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RESEARCH ARTICLES

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Does postural chain mobility influence muscular control in sitting ramp pushes?

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Abstract This study was conducted under the hypothesis that voluntary movement involves a perturbation of body balance and that a counter-perturbation has to be developed to limit the perturbation effects, which is a condition necessary to perform the movement efficiently. The stabilising action is produced in body segments that constitute the "postural" chain, and the voluntary movement by the segments said to constitute the "focal" chain. In order to deepen the understanding of how the postural chain contributes to the motor act, isometric transient efforts were considered. Seven adults in a sitting posture were instructed to exert bilateral horizontal pushes on a dynamometric bar, as rapidly as possible, up to their maximal force (Fx). Two sitting conditions were considered: full ischio-femoral contact (100 BP) and one-third ischio-femoral contact (30 BP), the latter being known to yield greater pelvis and spine mobility, that is greater postural mobility. Each session consisted of ten maximal pushes for each sitting condition. In order to explore the influence of postural mobility on muscular control and push force, surface EMGs of 14 postural and focal muscles were recorded. In addition, reaction forces (Rx) and displacement (Xp) of the centre of pressure (along the anteroposterior axis) were measured, as well as iliac crest acceleration ($\ddot{x}_h$ and $\ddot{z}_h$, along the anteroposterior and vertical axes, respectively). The results showed that push force varied abruptly during the task ramp effort. When the ischio-femoral contact was limited, push force was enhanced, as well as the rate of push force rise (Fx/Δt, Δt being the force rise duration), suggesting a greater perturbation to balance. Also, there were significant increases in the Rx reaction forces, indicating body segment acceleration: "dynamic" phenomena occurred in the articulated body chain in response to increases in Fx. In addition, even though muscular contraction was isometric, postural EMGs, as well as focal EMGs, were phasic, a feature which characterises transient force exertion. The Rx reaction forces were associated with backward displacement of the centre of pressure, Xp. The centre of pressure displacement was interpreted as a backward pelvis rotation, an interpretation which was confirmed by backward and upward iliac crest accelerations. When ischio-femoral contact was reduced, the backward pelvis rotation was significantly increased, resulting from an increased pelvis and spine mobility. Distinct focal and postural EMG sequences were found to be associated with the effort. Two different sets of muscles were observed when considering recruitment order, the focal and the postural muscles. The ankle muscles were activated before the pelvis, the back and the scapular girdle, with the upper limb muscles activated only after the onset of the primum movens of push action (serratus anterior): the activation process followed a distal to proximal progression order. Moreover, the postural EMG sequence was anticipatory, that is there were anticipatory postural adjustments (APAs). Modifying the ischio-femoral contact did not induce a change in either the postural muscle set or in the recruitment order. There were significant increases in the level of activation (integrated EMG) of the postural muscles when ischio-femoral contact was reduced. They did not result from an increase in EMG duration but only from a modulation of EMG amplitude, suggesting that postural control for different ischio-femoral contacts involves adapting the motor program according to the postural requirements, rather than changing the postural strategy. Moreover, as APA amplitude was increased when ischio-femoral contact was reduced, it could be assumed that the postural chain is programmed in relation to postural chain mobility. In addition, the increase in postural EMGs was interpreted as an increased counter-perturbation opposed to an increased push force. It is concluded that greater mobility of the postural chain favours a greater dynamic counter-perturbation, which, in turn, allows the development of a greater push force; the ability to develop such a counter-perturbation (termed PKC: posturo-kinetic capacity) is enhanced when postural chain mobility is greater. Postural

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chain mobility appears to be a task parameter, and postural control appears to involve adapting the motor program according to the postural requirements, rather than changing the postural strategy.

Keywords EMG pattern · Dynamic postural adjustments · Postural chain mobility · Posturo-kinetic capacity · Ramp efforts

Introduction

It has been proposed by Bernstein (1935) that motor tasks include a *postural* and a *focal* component. The focal component refers to the body segments that are mobilised in order to perform voluntary movement, such as upper limb flexions; in terms of biomechanics, these body segments are said to constitute the focal chain. The focal chain is controlled by the muscles which cause the movement, and, by definition, the primum movens is the muscle primarily responsible for causing the specified joint action, such as anterior deltoideus in this example. In the same task movement, the postural chain includes the body segments spanning from the shoulders to the feet, that is the rest of the body. It has been demonstrated that all the postural segments are accelerated when the upper limb is moved voluntarily (Bouisset and Zattara 1987; Lino 1995; Bouisset et al. 2000). In other words, inertial forces flow throughout the postural chain, since inertial forces are defined as acceleration times mass; this inertial force transmission underlies *dynamic* postural phenomena.

The theoretical reasons for these phenomena are well known. Indeed, the forces (and the torques) which are developed during the movement are transmitted through the articulated chain to more and more distant body parts down to the support surfaces, where reaction forces are produced. In other words, the intended focal movement involves a perturbation of body balance, as has been suggested by several neurologists since the turn of the last century (see, for example, André-Thomas 1940; Hess 1943).

In order to limit these perturbing effects, a counter-perturbation has to be developed, which allows the task to be performed efficiently (Hess 1943; Bouisset and Zattara 1981; Friedli et al. 1988). It has been proposed that the ability to develop such a counter-perturbation be called posturo-kinetic capacity (PKC; Bouisset and Zattara 1983).

As a consequence, if the perturbation is dynamic, the counter-perturbation must also be dynamic, that is, body postural segments must be accelerated. This suggests that PKC depends on functional mobility of the postural chain, which is related to joint range of movement amplitude and muscular torques at the postural joints (Bouisset and Le Bozec 2002). Therefore, it can be supposed that greater mobility of the postural chain allows greater counter-perturbation, which, in turn, favours performance (force, velocity and so on, according to the intended task). Consequently, if postural chain mobility is constrained, in one way or another, fewer postural segments could be accelerated, counter-perturbation would be limited and performance reduced.

Various experimental series have been undertaken in this sense on subjects performing a pointing task. It was established by Goutal et al. (1994) and confirmed by Van der Fits et al. (1998) that peak movement velocity was faster when subjects were standing normally than when they were seated. Lino et al. (1992) and Teyssèdre et al. (2000) have shown that a reduction of the ischio-femoral contact area elicits higher peak velocities as well as an increase in anticipatory postural adjustments (APAs). Under these conditions, the support base perimeter remains the same, but pelvis and spine mobility is reduced when ischio-femoral contact is increased (Vandervael 1956). More recently, isometric ramp efforts have been studied by Le Bozec et al. (1997, 2001). It was shown that isometric ramp efforts displayed dynamic postural adjustments and phasic EMGs in postural muscles.

These data were interpreted as supporting the view that any variation of the exerted force, whether the muscular contraction is "static" or "dynamic", perturbs the subject's balance, and that the maximal value of this force depends on the intensity of the dynamic postural counter-perturbation. In addition, it was hypothesised that increased postural mobility could induce increased performance and longer APAs.

Moreover, a biomechanical model of transient push efforts was recently proposed (Bouisset et al. 2002), with the aim of examining in greater depth the postural adjustments associated with voluntary efforts. To this end, its various terms have been recorded and evaluated from experimental data. They included *global* reaction forces and centre of pressure displacement, and the same quantities measured *locally* at the seat and foot levels. Based on a detailed examination of these terms of the model, it was concluded that transient muscular effort induces dynamics of the postural chain. These responses originate from the body supports and entail inertial forces in body segments. Also, these observations demonstrate that a postural counter-perturbation is associated with the motor act. More precisely, the results showed that the horizontal reaction forces and the horizontal centre of pressure displacement increased quasi proportionally with the push effort.

The aim of this paper was to explore the influence of postural chain mobility on muscular control. To this end, EMG patterns were considered in addition to performance, and specific biomechanical variables taken as reference. Maximal isometric ramp pushes were studied which were performed by subjects in two sitting postures, differing by the ischio-femoral contact with the seat. This task was chosen because it offers two major advantages: (1) isometric ramp efforts are assumed to yield transient perturbing forces, and, since the subjects are in quasi-static conditions, the dynamics should be located in the postural chain, that is between the feet and the shoulders; and (2) since the subjects are seated, the mobility of the postural chain is easy to manipulate, due to a change in the ischio-femoral contact with the seat: full ischio-femoral contact of the ischio-femoral length (100 BP) induces lesser lumbar spine and pelvis mobility than one-third contact (30 BP), as indicated above.

Materials and methods

Procedure

The subjects were seated on a custom-designed device (Lino 1995; Bouisset et al. 2002). In response to a verbal command "go" by the experimenter, they were instructed to perform two-handed horizontal isometric pushes on a dynamometric bar, from zero to maximal force as rapidly as possible, and to maintain it for 5 s. They wore shorts and socks, and the seat and foot-rests were covered with wood. Their posture was standardised before each trial, with the shank and trunk vertical, the thighs horizontal, the upper limbs stretched out and horizontal, and the hands gripping the bar, taken as reference for the biomechanical recordings. Therefore, during the whole motor act, the subject was in a quasi-static state as the contact with the bar, the seat and the foot-rests remained the same.

Full ischio-femoral contact of the ischio-femoral length (100 BP) and one-third contact (30 BP) were considered. The full ischio-femoral contact was adjusted so as to ensure that there was no contact between the edge of the seat and the lower leg. The one-third contact was determined from a measure of ischio-femoral length (buttock-popliteal length, according to anthropometric terminology), and induced a reduction of the contact of the thighs with the seat. Therefore, ischio-femoral contact with the seat was different from one condition to another, but the overall support contour drawn by the feet and the ischio-femoral contacts, that is the support base perimeter, remained the same. Each session consisted of ten isometric maximal pushes carried out according to a reaction time paradigm and the best seven based on a time criterion were considered. The series of pushes were separated by 3-min rest periods to prevent fatigue. They were permutated randomly from one experimental session to another. Seven right-handed male adults (mean body weight 69.7 ±10.3 kg) participated in the experiments. None of the subjects had a history of neurological or musculoskeletal disorder. The subjects gave their informed consent and the experiments were conducted in accordance with legal requirements (Huriet's law).

Apparatus

The dynamometric bar was equipped with two strain gauges (one at each side of the bar) to measure the horizontal force, Fx, exerted on it, and the Fx rise onset, t_0, was taken as the biomechanical reference of voluntary push. The custom-designed device included three rectangular force plates (a seat and two foot-rests), which were linked by a rigid frame, and measured the reaction forces exerted on the seat and on the foot-rests (Lino 1995; Bouisset et al. 2002). This investigation took into consideration global reaction forces (Rx along the anteroposterior axis) and global centre of pressure displacements (Xp along the anteroposterior axis). As three force plates were used, the global quantities were directly given by a computer program based on the biomechanical model which has been already presented (Bouisset et al. 2002). With the aim of supporting the argumentation based on dynamics, local kinematics were recorded at the iliac crest in six subjects. To this end, a tri-axial accelerometer (ENTRAN, ECG D, ±5 g) was used. It was fixed to an appropriately shaped splint, firmly attached at the right iliac crest by means of an elastic belt attached to another shaped splint located on the left iliac crest. The accelerometer's precise position was set so as to make its horizontal axis coincident with the laboratory horizontal plane given by a spirit-level. Backward pelvis rotation corresponds to simultaneous rear anteroposterior and upward vertical accelerations.

Surface EMGs were recorded from the dominant side by bipolar electrodes after a preliminary series of recordings, which included the systematic checking of bilateral trunk and lower limb muscles (Lesne 1997). A representative set of 14 of these 20 muscles was selected in order to eliminate muscles displaying the same pattern and less reliable activity. These muscles were distributed all over the body, and included muscles responsible for different actions in the horizontal push. Two muscle sets were termed focal muscles, insofar as they included the primum movens of the push effort and the upper limb muscles which transmitted the muscular action to the dynamometric bar. They were: (a) shoulder muscles [the primum movens: serratus anterior, SA; an agonist (and shoulder flexor): deltoideus anterior, DA; an antagonist (and shoulder fixator), trapezius superior, TS]; and (b) muscles of the focal limb which cross the elbow and wrist joints [biceps brachii, BB; triceps brachii (caput laterale), TB; extensor carpi radialis, ECR; flexor carpi radialis, FCR]. Two muscle groups were termed postural muscles, insofar as they controlled trunk and lower limbs, that is: (c) trunk and pelvis muscles (tensor fasciae latae, TFL; gluteus maximus, GM; erector spinae, ES; obliquus externus, OE; biceps femoris, BF); and (d) lower limb muscles crossing the ankle (tibialis anterior, TA; gastrocnemius lateralis, GL). Interelectrode impedance was less than 5 kΩ. The EMGs were amplified with differential amplifiers (frequency bandwidth from DC to 10 kHz). All the signals were recorded with an EMG sampling rate of 1,000 Hz. Timing and amplitude data were determined from individual trials.

Data processing

The Fx rise onset, t_0, was taken as the onset of voluntary push force increase. The onset times of Rx (termed 0 in Fig. 1) and Xp were determined when the corresponding values exceeded two standard deviations from pretest baseline. The onset time of EMG bursts for each respective muscle was determined when the EMG voltage exceeded two standard deviations from baseline activity, and τ_0, the EMG reference, was taken as the EMG onset of primum movens (SA). EMG signals were rectified in order to calculate the instantaneous mean voltage (iEMG) and the integrated EMG over the push duration (IEMG). The iEMGs and IEMGs were computed for each muscle during the time interval between EMG onset and the peak force time, and averaged for each subject. The value of the full-wave rectified EMG curve (iEMG) was also calculated at the SA burst onset (τ_0). To compare EMG data among conditions and across subjects, the iEMGs and IEMGs were normalised as follows: for each subject, the maximal IEMG value for a given muscle in 100 BP condition was taken as the reference value (100%).

All the data were collected, processed and stored on a PC. The processed data were analysed using a one-way repeated measures ANOVA technique. Where appropriate, paired *t*-tests were employed to compare data between two groups. Data were considered significantly different when the probability of error was 0.05 or less (P<0.05: significant; P<0.01: very significant; P<0.001: highly significant).

Results

Biomechanical data

All the biomechanical variables under study displayed transient variations. Therefore inertial forces were put into play, that is there were dynamic phenomena, according to

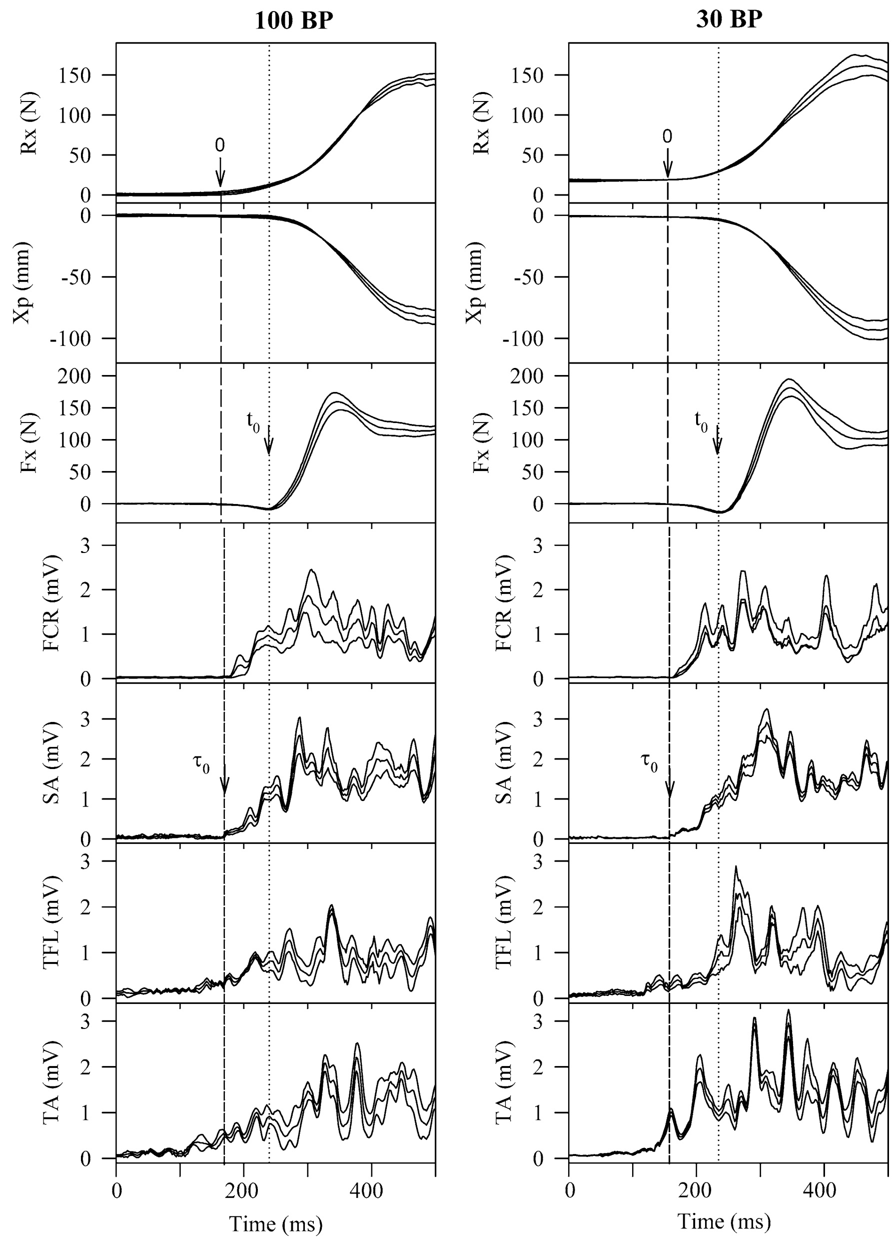

Fig. 1 Biomechanical quantities and EMGs. From *top* to *bottom*: anteroposterior component of reaction forces Rx (N); anteroposterior displacement of the centre of pressure Xp (mm); external force Fx (N); rectified EMGs from representative muscles of the four muscle groups under study, flexor carpi radialis (*FCR*) as representative of the focal muscles group, serratus anterior (*SA*) as the primum movens, tensor fascia latae (*TFL*) as representative of the pelvis group and tibialis anterior (*TA*) as representative of the ankle group. One standard deviation is shown above and below the mean (seven trials by the same subject). The two experimental conditions were: 100% of ichio-femoral contact (100 BP) and 30% of ischio-femoral contact (30 BP). $\tau_0 \downarrow$ EMG onset of primum movens (SA), $0 \downarrow$ onset of Rx, $t_0 \downarrow$ onset of voluntary push force increase. According to sign convention, Rx and Xp are positive when they are directed forward. Under the action and reaction law, the sign of anteroposterior body action on the supports is −Rx (as Rx is the reaction). Therefore, the force exerted by the subject on the bar (Fx) and his action on the supports (−Rx) are opposite in sign

mechanics. These variations were highly reproducible from one trial, one condition and one subject to another. More precisely, as shown in Fig. 1, Fx, the push force, and Rx, the horizontal reaction force, increased sharply during the ramp effort, while Xp decreased in the same way, indicating that the centre of pressure moved backwards (that is, the centre of pressure moved in the direction opposite to push force).

When ischio-femoral contact was reduced from 100 BP to 30 BP, the maximal force, Fx, obtained at the end of the ramp effort and the mean rate of Fx rise (Fx/Δt) were very significantly greater (Table 1), but the duration of force rise (Δt) was not significantly higher for 100 BP. In addition, the corresponding Rx and Xp values increased highly significantly for 30 BP. Moreover, Rx and Xp onsets ($[Rx]_0$ and $[Xp]_0$) preceded Fx onset occurring at t_0, indicating APAs. APA duration and amplitude values were higher in 30 BP than in 100 BP conditions (Table 2). The differences between 100 BP and 30 BP were highly significant (P<0.001) for both items under consideration. Finally, maximal ramp efforts displayed rear anteroposterior and upward vertical accelerations at the iliac crest, suggesting backward pelvis rotation. Only anteroposterior peak accelerations were very significantly superior in 30 BP as compared to 100 BP conditions (Table 3). The peak backward displacement was also significantly different, indicating increased backward pelvis rotation for 30 BP.

EMG data

All subjects displayed the same repeatable excitation pattern, despite intermuscle differences in EMG time courses: the EMGs increased rapidly, lasting until the end of the effort (Fig. 1). More precisely, the iEMG time courses displayed a monotonous increase (Fig. 2). The iEMGs of the postural muscles appeared greater for 30 BP, that is when peak force was greater. In order to compare the instantaneous excitation level and external force, iEMGs were plotted against Fx (Fig. 2), both variables being expressed as a percentage of their peak value. The comparison of the profiles showed that they were not different for the two ischio-femoral contact conditions. After an initial vertical increase indicating that they preceded the onset of Fx, they displayed curvilinear profiles. This feature was the same for the prime mover, SA, as well as for the other groups of muscles (lower limb, trunk, shoulder and upper limb muscles). The same shape was displayed by Rx and Xp (see Fig. 1). Table 4 further illustrates a comparison of EMG onsets in relation to the prime mover onset (SA).

Specifically, two different sets of muscles were observed when considering recruitment order: (1) *postural muscles*, including those muscles whose activation preceded SA significantly, that is from TA to GM (−42 to −17 ms for 100 BP, and −40 to −19 ms for 30 BP, prior to SA onset); and (2) *focal muscles*, that is the shoulder and upper limb muscles, including the DA to FCR sequence (−2 to 6 ms for 100 BP and −3 to 6 ms for 30 BP, in relation to SA onset). Thus, the postural muscles preceded the focal muscles. More precisely, the EMG onset of the last postural muscle (GM) started significantly before the primum movens (SA) (t=2.78; P<0.01 for 100 BP and t=3.72; P<0.001 for 30 BP). When the mean onset of the group of postural muscles was compared to that of the focal muscles, the differences were found to be highly significant [100 BP: F (1.6)=93; P<0.001; 30 BP: F (1.6)=147; P<0.001]. Moreover, the focal muscle group displayed significant differences between the shoulder (DA-SA-BB-TB) and the wrist muscles (ECR-FCR) for 100 BP [F (1.6)=33; P<0.001] and for 30 BP [F (1.6)=16; P<0.01].

The sequence was the same for both support conditions (Table 4): the sequence started with muscles of the postural chain and ended with elbow and wrist fixators, suggesting an ascending progression from the feet up in both support conditions. The onset differences between 100 BP and 30 BP were not significant for either muscle group under consideration (shoulder muscles, P=0.85; focal limb muscles, P=0.70; lower trunk and pelvis postural muscles, P=0.54; ankle postural muscles, P=0.36). In other words, APA durations did not display any difference between 100 BP and 30 BP (Table 5), insofar as they were defined by postural muscles antepositions. The value of the full-wave rectified EMG curve (iEMG), calculated for postural muscles at the SA burst onset (τ_0), which measured APA amplitude, was found to be very or highly significantly increased (except for ES) for the 30 BP condition (Table 5).

Table 1 Peak values of the biomechanical quantities for the two conditions of ischio-femoral contact. Mean values (*M*) and standard deviations (*S*) are given for external force increase (*Fx*), the rate of Fx rise (*Fx/Δt*) and its duration (*Δt*), anteroposterior resultant reaction force (*Rx*) and anteroposterior displacement of the centre of pressure (*Xp*). (*100 BP* 100% of ischio-femoral contact, *30 BP* 30% of ischio-femoral contact, *F* Snedecor test, *NS* not significant P>0.05)

	Fx (N)		Fx/Δt (N/s)		Δt (ms)		Rx (N)		Xp (mm)	
	M	S	M	S	M	S	M	S	M	S
100 BP	153.1	22.3	0.774	0.068	197	22	149.4	27.1	−126.3	27.2
30 BP	174.8	24.3	0.906	0.085	191	17	177.7	20.4	−164.4	26.3
F (1.6)	25.1**		30**		5 (NS)		86.1***		50.2***	

**P<0.01 (very significant)
***P<0.001 (highly significant)

Table 2 Anticipatory postural adjustments (APAs) for the two conditions of ischio-femoral contact. Mean values (*M*) and standard deviations (*S*) are given for anteroposterior resultant reaction force ($[Rx]_0-[Rx]t_0$) and anteroposterior displacement of the centre of pressure ($[Xp]_0-[Xp]t_0$). APA duration (*dAPA*) and amplitude (*pAPA*) are presented. $[Rx]_0$ and $[Xp]_0$correspond to the onset of Rx and Xp variations; $[Rx]t_0$and $[Xp]t_0$ to their values when Fx begins increasing. (*100 BP* 100% of ischio-femoral contact, *30 BP* 30% of ischio-femoral contact)

	dAPA				pAPA			
	$[Rx]_0-[Rx]t_0$(ms)		$[Xp]_0-[Xp]t_0$(ms)		$[Rx]t_0$ (N)		$[Xp]t_0$ (mm)	
	M	S	M	S	M	S	M	S
100 BP	−60	5	−62	6	11.6	1.9	1.4	0.9
30 BP	−67	4	−69	8	28.7	2.6	3.7	0.5
F (1.6)	128***		102***		295***		126***	

****P*<0.001 (highly significant)

In order to estimate the global excitation delivered to the muscles during the push rise, the area under the full-wave rectified EMG curve was calculated from EMG burst onset to peak force. Analysis of the data for individual muscles showed that when the ischio-femoral contact was reduced from 100 to 30 BP, the IEMGs of the postural muscles were enhanced (except for ES), whereas the primum movens and the other focal muscles (except for DA) did not display different values (Fig. 3). Changes in IEMG for TA, GL, BF, GM and OE, as well as DA, were highly significant (*P*<0.01, paired *t*-test). TFL (and TS) displayed only significant differences (*P*<0.01). In the focal muscles (except for DA), a general, albeit non-significant, tendency to increase was observed in the 30 BP condition. Finally, the postural muscle EMG durations, for all the muscles and subjects, varied from 210±16 to 184±11 ms in 100 BP and from 205±12 to 183±9 ms in the 30 BP condition. There was no significant difference [F (1.6)=3.7], which suggested that the IEMG increases were related to the EMG amplitude and not to its duration.

Discussion

The main results show that: (1) muscle recruitment includes postural in addition to focal muscles, and EMGs of both muscles sets are phasic; (2) the postural muscle sequence precedes the focal muscle sequence, and the postural biomechanical phenomena precede the onset of push force, defining APAs; (3) modifying the ischio-femoral contact area does not induce a change in either the

Table 4 EMG onset times. Means (*M*) and standard deviations (*S*) were expressed in relation to the prime mover (serratus anterior) onset (in ms). Negative values indicate anticipatory EMGs. The onset time of EMG bursts for each muscle was determined when the EMG voltage exceeded two standard deviations from baseline activity. (*100 BP* 100% of ischio-femoral contact, *30 BP* 30% of ischio-femoral contact)

Muscles	Abbreviation	100 BP		30 BP	
		M	S	M	S
Tibialis anterior	TA	−42	16	−40	12
Gastrocnemius lateralis	GL	−35	15	−34	10
Tensor fasciae latae	TFL	−33	13	−31	13
Erector spinae	ES	−25	9	−23	11
Biceps femoris	BF	−24	13	−27	14
Trapezius superior	TS	−23	10	−22	11
Obliquus externus	OE	−22	11	−24	12
Gluteus maximus	GM	−17	11	−19	9
Deltoideus anterior	DA	−2	10	−3	8
Triceps brachii	TB	−2	10	+2	9
Biceps brachii	BB	−1	9	+1	13
Serratus anterior	SA	–	–	–	–
Extensor carpi radialis	ECR	+4	11	+4	11
Flexor carpi radialis	FCR	+6	11	+6	13

Table 3 Peak values of the iliac crest accelerations and displacements for the two conditions of ischio-femoral contact. Mean values (*M*) and standard deviations (*S*) are given for anteroposterior ($\ddot{x}_h$) and vertical ($\ddot{z}_h$) peak accelerations and anteroposterior (x_h) peak displacement. The minus sign refers to negative acceleration and to backward displacement. (*100 BP* 100% of ischio-femoral contact, *30 BP* 30% of ischio-femoral contact, *NS* not significant *P*>0.05)

	$\ddot{x}_h(ms^{-2})$		$\ddot{z}_h(ms^{-2})$		x_h (mm)	
	M	S	M	S	M	S
100 BP	−14.02	1.7	0.64	0.1	−15.2	3.9
30 BP	−20.25	3.4	0.77	0.11	−22.9	3.6
F (1.5)	27.4**		5.3 (NS)		9.7*	

**P*<0.05 (significant)
***P*<0.01 (very significant)

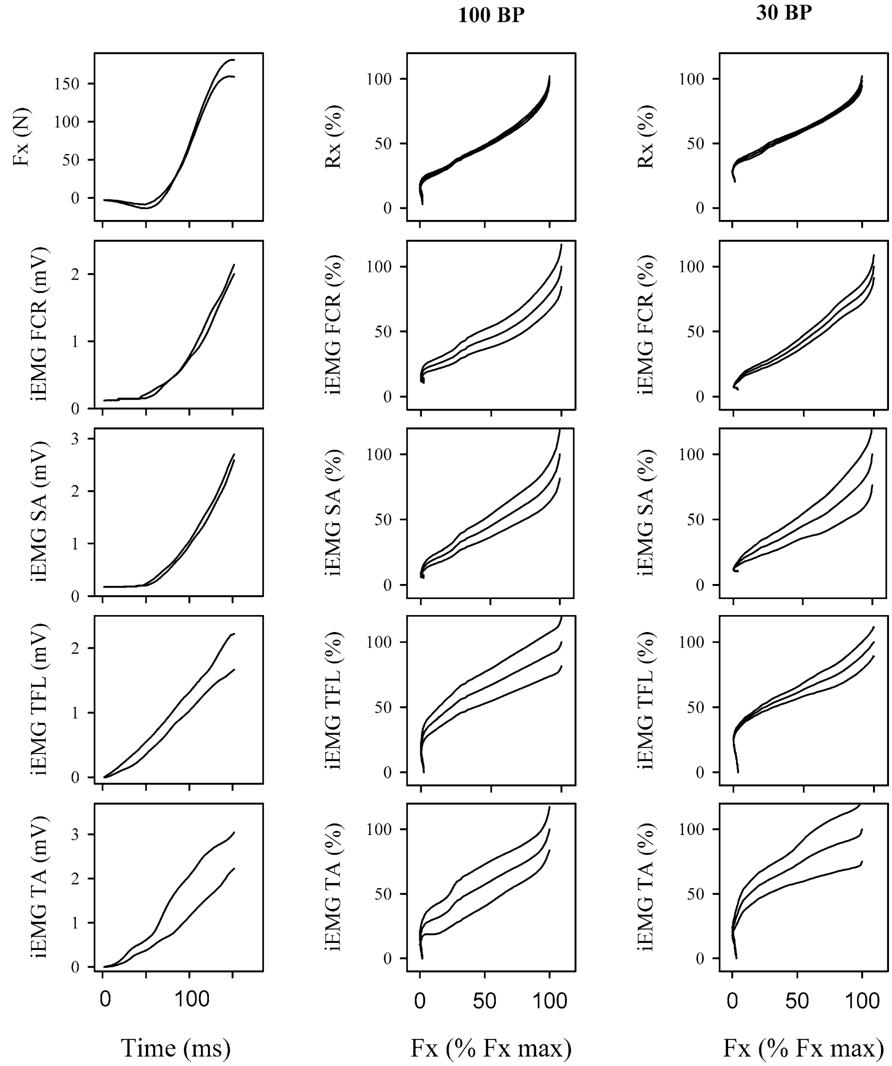

Fig. 2 Instantaneous variations of mean voltage EMGs (iEMGs) during maximal isometric ramp efforts. *First column*: iEMGs (mV) and Fx (N) as a function of time (ms). From *bottom* to *top*: a leg muscle (TA); a pelvis muscle (TFL); the primum movens (SA); a forearm muscle (FCR); and the push force (Fx), given as a reference. Same symbols as in Fig. 1. Mean profiles (seven trials by the same subject) are presented for the two experimental conditions: 100% of ischio-femoral contact (100 BP; *lower trace*) and 30% of ischio-femoral contact (30 BP; *upper trace*). *Second and third columns*: iEMGs (%) and Rx (%) as a function of Fx (%). The two experimental conditions were presented: 100% of ischio-femoral contact (100 BP) and 30% of ischio-femoral contact (30 BP). From *bottom* to *top*: a leg muscle (TA); a pelvis muscle (TFL); the primum movens (SA); a forearm muscle (FCR); and the anteroposterior reaction force (Rx), given as a reference. Same symbols as in Fig. 1. Mean profiles ± one standard deviation. Each item is expressed as a percentage of the maximal value. One standard deviation is shown above and below the mean (seven trials by the same subject). The initial iEMG increases are vertical because the onset of excitation precedes the onset of external force

Table 5 Influence of ischio-femoral contact length on postural muscles APA duration (*dAPA*) and amplitude (*pAPA*). Means (*M*) and standard deviations (*S*) were presented for 100 BP (100% of ischio-femoral contact) and 30 BP (30% of ischio-femoral contact). dAPAs were expressed in relation to the prime mover (serratus anterior; SA) onset (in ms). pAPAs are given by the value of the full-wave rectified EMG curve (iEMG) calculated at SA burst onset. Mean pAPAs were averaged across all subjects, for every muscle. Each value was expressed as a percentage of the maximal iEMG value, which was taken at the peak force instant and in the 100 BP condition. Muscle abbreviations as in Table 4. (*NS* Not significant P>0.05)

		TA		GL		TFL		ES		BF		OE		GM	
		M	S	M	S	M	S	M	S	M	S	M	S	M	S
dAPA (ms)	100 BP	42	16	35	15	33	13	25	9	17	11	24	13	22	11
	30 BP	40	12	34	10	31	13	23	11	19	9	27	14	24	12
	F (1.6)	1.4 (NS)		0.4 (NS)		0.6 (NS)		0.8 (NS)		0.9 (NS)		1.4 (NS)		2.2 (NS)	
pAPA (%)	100 BP	14.6	3.5	17.4	2.8	6.6	2.8	14.7	1.6	16.7	2.2	13.6	2.1	14.9	2.8
	30 BP	23.1	2.5	21.1	1.1	11.6	4.4	15.4	1.1	24.8	1.4	20.1	2.2	17.5	2.3
	F (1.6)	207***		27**		39***		4.9 (NS)		222***		156***		234***	

**P<0.01 (very significant)
***P<0.001 (highly significant)

postural muscle set or in recruitment order; there are significant increases in APA amplitude and in the level of activation of postural muscles when ischio-femoral contact is reduced; (4) performance, that is peak push force, is also enhanced under the same conditions.

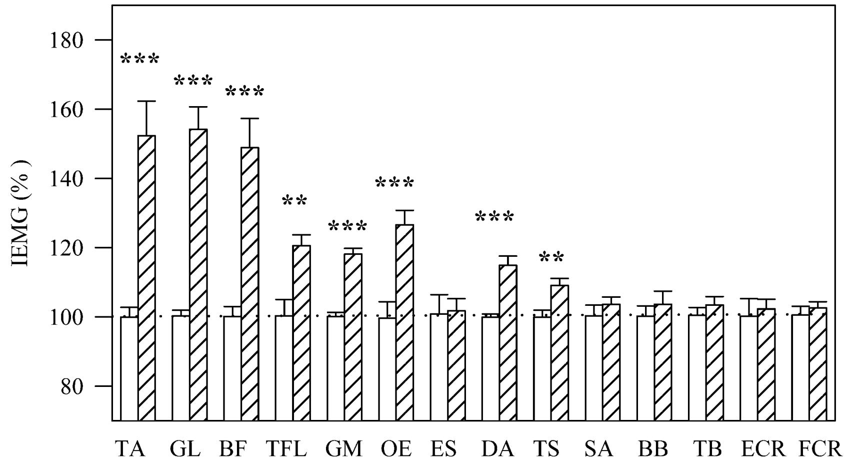

Fig. 3 Influence of ischio-femoral contact area on integrated EMGs. Mean integrated EMGs (IEMGs) are averaged across all subjects, for each muscle. They were presented with one standard deviation bar. Each 30 BP value (*hatched columns*) is expressed as a percentage of the value corresponding to the 100 BP condition (*white columns*). *** $P<0.001$; ** $P<0.01$; * $P<0.05$. *TA* Tibialis anterior, *GL* gastrocnemius lateralis, *BF* biceps femoris, *TFL* tensor fasciae latae, *GM* gluteus maximus, *OE* obliquus externus, *ES* erector spinae, *DA* deltoideus anterior, *TS* trapezius superior, *SA* serratus anterior, *BB* biceps brachii, *TB* triceps brachii (caput laterale), *ECR* extensor carpi radialis, *FCR* flexor carpi radialis

Muscular pattern and postural dynamics in ramp efforts

The present findings show that the push force, Fx, varies abruptly during the task ramp effort. Also, there are significant increases in the global reaction forces (Rx) indicating body accelerations. In other words, there are inertial forces (as inertial forces are, by definition, acceleration times mass): *dynamic* phenomena occur in the articulated body chain in response to Fx increase. Because the subject is in a fixed posture, with the upper limbs outstretched and the hands grasping the bar, body link accelerations can only originate from the body parts interposed between the scapular girdle and the ischio-femoral and foot contact areas, that is from the postural chain, in agreement with previous results (Le Bozec et al. 1997; Bouisset et al. 2002). In the present study, dynamic behaviour of the postural chain is proven to be at a global level by the displacement of the global centre of pressure (Xp) and confirmed at a local level by the observed hip accelerations.

EMG data establish that dynamic phenomena originate from postural muscles in both ischio-femoral contact conditions. First, muscle recruitment includes muscles crossing the main postural joints, that is, in order, the ankle, knee, hip and trunk, before the focal muscles: postural muscles are associated to focal muscles, and are activated prior to them. These results are in agreement with global biomechanical data, which establish that Rx and Xp onsets precede the onset of Fx (Table 2). Second, EMGs are phasic even though muscular contraction is isometric. Phasic EMGs in focal muscles have been reported for ramp efforts by many authors since Simons and Zuniga (1970), Gottlieb and Agarwal (1971) and Bouisset et al. (1973). This type of phasic feature is known to underlie a transient force increase, that is "anisotonic isometric" contractions in this study. In other words, there is no tonic postural adaptation to the ramp effort.

When upper limb task movements, such as pointing tasks, are performed while standing, postural muscle activities are observed at the ankle, knee, hip and trunk (for review, see Bouisset and Le Bozec 2002). When these movements are performed while sitting, authors have reported EMGs in the trunk, pelvis and lower limb muscles (Bouisset et al. 1964; Son et al. 1988; Crosbie et al. 1995; Lino 1995; Tyler and Hasan 1995; Teyssèdre et al. 2000). Oddsson and Thorstensson (1987) proposed classifying postural disturbance caused by voluntary movement as either dynamic or static. In ramp efforts, the disturbance provoked by the increase in push force is dynamic even though the upper limbs do not move, so that the "movement" is static. However, as in other categories of movement, the dynamic disturbance requires postural muscle activity.

In conclusion, these results extend to task ramps those reported for dynamic task movements. In particular, they give a complete account of the muscular synergy associated with ramp efforts to maintain body balance, and they identify the timing of the principal postural EMG onset for both ischio-femoral contacts. They complete and extend previous results (Le Bozec et al. 2001).

Effects of ischio-femoral contact on muscle recruitment

A reproducible EMG pattern was found to precede and accompany ramp efforts performed in the sitting posture. Biomechanical data can help to interpret muscular actions. The first muscles to be recruited (TA) raise the forefeet and the second muscles (GL) raise the heels (and fix the thighs); their actions can explain the backward centre of pressure displacement (and the backward body action exerted at the foot level, according to Bouisset et al. 2002). Then, as internal rotators, both TFL contribute to lock the pelvis, that is to anchor the pelvis to the seat; they also tend to flex the thigh, decreasing ischio-femoral contact with the seat. ES contraction flattens the hypolordotic lumbar spine position associated with the initial posture. The BF fix the thighs and, in cooperation with GM, extend the pelvis, while both OE flex the trunk. As SA antagonists, both TS contribute to a stiffening of the thoracic spine. Consequently, BF, GM and OE cooperate to induce backward pelvis rotation, which is suggested by the Xp backward displacement and is confirmed by accelerometric data.

Afterwards, SA, primum movens, and its co-agonist, DA, develop the push force and contribute to shoulder fixation, in cooperation with BB and possibly TB. The transfer of push force from the shoulder to the bar requires that the upper limb joints also be fixed. Indeed, upper limb muscles fix the elbow (TB, BB) and wrist (ECR, FCR) joints. The sequence is reproducible within each muscle set, whether focal or postural.

Changing the ischio-femoral contact area from 100 BP to 30 BP does not induce a change in either the postural muscle set or in the recruitment order. In particular, the bottom-up sequence was maintained. This type of sequence has been described in many manual task movements since Belenkii et al. (1967), inasmuch as muscular deactivations are considered in addition to activations. However, it cannot be excluded that the "distal to proximal" order (Cordo and Nashner 1982) is not an absolute rule, contrary to what has been supposed. Indeed, Lino (1995) and Teyssèdre et al. (2000), who considered unilateral pointing tasks performed by sitting subjects, reported that the anticipatory EMG sequence started with a deactivation of the ipsilateral trunk extensors (when they were active in the initial posture), followed by an excitation of the pelvis, thigh and, only thereafter, ankle muscles. Therefore, one might wonder whether the postural muscle pattern would depend primarily on the role played by the various supports according to task requirements.

On the other hand, the level of activation of postural muscles is significantly greater for 30 BP, as well as the push force (Fx is increased) and the global dynamic phenomena (Rx is increased). Also, the IEMG of the DA was increased, as was the IEMG of the TS, a shoulder fixator. However, there are no significant increases for SA and the upper limb muscles. Indeed, the IEMGs of the focal muscles displayed only a relatively consistent tendency to increase in the 30 BP condition, with the exception of DA, and this is likely to be related to its flexion action at the shoulder (which induces the vertical component of push effort reported by Bouisset et al. 2002). However, non-significant EMG increases in the focal muscles do not necessarily mean that muscle torques, and particularly SA torque, are not increased. Indeed, it is possible that the SA is lengthened during the push effort, which could result in increased muscle force for the same EMG level, in accordance with the well-known muscle tension-length relationship. However that may be, the results show that external force is primarily enhanced at the expense of a significant increase in the level of activation of the postural muscles. These increases do not result from an increase in EMG durations. Indeed, there is only a modulation of EMG amplitude, which corresponds to a speed-sensitive strategy according to Corcos et al. (1990). Thus, postural control during ramp efforts for different ischio-femoral contacts involves adapting the motor program according to the postural requirements, rather than changing the postural strategy. Latash and Anson (1996), Gantchev and Dimitrova (1996) and Teyssèdre et al. (2000) have reported similar results in their studies on pointing movements.

In other words, there was no change in how the subjects planned to perform the motor task when the surface of the ischio-femoral contact with the seat was reduced. Therefore, as has been proposed by Aruin et al. (1998), control of posture would involve adapting the general motor programs for postural regulation according to current postural requirements.

Reduction of ischio-femoral contact and performance

The present results establish that the maximal force reached at the end of the ramp effort is greater when the ischio-femoral contact area is reduced (30 BP), whereas the overall support contour, that is the support base perimeter, remains unchanged. This might be surprising, though only at first glance. Indeed, performance enhancement is associated with increased dynamics, which are proven by the corresponding Rx and Xp maximal values: the increase in the Rx reaction forces corresponds to an increase in the backward displacement of the centre of pressure, Xp. This interpretation is reinforced by the kinematic data, which confirmed that there is a backward pelvis rotation.

It is known that pelvis mobility is modified by a reduction in the seat contact area from 100 BP to 30 BP. Indeed, when ischio-femoral contact is limited, such as in the 30 BP posture, the pelvis can rotate in relation to an axis passing through the contact of the ischiatic tuberosities with the seat, in addition to an axis passing through the femoral heads (Vandervael 1956). On the other hand, when ischio-femoral contact is complete, that is in the 100 BP posture, the thighs are in close contact with the seat and cannot be displaced: the pelvis can only move about an axis passing through the femoral heads. Therefore, pelvis mobility is less in the 100 BP condition than in 30 BP. In addition, lumbar lordosis flattens in proportion to pelvis rotations, which favours trunk flexion associated to push effort (Gaughran and Dempster 1956). Consequently, the pelvis and lumbar column contribute to postural chain mobility, which is less in the 100 BP than in the 30 BP condition.

According to the PKC theory (Bouisset and Zattara 1983; Bouisset and Le Bozec 2002), since movement induces a dynamic perturbation, the counter-perturbation must be dynamic as well. Now, given that isometric ramp efforts induce dynamics, the postural counter-perturbation must also be dynamic, in order to reach the intended performance. Consequently, if postural chain mobility is constrained in one way or another, fewer postural segments could be accelerated, counter-perturbation would be limited and performance reduced. In other words, the increased mobility of the postural chain favours postural dynamics, and hence PKC, which produce greater force at the end of the effort.

These results generalise to ramp efforts those obtained by Lino et al. (1992) for pointing movements performed under the same two support conditions. When ischio-femoral contact is reduced, performance (that is maximal velocity in pointing movement) increases significantly and dynamic postural phenomena have been reported. Thus, it does not matter if the effort is “dynamic” as in the Lino et al. (1992) study, or “static” but “anisotonic”, as in this one. In both conditions, the effect is associated with a variation of muscular force: performance is increased when the contact area is reduced under both static and dynamic conditions, insofar as postural chain mobility is greater. In a complementary study, the effect of postural conditions on peak velocity was examined in greater detail (Goutal et al. 1994). The main result showed that velocity was faster when the subjects were standing normally than when they were sitting, even in the 30 BP condition. These results agree with Van der Fits et al. (1998), who reported that larger support surfaces induce a peak velocity limitation, with velocity decreasing stepwise from standing to upright sitting, long-leg sitting, and semi-reclined sitting.

In conclusion, postural compensation to ramp effort perturbation depends not only on the support base perimeter, that is stability area, but also on postural chain mobility, that is on the free play of postural joints. In this study, it is a function of pelvis and lumbar column *mobility*, which appears to be a key PKC factor.

Generation of force and task programming

It was recently suggested (Latash 1993; Aruin and Latash 1995) that there is a single control process for a whole-body movement, leading to these two distinct peripheral patterns classified as focal and postural. Some of the results presented in this study support such an assumption. Indeed, it has been observed that the iEMG time courses displayed a curvilinear increase (Fig. 2 *left column*). This feature was the same not only for the focal muscles, which is in accordance with previous data (see, for example, Simons and Zuniga 1970), but also for the postural muscles: the postural muscles display the same kind of relationship as the focal muscles. The slight intermuscle differences are likely to depend on the proportion between fast and slow motor units, as claimed by Woods and Bigland-Ritchie (1983). However, the iEMG increases were steeper and the peak values greater for 30 BP, that is when the peak push force was also greater. When iEMGs were plotted against Fx (Fig. 2 *middle and right columns*), the comparison of the profiles showed that they were not different for the two ischio-femoral contact conditions. Therefore, the excitation pulses addressed by the alpha motoneurons pools to focal and postural muscles could be assumed to be common, or at least controlled by the same rule for height and duration. This rule would be defined by a constant pulse duration and a modulated height in relation to the rate of force rise, that is “a control pulse height policy” (Gordon and Ghez 1987), or, which is equivalent, a “speed insensitive strategy” (Corcos et al. 1990). The excitation pulse to the postural muscles would be scaled according to postural chain mobility.

The assumption that postural mobility is a task parameter appears to be supported by APA data. Indeed, the global biomechanical variables (Rx and Xp) onset precedes that of external force rise, and the APAs are longer and greater when postural chain mobility is increased. The postural EMG sequence was anticipatory, and the APA amplitudes are increased when ischio-femoral contact was reduced, as was the level of activation (IEMG) of the postural muscles. Consequently, it can be assumed that the postural chain is programmed in relation to this parameter, that is postural chain mobility can be considered a task parameter.

To summarise, it could be surmised that there is a single control process for a whole-body movement, leading to these two distinct peripheral patterns classified as focal and postural. The postural command appears to be scaled in proportion to postural chain mobility.

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