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Does postural chain muscular stiffness reduce postural steadiness in a sitting posture?

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Abstract

This study investigated the effect of postural chain muscular stiffening on postural steadiness when it is rhythmically perturbed by respiration. It consisted of an analysis of centre of pressure (CP) displacements when constant sub-maximum pushing efforts were performed in a sitting posture. Muscular stiffness, assessed by surface electromyography (iEMG), was imposed at two controlled levels, using two intensities of pushing effort (20% and 40% of the maximum voluntary contraction: 20MVC and 40MVC). Lumbo-pelvic mobility was varied using two different support areas at the seat contact (100% and 30% of the ischio-femoral length: 100BP and 30BP). Respiratory disturbance to posture was varied using two respiratory rate conditions (quiet breathing (QB), which is the spontaneous rate, and fast breathing (FB) at a rate imposed by a metronome).

The results demonstrated that an increased push effort was associated to a higher iEMG level, and induced greater mean deviation ($\bar{X}_p$) and sway path (SP) of antero-posterior CP displacements. It was concluded that postural muscle stiffness reduces postural steadiness. It was suggested that it could be related to a weaker compensation of respiratory disturbance to body posture.

Keywords: Posture; Steadiness; Respiration; Muscular stiffness; Spine mobility

1. Introduction

According to Newton's laws, a physical system is in static equilibrium if the sum of the external forces and that of their moments are equal to zero. In human posture, this means that the sum of gravity and support reaction forces amounts to zero. Unlike gravity, support reaction forces vary continuously, as internal forces, such as those inducing respiratory or cardiac cyclic movements, are transmitted from their origin to the body contact surfaces. This is why, when a given posture is maintained, the body is considered to be in dynamic, and not static, equilibrium. In other words, the internal perturbing forces must be compensated for at all times to maintain postural equilibrium, i.e. to keep the projection of the centre of gravity within the boundaries of the support base.

Posturo-kinetic capacity (PKC) was defined as the capacity to develop a counter-perturbation to the posture perturbation and therefore to limit its negative effects on body stability [1]. It was recently assumed to be a dynamic process that depends on postural chain mobility [2], and that postural chain mobility results from anatomical and physiological factors. From an anatomical viewpoint, the mobility of an articulated chain is a function of the range of individual joint movements, which results from their anatomical structures, primarily from joint structural stiffness, a "passive" biological characteristic: it determines the range of motion capacity. From a physiological viewpoint, the *dynamic* mobility of an articulated chain is a function of the muscular properties, which, for a given excitation pattern, results in muscular tension (and stiffness), and then corresponds to an "active" property. Joint movements during postural maintenance have been reported

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to be less than one degree [3]. Hence, the anatomical range of joint motion reduction could impair counter-perturbation mechanisms in very specific conditions. In contrast, a stiffening of the postural chain could impair the dynamic counter-perturbing movements. More precisely, it could restrict respiratory disturbance compensation, which was reported to involve spine mobility [3–5]. In their study of low back pain patients, Hamaoui et al. [6] examined the question in depth, and proposed to explain the lesser compensation observed in the patients [7,8] by an increase in muscular tension. This study aimed to clarify this hypothesis, and assessed the postural chain muscular stiffening effect on postural stability, in relation to respiratory disturbance compensation.

2. Methods

2.1. Subjects

Ten healthy male subjects (mean age: 25 ± 5 years; mean weight: 650 ± 5 N; mean height: 177 ± 5 cm) participated in the experiments. None had suffered any musculo-skeletal, neurological, respiratory or vestibular disorder. They gave their informed consent and the experiments were conducted in accordance with legal requirements (Huriet's law).

2.2. Experimental set-up

A custom-designed seat, composed of three six-channel force plates (one located under the seat, and two under the feet) linked by a rigid frame, which measured reaction forces along three Galilean axes, was used to calculate the antero-posterior co-ordinates of the centre of pressure (CP) at the seat and foot levels. A dynamometric bar, equipped with force transducers, measured the horizontal force (F_x) exerted by the subjects. An oscilloscope connected to the force transducers was installed on the frame, at the subject's eye level to supply visual feedback. Respiratory kinematics was measured by inductive plethysmography (Respitrace Plus), assessing thoracic and abdominal perimeters with two sensing belts. This method, which uses electromagnetic recording, has been described in detail in another article [9]. Data were sampled at 50 Hz with an A/D converter and stored on a PC for off-line analysis.

In order to check muscular tension along the postural chain, six subjects underwent EMG recordings of five postural muscles, located at the trunk and the thigh: erectores spinae at the lumbar (ES-L) and thoracic (ES-T) levels, serratus anterior (SA, primum movens), rectus abdominis (RA) and rectus femoris (RF). They were assumed to constitute a representative sample of postural muscles, according to previous data [10]. EMG signals were recorded from the dominant side by bipolar surface electrodes. Inter-electrode impedance was less than 5 kΩ. EMGs were amplified with differential amplifiers (frequency bandwidth from dc to 10 kHz). Individual EMGs were full-wave rectified, and digitized with a sampling rate of 1000 Hz. EMG signals were rectified in order to calculate the integrated EMG (iEMG) over 2 s successive intervals for each 30 s trial. To allow comparison between the two levels of force and subjects, iEMGs were expressed as a percentage of the values displayed during the maximum voluntary contraction (MVC) effort, which was assessed during the pre-test series. The data were analysed using one-way repeated measures analysis of the variance technique (Sigmastat® software), after passing normality test of Kolmogorov-Smirnov.

2.3. Procedure

A paradigm based on bilateral isometric pushes in sitting posture was used, because it offers two main advantages: (i) as the hands are gripping a dynamometric bar, the upper body constitutes a closed-chain configuration, which results in a reinforcement of the role of the spine and pelvis in the respiratory disturbance compensation; (ii) the push effort involves an increase in postural muscles activity [10], and then, as the effort is isometric, their tension is increased, which is easily evaluated by surface EMG.

The experimental factors referred to breathing rate, push effort and seating conditions. Two breathing rates were considered: spontaneous quiet breathing (QB), and fast breathing (FB) imposed at 0.33 Hz, with the help of a metronome. Fast breathing was used to highlight weaker postural steadiness by an increase of respiratory disturbance to posture. Two submaximal levels of isometric push effort were used: 20% (20MVC) and 40% (40MVC) of the maximal voluntary contraction (MVC). Two seating conditions were imposed: 100% (100BP) and 30% (30BP) of ischio-femoral contact with the seat, given that the latter was known to provide greater mobility of the pelvis [11], and thus of the lumbar spine with which it articulates.

The subjects were requested to sit upright on the custom-designed seat, which is adjustable to individual anthropometric characteristics, with their thighs horizontal, trunk and legs vertical, upper limbs stretched out horizontally and hands gripping a dynamometric bar located at shoulder level. First, they had to exert three brief maximal voluntary pushing efforts separated by 120 s rest intervals, in 100BP and in 30BP, in order to determine their MVC. Then, they were asked to perform series of five 30 s trials of pushing effort in the six conditions of effort (20MVC/40MVC), respiratory rate (QB/FB) and seat contact area (100BP/30BP). They were instructed to exert a constant pushing effort, using the visual control of the oscilloscope, and to breathe at the imposed condition. The recording was set off after the pushing effort and the breathing rate were stabilized, to exclude the transient phase of the effort. The rest time was 1 min between trials, and 5 min between series. The order of the experimental series was randomly assigned to prevent an order effect.

2.4. Data analysis

Two classical posturographic parameters, representing global (seat plus feet) CP displacement were calculated: the mean CP deviation along the antero-posterior axis ($\bar{X}_p$), and the CP total excursion along the antero-posterior axis, that is the sway path (SP). Standard deviation of the antero-posterior force exerted on the dynamometric bar (ΔF_x) was also calculated.

The data were analysed using a factorial analysis of variance (ANOVA), after passing normality test of Kolmogorov-Smirnov. ANOVA was carried out for each experimental factor, i.e. ischio-femoral contact area, push effort and respiratory rate, using Statistica® software. Significant statistical difference was set at a minimum of $p < 0.05$.

3. Results

The most striking characteristic is that SP (Fig. 1), $\bar{X}_p$ and ΔF_x, were all sensitive to push force, displaying greater values when the push effort was greater (40MVC versus 20MVC) (Table 1). Raw data of CP displacements exhibited larger displacements, mainly along the antero-posterior axis (Fig. 2). Conversely, neither postural indices nor ΔF_x, significantly differed when the ischio-femoral contact was modified (100BP versus 30BP).

$\bar{X}_p$ and ΔF_x did not vary with respiratory rate, while SP displayed greater values in fast breathing ($p < 0.05$) (Table 1). In addition, the analysis of variance showed no interaction effect between the three experimental factors.

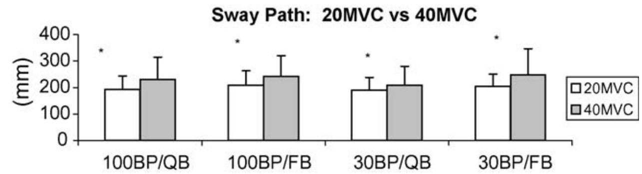

Fig. 1. Sway path (SP; mm) of antero-posterior CP displacements, pushing at 20% (20MVC) vs. 40% (40MVC) of the maximum voluntary contraction, in four experimental conditions: 100BP and 30BP represent 100% and 30% of ischio-femoral contact, QB and FB represent quiet breathing and fast breathing. Mean and standard deviation are represented. The asterisk (*) indicates a statistically significant difference at $p < 0.05$.

The EMG data demonstrated that the excitation level of five muscles increased in 40MVC as compared to 20MVC condition ($p < 0.01$ for ES-T, <0.05 for RA, <0.05 for SA and <0.001 for RF, non-significant for ES-L) (Fig. 3). Fig. 4 represents a typical full-wave rectified EMG of ER-T and RA in the two pushing effort conditions.

4. Discussion

4.1. Postural steadiness and muscular stiffness

This study offers original results, which can be discussed in relation to the influence of postural chain stiffening on body steadiness. First of all, it is clear that postural chain stiffening occurs during constant isometric pushes. On the one hand, electromyographic activity has been shown to increase continuously in postural muscles during transient push efforts performed up to the maximum, i.e. up to MVC [10]. Our EMG data confirm this result for two submaximal constant pushes: postural muscles are stimulated and the excitation level increases when the external force is

Table 1
Effect of ischio-femoral contact area, pushing effort and breathing rate

	100BP			30BP		
	20MVC	40MVC	p	20MVC	40MVC	p
$\bar{X}_p$ (mm)						
QB	1.05 ± 0.55	1.68 ± 0.94	*	1.19 ± 0.51	1.65 ± 0.90	*
FB	1.09 ± 0.53	1.60 ± 1.00	**	1.24 ± 0.53	1.72 ± 0.86	*
p	NS	NS		NS	NS	
SP (mm)						
QB	192 ± 51	230 ± 85	*	190 ± 48	219 ± 66	*
FB	209 ± 54	242 ± 78	*	205 ± 46	233 ± 58	*
p	*	*	*	*		
ΔF_x (N)						
QB	0.08 ± 0.04	0.10 ± 0.04	*	0.08 ± 0.02	0.10 ± 0.03	*
FB	0.08 ± 0.03	0.10 ± 0.03	*	0.08 ± 0.03	0.11 ± 0.03	*
p	NS	NS		NS	NS	

Experimental conditions: 100BP and 30BP represent sitting with 100% and 30% of ischio-femoral contact, 20MVC and 40MVC represent pushing effort at 20% and 40% of the maximum voluntary contraction, QB and FB represent quiet breathing and fast breathing. *Indices*: $\bar{X}_p$ (mm), mean deviation of the antero-posterior center of pressure (CP) displacements; SP (mm), sway path of the antero-posterior CP displacements; ΔF_x (N), standard deviation of the horizontal pushing effort. Mean ± standard deviation are represented. NS: non-significant.

* Statistically significant difference, $p < 0.05$.

** Statistically significant difference, $p < 0.01$.

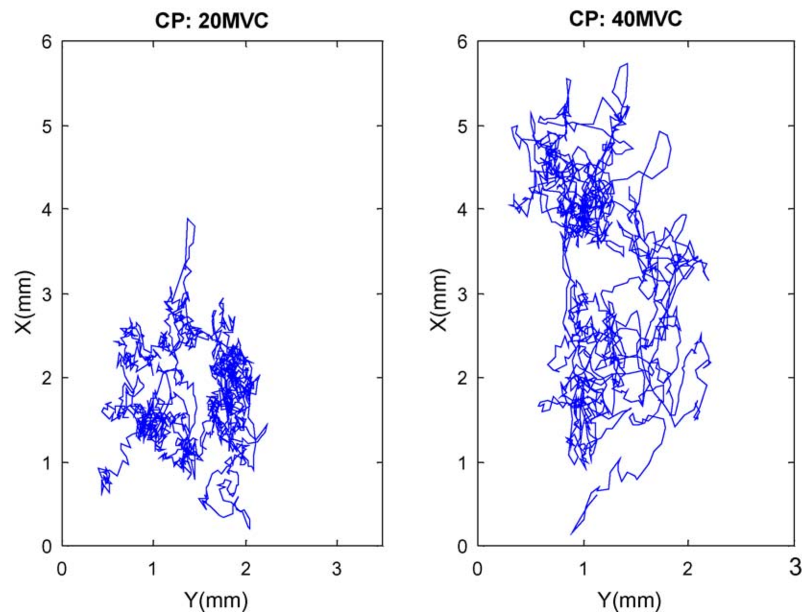

Fig. 2. Center of pressure (CP) displacements in 20MVC and in 40MVC, for a representative subject. X represents the antero-posterior axis and Y the transversal axis. Recordings were taken in 100BP and quiet breathing condition. Note the increase of CP displacements along the antero-posterior axis in 40MVC.

increased. On the other hand, an EMG increase results from an increase in muscle activation, and induces an increase in muscular force. When the contraction is isometric (and stationary) as in this study, the force exerted by a muscle is a function of the EMG it displays. In other words, an EMG increase in postural muscles reflects a force increase in each of these muscles. Hence, it could be assumed that a greater push effort induced active muscular stiffening along the postural chain.

With respect to the posturographic parameters ($\bar{X}_p$ and SP), the main result refers to the influence of postural chain stiffening. Indeed, both parameters increased significantly from 20MVC to 40MVC, that is, when stiffening is increased, independent of the respiratory rate (QB or FB) or the ischio-femoral contact area (100BP or 30BP). Therefore, it can be said that body steadiness is lower when stiffening is higher. This phenomenon could be first explained by a slowing down

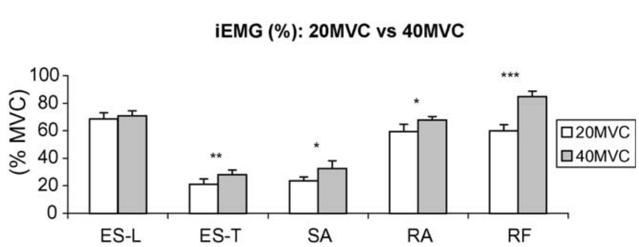

Fig. 3. Integrated EMG (iEMG) of postural chain muscles, as a percentage of the value in maximum voluntary contraction (MVC): pushing at 20% (20MVC) and 40% (40MVC) of the MVC. The records were taken in quiet breathing condition, with 100% ishio-femoral contact. Postural chain muscles: lumbar erectores spinae (ES-L), thoracic erector spinae (ES-T), serratus anterior (SA), rectus abdominis (RA), rectus femoris (RF). The asterisks indicate a statistically significant difference: $^{*}p < 0.05$; $^{**}p < 0.01$; $^{***}p < 0.001$.

effect of the increased muscular tension upon the counter-perturbing movements, making them less efficient. Second, the muscular stiffening could also extend the transmission of the disturbances toward the extremities of the postural chain, inducing larger CP displacements at its lower part and greater F_x variations at its upper part.

Furthermore, it is well known that body balance can be perturbed by autonomous functions, such as respiration and heartbeat [12]. Now, the sway path is sensitive to both respiratory rate (it is higher in fast breathing condition) and push forces, suggesting that respiratory disturbance is involved in lower body steadiness. Such an assumption is not supported by $\bar{X}_p$ data, which do not vary with respiratory rate. Nevertheless, it appears to be in accordance with previous results [13,5], which showed that, contrary to mean CP deviation, sway path is very sensitive to the transient regulation of postural phenomena, such as those induced by respiration, i.e. to steadiness.

The muscular stiffening effect on CP displacements, using experimental variation, is consistent with increased postural sway in Parkinson's disease [14], of which muscular rigidity is one of the main symptoms. In low back pain, it reinforces the hypothesis that increased postural sway is linked to greater back muscular tension [6]. These authors have demonstrated that increased CP displacements cannot be ascribed to a loss of the dorso-lumbar range of motion, which was assumed to represent spine structural stiffness. Conversely, it was suggested that the involvement of greater active muscular tension was related to a pain muscular reflex spasm or to fear avoidance attitude. Hence, treating increased muscle tension along the whole postural chain in the frame of pathology could be considered a relevant strategy in improving balance function impairments.

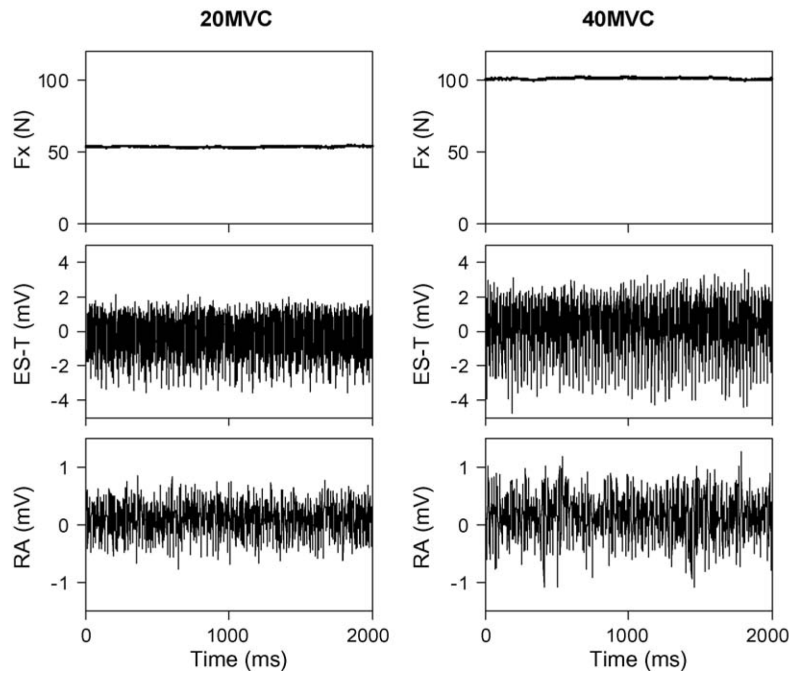

Fig. 4. Representative EMGs of thoracic erectores spinae (ES-T) and rectus abdominis (RA) during two submaximal isometric push efforts (20MVC and 40MVC). From top to bottom: push force (F_x), EMG from ES-T and from RA. Left column, pushing at 20% of the maximum voluntary contraction (20MVC); right column, pushing at 40% of the maximum voluntary contraction (40MVC). Recordings were taken in 100BP and quiet breathing condition.

4.2. SP and ischio-femoral contact area

Surprisingly, SP was not sensitive to ischio-femoral contact area (100BP versus 30BP), which was assumed to vary pelvic mobility [11]. Indeed, it was shown in another study that SP exhibited a significant increase in 100BP when maintaining a quiet sitting posture (arms loosely hanging at the sides) [5]. Such a difference could be ascribed to the upper limbs fixation, which stabilizes the upper part of the postural chain, and possibly reduces the role of pelvic mobility in sitting posture maintenance. Moreover, increased muscular tension resulting from the push effort should reduce pelvic mobility, and consequently its variations related to different ischio-femoral contact areas.

4.3. $\bar{X}_p$ as an estimation of ΔF_x

Lastly, it seems interesting to stress that $\bar{X}_p$ and ΔF_x vary in relation to the same factors. It can easily be explained from a biomechanical standpoint. Indeed, in these experiments, the body constitutes a closed-chain configuration, as the hands are gripping the dynamometric bar and the body is in contact with the seat and the foot rests. As stated by Gaughran and Dempster [15], the push force in isometric efforts is proportional to the difference $X_G - X_P$ (where X_G is the position of the centre of gravity along the antero-posterior axis). Now, it has been shown, in this experimental condition, that X_G is negligible in comparison to X_P [16]. Therefore, it is not surprising that $\bar{X}_p$ and ΔF_x vary in parallel. Finally, one might ask whether ΔF_x does not represent an estimate of $\bar{X}_p$ when closed-chain configurations are considered.

5. Conclusion

In conclusion, this study demonstrates that increased muscular spine tension reduces postural steadiness in a sitting posture, possibly through weaker compensation of respiratory disturbance. In a conceptual point of view, it reinforces the hypothesis that posturo-kinetic capacity is dependent on the whole postural chain mobility.

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