity for the brachioradialis was determined at maximum angular acceleration, maximum angular velocity, and maximum angular deceleration of the elbow. For the deltoid posterior EMG-activity was determined at maximum angular acceleration, maximum angular velocity, and maximum angular deceleration of the shoulder. The reason for capturing the EMG activity in this way is that an increase in antagonist activity was assumed to result in a larger co-contraction at the specific joint, because that activity is “not needed” to drive the arm in the forward direction. It merely serves to increase the stiffness of the arm.

To find out whether the CP participants showed increased co-contraction in their least affected arm as compared to the control participants, we calculated the cross-correlation coefficients between the normalized EMG-traces of the brachioradialis and the triceps lateral head and between the normalized EMG-traces of the anterior and posterior deltoid for all 40 trials of each participant separately. Following Fisher-Z transformations, the median coefficients across all 40 trials were determined for each participant.

Table 3
Results of the repeated measures ANOVAs

	Trunk involvement[a]	Duration of cocontraction elbow	Duration of cocontraction shoulder	EMG-activity brachioradialis	EMG-activity delt. posterior
Group	n.s.	[b]	[b]	[b]	[b]
Accuracy	$F(1, 9) = 3.83$, $P = 0.082$	n.s.	$F(1, 9) = 10.28$, $P = 0.011$	n.s.	$F(1, 9) = 9.90$, $P = 0.012$
Trunk		n.s.	n.s.	n.s.	n.s.
Group × accuracy	n.s.	n.s.	n.s.	n.s.	n.s.
Group × trunk		n.s.	$F(2, 9) = 6.45$, $P = 0.018$	n.s.	$F(2, 9) = 8.12$, $P = 0.010$
Accuracy × trunk		n.s.	n.s.	n.s.	n.s.
Group × accuracy × trunk		n.s.	n.s.	$F(2, 9) = 6.59$, $P = 0.017$	n.s.
Accuracy × moment[c]				n.s.	$F(2, 18) = 6.47$, $P = 0.008$

For results of step-down analyses see text. Empty cell: not applicable; n.s.: not significant.
[a] Only results of non-fixed trunk conditions were included in the analysis of trunk involvement.
[b] Because of the median-based normalization technique we used, group differences could not be evaluated.
[c] The only relevant interaction with the factor moment that was statistically significant.

2.5. *Statistical analysis*

Means of the dependent variables across the replications of each condition were analysed using repeated measures ANOVAs (see Table 3). The design consisted of one between-subjects factor group (tetraparetic, hemiparetic, and control) and two within-subject factors, namely accuracy (high versus low) and trunk (fixed versus non-fixed). For the dependent variable trunk involvement only the results for the non-fixed trunk conditions were used in the analyses. Separate ANOVAs involving the additional within-subject factors movement (first versus last) and moment (i.e., maximum joint angle acceleration, velocity, and deceleration) were conducted in the analysis of trunk involvement and the second measure of co-contraction, respectively. Requirements for homogeneity of variances (Levene's test) were met for all dependent variables. Step-down analyses of statistically significant interactions were performed by means of contrasts. To compare the groups with regard to co-contraction, one-tailed *t*-tests were performed on the cross-correlations between the EMG-traces of the muscles within an antagonistic muscle pair. An alpha level of 0.05 was used for all statistical tests.

3. Results

3.1. *General task performance*

All participants were able to perform the task. Hardly any trials (up to a maximum in one participant of only four out of 40 trials) had to be repeated because of spilling of water. As can be seen in Table 2, the control participants proved most successful in moving at the pace that was indicated by the metronome.

A typical example of normalized EMG-patterns observed in a tetraparetic participants can be seen in Fig. 1. As expected, brachioradialis activity increased during backwards movements whereas triceps and anterior deltoid activity rose during forward movements. The out-of-phase activity of the elbow muscles was reflected by a negative cross-correlation between the time series of this particular example ($r = -0.36$). A different muscle activation pattern can be observed in Fig. 1 for the posterior and anterior deltoid. The antagonistic parts of this shoulder muscle show a more synchronized activity pattern. This co-activation was reflected by a positive cross-correlation ($r = 0.56$). These data demonstrate the general finding that muscle activation patterns at the shoulder showed larger in-phase activity, and thus co-contraction, than at the elbow. No significant difference was found between the groups for the cross-correlation between the normalized EMG-traces of the brachioradialis and the triceps lateral head (control group: mean + 0.10, S.D. 0.26, range −0.15 to +0.48; hemiparetic group: mean + 0.22, S.D. 0.19, range +0.07 to +0.49; tetraparetic group: mean + 0.02, S.D. 0.28, range −0.34 to +0.27). Therefore, we conclude that that the level of co-contraction at the elbow was similar among the three groups. Also, no difference was found between the hemiparetic and tetraparetic group as regards the cross-correlation between the normalized EMG-traces of the anterior and posterior deltoid. However, trends in the expected direction were found when we compared each CP group with the control group on this measure (for both comparisons, $t(6) = -1.4$, $P = 0.10$; control group: mean + 0.06, S.D. 0.44, range −0.47 to +0.50; hemiparetic group: mean + 0.39, S.D. 0.18, range +0.15 to +0.57; tetraparetic group: mean + 0.46, S.D. 0.34, range −0.02 to +0.75). It must be noted that these differences might not have reached significance because of the large between-subjects variability. The individual medians within the control group were rather low

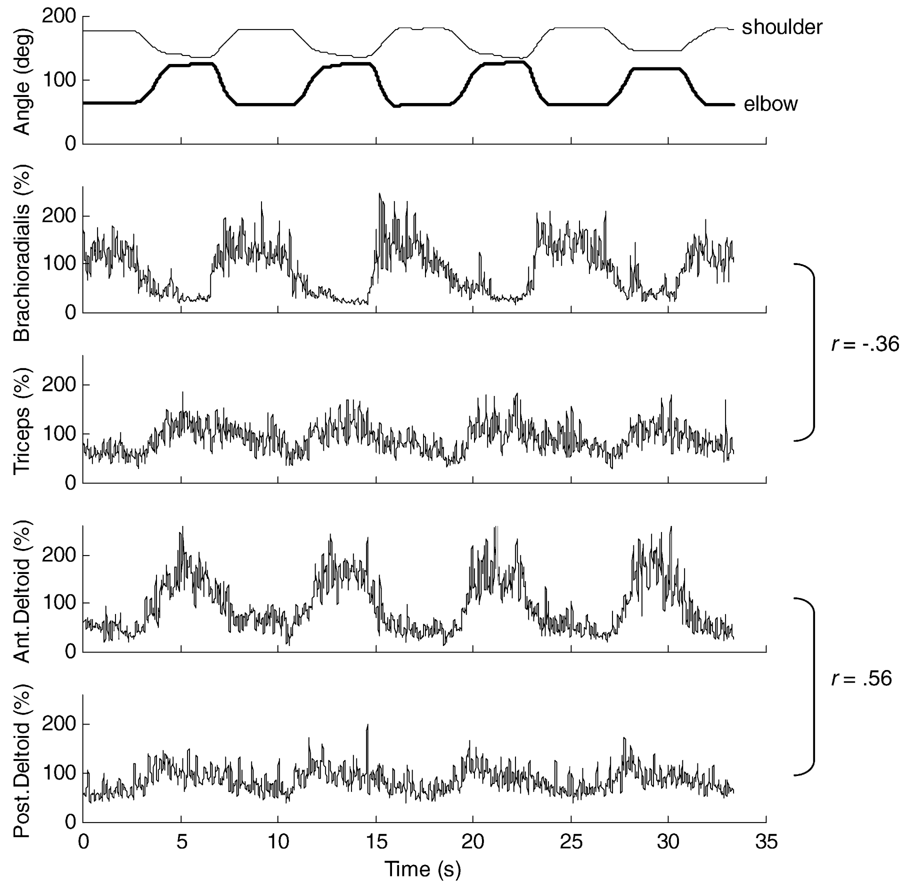

Fig. 1. Examples of normalized EMG-patterns found for a tetraparetic participant (nr. 1) during a trial, in which she moved four times back and forth between the targets with a full spoon. Trunk motion was not blocked. In the top panel the elbow and shoulder angles are depicted as a function of time, *r*: cross-correlation coefficients between the normalized EMG-traces.

or even negative for three out of four participants (−0.47, −0.13, and +0.33), while in the hemiparetic group and especially the tetraparetic group the median coefficients were higher for three out of four participants (hemiparetic group: +0.42, +0.43, and +0.57; tetraparetic group: +0.45, +0.65, and +0.75). When the data of both CP groups were collapsed and compared to the data of the control group, the difference did reach statistical significance (t (10) = −1.9, $P < 0.05$). These findings suggest that co-contraction at the shoulder is higher in the CP groups as compared to the control group.

It must be noted that tri-phasic EMG-patterns were not likely to be observed, since the movements had to be performed at a slow speed (2 s; see also Section 2.3). Models of multijoint arm movements (Lan, 1997; Stroeve, 1997) have demonstrated why slow movements are generally accompanied by high levels of co-contraction. During the prolonged deceleration in slow movements, considerable damping in the muscles and joints is caused by an increased overall stiffness of the arm rather than the phasic stiffness changes in antagonists that occur during fast movements. These findings are in line with experimental results (Latash & Gottlieb, 1991; Lestienne, 1979).

3.2. Trunk involvement

In the trunk-free conditions, no group effects as regards trunk involvement were found, $F(2, 9) < 1$, ns. In the entire data set of forward movements performed in the trunk-free conditions, accuracy proved to have a significant effect on trunk involvement, $F(1, 9) = 5.36$, $P = 0.046$. However, this effect proved to be due to the remaining speed variations as a function of the imposed accuracy constraints despite the external pacing of the metronome, particularly in the CP participants. In the speed-controlled subset of the data, the effects of accuracy on trunk involvement almost completely disappeared. Only a marginal trend remained in the direction of an increase in trunk involvement with increases in accuracy demands, $F(1, 9) = 3.83$, $P = 0.082$ (see Fig. 2). These results confirmed our hypothesis that movement speed and trunk involvement are correlated.

Because of the quasi-cyclical character of the task, we reasoned that after the first forward movement the participants with CP might have returned to the original posture at the start of each trial less than the control participants did, as a possible strategy to actively decrease the overall reaching distance. To test for this possible confounding effect we

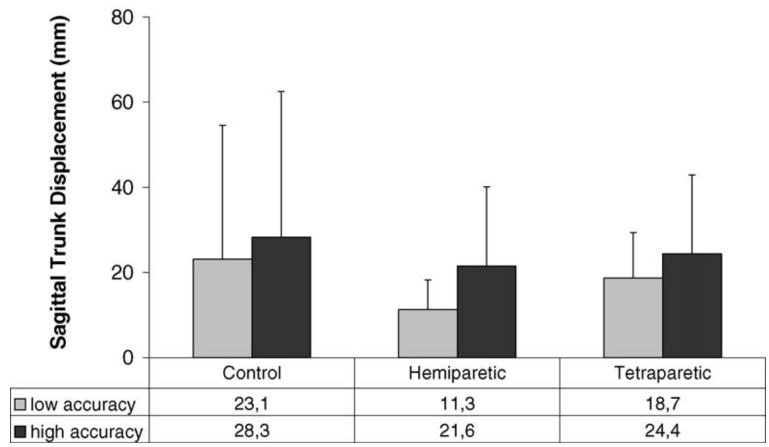

Fig. 2. Sagittal trunk displacement (in millimeters) for forward movements, as a function of accuracy and group. Error bars represent between-subjects variability (standard deviations).

examined the first and the last forward movement within each trial separately. This analysis showed neither group nor movement related effects (first versus last), thereby falsifying the above suggestion.

3.3. Effects of accuracy demands on phasic co-contraction

No main effect of accuracy on the duration of phasic co-contraction of the brachioradialis-triceps muscle pair was found (Fig. 3A). However, we did find an effect of accuracy on the duration of phasic co-contraction at the shoulder, $F(1, 9) = 10.28$, $P = 0.011$ (see Fig. 3B). The phasic co-contraction of the deltoid anterior and the deltoid posterior was prolonged when the spoon was filled with water as compared to when it was empty for all three participant groups.

For our second measure of co-contraction we determined the time-course of the EMG-activity of the brachioradialis and deltoid posterior during forward movements, when these muscles act as antagonists at the elbow and shoulder, respectively (see Fig. 4 for mean EMG signals). No effect of accuracy on the EMG-activity of the brachioradialis was found, indicating that co-contraction at the elbow did not increase when the spoon was filled with water as opposed to when it was empty. However, we did find an interaction between moment and accuracy on the EMG-activity of the deltoid posterior, $F(2, 18) = 6.47$, $P = 0.008$. During the high-accuracy task, as compared to the low-accuracy task, the activity of this muscle was higher at maximum shoulder angular velocity, $F(1, 9) = 9.61$, $P = 0.013$, and at maximum shoulder angular deceleration, $F(1, 9) = 15.17$, $P = 0.004$. This indicates that under high accuracy demands, the co-contraction at the shoulder increases at these moments, but not at maximum shoulder angular acceleration. Again, this effect was present for all three participant groups.

3.4. Effects of trunk restraint on phasic co-contraction

Thus far, the results indicate that the regulation of co-contraction at the shoulder serves as a mechanism to deal with increased accuracy demands in CP. It appears that trunk involvement does not serve a similar goal, although a trend towards larger trunk involvement was found in the high

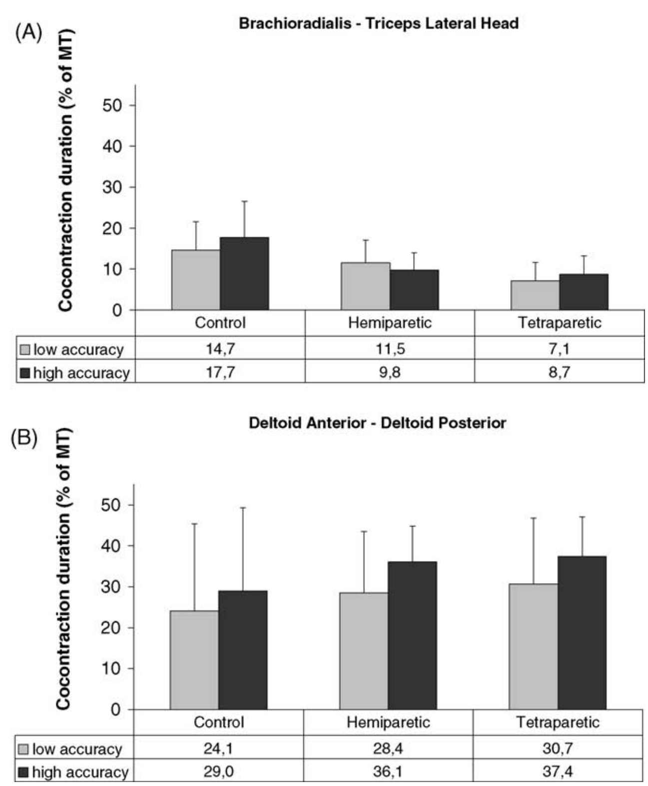

Fig. 3. The duration of co-contraction (as percentage of movement time (MT)) for (A) elbow flexor and extensor and (B) shoulder flexor and extensor for forward movements. Effects of accuracy demands. Error bars represent between-subjects variability (standard deviations).

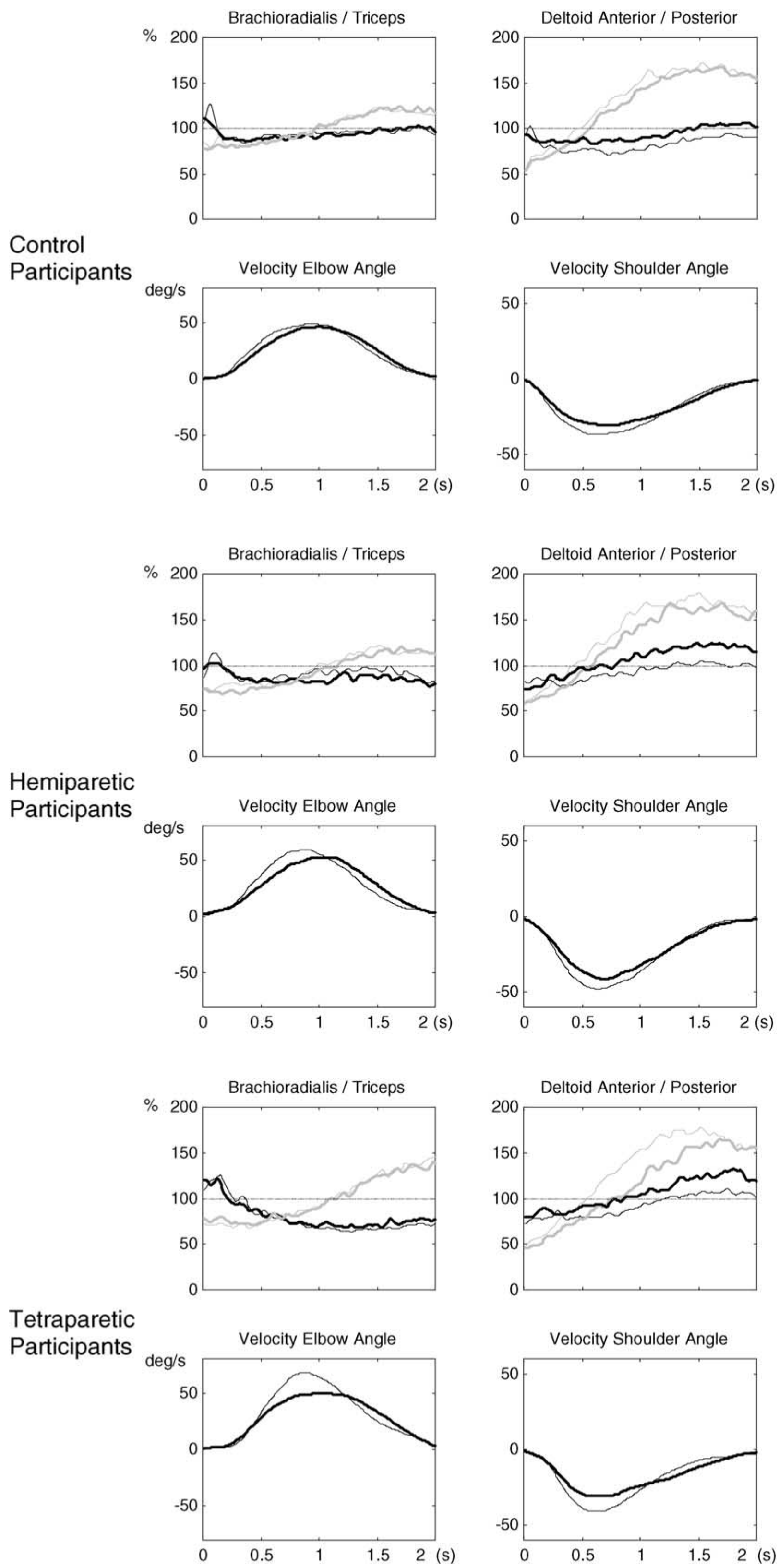

Fig. 4. Mean normalized EMG-signals and elbow and shoulder angle velocity profiles of the forward movements. Black signals depict mean normalized activity of the brachioradialis and deltoid posterior (antagonists). Grey signals depict mean normalized activity of the agonists: the triceps lateral head and the deltoid anterior. Thick lines represent ‘filled spoon’ conditions, thin lines represent ‘empty spoon’ conditions. Note that a difference was found in each participant group at maximum shoulder angular velocity and maximum shoulder angular deceleration between the thick and thin black lines in the top right panels, representing the activity of the deltoid posterior during high and low accuracy conditions, respectively.

accuracy condition as compared to the low accuracy condition. As a further test of the role of trunk involvement for accuracy control we examined the co-contraction in those conditions in which trunk motion was made impossible. If adjustment of trunk involvement was indeed *not* a primary regulatory mechanism to control movement accuracy, we expected that under high accuracy conditions trunk fixation would not lead to an increase in co-contraction at the shoulder.

No effects of trunk restraint were found on the duration of phasic co-contraction at the elbow joint. However, a significant trunk × group interaction was found for the duration of co-contraction at the shoulder joint, $F(2, 9) = 6.45$, $P = 0.018$. Step-down analyses of this interaction revealed that for the hemiparetic group the duration of co-contraction around the shoulder was shorter when the trunk was fixed to the chair (26% of MT) as compared to when the trunk was not fixed (38% of MT), $F(1, 3) = 13.65$, $P = 0.034$. For the control and tetraparetic participants no significant main effect of trunk fixation was found on the duration of co-contraction at the shoulder.

Our second measure of co-contraction, i.e., the time-course of the EMG-activity of the brachioradialis and deltoid posterior, revealed the following effects of trunk restraint. For brachioradialis activity an interaction between trunk, accuracy and group was found, $F(2, 9) = 6.59$, $P = 0.017$. Step-down analyses showed that trunk fixation led to an increased brachioradialis activity only for the hemiparetic group in the low-accuracy conditions (empty spoon), $F(1, 3) = 21.33$, $P = 0.019$. For the activity of the deltoid posterior, a trunk × group interaction was found, $F(2, 9) = 8.12$, $P = 0.01$. It appeared that the control participants increased the activity of this muscle when the trunk was not allowed to move, $F(1, 3) = 12.43$, $P = 0.039$, while the hemiparetic participants tended to decrease the activity in this particular condition, $F(1, 3) = 9.89$, $P = 0.051$. For the tetraparetic participants, no significant effects were found.

4. Discussion

The aim of the present experiment was to examine whether trunk involvement and co-contraction of antagonistic muscles at the elbow and shoulder joint are exploited by individuals with CP as strategic mechanisms to deal with accuracy constraints of a task. Since we wanted to study effects of accuracy demands and trunk restraint on muscular co-contraction *during* a movement, we used a spoon-handling task, as this task allowed us to manipulate accuracy demands during task performance in a continuous fashion.

Our rationale was that if individuals with CP are able to regulate these disorder-related phenomena to variations in accuracy demands, then these symptomatic features might at least in part be regarded as adaptive, instead of as merely primary symptoms of the neurological damage.

As a first result, we showed that the commonly reported larger, or excessive, trunk involvement in cerebral palsy (Van Roon et al., 2004; Van Thiel & Steenbergen, 2001) almost completely disappeared when healthy participants were instructed to move at a similar speed as the participants with CP. From this we conclude that the increased trunk use in CP might for a large part be explained by the general slow movements in this group. In a similar vein, the effect of increased trunk use with increasing accuracy demands that we found in our previous study (Van Roon et al., 2004) disappeared when we controlled for movement speed. Again this suggests that the decrease in movement speed that accompanies high accuracy tasks might be responsible for the increased displacement of the trunk, rather than the increased accuracy demands per se. However, we did find a remaining weak tendency to increase the displacement of the trunk when accuracy constraints were larger (filled spoon) even when movement speed was partialled out. This suggests that trunk motion was indeed not used as the principal regulatory mechanism to enhance the accuracy of moving. The finding that trunk restraint had no additional effect on the phasic co-contraction in the high accuracy condition further substantiates this claim. Unexpectedly, a trend was found for the co-contraction at the shoulder to decrease in the hemiparetic group in the trunk-fixed conditions. This finding may exemplify that the task might have been easier for them to execute, because there is potentially one degree of freedom less to control. In the ‘trunk-free’ condition, the affected arm of the hemiparetic participants serves as a relatively insecure base of support for the trunk. Relieve of this base of support by fixing the trunk may therefore facilitate control of the unaffected limb. The finding that co-contraction was unaffected by trunk restraint in the tetraparetic group may suggest that stiffness of the limb in this particular condition would become too high to perform the task successfully. However, these speculations demand further research.

Two notes on trunk involvement should be made. First, the difference in trunk displacement between the high and low accuracy task in the present study was comparable to or even larger than the differences found in our previous study (Van Roon et al., 2004). Therefore, our conclusion about the role of the trunk should be made with some caution and needs to be verified in a larger group of participants. Second, the fact that participants rested their non-reaching hand on the tabletop may have confounded the findings. In such a posture, the non-reaching hand may function as a stabilizer for the trunk and the need for trunk regulation is altered. Still, in our previous study (Van Roon et al., 2004), in which the non-reaching hand also rested on the tabletop, we did find an effect of accuracy demands on the involvement of the trunk. Nevertheless, to evaluate the naturally preferred recruitment of the trunk, reaching movements from a posture in which the other hand is left alongside the trunk should be evaluated. Although studies in which such a posture had to be adopted also reported larger forward trunk displacements (Levin et al., 2002; Michaelsen, Luta, Roby-Brami, & Levin, 2001) we chose to make the task as natural as possible and therefore allowed the participants to use the non-moving hand as a stabilizer.

Next to trunk involvement co-contraction of antagonist muscles in the arm was examined for its potential to regulate movement accuracy. Despite the fact that we only tested a relatively small number of participants, we found significant effects of accuracy demands on our measures of co-contraction. Not only in the control group, but also in the hemiparetic and tetraparetic group the duration of phasic co-contraction at the shoulder joint was consistently extended in all participants when the accuracy demands of the task were higher (moving a spoon filled with water versus an empty one). Our second measure of co-contraction, the time-course of the EMG-activity of the antagonists, revealed an increased activity of the deltoid posterior at peak angular velocity and at peak angular deceleration of the shoulder under increased accuracy demands in all participants. This indicates that the increase in co-contraction at the shoulder becomes particularly evident in the second part of the movement. These findings are in line with those found in a repetitive pointing task by Laursen, Jensen, and Sjøgaard (1998). In a group of healthy participants, Laursen et al. (1998) showed increased levels of EMG activity at the shoulder when demands for accuracy control increased. The present study extends these findings to participants with CP.

Contrary to these effects of accuracy on co-contraction regulation at the shoulder, co-contraction at the elbow was unaffected by accuracy demands of the task. This contrasts with recent findings of Gribble et al. (2003) in which effects of accuracy on the level of co-contraction at the elbow were found. Gribble et al. asked participants to make rapid pointing movements to small and large targets in the horizontal plane, while the arm was supported against gravity by means of air-sleds. In the present set-up, however, a functional multi-joint task was used during which the moving arm was not supported against gravity by means of air-sleds. It may be assumed that in this situation it is the shoulder that is used to control the stability of the arm more than the elbow.

We now turn to two inherent limitations of the co-contraction analyses that we applied in the present study. The first limitation concerns the consequences of the EMG normalization procedure that was applied. We did not use controlled maximum-voluntary-contraction (MVC) measurements to convert the pre-processed EMG levels. Instead, we used individual and muscle-based median EMG-levels for this purpose. Due to this normalization procedure, our two measures of co-contraction that were used to determine the effects of accuracy demands and trunk restraint could not inform us about possible systematic CP-related elevations of phasic EMG activity. In a pilot-study, in which we did use MVC measurements to normalize EMG-data, we incidentally observed larger EMG values during task performance as compared to the MVC measurements, deeming the MVC measurements unreliable for normalization purposes. Indeed, it has been reported that whereas healthy individuals are well able to produce MVCs in isolated muscles, this ability is diminished in individuals with brain damage such as CP (Damiano et al., 2000).

Second, as our two measures of phasic co-contraction were computed from surface EMG measurements, we cannot exclude that any differences related to muscle moment arms, or in muscle force-generating ability that may be most prominent in the paretic muscles of the participants with CP, or any potential contributions to the observed co-contraction from other muscles that were not recorded in this experiment (e.g., trunk stabilizing musculature) have played a role. In addition, as we chose to use a functional, hence dynamic task, relations between applied muscle force on the one hand and measured EMG and movement kinematics on the other, may not be unequivocal as is known in static conditions (cf. Ostry & Feldman, 2003; Osu & Gomi, 1999). This matter also warrants further investigations along the lines pursued here.

Although these limitations may have affected the results of the present study, we feel confident that the measures used here may be useful as an estimate of how co-contraction in elbow and shoulder joint changes in participants with CP as a function of variations in task-accuracy constraints. In advancing our knowledge on the control signals used by the CNS for multi-joint movement control in individuals with brain damage, determinants of co-contraction should be examined in detail in this group. This study provides a first step into this endeavour. Here, we only tested the least affected arm. In future research, it should be examined whether similar phenomena can be observed for the most affected arm, as well. In order to do so, a considerably easier task should be designed, as the present task was too complex to be performed with the most affected arm.

A final note should be made. No interactions between group and accuracy were observed when examining the effects of accuracy demands on the co-contraction at the shoulder. We recognize that such interactions may have been obscured by the small group sizes of our study.

In sum, the present findings show that the commonly reported increase in trunk involvement and high co-contraction levels in CP should not exclusively be regarded as symptom of the neurological damage. Our results provide a first indication that also in individuals coping with the consequences of CP modulation of co-contraction at the shoulder is a likely mechanism to deal with variations in accuracy constraints, as suggested by several other researchers for people without neurological disorders (Gribble et al., 2003; Laursen et al., 1998). Increased trunk involvement proves a secondary reaction to these constraints.

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