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Micro movements of the upper limb in fibromyalgia: The relation to proprioceptive accuracy and visual feedback

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ABSTRACT

The purpose of this study was to explore the role of visual and proprioceptive feedback in upper limb posture control in fibromyalgia (FM) and to assess the coherence between acceleration measurements of upper limb micro movements and surface electromyography (sEMG) of shoulder muscle activity (upper trapezius and deltoid). Twenty-five female FM patients and 25 age- and sex-matched healthy controls (HCs) performed three precision motor tasks: (1) maintain a steady shoulder abduction angle of 45° while receiving visual feedback about upper arm position and supporting external loads (0.5, 1, or 2 kg), (2) maintain the same shoulder abduction angle without visual feedback (eyes closed) and no external loading, and (3) a joint position sense test (i.e., assessment of proprioceptive accuracy). Patients had more extensive increase in movement variance than HCs when visual feedback was removed ($P < 0.03$). Proprioceptive accuracy was related to movement variance in HCs ($R \geqslant 0.59$, $P \leqslant 0.002$), but not in patients ($R \leqslant 0.25$, $P \geqslant 0.24$). There was no difference between patients and HCs in coherence between sEMG and acceleration data. These results may indicate that FM patients are more dependent on visual feedback and less reliant on proprioceptive information for upper limb posture control compared to HCs.

1. Introduction

Fibromyalgia (FM) is a chronic pain syndrome characterized by widespread pain and a range of somatic, psychological and functional symptoms. A common symptom is alterations in the control of movements; including deficits in manual dexterity (Perez-de-Heredia-Torres et al., 2013), postural control (Jones et al., 2011), balance, and coordination (Watson et al., 2009). Moreover, FM patients have a distinctly different pattern of upper limb micro movements compared to healthy controls (HCs) (Bardal et al., 2012). Micro movements are low amplitude, rhythmic limb oscillations that can be recorded when maintaining a steady limb posture against gravity. Patients with FM had a higher share of their upper limb oscillations in low frequencies (1–3 Hz) compared to HCs, where the majority of the micro movements were found in the 8–12 Hz range (Bardal et al., 2012). However, the underlying cause of this divergent movement pattern in FM is not clear.

Micro movements with frequencies below 3 Hz are commonly referred to as control error (Endo and Kawahara, 2010), and are associated with voluntary control based on visual cues (Loncharich and Newell, 2012). Higher frequencies between 6 and 15 Hz are often considered to be involuntary limb oscillations, i.e., physiological tremor, representing a part of the cortical drive to the muscles (McAuley and Marsden, 2000; Raethjen et al., 2002). The frequency distribution of micro movements is however a result of a complex and still debated set of biomechanical and physiological processes, including oscillatory reflex loops (Durbaba et al., 2005) and mechanical factors (Takanokura and Sakamoto, 2001), which will vary between muscles and limbs and thereby complicate the interpretation of these movements.

Both amplitude (Carignan et al., 2009) and frequency distribution (Takanokura and Sakamoto, 2001) of micro movements can be modified by altered sensory information. FM is associated with several sensory alterations such as amplification of sensory information (Berglund et al., 2002) and disruption of somatosensory processing (Pujol et al., 2014). The dominance of low frequency limb oscillations in FM (Bardal et al., 2012) may therefore reflect a sensory deficit in processing of afferent feedback with a biased dependency upon visual information in upper limb posture control. However, there is currently no evidence to support this hypothesis. Micro movements are also affected by the mechanical properties of the limb (Takanokura and Sakamoto, 2001). An alternative explanation of the observed difference in micro movement

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patterns between FM patients and HCs is therefore the different contribution from neural and mechanical factors.

The main objective of the present study was to explore the role of visual and proprioceptive feedback in upper limb posture control in FM. A second objective was to assess the neural contribution to the upper limb micro movements by calculating the coherence between upper limb acceleration and shoulder muscle activity in trapezius and deltoid during a steady upper limb posture.

2. Methods

Twenty-five female patients with FM and 25 sex- and age-matched (±4 years) HCs were included in the study (Table 1). All eligible patients were examined by a physician to verify FM diagnosis as defined by the American College of Rheumatology (Wolfe et al., 1990). The procedure for enrolment and exclusion criteria have been described elsewhere (Bardal et al., 2015).

The study protocol was approved by the Regional Committee for Ethics in Medical Research (project no. 4.2008.2115) and all participants signed an informed consent before enrolment. The study was carried out according to the Declaration of Helsinki.

2.1. Protocol and equipment

Subjects performed three precision motor tasks; Task1: maintain a steady shoulder abduction angle of 45° guided by visual feedback and while supporting external loads of 0.5, 1, or 2 kg. Task2: maintain the same shoulder abduction angle without visual feedback (eyes closed) and no external loading. Task3: actively reproduce the same shoulder abduction angle without visual feedback (eyes closed), i.e., a shoulder joint position sense (JPS) test. Subjects also performed three maximal voluntary contractions (MVCs) at 45° shoulder abduction. The contractions lasted 5 s and were separated by 1 min rest. Subjects were seated in an adjustable chair during all tasks and were strapped across the pelvic and upper body to restrict torso movements. All tasks were performed with forearms pronated and in parallel to the sagittal plane. Elbows were flexed at 90°. The examiner placed the arm in the correct position for each trial.

In task1, subjects performed three bilateral isometric shoulder abduction contractions where the goal was to match the upper arm posture with the set target of 45° shoulder abduction. Each contraction lasted 45 s separated by 1 min rest. Target position and real time feedback was provided on an 18.5″ computer screen placed at ~75 cm distance at eye level. The target position was a horizontal line in the centre of the screen while the feedback signal was presented as a tracking line giving shoulder abduction angle. The subjects were instructed to cover the target line with the feedback signal. External loads were attached to a forearm orthosis ~8 cm from the elbow rotation centre. This protocol was used to replicate the set-up used by Bardal et al. (2012) and to verify the findings in a new study sample. In task2, the subjects held the steady limb posture from task1 unloaded and with eyes closed. The subjects were instructed to maintain a steady limb posture until 5 s of stable signal was obtained. Task3 was a shoulder JPS test with *passive* positioning and *active* repositioning. In the passive positioning the examiner moved the subjects' dominant arm with ~10°/s from the starting position (0° shoulder abduction) to the target position (45° shoulder abduction). The subjects were then asked to keep this limb posture for 5 s and try to remember it. The subjects were then asked to actively reproduce the same limb posture. Eyes were closed during both trials. The positioning error between the passive positioning and active repositioning is defined as proprioceptive accuracy in the remaining text.

Upper limb micro movements were measured by a three-axis acceleration sensor (Delsys Inc., Boston, USA; range: ±2 g, resolution: 0.006 g, bandwidth: 0–50 Hz) placed on the mid-acromiale-radiale line of the lateral surface of the dominant upper arm. Muscle activity of deltoid and upper trapezius was recorded by surface electromyography (sEMG). The electrodes were bipolar (1 cm spacing), single differential, Ag, polycarbonate electrodes with a detection area of 10 mm^2, placed following SENIAM guidelines (Hermens et al., 2000). The reference electrode was placed on C7. The signals were amplified with a gain of 1000 and band-pass filtered (20–450 Hz) with a Bagnoli 16-channel sEMG system (Delsys Inc., Boston, USA) prior to sampling. Force output was recorded by two force transducers (Interface Inc. Scottsdale, USA) attached to the forearm orthosis with non-elastic polyester bands. Sample rate was 1000 Hz for all measurements. Pain in the neck/shoulders and low back was scored on a visual analogue scale (VAS) prior to the tasks and after the completion of the MVCs.

2.2. Data analysis

All analysis was computed in Matlab (Mathworks, Nattick, USA). For limb posture analysis, 30 s of stable signal was used. A stable signal was defined as a signal without noticeable limb movements assessed by visual inspection. To assess the effect of visual feedback, the first 5 s of stable signal from task1 (0.5 kg) was compared with 5 s of the stable signal from task2. The vertical acceleration signal was filtered with an orthogonal wavelet filter, and the Welch's averaged, modified periodogram method (Welch, 1967) was used to calculate the 1–60 Hz power spectrum of the detrended time series using bin sizes of 1 Hz. Power in low frequency band (1–3 Hz) and a high frequency band (6–15 Hz) were calculated (Raethjen et al., 2002). For between-group comparisons the power was normalized to the total spectral power, while absolute power was used for within-group comparisons. Total power was used as a measure of movement variance.

The digitalized sEMG signals were band-pass filtered (10–350 Hz) with an 8th order Butterworth filter. Root-mean-square (RMS) was calculated in 200 ms non-overlapping windows and normalized to maximum RMS obtained during the MVCs. Further, the median frequency of the sEMG signal was calculated for task1. For the coherence analysis, sEMG data was band-pass filtered (50–350 Hz) with a 4th order Butterworth filter to remove movement artefacts while keeping the tremor bursts in the rectified sEMG. Calculation of the power spectrum of the rectified sEMG was performed using the same method as for the acceleration signal. Coherence was calculated as:

$$\text{Coherence} = \frac{|\text{cross PSD}|^2}{\text{PSD-sEMG} \times \text{PSD-Acc}}$$

where cross power spectral density (PSD) is the cross spectrum of the sEMG and accelerometer signal, PSD-sEMG is the power spectral

Table 1
Subject characteristics of patients with fibromyalgia (FM) and healthy controls (HCs). Values are mean ± standard deviations.

	FM (n = 25)	HCs (n = 25)	P[a]
Age (years)	55.8 ± 6.8	51.8 ± 8.3	0.12
Body mass index (kg/m^2)	28.8 ± 4.1	25.2 ± 3.5	0.71
Pressure pain threshold (kPa)	158 ± 52	261 ± 86	0.001
Neck/shoulder pain before testing (VAS)	14.4 ± 14.3	3.4 ± 7.1	0.001
No. of tender points	14.1 ± 2.3	NA	
Years since diagnosed	9.7 ± 6.1	NA	
FIQ score	46.7 ± 18.3	NA	

FIQ, fibromyalgia impact questionnaire; NA, not applicable; VAS, visual analogue scale (0–100 mm).
[a] Independent samples t-test.