Proprioceptive postural control strategies differ among non-injured athletes.

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ABSTRACT:

Postural control during complex tasks requires adequate sensory integration and somaesthetic reweighting: suboptimal postural strategies can lead to injury. We assessed the ability of healthy athletes to reweight somaesthetic signals during postural perturbations on different surfaces. Thirty-five young (16±1 years), healthy, elite handball players participated in this cross-sectional study. Proprioceptive reweighting was evaluated via vibration of the triceps surae and lumbar muscles on *firm* and *foam* surfaces. Postural variables and the electromyographic activity of the gluteus medius (GM), semitendinosus (ST) and fibularis longus (FL) were recorded during the PRE (10s), VIBRATION (20s) and POST (20s) periods. Ankle proprioception was predominantly used on the *firm* compared to *foam* support. However, two opposing behaviours were observed: a *"rigid"* strategy in which reliance on ankle proprioception increased on the *foam*, and a *"plastic"* strategy that involved a proximal shift of proprioceptive reliance (p<.001). The *plastic* strategy was associated with a more effective recovery of balance after vibration cessation (p<.05). ST activation was higher during POST in the *rigid* strategy and did not return to the PRE level (p<.05) whereas it did in the *plastic* strategy. Proprioceptive strategies for postural control are highly variable and future studies should evaluate their contribution to injury.

KEY WORDS:

proprioceptive reweighting, ACL injuries, plasticity, sensory integration

Introduction

Efficient postural control involves the integration of visual, vestibular and somaesthetic/proprioceptive afferent signals, as well as the dynamic reweighting of sensory sources [1]. Sensory reweighting is defined as the alteration of the weight assigned to a signal in order to adapt to the environmental conditions and available sensory information [2]. Reweighting may occur within a single sensory modality, e.g. the weight assigned to somaesthetic signals from different anatomical locations may be adjusted according to their reliability at a given time/condition [3]. For example, ankle proprioceptive signals are predominant on firm surfaces [4] while the reliance is shifted to lumbar signals on unstable surfaces due to the loss of reliability of ankle signals [5].

Reweighting processes are heterogenous among individuals [6,7] and are affected by to natural and long-term (static) modifications such as muscle fatigue [8] or normal aging [9]. Rapid reweighting is essential to ensure optimal postural control during short term (dynamic) or sudden changes in the environment, for example during sport [10], or for balance recovery after a perturbation [11–13]. Suboptimal reweighting can reduce postural and motor performance and increase the risk of injuries [10,14].

Anterior cruciate ligament (ACL) rupture is a common non-contact injury especially among young athletes [15], and unanticipated cutting movements are the most frequent injury situation [16]. Inadequate sensorimotor control and poor postural control are a risk factor for ACL injuries [15,17–19] and alterations in sensory reweighting processes have been reported before a contralateral rupture [20]. There appears to be a direct association between alterations in the activation of neural pathways and high-risk biomechanical factors [21] causing a sensorimotor mismatch and leading to ACL injury [15,22]. During unplanned situations, the cognitive load associated with the visual demand required to manage the opponent's behaviour could lead to poor knee-stabilising strategies, increasing the risk of ACL injury [23]. Thus, in this particular situation, reweighting in favour of proprioceptive signals is crucial to ensure optimal postural control and avoid sensorimotor mismatch [10,19,24].

Central sensory integration processes are limited and an increased attentional demand decreases the ability to rapidly integrate proprioceptive cues and respond with efficient motor strategies [1,2]. The permanently changing profile of sensory sources, especially during sports tasks, requires efficient sensory reweighting to ensure optimal motor control for knee stability [12,25]. Alterations in brain regions responsible for proprioceptive reweighting in individuals with ACL rupture have been reported [18,26,27], however the exact mechanisms behind postural plasticity remain to be established [10].

The aim of this study was to assess proprioceptive reweighting in young, elite handball players without ACL injury as a first step to increase understanding of sensory integration in at-risk populations. To this purpose, we evaluated the effects of a vibration-induced postural perturbation on different surfaces. We hypothesized that dynamic proprioceptive reweighting would be sub-optimal in a proportion of these healthy individuals.

Material and methods

A priori calculation of the number of subjects required to obtain a statistical power of 0.80 and type 1 error of 0.05, showed that at least 22 subjects were needed [5]. Inclusion criteria were elite handball players aged between 14 and 18 years. Exclusion criteria were known neurological disorders, vestibular impairment and trunk or lower limb injuries within the previous 3 months. All subjects provided written informed consent and the protocol was approved by the Ethical Committee of Savoie Mont-Blanc University.

To ensure proper reliability, the procedure was the same detailed by Kiers et al. [28]. Participants stood barefoot in bipedal stance with their arms relaxed by their sides and their head in neutral. Foot position was standardised, and vision was prevented using opaque goggles. Two conditions were evaluated: "*firm*" (standing on the force plate) and *"foam"* (Physiopad®; 50x41x5cm; 52kg/m^3).

Four muscle vibrators (VB115, Techno Concept, France) were placed bilaterally on the triceps surae (TS) and the lumbar paravertebral muscles (LPM). Vibration frequency was set at 80Hz with an amplitude of 0.5mm to stimulate the muscle spindles [29]. To avoid a learning effect [28], computer software automatically and randomly triggered the vibrators. This also ensured that neither the participant nor the experimenter could anticipate the next site of vibration. Each trial lasted for 50 seconds. Recordings began 10s prior to vibration ("PRE"), vibration was applied for 20s ("VIB") and the recording continued for another 20s during the re-stabilization period ("POST").

Electromyographic (EMG) activity was recorded from three muscles of the dominant limb (i.e. the preferred push-off limb) using surface electrodes positioned by the same experimenter according to the surface EMG for non-invasive assessment of muscles (SENIAM) guideline [30], after the skin was shaved and cleaned. We evaluated the gluteus medius (GM) and semi tendinous (ST) because sub-optimal activity of these muscles is related to the risk ACL injury [16], and fibularis longus (FL) because it plays an important role in postural control on different surfaces [31].

A force platform (AMTI, model BMS464508, Watertown, MA, 1000Hz) connected to a measurement card (PCIM-DAS16 card, measurement computing, A/D conversion 16bits) was used to record centre of pressure (CoP) displacement. Signals were stored for subsequent analysis using DColl software (GRAME, Laval University, Quebec, Canada) and filtered using a Butterwoth low-pass, fourth order filter with a cut-off frequency of 10Hz. Anterior/Posterior centre of pressure displacement (*d*CoP) and velocity (*v*CoP) were calculated using custom software developed in Matlab (GRAME, Laval University, Quebec, Canada). EMG signals were recorded with pre-amplified electrodes (type SX230-1000, Biometrics Ltd, Newport, UK, 1000Hz) placed 2cm apart on the muscle bellies. The EMG signal was band-pass filtered (15-450Hz) close to the recording site.

We analysed variables that have been shown to be the most reliable indicators of the response to muscle vibration for the evaluation of proprioceptive weighting and balance recovery [28]. Relative Proprioceptive Weighting (RPw) is the ratio between the effects of vibration of the TS and the LPM (absolute TS/(absolute TS+absolute LPM)) on CoP displacement (*d*RP_W) and velocity (*v*RP_W). It provides a reliable indication of individual proprioceptive strategies [5,8,28,31–33]: an RP_W of 1 indicates 100% reliance on ankle afferences while an RP_W of 0 indicates a 100% reliance on lumbar afferences.

The change in the *d*RP_W between the *firm* and *foam* supports (expressed as a percentage of the *d*RPw on the firm support) was used to dichotomize the sample according to the relocation of proprioceptive predominance on each support [5]. A change <100% corresponded to a reallocation of signals from ankle to lumbar and reflected a *plastic* proprioceptive profile. Conversely a change ≥100% indicated a *rigid* profile with no anatomical shift in proprioceptive reliance and the continued use of an ankle steered strategy [5,32,33].

Proprioceptive reintegration was evaluated by quantifying balance recovery after muscle vibration cessation. At the instant of ankle vibration cessation, an abrupt forward motion (anterior “overshoot”) of the subject occurs [28], the magnitude of which reflects sensory reweighting processes. We calculated this as the distance between the mean CoP position during the last 5s of vibration and the maximum anterior "peak" of the CoP position after vibration cessation [34].

We expressed absolute balance recovery during the first 20 seconds after vibration cessation ($REC_{abs\,(30\text{-}40)}$ and $REC_{abs\,(40\text{-}50)}$) as the mean CoP position compared to PRE [28]. The slope of the recovery (REC_{slope}) was calculated by the distance between the maximum anterior "peak" of the CoP position and the mean CoP position during the last ten seconds of the trial (Figure 1). To calculate muscle activity during each period, we normalised the RMS values of the EMG signals using the "PRE" activity (100%) and compared this to the VIB and POST conditions [31].

Normality was assessed using the Shapiro–Wilk test. The level of significance was fixed at $p<0.05$. Postural variables were compared between the *firm* and *foam* surfaces using the paired t-test. Analysis of Variance (ANOVA) was used to compare EMG activity between PRE, VIB and POST. Effect sizes (Cohen's d or partial η^2) were calculated for all comparisons and compared using the Hopkins scale. Post-hoc pairwise comparisons were performed as necessary. The statistical analysis was performed using JASP (Amsterdam 0.12.2.0).

Results

Thirty-five elite handball players were included (Table 1). dRP_W and vRP_W were significantly lower on *foam* compare to *firm* surfaces in the total sample (0.71±0.18 vs 0.82±0.11; $p=0.004$; $d=0.521$ and 0.76±0.16 vs 0.65±0.22; $p=0.018$; $d=0.419$ respectively) (Figure 2a).

There was a large inter-participant variability in the change in dRP_W (Figure 2a). Twenty participants used a *plastic* and 15 used a *rigid* strategy. dRP_W was significantly higher in the *rigid* than the *plastic* group on the *foam* surface (0.84±0.09% vs 0.61±0.16%; p<0.001; d=-1.648), indicating a large predominance of ankle signals whereas the *plastic* strategy involved more lumbar signals (Figure 2b).

During the recovery period (POST), "overshoot" and "REC_{slope}" values were significantly higher in the *plastic* than the *rigid* group (7.82±3cm vs 5.64±1.91cm; p=0.019; d=0.840 and 0.36±0.29cm vs 0.19±0.11cm; p=0.05; d=0.696 respectively). There were no between group differences for $REC_{abs(30-40)}$ or $REC_{abs(40-50)}$.

ST activity differed significantly during TS vibration on *firm* surface between the two groups (p<0.001, η^2_p=0.346) (Figure 3). Post-hoc comparisons showed that ST activity was higher during POST compared to VIB (p<0.001) and PRE (p=0.002) for the total sample and was higher in the *rigid* than the *plastic* group (p=0.027, η^2_p=0.152). ST activity was higher in POST compared to VIB in both groups (169.2±86.1% vs 122.2±64.5%; p=0.02 and 124.8±32.8% vs 89.84±21.9%; p=0.031 for *rigid* and *plastic* respectively). However, the difference between PRE and POST was only significant for the *rigid* strategy (p<0.001*)*. ST activity was also higher in the *rigid* than *plastic* group during POST (169.2±86.1% vs 124.5±32.8%; p=0.048), with no between group difference during VIB. There were no between group differences for the LF or GM.

Discussion:

The results of this study confirm that healthy individuals use an ankle proprioceptive strategy on firm surfaces and a lumbar strategy on foam surfaces [5,31,35]. Comparison of the $d\mathrm{RP_W}$ values with other studies (Table 2) showed that proprioception was more predominantly ankle steered in our population. This could be due to the young age and athletic nature of the sample since both age [3,36] and sport [10,24] influence ankle proprioceptive integration. As a learning effect exists for the TS vibration, an attenuation of the effect on *foam* conditions may also exist in other studies [28] since no trials randomization was performed. Comparisons between studies based on raw RP_W values are difficult because the exact foam densities used in the procedures are rarely described. The change in RP_W between the two conditions seems more relevant for analyzing the behavior and proprioceptive plasticity of the participants (Figure 2b).

The main finding of this study is the high degree of variability in proprioceptive strategies among the healthy athletes. Since plasticity is defined as the CNS ability to change its activity in response to extrinsic stimuli [37], we categorized two opposing behaviours: a *plastic* strategy that involved a switch from ankle to lumbar proprioception when necessary and a *rigid* one that relied on ankle signals regardless of their accuracy. These results confirm the heterogeneity that exists in the weighting of sensory inputs by individuals to control their balance [2,6]. The exact reason of this discrepancies remain to be defined but natural predisposition could explain both postural skills and proprioceptive abilities in athlete [10,24].

The RPw values in the *rigid* group were similar to those in pathological populations on foam surface (Table 2); suggesting that this strategy involves the overuse of unreliable signals and sub-optimal sensory integration processes [5,32]. Indeed, similarly to the rigid group, an ankle steered strategy on foam surface was observed among LBP patients [5,32,33] or after acute back muscle fatigue [8] (i.e higher RP_W compare to healthy individuals or to firm condition). *Rigid* behaviour might be therefore considered as less appropriate when postural demand increase [4,5,33]. Furthermore, reintegration of proprioceptive signals after the perturbation was more efficient in the *plastic* group, suggesting that plastic proprioceptive postural control is more optimum [3,12,14]. We were unable to ascertain if the overshoot and recovery slope values for the rigid group were similar to values in pathological populations or those at risk of injury because these variables have not yet been used for that purpose.

Concerning EMG signals, ST activity increased during the re-stabilization (POST) period compared to the VIB period in both groups. However, it was only significantly higher than the PRE period in the *rigid* group. This higher level of activity during the recovery phase could indicate difficulty recovering from the balance perturbation and less efficient postural control [38]. Further studies should investigate quadriceps activity to evaluate the agonist/antagonist ratio and motor control associated with different sensory integration strategies. Indeed, an increase in muscle co-contraction is a maladaptive postural control strategy that has been found in older subjects [39] and individuals with an increased risk of falls [40].

Rigid postural patterns indicate a lack of flexibility in response to perturbations [41]. Neuromuscular flexibility may protect from injury as greater degree of movement variability reduced the risk of ACL injury after reconstruction [41]. Impaired activation and functional connectivity between the left primary and secondary somatosensory cortex and the cerebellum has also been associated with future ACL rupture [20,22,42]. As optimal postural stability requires appropriate proprioceptive reweighting [1,8,32,33], and poor balance control and inadequate hamstring recruitment are associated with an increased risk of ACL tear [16,17], further studies are needed to assess the relationship between proprioceptive strategies and injury risk.

Conclusion

The modulation of somaesthetic integration appears to be heterogenous among young, healthy, elite handball players. Two distinct behaviours were found: a *plastic* strategy that involved a proprioceptive reweighting from the ankle to lumbar region, and a *rigid* strategy involving different muscle activation and impaired balance recovery. Whether the use of different proprioceptive strategies predisposes athletes to injuries such as ACL rupture remains to be defined.

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Tables

Table 1. Mean (±SD) baseline characteristics

	Females (n=14)	Males (n=21)
Age (years)	14.4 (0.8)	15.6 (0.8)
Height (m)	1.7 (0.1)	1.9 (0.1)
Weight (kg)	58.6 (8.1)	74.2 (7.9)
BMI ($kg.m^{-2}$)	20.4 (1.9)	21.4 (1.6)

Table 2. RP_W values from previous studies. LPB= Low Back Pain

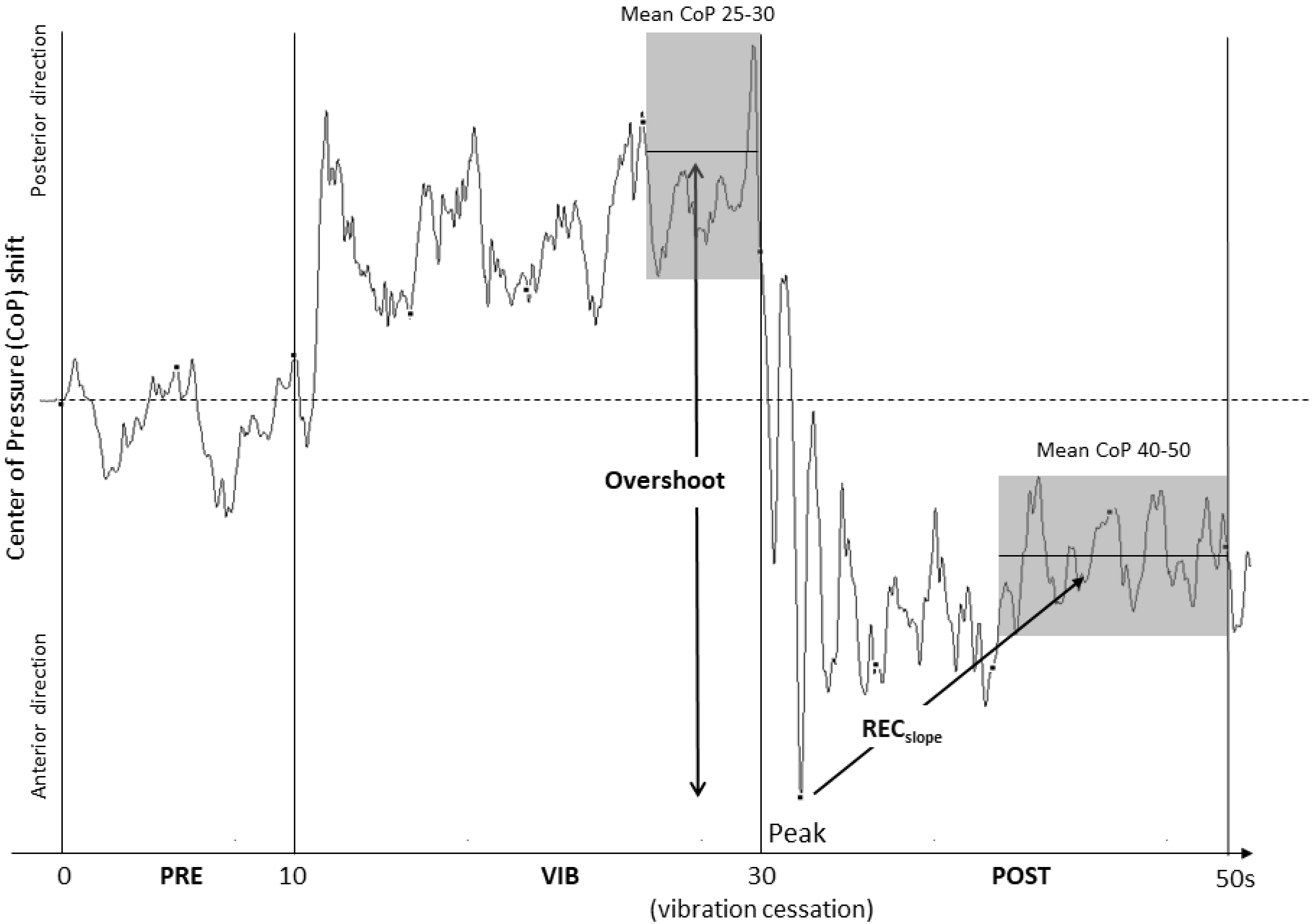
Study *(sample size; age in years ±SD)*	**Support condition**							
	Firm				**Foam**			
This study:	Rigid: 0.77	Plastic: 0.86	Overall: 0.82		Rigid: 0.85	Plastic: 0.62	Overall: 0.71	
Brumagne et al. [5]: *LBP (21; 23.5±1)* *Healthy (24; 23±1.6)*	LBP: 0.82		Healthy: 0.68		LBP: 0.85		Healthy: 0.45	
Claeys et al. [33]: *LBP (106; 18.5±0.5)* *Healthy (50; 19.6±1.6)*	LBP: 0.75		Healthy: 0.62		LBP: 0.55		Healthy: 0.40	
Johanson et al. [8] *LBP (16; 22 ±1.1)* *Healthy (16; 22.7 ±1.7)*	LBP Normal: 0.85	LBP Fatigued: 0.86	Healthy Normal: 0.73	Healthy Fatigued: 0.78	LBP Normal: 0.86	LBP Fatigued: 0.86	Healthy Normal: 0.52	Healthy Fatigued: 0.72
Claeys et al. [32] *No LBP-No LBP (22;20.5 ±3.8)* *No LBP-LBP (30; 20.5 ±2)* *LBP-No LBP (9; 21 ±1.9)* *LBP- LBP (29; 19.9 ±0.9)*	No LBP-No LBP: 0.68	No LBP-LBP: 0.76	LBP-No LBP: 0.72	LBP-LBP: 0.72	No LBP-No LBP: 0.42	No LBP-LBP: 0.55	LBP-No LBP: 0.5	LBP-LBP: 0.52
Forestier et al. [31] *(10; 23.5 ±3)*	Healthy: 0.66				Healthy: 0.47			

Figures

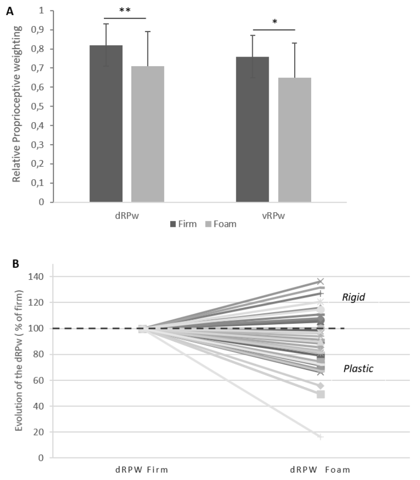
Figure 1. Representative example of sagittal displacement of the centre of pressure on the firm surface with vibration of the TS.

Figure 2. (A) Mean (±SD) dRP_W and vRP_W values for the foam and firm conditions. $*p<.05$ and $**p<.01$ (B) Individual change in dRP_W from the firm to the foam condition: dRP_W increased for the *rigid* strategy (>100%) and decreased for the *plastic* strategy (<100%).

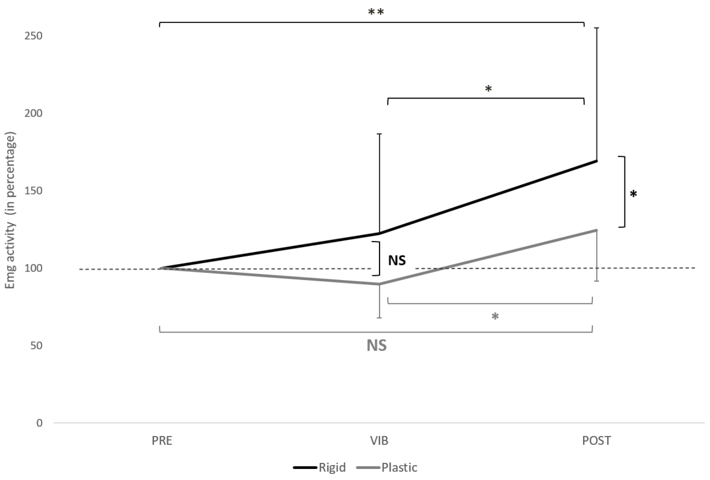
Figure 3. Mean EMG activity (normalised to the pre period) of the ST for the firm condition during PRE, VIB and POST during TS vibration. The dark line represents individuals with a rigid strategy and the grey line a plastic strategy. (NS) non-significant, $*p<.05$ and $**p<.001$



Mean CoP 25-30
Posterior direction
Center of Pressure (CoP) shift
Anterior direction
Overshoot
Mean CoP 40-50
REC$_{slope}$
Peak
0
PRE
10
VIB
30
(vibration cessation)
POST
50s



A
**
*
Relative Proprioceptive weighting
1
0,9
0,8
0,7
0,6
0,5
0,4
0,3
0,2
0,1
0
dRPw
vRPw
Firm
Foam
B
Evolution of the dRPw (% of firm)
140
120
100
80
60
40
20
0
Rigid
Plastic
dRPW Firm
dRPW Foam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*
*
NS
*
NS
Emg activity (in percentage)
250
200
150
100
50
0
PRE
VIB
POST
Rigid
Plastic