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The Effects of a Temporal Warning Signal on the Biomechanical Preparations for Sudden Loading

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Summary: An experiment was performed to evaluate the biomechanical preparations exhibited in anticipation of sudden loading. Four experienced subjects received sudden loads at 1 min intervals. An analogue display was used to convey the time remaining in the 1 min intervals. The dependent measures included the electromyographic (EMG) data obtained from eight trunk muscles and the changes in body posture. These data were compared with data from a baseline session in which no timing display was available. In both sessions, when loads were anticipated the back muscles were tensed. However, with the timing display available there was an alteration in the preparatory co-contraction of the trunk muscles. The change in co-contraction was primarily due to the increased torque generated by the erector spinae (ES) as opposed to a decrease in the torque generated by the anterior muscles. This indicated that there was less stiffening of the torso during preparation when temporal information was available. During the sudden loading three of the four subjects reduced the peak compressive forces on the spine predicted via an EMG driven model while maintaining consistent levels of trunk stability.

Key Words: Sudden loading—Preparation—Low back disorders—Perturbation—Electromyography—EMG—Ergonomics.

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INTRODUCTION

Back injuries continue to plague a sizeable percentage of the population at any one time[1]. While many of these injuries may be considered cumulative trauma in nature, the onset of pain from these injuries is frequently associated with sudden unanticipated loads or sudden unanticipated body motions as in slips and falls[14,15]. When anticipated, such perturbations are less likely to trigger the sudden exertion associated with pain onset. Magora[14] suggested that the 'degree of preparedness' prior to sudden maximal efforts was a critical factor affecting the onset of occupational injuries.

Biomechanical studies have shown that the trunk muscles become quite active in resisting perturbations which affect the body's stability[5]. These studies have shown the muscle contraction patterns to be specific with regard to the direction of the applied load both in the torso[5] and in the lower extremities[4]. When subjects could anticipate the sudden loads generated by dropping weights into a box held in the hands, Marras et al.[17] found that the magnitude of the peak trunk muscle responses were on average 35% greater than that required to support the load under static conditions. This result

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showed the effects of dynamic loading. However, under conditions in which no warning was available these authors reported that the muscle responses increased by up to 300% over that observed when the sudden loading could be anticipated[17]. In another study, the peak EMG activities were found to increase by 45, 84 and 99% as the warning time decreased from 400 ms to 200 ms, to 100 ms and to 0 ms, respectively[13]. Similar findings have been reported more recently by Cresswell[7]. Greenwood and Hopkins[8] used the term 'startle response' to describe the large burst of muscle activity as subjects were dropped unexpectedly.

This startle response was absent when these subjects were dropped at an anticipated point in time. Based on the data summarized above, it was theorized that when individuals had adequate warning they would prepare themselves in such a way that the adverse biomechanical consequences of the loading or perturbation were minimized. Anticipatory postural adjustments have been measured as individuals responded to perturbations affecting the body's postural stability[3,6,18]. These studies have shown that individuals would adjust their body so as to counter the anticipated forces and moments associated with the upcoming perturbation[3].

In addition to postural adjustments, Bouisset and Zattara[4] have shown with EMG that the muscles immediately affected by the upcoming perturbation are tensed during this anticipatory period. Houk[9] suggested that this stiffening increased the responsiveness of the muscle. Alternatively, the rise in EMG possibly reflected a stiffening of the joint which was about to be loaded. Lavender et al.[13] reported that the muscles were recruited prior to the loading and that the time at which these muscles were recruited was a function of the warning time available. Similar results have been reported by Omino and Hayashi[21].

In a more complete study focussing on the development of preparatory responses, Lavender, Marras and Miller[11] exposed subjects to repeated sessions in which they received 30 sudden loads at 1 min intervals. The subjects were given no other temporal information. The subjects developed preparatory strategies which tended to increase the activation of the posterior trunk muscles prior to the load application. Elevated EMG activity was also seen in the abdominal muscles of some of the subjects. This preparatory co-contraction would potentially provide a stiffened torso response to the sudden loading. However, no consistent anticipatory postural adjustments were observed. Comparisons between the initial and final sessions indicated that these preparatory strategies were successful in increasing the stability of the torso and in reducing the estimated peak compression acting on the spine during the actual load application.

The strategies described above were developed under conditions in which the temporal uncertainty regarding the onset of the sudden loading was quite large. Several questions remained as to how these preparatory strategies would change with more accurate temporal information. Specifically, this study addressed the following two hypotheses:

1. The preparatory strategies which include muscle activations and postural adjustments will be of greater magnitude under conditions where accurate information is available which describes when the loading will take place.
2. The preparatory strategies developed with the temporal information will further increase the stability of the torso and decrease the estimated spinal compressive forces during the sudden loading.

METHODS

Approach

The current experiment used a variation of a sudden loading paradigm described in detail by Lavender et al.[11]. This paradigm allows the onset of the sudden loading to be controlled by the experimenter. The sudden loads in this experiment were applied at 1 min intervals by dropping a weight through a chute and into a bucket held in the subjects' hands. Subjects participated in multiple sessions with each session containing 30 loading cycles. The postural and EMG (muscle activation) data were sampled both prior to and during the loading.

Subjects

Four male subjects participated in the study. All were between the ages of 22 and 44. The mean (and range) height and weight were 177.0 cm (173.3–182.4 cm) and 78.0 kg (68.0–84.8 kg), respectively. The subjects each signed consent forms approved by the university's human subjects review committee and were financially compensated for their efforts.

Experimental Design

The experiment consisted of two phases. In the first phase the subjects were trained on the experimental protocol in at least five sessions to determine a baseline performance. In each training session the subjects received 30 sudden loadings separated by 1 min intervals. These loadings were comprised of a 53.4 N weight dropped 1.1 m into a bucket held by the subjects. The empirically determined force at contact was approximately 800 N, although this is largely affected by the stiffness of the individual's response. During these initial training sessions no auditory or visual cues marked the end of the interval, only the delivery of the weight. The final session from this phase served as the baseline data to which the current experimental manipulation was compared. A detailed description of the changes observed during this training phase has been published elsewhere[11]. In the second phase the same loading paradigm was continued, however an analogue timing display was used to provide visual cues as to the onset time of the next loading. This phase was continued for two sessions. The data from the second session were compared with the baseline data obtained previously to determine the effects of the warning signal. Hence, each subject served as their own control in that baseline data from the first phase were compared with the data obtained from the second phase in which the warning signal was used.

The dependent variables consisted of the EMG data sampled from eight of the primary muscles supporting the torso at the level of L5/S1. These muscles included: the left and right latissimus dorsi (LD), the left and right erector spinae (ES), the left and right rectus abdominus (RA) and the left and right external oblique (EO). Postural measures of the trunk and whole body displacements were obtained via electrogoniometer and force platform data.

Apparatus

Subjects stood in a three-sided booth-shaped structure on top of a force platform as shown in Figure 1. A bag containing 5.4 kg of lead shot served as the weight. This weight was dropped through a stove pipe into a bucket (0.77 kg) held in the hands. The level of the force platform was adjusted for each subject such that the top of the bucket was at the level of a momentary push button switch (defined as the 'ready' switch) when the elbows were at a right angle and the wrists were in a neutral posture. The total vertical distance travelled by the weight was 110 cm. A photocell affixed to a wooden frame surrounding the top of the bucket provided an analogue pulse approximately 60 ms prior to the weight contacting the bottom of the bucket.

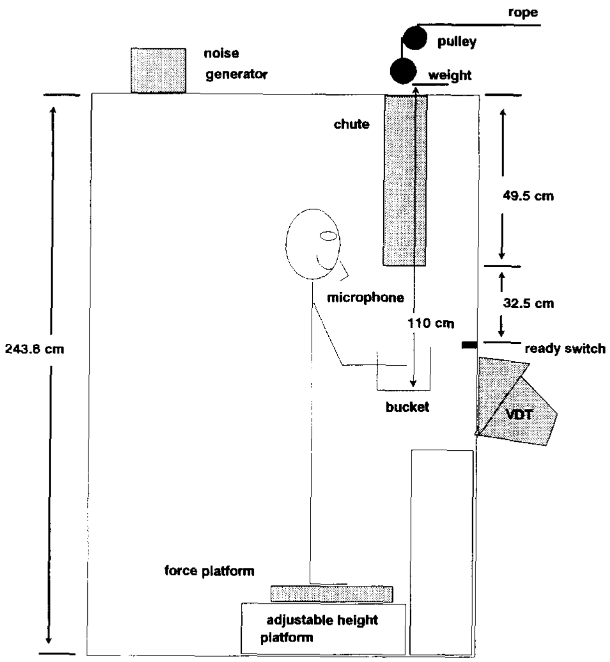


FIG. 1. The booth in which subjects stood during the experiment. The monitor in front of the subject was used to display the tracking task and the analogue warning signal (during the appropriate sessions).

Below the ready switch, a computer monitor displayed a simple tracking task and the analogue warning signal. The sole purpose of the tracking task was to occupy the subjects attention during the interval between load applications. It was anticipated by the experimenters that the tracking task would create a consistent attention load for the subjects between each loading and from one session to the next. The one-dimensional task required subjects to keep a disc-shaped object centred on the monitor's screen. Control was achieved via a voice recognition system programmed to recognize verbal directional commands. Beneath the monitor a wooden shelf allowed the subjects to rest the bucket between loadings.

A noise generator masked audio cues which could indicate when the weight would fall. Subjects were also given industrial hearing protective devices to further assure the removal of auditory cues.

EMG data were collected using Beckman bipolar surface electrodes with a 3 mm diameter sensor. The signals were passed to small preamplifiers with a gain of 1000 attached to a belt worn by the subjects. These were connected via a cable to an amplifier which had a gain of 57. The amplified signals were then rectified and integrated with a time constant of approximately 100 ms. Ground reaction forces were used to determine overall displacements in the body's centre of gravity and were collected using an Advanced Mechanical Technology Inc. (AMTI) OR6-5-1 series force platform (Figure 1). Trunk posture data were collected with the lumbar motion monitor (LMM). This device is an electrogoniometer which measures the kinematics of the torso in the sagittal, frontal and transverse planes. The unit was strapped over the spine by means of a rigid harness system. The final data sources were event markers wired to provide an analogue pulse to mark: (a) when the subject pushed the ready switch, and (b) when the weight broke the photobeam on its way into the bucket. All signals were passed through the A/D board and sampled by the computer at a rate of 100 Hz and with a 5 mV resolution.

Procedure

Electrodes were applied to each of the four bilateral muscle pairs along the line of action for a given muscle with an inter-electrode spacing of 2 cm. The specific placements were as follows: (a) the ES electrodes were centred halfway between the L3 and L4 spinous processes and approximately 3 cm lateral from the midline; (b) the LD electrodes were placed over the belly of the muscle at the level of T7 and approximately 13–15 cm lateral from the midline; (c) the RA electrodes were placed at the level of the umbilicus 2 cm lateral from the midline; (d) the EO electrodes were placed at the level of the umbilicus and approximately halfway between the iliac crest and the anterior superior iliac spine.

Subjects received instructions which stressed the relative emphases on the two experimental tasks: receiving the periodically falling weights and the tracking task. Subjects were told that every minute a weight would fall. Their task was to allow the weight to drop into the bucket and hold for approximately 3 s. One minute following the presentation of the previous weight another would be delivered. In between receiving weights the subjects were asked to perform the verbally controlled tracking task. Where the warning signal was not present, subjects were instructed to estimate when most of the 1 min time interval had elapsed (the subjects were not permitted to wear watches). Both with and without the warning signal the secondary tracking task was stopped by saying 'pause' into the microphone, the subjects were instructed to press the ready switch with the front of the bucket and to prepare for the falling weight. The ready switch sent a pulse to the computer indicating when the subject began preparing for the load.

Unknown to the subjects, data were only collected and stored during the final session of each phase. In order to conceal which sessions were the actual data collection sessions, the EMG electrodes were applied to standard muscle sites as described in Lavender et al.[12] at the beginning of every session. The distance between electrodes was approximately 2 cm along a line parallel with the muscle fibres. The electrode collars were circled with magic markers so similar pick-up sites could be used each day.

During the final testing sessions with and without the warning signal the subjects were asked to perform two isometric maximal voluntary contractions (MVCs). The EMG data obtained during these attempted trunk flexions and trunk extensions (while in an upright posture) were used in normalizing the EMG values from their respective sessions. Resting EMG values were obtained as subjects stood in a relaxed posture.

Prior to each session the subjects were fitted with the LMM. Reference values were obtained by having the subject stand in an upright neutral posture.

After the setup procedures the subject was asked to step into the experimental chamber. At this time the subject was given the hearing protective devices and the subject put on the microphone headset. The subject received a signal indicating when the first 1 min interval started. The subject then initiated the tracking task with a verbal command and for the next 30 min received the weight at 1 min intervals. It should be noted that none of the subjects reported fatigue in the trunk muscles following any of the sessions.

Data Treatment

Data were divided into two stages, the pre-load stage (PLS) and the sudden loading stage (SLS). The data were collected from both phases by having the computer initiate sampling at 48 s into the 1 min period. The sampling continued through the weight drop at 60 s into the SLS for an additional 3 s. This is shown schematically in Figure 2.

The EMG data were converted into sagittal plane muscle torques. This representation of the EMG data was selected because it facilitated the comparison of muscle activation levels between different muscle groups within an individual. By taking into account the physiological parameters such as cross-sectional area, moment arm and lines of action, this expression of the data provides an estimate of the overall contribution to the net restorative moment made by each muscle group. Thus, the co-contraction can be better described. Moreover, since this procedure uses anthropometric data (trunk breadth and trunk depth) in computing the moment arms and cross-sectional areas, some of the variation inherent in EMG data due to anthropometric differences between the subjects has been removed. It is important to note that while this conversion of the EMG data to torque values may introduce some error, the error is consistent within each individual.

The first step of this conversion required that the EMG signal first be normalized as shown in Eqn 1.

$$\mathrm{NEMG}_{ij} = (\mathrm{obs}_{i,j} - \mathrm{rest}_{i,j})/(\mathrm{max}_{i,j} - \mathrm{rest}_{i,j}) \quad (1)$$

where i = muscle 1 to muscle 8, j = test phase: baseline or warning signal, $\mathrm{obs}_{i,j}$ = the current IEMG value of muscle i, $\mathrm{rest}_{i,j}$ = the minimum resting IEMG value of muscle i, $\mathrm{max}_{i,j}$ = the maximum IEMG value from muscle i obtained from either MVC.

The NEMG were converted to muscle torques by using the force limitation of 50 N cm^{-2} for muscle tissue (upper bound used by McGill, Patt and Norman)[19] in the following expression:

$$\mathrm{torque}_{i,j} = \mathrm{NEMG}_{i,j}{*}\ \mathrm{Xsect}_i{*}50\ \mathrm{N\ cm}^{-2}{*}\mathrm{MA}_i{*}\mathrm{Cos}(\mathrm{X}_i) \quad (2)$$

where i = muscle 1 to muscle 8, j = test phase: baseline or warning signal, $\mathrm{NEMG}_{i,j}$ = normalized EMG signal computed from Eqn. 1, Xsect_i = muscle i's cross-sectional area which was computed based on coefficients employing torso anthropometric data and constants provided in the literature[22]. MA_i = the moment arm for muscle i (also obtained using the models of Schultz et al.[22]. X_i =

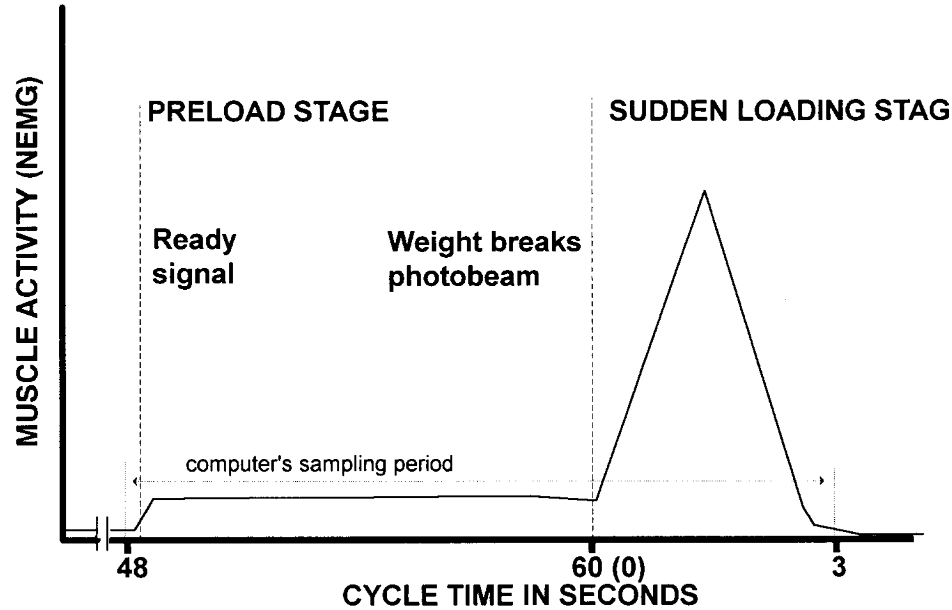


FIG. 2. An example of the data obtained from a single muscle for one loading cycle. The preparatory stage was initiated when the subject pushed the bucket against the momentary switch ('ready signal'). During this stage typically the muscle response was elevated above the resting value and was relatively constant. This was maintained until the sudden load was applied as indicated by the break in the photobeam.

the angle of a muscle *i*'s line of action relative to the upward vertical.

This relationship assumes a linear relationship between the NEMG and torque. Others have reported a relatively linear relationship between predicted ES muscle tension and myoelectric activities up to moderate exertion levels[23]. The torque estimated for each muscle is a linear transformation of the muscle force data. Although recent data indicate that the RA force may be best described with a power function in which the EMG values are raised to an exponent slightly less than 1.0[24].

The muscle torque during the PLS was characterized by a steady state value which was elevated relative to baseline values (prior to the PLS). Therefore, the individual muscle torques were averaged over the PLS. As the loadings were symmetric about the body's mid-sagittal plane, the corresponding muscle torques from the bilateral muscle pairs were summed. These sums are presented here. The SLS was characterized by a sharp increase in the muscle torques immediately following the weight contact. The EMG data from the SLS were used in an EMG driven model developed by Marras and Sommerich[16] to predict the peak compression acting on the spine during the loading.

The sagittal plane centre of gravity (COG) displacement was obtained as follows:

$$\text{sagittal COG}_t = \text{My}_t/\text{Fz}_t \qquad (3)$$

where My_t = the moment measured by the force-platform about an axis parallel with the body's frontal plane as a function of time, Fz_t = the vertical reaction force measured by the force-platform as a function of time.

The changes measured here could be due to the displacement of any body segment or a combination of body segments. PLS data from the force platform and the LMM were obtained from when the ready button was pressed to the point at which the weight broke the photobeam. For a given loading cycle the COG displacement and the lumbar position changes were computed by taking the difference between the final and initial values. During the SLS the change in these measures was defined as the difference between the final value of the PLS and the most extreme point immediately following the weight contact.

Only the data from the even numbered trials were used in the present analysis. Therefore, from each session, 15 trials were potentially available for analysis. Since this study focused on preparatory responses, trials with less than a 2-s preparatory period (without the warning signal) or less than a 1-s preparatory period (with the warning signal), as defined by when the subject pushed the ready switch, were not included in the analysis.

Due to the small number of subjects and the individual differences observed in the preparatory responses, each subject's data were evaluated separately. Multivariate and univariate analyses of variance (MANOVAs and ANOVAs) were conducted on the net muscle torques and the postural measures within each subject to determine whether there were significant changes over the experimental sessions. Similarly, ANOVAs were conducted on the data collected following the sudden loading to compare whether these responses were changed with the presence of the warning signal.

RESULTS

The NEMG and LMM data for a typical loading event when the warning signal was present is shown in Figure 3. Note the elevated EMG activities of the ES and the left EO prior to the weight breaking the photobeam. Also note that the subject, while originally leaning forward approximately 12°, shows a small trunk extension motion prior to the weight contacting the bucket.

A summary of the statistical analyses performed on the PLS muscle torques is shown in Table 1. The MANOVAs in which the four muscle torques were combined showed highly significant effects for all subjects due to the addition of the warning signal. This indicates that all of the subjects changed their muscular preparation strategies in response to the warning signal display. The form of this alteration in the muscle activation strategies is best identified by the outcome of the univariate ANOVAs also listed in Table 1.

The magnitude of the ES muscle torques were quite variable across the four subjects (Figure 4a). The values ranged from just over 2 Nm to greater than 12 Nm. The standard deviations indicated that the variability within a condition for each subject was quite small. In response to the warning signal all subjects showed a similar trend in which the pre-load ES muscle torque increased. This change was statistically significant ($P < 0.01$) in three of the four subjects.

While statistically significant changes were observed in the LD torques, the overall magnitude of these muscle torques shown in Figure 4b, were

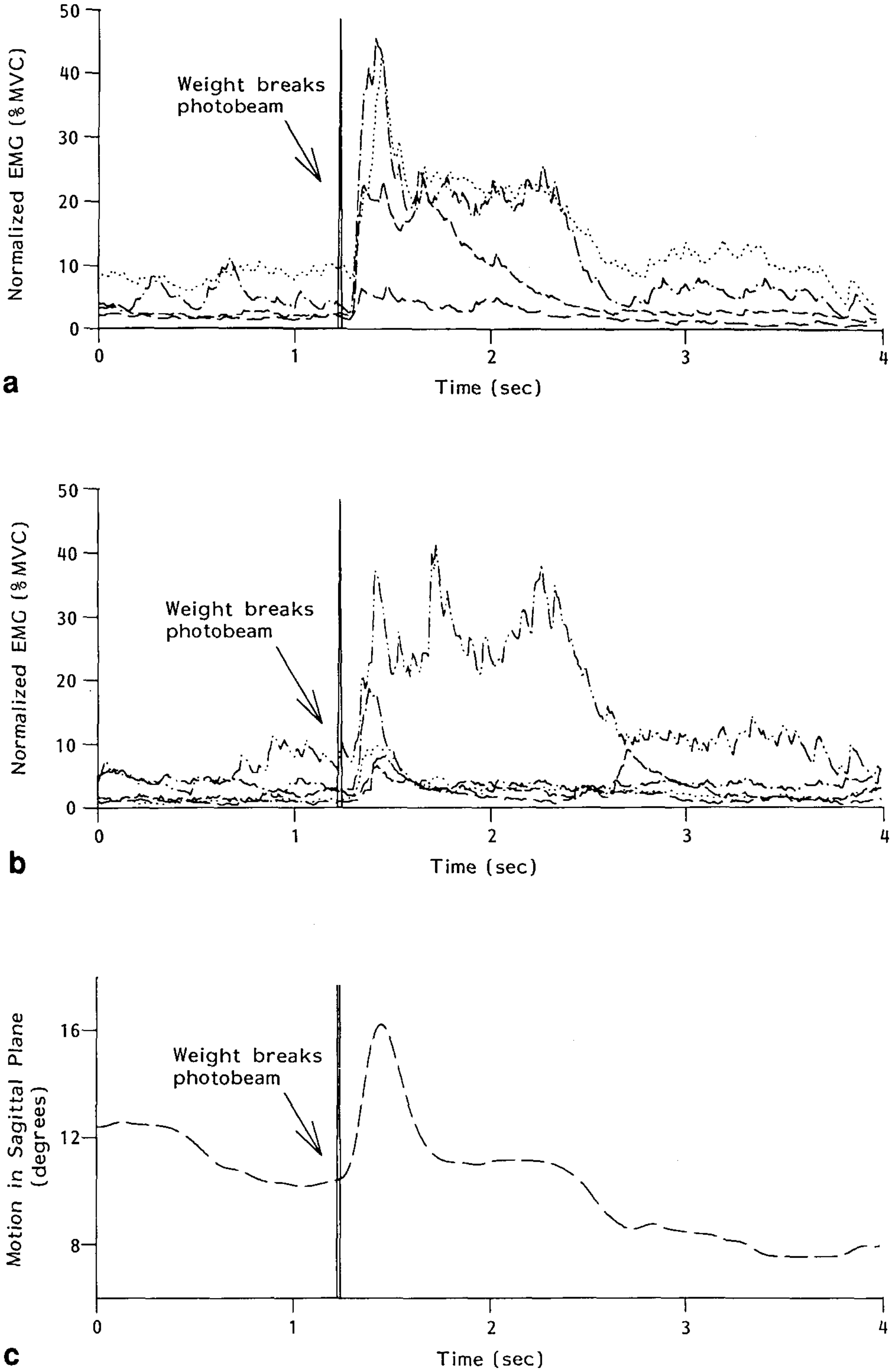


FIG. 3. Typical data from one of the subject 3's responses to the sudden load when the warning signal was available. **a**, Normalized EMG response from the four posterior muscles. The spike at approximately 1.2 s represents the weight breaking the photobeam. ——— right latissimus dorsi; – – – left latissimus dorsi; · · · · right erector spinae; – · – · – left erector spinae. **b**, data from the four anterior muscles. – – – right rectus abdominus; · · · · left rectus abdominus; – · – · – right EO; – ·· – ·· – left EO; **c**, sagittal plane motion obtained from the LMM over the same time period.

TABLE 1. *Summary of the results from the MANOVA and ANOVA procedures performed on each subjects data*

MANOVA	Subject 1	Subject 2	Subject 3	Subject 4
F ratio	13.6	13.8	42.7	177.9
df	4, 18	4, 14	4, 22	4, 16
P value	<0.001	<0.001	<0.001	<0.001
Erector spinae				
F ratio	45.94	2.58	62.76	22.18
df	1, 21	1, 17	1, 25	1, 19
P value	<0.001	ns	<0.001	<0.001
Latissimus dorsi				
F ratio	28.39	16.76	25.21	9.29
df	1, 21	1, 17	1, 25	1, 19
P value	<0.001	<0.001	<0.001	<0.007
Rectus abdomini				
F ratio	28.29	30.80	129.36	398.65
df	1, 21	1, 17	1, 25	1, 19
P value	<0.001	<0.001	<0.001	<0.001
External oblique				
F ratio	0.09	10.89	25.68	5.83
df	1, 21	1, 17	1, 25	1, 19
P value	ns	<0.005	<0.001	ns
Centre of gravity				
F ratio	15.25	0.08	22.86	0.88
df	1, 21	1, 17	1, 25	1, 19
P value	<0.001	ns	<0.001	ns
Trunk extension				
F ratio	76.78	0.95	2.38	0.53
df	1, 21	1, 17	1, 25	1, 19
P value	<0.001	ns	ns	ns
Estimated compression during SLS				
F ratio	28.79	6.63	0.40	7.41
df	1, 21	1, 19	1, 25	1, 19
P value	<0.001	<0.020	ns	<0.020
Trunk flexion during SLS				
F ratio	11.07	0.31	45.60	0.01
df	1, 21	1, 17	1, 25	1, 19
P value	<0.005	ns	<0.001	ns

Each statistical test is summarized by the *F* ratio, degrees of freedom (*df*) and by the significance level (ns = nonsignificant or $P > 0.05$). All tests on data from the pre-load stage (PLS) except where indicated.

very low. In fact, these values averaged only 3.6% of the ES preparatory torque. The greatest LD torque was observed in subject 2 during the warning signal condition, although the variability in this response was also quite high.

The torques from the anterior muscles indicated that these muscles were co-contracted during the PLS. The activation of the RA muscles (Figure 4c) was affected by the warning signal in all four subjects, although no consistent trend could be distinguished. Even after a 23% reduction in RA torque in the warning signal condition, subject 1's response was still better than twice that observed in any of the other subjects when the warning signal was present.

A consistent trend was not seen in the EO muscles' recruitment (Figure 4d). While statistically significant changes were observed, only two of the four subjects appeared to have activated these muscles to any level of consequence during the PLS. Of these, one subject reduced his EO torque and the other subject's EO torque was unchanged when the warning signal was present.

The changes in the postural measures between the baseline and the warning signal conditions are shown in Figure 5. The COG displacements showed a posterior trend in three of the four subjects, although this change was statistically significant (see Table 1) in only two of these subjects (Figure 5a). In part this posterior shift in the COG may have

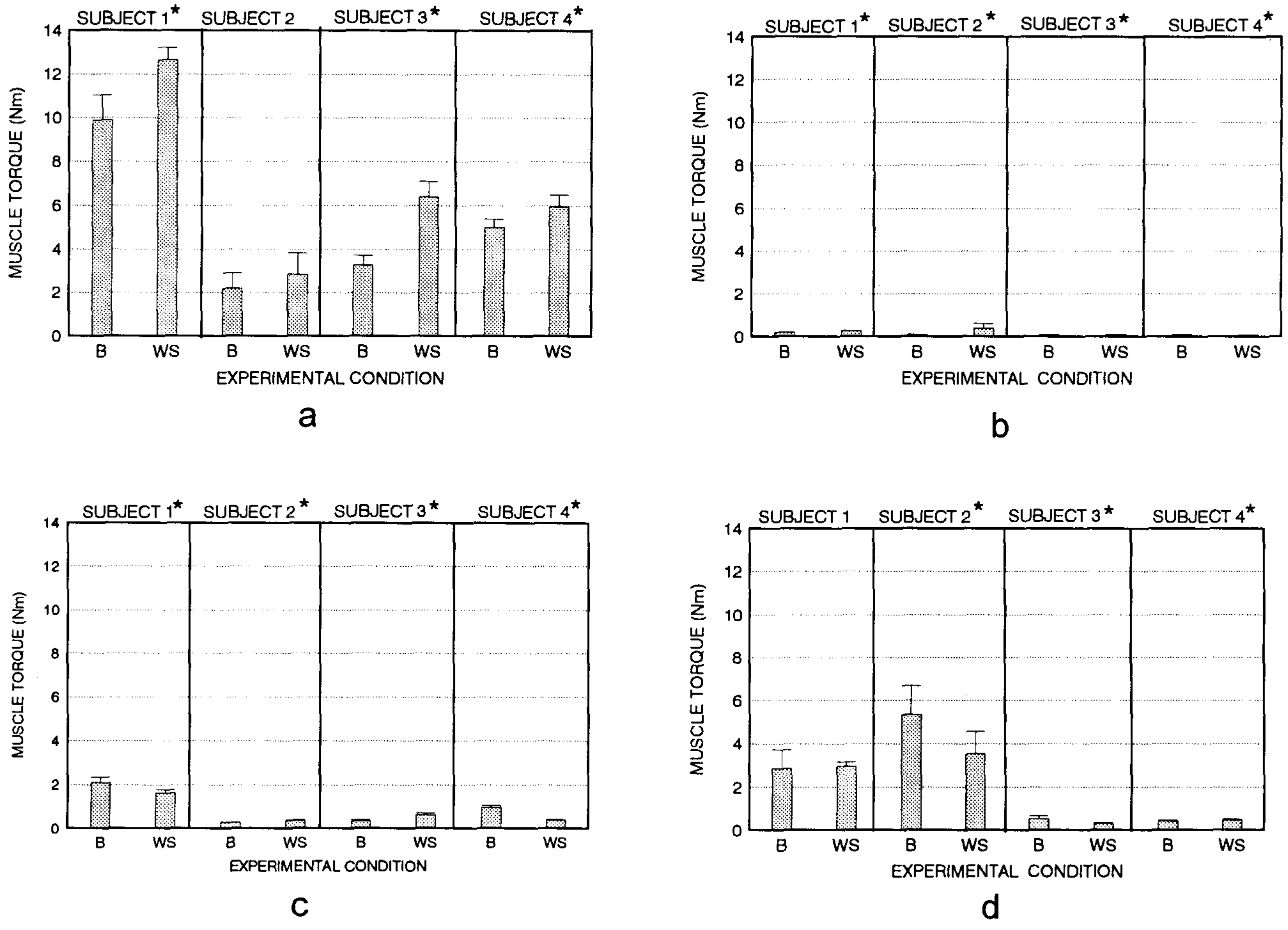


FIG. 4. The mean and standard deviation of the summed left and right muscle torques from **a**, the erector spinae, **b**, the latissimus dorsi, **c**, the rectus abdomini, and **d**, the external oblique for each subject during the pre-load stage. The vertical lines separate each subject's data. The subjects which showed statistically significant differences ($P < 0.01$) between the baseline (B) and the warning signal (WS) conditions are marked with an asterisk.

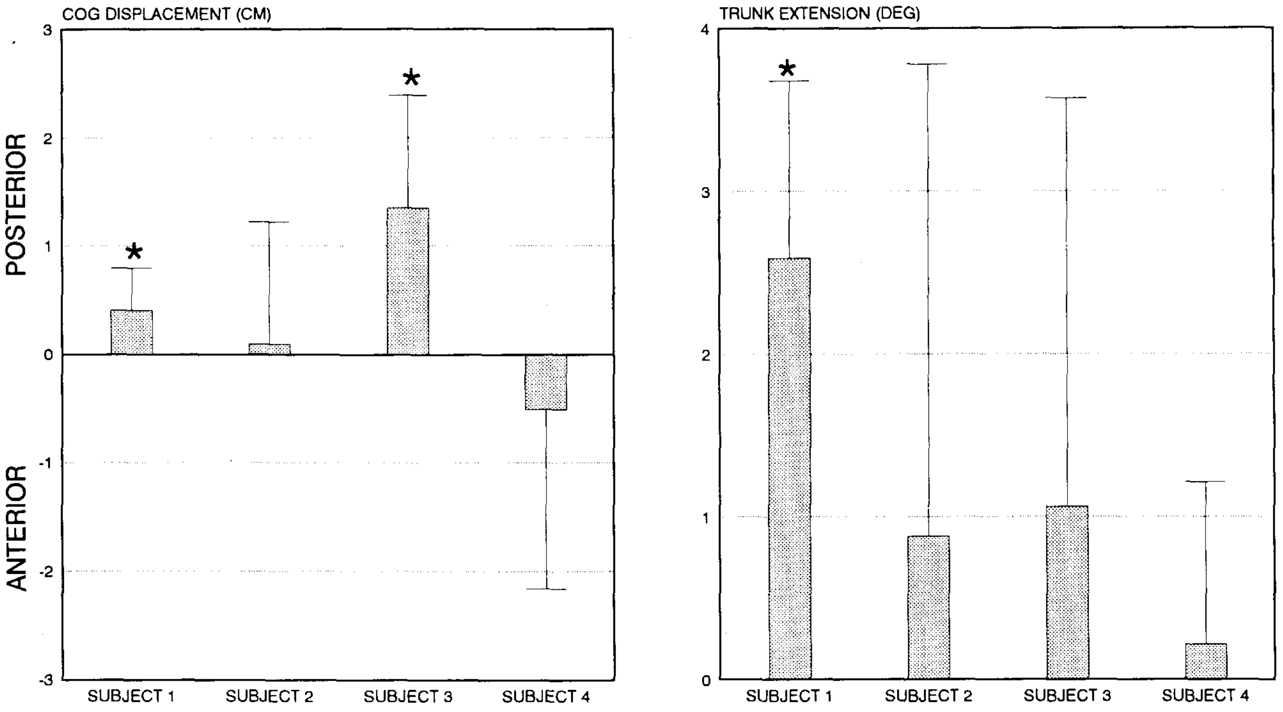
been due to increased trunk extension (Figure 5b). However, the variability in the postural measures was quite large. This was especially true for the trunk extension data.

The effectiveness of these preparations was evaluated in terms of the peak spine compression and trunk stiffness (as determined by the trunk flexion) during the sudden loading stage. Figure 6a shows the peak compression estimated with the Marras and Sommerich EMG driven model. Note that in three of the four subjects the estimated compression forces at L5/S1 decreased significantly (see Table 1). These values were reduced by 16% for subject 1 and by 11% for subjects 2 and 4.

Most of the mean trunk flexion values during the SLS were between 4 and 6° regardless of the experimental condition (Figure 6b). There appeared to be no consistent trend in the trunk flexion data (trunk stiffness) due to the presence of the warning signal. Subject 3 significantly decreased his SLS flexion response to resemble that of the other subjects when the warning signal was available, however, subject 1 showed a small but statistically significant increase in trunk flexion under the same condition. The remaining two subjects showed little change in this measure.

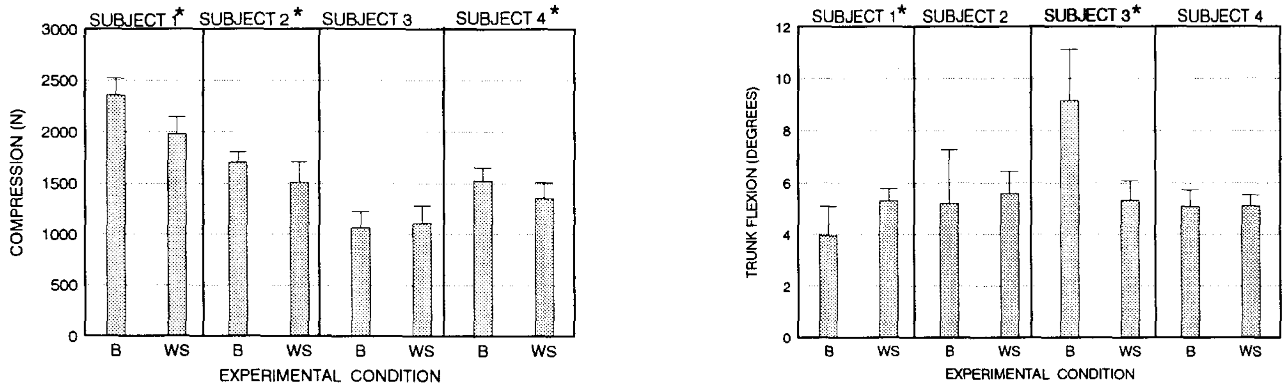
DISCUSSION

The results indicate that the availability of accurate temporal information did indeed produce changes in the preparatory strategies employed by the four subjects. The most pronounced changes were in the torques developed in the ES muscles during the PLS. This increase in the posterior muscle torque during the warning signal condition combined with



a b

FIG. 5. **a**, The mean change in the pre-load COG positioning in cm, and **b**, pre-load trunk extension in degrees between the baseline and the warning signal conditions. The standard deviations of these differences are also shown. Subjects showing significant ($P < 0.01$) changes are marked with an asterisk.



a b

FIG. 6. Evaluation of the torso's response to the sudden loading. **a**, The estimated peak spine compression during the SLS was obtained by using the NEMG data in the Marras and Sommerich (1991) EMG driven model. **b**, The peak forward trunk flexion was detected with the LMM during the SLS. Subjects showing significant ($P < 0.05$) changes are marked with an asterisk.

a small decrease in the anterior torque suggest an overall shift in the preparatory strategy employed. It is theorized that the effectiveness of employing these new strategies during the PLS is suggested by the significant reduction in the peak spine compression experienced by three of the four subjects during the SLS. Although, mixed effects were seen with regards to the torso stability.

Across the four subjects the ES output in the warning signal condition increased between 19 and

98%. We conclude that it was primarily this increased ES torque (Figure 3a), as opposed to the small decrease in RA and EO torques (Figures 3c and 3d), which was responsible for the decrease in the co-contraction of the trunk flexors relative to the trunk extensors. A simultaneous activation of the agonist and antagonist muscles would likely serve to stiffen the spine. However, the current findings suggest that with the temporal information available from the warning signal display, subjects shifted their muscle recruitment toward a strategy in which there was less preparatory stiffening (i.e. less co-contraction). This contrasts with Cresswell's[7] findings which did find greater anterior muscle activity and co-contraction during sudden loadings triggered by the subject relative to loadings where the temporal onset could not be anticipated.

The question remains as to the function of the increased ES torque. The ES activation is most likely a reflection of the anticipated flexion moment. Some of the resulting muscle torque could be used to alter the trunk posture prior to loading. However, if this were true then torso extension motions accompanied by a posterior shift in the body's COG should have been consistently observed. The changes in COG and trunk displacements in Figure 5 do not provide much support for this conclusion. Therefore, we hypothesize that this torque is elevated to reduce the response latency of the primary muscles resisting the expected perturbation.

The limited use of postural preparations during the PLS is likely in part due to the nature of the experimental task. The literature suggest that a rearward shift should be observed in the body's COG in response to the anterior loads[4]. The loading task used in the current study required the bucket to be held under a chute. Any rearward shift in the torso would increase the extension of the shoulders in order to maintain the bucket position. Such a change would have two adverse effects. First, the shoulders would be held in a more extended static posture. And second, the moment imposed by the sudden loading would increase as the distance between the load and spine increased. It is also possible that the differences between subjects could be attributed to variations in the initial posture. The postural change was computed based on the difference between the posture at the moment the ready switch was pressed and the sample prior to the weight contact. If subjects had adjusted their posture prior to indicating they were ready this change would not have been detected.

The minimal recruitment of the LD muscle groups during the PLS was not anticipated (Figure 4b). These muscles have been shown to be very active during the SLS in earlier studies[13,17]. This discrepancy is likely due to the dual role these muscles play in shoulder and trunk extension. If these muscles were to contract during the PLS to stabilize the torso, they would increase the load on the shoulder as additional force from the shoulder flexors would be required. These muscles are only well suited to function as trunk extensors when the posture of the shoulder is stabilized or when shoulder extension is also required, for example, when holding a load against the body or during a pulling task. The task employed in the current study required subjects hold the bucket under the chute when the load was anticipated. This did not allow for the arm posture to be fixed without significant co-contraction in the shoulder, hence nominal LD pre-load torque.

There were limitations in the experimental approach which need to be addressed. First, the extent to which crosstalk affected the EMG recordings was not specifically determined. During the PLS it is believed that this effect was minimal for the posterior muscles since the root mean square (rms) signals obtained from the ES were of a substantially greater amplitude than those of the LD. Similarly, the rms values from the EO appeared to be independent of the RA activations. During the SLS it is possible that crosstalk contributed more significantly to the results. Surface EMG data published by Koh and Grabiner[10] for the vastus lateralis (VL) and the lateral hamstrings (LH) suggest that the crosstalk should be less than 10% when the muscles are contracted at submaximal levels. Second, this study does not compare the effects of a temporal warning signal with the situation in which the subject receives no warning at all. In the control condition the 60 s loading cycle allowed individuals to estimate approximately when the loading would occur and initiate their preparatory responses prior to the loading, although there was temporal uncertainty. However, the effects of a sudden unexpected loading have been described previously[13]. And third, in this paper a linear relationship between EMG and torque is assumed. The use of the linear function has been shown to be a reasonable approximation with submaximal loads[23]. Given that the true relationship is curvilinear for most muscles[2], assuming a linear relationship between EMG and torque did introduce some error in the muscle force estimation and therefore in the muscle torques as well. The linear approximation, however, will have only a minimal effect on the muscle torques computed from the low amplitude EMG signals observed during the PLS.

Even though the data presented here shows a large degree of variation between individuals in the magnitude of their preparatory responses, there were many trends which indicated that the strategies were converging during the warning signal session. For example, the ES torque increased in all subjects; all subjects showed very low LD torque; the subjects with the higher RA torque showed decreases and vice versa; the one subject who used much greater EO torque during the baseline session than the others showed a very significant reduction and most subjects showed a similar trend in their postural preparation towards a more posterior COG displacement (Figure 5). Furthermore, the reduced co-contraction between the anterior and the posterior muscles seen across the four subjects indicates a more global change in the preparatory strategy. More specifically, the data suggest that the preparatory muscle recruitment may support a strategy similar to those employed with rapid ballistic motions or skilled activities. These exertions are frequently associated with more of a sequential pattern of agonist-antagonist contraction as opposed to a simultaneous activation of the antagonistic muscle groups[20,25]. In summary, this change in strategy suggests that in the presence of an external temporal warning signal, the biomechanical system supporting the spine is more efficiently prepared to interact with the external load. Furthermore, these results indicate that the variance in preparatory strategies between individuals is reduced when accurate warning information is available. This is promising news for those designing systems in which biomechanical preparations may be required to perform a given task safely.

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