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On the nature of the electromyographic signals recorded during vibration exercise

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Abstract

Purpose Surface electromyography (EMG) has been widely used to measure neuromuscular activity during vibration exercise (VE) to investigate the underlying mechanisms elicited by VE. However, the EMG spectrum recorded during VE shows sharp peaks at the vibration frequency whose interpretation remains controversial. Some authors considered those peaks as a result of motion artifacts, while others interpreted them as due to vibration-induced neuromuscular activity. The aim of the present study is to clarify the nature of those sharp peaks observed during VE.

Methods Three independent EMG measurements were performed during VE: in vitro (IVT), in vivo at rest (IVV_R), and in vivo during voluntary contraction (IVV_C). The amplitudes of the EMG vibration frequency components (A_{VF}) were extracted for all measurements. The conduction velocity (CV) of the vibration frequency components and the full EMG spectrum were also estimated during voluntary contraction.

Results Our spectrum analysis revealed small A_{VF} for IVT and IVV_R, accounting for only 3.3 and 7.6 % of that obtained from IVV_C. Moreover, the CV estimation indicated the EMG vibration components to propagate along the muscle fiber with $CV \approx 6.5$ m/s, comparable to the CV estimated using the full EMG spectrum (5.7 m/s).

Conclusion We may therefore conclude that the sharp spectral peaks observed during VE are mainly due to vibration-induced muscle activity rather than motion artifacts.

Keywords Electromyography · Vibration exercise · Motion artifacts · Tonic vibration reflex · Conduction velocity

Abbreviations

A_{VF}	Amplitude of the vibration frequency components
CV	Conduction velocity
CV_{ACC}	CV of the acceleration signals
CV_{EMG}	CV of the full EMG spectrum
CV_{VF}	CV of the vibration frequency components
EMG	Electromyography
IVT	In vitro
IVV _C	In vivo during voluntary contraction
IVV _R	In vivo at rest
MF	Mean frequency
MVC	Maximum voluntary contraction
TVR	Tonic vibration reflex
VE	Vibration exercise
WBV	Whole body vibration

Introduction

It is extensively reported that vibration exercise (VE) can have a beneficial effect on muscle strength and power performance, possibly due to enhanced neural excitation on skeletal muscles (Delecluse et al. 2003; Cardinale and Bosco 2003; Rees et al. 2008; Cochrane and Stannard 2005; Rittweger et al. 2003). Various forms of vibration training have been proposed in the last decade. Whole

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body vibration (WBV) platforms have been popular training devices in recreational and athletic training as well as for rehabilitation purposes (Cardinale and Bosco 2003; Cochrane and Stannard 2005; Cardinale and Lim 2003; Rees et al. 2008). The acute and long-term effects of WBV training have been investigated by several authors (Abercromby et al. 2007; Cochrane and Stannard 2005; de Ruiter et al. 2003a; Rittweger et al. 2000, 2003; Blottner et al. 2006; de Ruiter et al. 2003b; Kvorning et al. 2006). Vibrating dumbbells (Bosco et al. 1999), vibrating pulley-like devices (Issurin et al. 1994; Issurin and Tenenbaum 1999), and vibrating barbells (Poston et al. 2007) have been proposed for the upper limbs. Other devices producing a sinusoidal force-modulated stimulation (Kinser et al. 2008; Sands et al. 2006; Mischi and Cardinale 2009; Xu et al. 2012) have also been shown to acutely enhance strength, power, and flexibility, mainly due to the neuromuscular demands placed by such force modulation on skeletal muscles.

The beneficial effects of VE have been previously attributed to a specific reflex mechanism named tonic vibration reflex (TVR) (Eklund and Hagbarth 1966; Hagbarth and Eklund 1966; Hagbarth et al. 1976; Bongiovanni et al. 1990). TVR derives from the stimulation of Ia-afferents when applying a sinusoidal stimulation directly to a muscle or a tendon (Bongiovanni et al. 1990; Fromm 1976; Burke et al. 1976). However, TVR seems also to be modulated by alterations in spindle sensitivity through γ control (Matthews 1966; Ribot-Ciscar et al. 2000). By applying a vibration directly to the tendon, previous studies have suggested TVR to be composed of motor unit activity synchronized and unsynchronized with the vibration cycle (Hori et al. 1989; Romaiguère et al. 1991). Studies have also indicated that TVR may contribute to muscle fatigue and/or increase the risk of cumulative trauma disorders observed after repetitive exposure of the hand to vibration (Park and Martin 1993).

The underlying mechanisms explaining the effect and risk of VE are still unclear, limiting the possibilities to introduce and exploit vibration training in various populations. To better understand those mechanisms, surface electromyography (EMG) is widely used to measure the level of neuromuscular activity in the targeted muscles exposed to VE. In general, the EMG amplitude can be estimated by the root mean square value of the EMG signal and used as an indicator of force production (Staudenmann et al. 2010). The mean frequency (MF) and average conduction velocity (CV) of the surface EMG signal are indicators of myoelectric muscle fatigue (Naeije and Zorn 1982; Moritani et al. 1986). The MF can be estimated as the first statistical moment of the EMG amplitude spectrum (Merletti et al. 1990). The CV of the motor unit action potentials along the muscle fibers can be estimated using two or more recording

electrodes placed on the skin along the muscle fibers with known inter-electrode distance (Hunter et al. 1987; Farina and Merletti 2004; Farina and Negro 2007; Mischi et al. 2010; Rabotti et al. 2010).

During VE, sharp peaks can be observed at the vibration frequency and its harmonics in the spectrum of the surface EMG recordings. In some studies, these peaks have been interpreted as the result of motion artifacts (Abercromby et al. 2007). These results were confirmed by other authors using dummy-like electrodes, i.e., electrodes placed on the patella (Fratini et al. 2009). Possible mechanisms contributing to the generation of motion artifacts were previously suggested to be variations in the half-cell potential caused by ionic gel movement at the metal–electrolyte interface and the skin–electrode interface (Kahn 1965), skin potential variations caused by skin stretch (Tam and Webster 1977; Ödman and Oberg 1982; Ödman 1981; de Talhouet and Webster 1996), and movement of the recording cables (Huhta and Webster 1973; Simakov and Webster 2010). Based on these findings, band-stop filters (Mischi and Cardinale 2009; Mischi et al. 2010) and accelerometry-based adaptive filters (Xu et al. 2013) were applied to the EMG signals recorded during VE to suppress those frequency peaks.

On the contrary, some studies analyzed the frequency spectrum of the EMG signals recorded during resting condition and voluntary contraction, while vibration was applied perpendicularly to the distal tendons of the hand flexor muscles (Martin and Park 1997). The spectra were found to be flat while resting and to exhibit prominent peaks correlated with the vibratory stimulus during voluntary contraction. The prominent peaks observed during voluntary contraction were therefore interpreted as TVR, representing the degree of motor unit synchronization. Some other studies compared the EMG signal recorded by contacted electrodes placed over the soleus muscle with those recorded by dummy electrodes placed on insulating tape fixed on top of the skin (Ritzmann et al. 2010). They found an average EMG amplitude ratio of 7 % between the dummy electrodes and the contacted electrodes at the vibration frequency. In these studies the spectra and latencies of the EMG signals recorded during WBV matched those of the stretch reflex mechanically evoked by an ankle ergometer (Ritzmann et al. 2010). Based on these findings, the same conclusion as in (Martin and Park 1997) was drawn.

It is evident that the interpretation of the nature of the EMG recordings during VE is controversial. Indeed, before drawing conclusions based on the EMG recordings, their nature has to be examined carefully. The aim of the present study is to clarify whether the sharp peaks at the vibration frequency and its harmonics during VE consist primarily of motion artifacts or stretch reflex. To this end, three measurements were carried out independently.

First, *in vitro* measurements (IVT) were employed to test whether the movement of the recording cable and electrodes during vibration causes motion artifacts in the electrical signal.

Secondly, *in vivo* measurements at resting condition (IVV_R) were carried out to provide a signal that would mainly be influenced by motion artifacts, with only little interference from muscle signals. This signal can be used to examine motion artifacts caused by movement of the ionic gel, causing variations of the half-cell potential, skin potential variations, and variations in skin and electrodes impedance.

Finally, *in vivo* measurements during voluntary contraction (IVV_C) were performed on eight subjects. It is widely accepted that, during muscle contraction, motor unit action potentials propagate along the muscle fiber with a velocity in the order of few m/s (Davies and Parker 1987; Li and Sakamoto 1996). Therefore, it is reasonable to hypothesize that if the components at the vibration frequency and its harmonics consist mainly of stretch reflex, they should propagate along the muscle fiber similar to other EMG components. Conversely, if they do not propagate or propagate with different characteristics, it could be speculated that they are due to motion artifacts. Our IVV_C measurements were designed to test this hypothesis. All the measurements were performed by a fully controlled dedicated setup (Xu et al. 2013).

Method

Experimental setup

Actuator

The adopted actuator for VE was previously described in (Xu et al. 2013), whose scheme is shown in Fig. 1. Briefly, a three-phase permanent magnet motor (MSK060C, Bosch Rexroth, Boxtel, The Netherlands) is employed to generate a force consisting of a constant baseline force with superimposed sinusoidal vibrating force. An example of such vibrating force is shown in Fig. 2. A 50-cm aluminum bar is mounted perpendicular to the motor output shaft to apply the generated vibrating force pulling down a rope. The position of the bar is measured by a rotary encoder embedded in the motor, providing a visual feedback to the training subject. The driving voltage to generate sinusoidal force patterns is generated by a National Instrument card (PCI-5402, National Instruments, Austin, TX, USA), which is connected to a PC and controlled by dedicated software implemented in LabView® (National Instruments, Austin, TX, USA).

Fig. 1 Scheme of the actuator

Fig. 2 An example of vibrating loading function consisting of a baseline force with superimposed sinusoidal vibrating force of 40-N amplitude

The force generated by the actuator was calibrated by means of a load cell LCAE-35 kg (Omega Engineering Inc., Stamford, CT, USA), which uses resistance-based force sensors (strain gauges) in full bridge configuration and is mounted on top of the bar. The measured voltage and strain-gauge data were fitted into a second-order linear model, as described in (Xu et al. 2013). The calibration revealed a high correlation ($r = 0.98$) between the second-order model fit and the measured data, ensuring accurate control on the applied force. According to our calibration,

Fig. 3 EMG Recording: three unipolar signals were recorded from the biceps brachii. Two accelerometers were placed aligned with the electrodes to monitor their movement

forces up to 300 N and vibration frequencies up to 70 Hz could reliably be produced by this actuator.

EMG recording

Surface EMG was recorded from the biceps brachii by three circular Ag–AgCl electrodes of 1-cm diameter (3M RedDot, Nadarzyn, Poland). The electrodes were aligned with the muscle fiber direction and placed between the tendon (at the elbow side) and the muscle belly. The inter-electrode distance was 2.5 cm. Two accelerometers were placed aligned with the recording electrodes to measure the movement of the electrodes, as shown in Fig. 3. All the measured signals were amplified and digitized using a 72-channel Refa amplifier (TMS International, Enschede, The Netherlands). The sampling frequency for each channel was 2,048 Hz. Active shielding and grounding of the cables was used to minimize the powerline (50 Hz) interference. The ground electrode was placed on top of the clavicle on the right side of the subject.

Subjects

Eight male subjects (age = 28 ± 5 years) volunteered for the experiment. All the subjects were right-handed. They were healthy with no previous neurological disorder or injuries of the upper limbs. Written informed consent was obtained from every subject prior to participation in the study, and the test procedures were approved by the board of the Student Sport Centre of the Eindhoven University of Technology.

Fig. 4 Setup configuration for the three measurements: **a** IVT; **b** IVV_R; **c** IVV_C

Measurement protocol

In vitro measurements

IVT measurements were designed to test the motion artifacts caused by movement of the electrodes and the recording cables during vibration. They were carried out by connecting the sensing surface of the two electrodes together and then mounting them on top of the aluminum bar of the experimental setup, as shown in Fig. 4a. Six trials of 12-s sinusoidal vibration with different frequencies (0, 25, 35, 45, 55, and 65 Hz) were then applied to the electrodes through the bar in a random order. The adopted vibrating force was ± 40 N (Xu et al. 2013; Mischi and Cardinale 2009) around a baseline force equal to zero. An accelerometer was placed close to the electrodes to assess the resulting vibration of the bar. The bipolar electrode signal was recorded during vibration. Five repetitions were performed, and the average amplitude spectrum was derived and used for future analysis.

In vivo measurements at resting condition

IVV_R measurements aimed at investigating the motion artifacts caused by movement of the gel, skin potential variations, and skin and electrode impedance variations. The subjects were seated comfortably with their right wrist positioned on top of the aluminum bar of the experimental setup, as shown in Fig. 4b. The elbow was supported at a 90° angle to achieve a comfortable arm posture and optimal relaxation. The same 12-s vibrations used in IVT, namely 0, 25, 35, 45, 55, and 65 Hz, were applied to the wrist by the experimental setup. During vibration, EMG and acceleration signals were recorded from the biceps brachii (Fig. 3).

In vivo measurements during voluntary contraction

IVV_C measurements were designed to analyze the propagation of the EMG vibration component and the full EMG spectrum. For the IVV_C measurements, the same experimental setup and electrode position as for the IVV_R were used. Only the relative position between the elbow and the body needed to be changed to allow the subjects to correctly use the setup and perform voluntary contraction of the biceps (Fig. 4c). The subjects were instructed to sit on the fitness bench positioned on the actuator and hold the handle with their elbow at 90°. Their position was monitored by the visual feedback provided by the rotary encoder embedded in the motor. The maximum voluntary contraction (MVC) was then estimated according to the following protocol. An increasing force was applied by the actuator until the subject could not sustain it. For each subject, the test was repeated three times, separated by 3-min recovery.

The subject MVC was then determined as the maximum value of the three repeated tests. After establishing the MVC, the subjects performed 12-s isometric contractions. The force, applied by the actuator to the elbow flexor muscles during the entire 12-s contraction, was the combination of a steady force at 60 % of the subject MVC and a superimposed vibration with a constant amplitude of 40 N. During the 12-s contraction, EMG and acceleration signals were recorded from the biceps (Fig. 3) by the amplifier. Six trials with different frequencies, the same as for IVT (0, 25, 35, 45, 55, 65 Hz), were performed in random order, each one followed by a recovery period of 5 min.

Our preliminary test revealed different accelerations for the bar, the relaxed arm, and the contracted arm under the same vibrating force (40 N). This can be explained by the effects of tissue compliance. Soft tissue has been reported to act as wobbling masses that vibrate in a damped manner in response to mechanical vibration (Wakeling and Nigg 2001). Changes in muscle tension are able to alter the resonance and damping coefficient of soft tissues. The use of the accelerometers was therefore essential to monitor the acceleration and provide a scaling factor to the EMG signals recorded at different test conditions to compare them (indicated in Fig. 3).

Signal processing

Spectrum analysis

To compare the amplitudes of the vibration frequency components (A_{VF}) under different test conditions, EMG spectrum analysis was performed. For each measurement, two bipolar signals were derived by subtracting the adjacent electrode signals (1–2 and 2–3, Fig. 3). The obtained bipolar signal was first band-pass filtered between 20 and 450 Hz by a finite impulse response filter. The time interval between 5 and 8 s of the vibration period was adopted to calculate the Fourier transform, providing a frequency resolution of 0.25 Hz and allowing the extraction of the A_{VF} exactly at the vibration frequency.

Due to the effect of varying tissue compliance, direct comparison of the A_{VF} extracted from different test conditions is not appropriate. A suitable solution was therefore to normalize them with the corresponding acceleration signals, assuming that motion artifacts are linearly related to the acceleration (Ödman and Öberg 1982). The average of the two normalized A_{VF} derived from the two bipolar signals was then used for the analysis.

CV estimation

The CV of the components at the vibration frequency without harmonics (CV_{VF}), the CV of the entire EMG signals

(CV_{EMG}), and the CV of the acceleration signals (CV_{ACC}) were estimated only on the signals recorded during IVV_C measurements. The same epoch used for the spectrum analysis was also used for all the CV estimation. The recorded three unipolar signals were first normalized to obtain zero mean and unit variance. Two bipolar signals were then obtained by subtracting the normalized signals from adjacent electrodes (1–2 and 2–3, Fig. 3) and band-pass filtered between 20 and 450 Hz, similarly to what we did for the spectrum analysis.

To estimate the CV_{VF} at a single frequency, the phase method described in (Hunter et al. 1987) was employed in this study. The cross-correlation function $R_{xy}(k)$ of the two bipolar signals, x and y , was then calculated as

$$R_{xy}(k) = \frac{1}{N} \sum_{i=1}^N x(i)y(i+k), \quad (1)$$

where k is the correlation function lag and N the number of samples.

The auto-correlation function $R_{xx}(k)$ of one of the bipolar signals, x , was calculated as

$$R_{xx}(k) = \frac{1}{N} \sum_{i=1}^N x(i)x(i+k). \quad (2)$$

The auto- and cross-correlation functions are related via

$$R_{xy}(k) = \sum_{j=0}^M h(j)R_{xx}(k-j), \quad (3)$$

where $h(j)$ is the impulse response function between R_{xx} and R_{xy} , and $M = N - 1$.

The Fourier transform of the impulse response function $h(j)$ provides an estimation of the transfer function $H(f)$. The frequency response of the phase $\phi(f)$ can be calculated as

$$\phi(f) = \arctan \frac{\text{Im}[H(f)]}{\text{Re}[H(f)]}, \quad (4)$$

where $\text{Im}[H(f)]$ and $\text{Re}[H(f)]$ are the imaginary and real parts of $H(f)$. The time delay of the vibration components, τ_{VF} , can be calculated as

$$\tau_{VF} = \phi(f_v)/(2\pi f_v), \quad (5)$$

where $\phi(f_v)$ is the phase delay at the vibration frequency f_v . The CV of the vibration components, CV_{VF}, can then be calculated as

$$\text{CV}_{VF} = d/\tau_{VF}, \quad (6)$$

where $d = 25$ mm is the inter-electrode distance.

The implemented phase method is based on the assumption that the two bipolar signals differ by pure time delay

Fig. 5 Example of coherence function of two bipolar signals

without shape variation (Farina and Merletti 2004). However, the assumption of pure temporal delay is not completely fulfilled and changes in the shape of the detected surface signals along the muscle fiber direction can be expected (Farina and Merletti 2004). The coherence function $C_{xy}(f)$ (Bendat and Piersol 1980) was therefore employed as a measure of the degree to which y is linearly related to x as a function of frequency f ,

$$C_{xy}(f) = \frac{|P_{xy}(f)|^2}{P_{xx}(f)P_{yy}(f)}, \quad (7)$$

where $P_{xx}(f)$ and $P_{yy}(f)$ are the power spectral densities of x and y , and $P_{xy}(f)$ is the cross power spectral density of x and y . Previous studies have analyzed the EMG signals recorded on the biceps using stainless steel electrodes and reported low coherence for the components below 50 Hz (Hunter et al. 1987). Our measurements also indicated low coherence at lower frequency calculated using Welch's averaged modified periodogram method (Welch and Peter 1967), as shown in Fig. 5. Using coherence as an exclusion criterion, only those CV estimates with high coherence (>0.8) were taken into consideration.

The CV of the acceleration signals, CV_{ACC}, at the vibration frequencies was also estimated using the phase method, providing a reference for the mechanical vibration waves.

Furthermore, to compare the propagation of the vibration components with normal EMG signals, the average CV of the full EMG spectrum, CV_{EMG}, between 20 and 450 Hz was estimated using the cross-correlation function method in the time domain (Hunter et al. 1987). In addition, the CV of the EMG signals recorded without vibration (vibration frequency = 0 Hz) was estimated. However, the

cross-correlation function is discrete in the time lag, limiting the resolution of the estimated delay. A suitable solution to estimate the maximum value of the cross-correlation function was an interpolation around its peak value (± 3 points in this study) by a second-order polynomial, which has been reported to provide performance comparable with those obtained by ideal interpolation over all the available values (Farina and Merletti 2000).

All the signal analysis was implemented in a custom-made Matlab® (Matlab 2011a, The Mathworks, Natick, MA, USA) program.

Statistical analysis

All data are presented as mean and SD. Paired Student's *t* tests (one tail, different variance) were performed to reveal the difference in A_{VF} between different test conditions. Furthermore, CV_{VF} was compared with CV_{EMG} and CV_{ACC} , using paired Student's *t* tests to establish the significance level (*p* value) of the mean values. For all the analyses, alpha was set at 0.05.

Results

Spectrum analysis

After normalization with respect to the corresponding acceleration signal, our spectrum analysis revealed a very small A_{VF} ($0.10 \pm 0.05 \mu V/(m/s^2)$) for the IVT measurements. A significantly higher value ($p < 0.05$) was found in the IVV_R measurements ($0.23 \pm 0.37 \mu V/(m/s^2)$). However, this value corresponded to only 7.6 % of that estimated from the IVV_C measurements ($3.03 \pm 2.79 \mu V/(m/s^2)$), and their difference was significant ($p < 0.05$). Furthermore, during voluntary contraction, the normalized A_{VF} increased for increasing vibration frequency. A representative example of the normalized spectrum for all the three test conditions is shown in Fig. 6, and the average results for all the subjects are reported in Table 1.

CV estimation

For the IVV_C measurements, CV_{VF} , CV_{EMG} , and CV_{ACC} were estimated and selected according to the value of the coherence function of the two recorded bipolar signals, whose average value is shown in Fig. 7. CV estimates with coherence lower than 0.8 were excluded. After this exclusion, two subjects were included at 25 Hz and one at 35 Hz. For the other vibration frequencies, six subjects were included.

The CV estimation revealed an average value of 6.5 ± 2.4 m/s for CV_{VF} , calculated by only using the components

Fig. 6 A representative example of the normalized spectra for all the three test conditions ($f_v = 55$ Hz): **a** IVT; **b** IVV_R ; **c** IVV_C

at the vibration frequency without harmonics. A slightly lower value was estimated for CV_{EMG} (5.7 ± 1.5 m/s). An example indicating the propagation of the two EMG

Table 1 Normalized A_{VF} for different test conditions ($\mu V/(m/s^2)$)

f (Hz)	IVT	IVV _R	IVV _C
25	0.11 ± 0.07	0.24 ± 0.30	2.00 ± 1.28
35	0.10 ± 0.06	0.40 ± 0.74	2.35 ± 1.69
45	0.11 ± 0.06	0.28 ± 0.25	2.63 ± 1.40
55	0.10 ± 0.06	0.10 ± 0.10	3.98 ± 3.46
65	0.11 ± 0.05	0.14 ± 0.13	4.21 ± 4.60
Average	0.10 ± 0.05	0.23 ± 0.37	3.03 ± 2.79

Data are given as mean $\pm$ SD

Fig. 7 Average coherence of the EMG signal at adjacent electrodes for increasing frequencies **Fig. 8** Two EMG signals recorded during IVV_C ($f_v = 55$ Hz) at two adjacent electrode pairs, indicating EMG propagation

signals is shown in Fig. 8, which is derived from the same signal used in Fig. 6c. In the absence of vibration, the average CV_{EMG} over the six subjects was 5.6 ± 1.2 m/s.

Table 2 Averaged CV estimates during voluntary contraction

f (Hz)	CV_{VF} (m/s)	CV_{EMG} (m/s)	CV_{ACC} (m/s)
25	6.9 ± 0.2	6.2 ± 0.1	37.8 ± 28.7
35	4.6 ± 0	6.2 ± 0	16.0 ± 0
45	6.7 ± 2.0	5.7 ± 1.1	247.1 ± 492.0
55	7.2 ± 2.3	6.1 ± 0.6	57.3 ± 56.2
65	6.7 ± 2.2	6.1 ± 0.6	138.4 ± 164.2
Average	6.5 ± 2.4	5.7 ± 1.5	127.3 ± 273.4

Data are given as mean $\pm$ SD

Furthermore, a significantly ($p < 0.05$) higher CV was found for the acceleration signals (127.3 ± 273.4 m/s). The average CV for all the subjects is reported in Table 2.

Discussion

Spectrum analysis

IVT measurements were performed to test motion artifacts caused by cable and electrode movement, while IVV_R measurements were performed to test motion artifacts generated by the skin–electrode interface during vibration. The small A_{VF} observed in both measurements indicate minor contribution of motion artifacts to the EMG signal during VE. Some studies suggested that the highest motion artifacts may originate from the deformation of the skin under the electrode (Tam and Webster 1977; de Talhouet and Webster 1996). We performed IVV_R measurements with stretch deformation induced by pinching two clips to the skin on both sides of the recording electrodes (inner and outer arm, separated by 2 cm) and revealed no significant difference in the A_{VF} , confirming the limited contribution of motion artifacts by skin stretch deformation to the EMG signal recorded during VE.

Due to the effects of tissue compliance in relation to muscle contraction, higher accelerations were observed in IVV_C measurements as compared to IVV_R measurements. Assuming motion artifacts to increase linearly with an increase in acceleration (Ödman and Öberg 1982), the A_{VF} extracted from different test conditions were normalized based on the corresponding acceleration signals. However, the linear relation between motion artifacts and acceleration should be studied more extensively. Some studies suggested that skin-stretch-related motion artifacts increase with the pressure applied to the skin in a logarithmic fashion (de Talhouet and Webster 1996). A logarithmic relation between motion artifacts and acceleration could therefore be expected. This would in fact lead to an even larger difference in A_{VF} between IVV_C and IVV_R measurements.

Our spectrum analysis results indicate that the sharp spectral peaks observed during VE may mainly consist of

Fig. 9 Average A_{VF} without normalization for the IVV_C measurements

stretch reflex, such as TVR, rather than motion artifacts. TVR has been suggested to originate from synchronized motor units firing during the vibration cycle (Martin and Park 1997). It has been reported that the motor unit synchronization increases with increasing levels of initial muscle tension (Martin and Park 1997; Mischi and Cardinale 2009). Therefore, it is reasonable to expect the relative contribution of motion artifacts to the vibration spectral peaks to decrease for higher levels of muscle contraction.

Furthermore, Martin and Park (1997) previously found an increase in TVR with increasing vibration frequencies up to about 100 Hz in wrist and finger flexors muscles exposed to vibration. The increase in TVR was ascribed to an increase in motoneuron depolarization with the firing frequency of Ia-afferents, which leads to a recruitment of motor units with increasing threshold (Martin and Park 1997). In our IVV_C measurements, the A_{VF} without normalization were found to increase for increasing vibration frequency up to 55 Hz and then decrease (Fig. 9), supporting the hypothesis that those components at the vibration frequency may mostly consist of TVR. However, the cutoff frequency was lower in our measurements. This could not be explained by misbehavior of the Ia-afferents as in (Martin and Park 1997) since Ia-afferents were reported to be able to respond in 1:1 synchrony up to about 100–150 Hz (Burke et al. 1976). One possible explanation may relate to the vibration transmission chain in our measurements, which was more complicated as compared to (Martin and Park 1997), possibly resulting in a lower mechanical cut-off frequency. Wrist, elbow, and muscle–tendon systems behave like a low-pass filter and can attenuate the transmission of vibration to the spindles, resulting in a decrease in driving Ia-afferents.

CV estimation

The adopted phase method provides an estimation of the CV_{VF} at a single frequency, without resolution limitation (Hunter et al. 1987). Moreover, by use of the phase method, band-pass filtering to select the peaks corresponding to the vibration frequency is no longer necessary, therefore avoiding possible phase distortion introduced by filtering (Ritzmann et al. 2010). However, the phase method presents some limitations for the assessment of conduction velocity in the presence of shape changes between EMG signals recorded in adjacent channels (Farina and Merletti 2004). This was the reason why the phase method was not adopted for the estimation of the CV for the entire EMG signals (CV_{EMG}), but only for the components at the vibration frequency, CV_{VF} .

In fact, during IVV_C , the signal to noise ratio (SNR) at the vibration frequency is much larger as compared to other frequency components (Fig. 6c), especially at higher vibration frequencies. Moreover, to estimate CV_{VF} , instead of directly calculating the phase difference between the two bipolar signals (Farina and Merletti 2004), we estimated the phase of the impulse response function between the auto-correlation and cross-correlation (Eq. 3), which has been reported to exhibit a sharper peak at the true delay between the two bipolar signals (Hunter et al. 1987). Finally, by employing the coherence function as a measure of the linear relation between the two signals, only the estimated CV_{VF} with corresponding coherence higher than 0.8 were taken into consideration, further supporting the accuracy of our results.

The CV estimation indicates that the EMG vibration components (without harmonics) propagate along the muscle fiber with a velocity of approximately 6.5 m/s, which is significantly different from the measured propagation of the motion artifacts, as indicated by the acceleration signals. Furthermore, CV_{VF} is comparable to CV_{EMG} with and without vibration and all of them are in the range reported in the literature for EMG CV (Davies and Parker 1987; Li and Sakamoto 1996). It is therefore reasonable to speculate that the EMG vibration components consist mainly of vibration-induced muscle activity. Two mechanisms may explain the small difference between CV_{VF} and CV_{EMG} .

First, the higher CV_{VF} may result from motoneuron recruitment elicited by VE, involving primary spindle endings. As mentioned above, the increase in TVR during vibration was ascribed to the recruitment of motor units with increasing threshold, resulting from the increase in motoneuron depolarization with the firing frequency of Ia-afferents (Martin and Park 1997). In addition, motor units with high threshold have been suggested to have higher CV (Houtman et al. 2003).

Aside from the recruitment process of motor units, an alternative explanation can be proposed. As indicated in our results, even if the contribution of motion artifacts is small, they may still account for about 7.6 % of the A_{VF} derived from IVV_C . Our results also revealed a higher (about 10–20 times) CV for motion artifacts, as indicated by the acceleration signals. Therefore, combining both EMG signals and motion artifacts, the components at the vibration frequency may show a higher propagation velocity than EMG signals alone.

Comparison with literature

In contrast to our results, Fratini et al. (2009) suggested the sharp EMG spectral peaks observed during WBV to result from motion artifacts. This discrepancy can most likely be ascribed to methodological constraints in (Fratini et al. 2009). Fratini and co-authors studied the relevance of motion artifacts during WBV by placing electrodes on the patella, assuming that any signal recorded with those electrodes related to motion artifacts. However, the electrodes placed on the patella may also register cross talk from surrounding muscles. In fact, a considerable amount of cross talk has been recorded from electrodes placed above osseous structures (Ritzmann et al. 2010).

It should be noted that the experimental setup in the present study is different compared to standard WBV devices. Our device provides force control, and vibration consists in the sinusoidal force modulation applied to the muscle rather than a varying displacement. However, a displacement can still be appreciated as a result of the force modulation and may vary depending on muscle tension and vibration frequency. For instance, for a 40-N force modulation at 45 Hz we measured a displacement of 1 mm.

Ritzmann et al. (2010) was the first gathering evidence that the sharp EMG spectral peaks observed during WBV are due to stretch reflex. To this end, dummy electrodes were used and EMG signals recorded during WBV were compared with those evoked by an ankle ergometer. However, dummy electrodes are not adequate to record motion artifacts, since they do not take the skin into consideration, which has been reported as the most important source for motion artifacts (Kahn 1965; Tam and Webster 1977; Ödman and Öberg 1982; Ödman 1981; de Talhouet and Webster 1996). In our study, IVV_R measurements tested all the possible sources of motion artifacts and, indeed, revealed significantly higher artifacts as compared to IVT measurements. Furthermore, the match of the spectrum and the latencies between the EMG signals recorded during WBV and those evoked by the ankle ergometer in (Ritzmann et al. 2010) may not necessarily lead to their conclusion. The signals evoked by the ankle ergometer could be either motion artifacts or stretch reflex, the same as the

EMG signals recorded during WBV, whose nature was under investigation in that study (Ritzmann et al. 2010). In contrast, our CV estimation indicated that the EMG vibration components propagate along the muscle fiber with a velocity comparable with normal EMG signals, providing strong evidence for the minor contribution of motion artifacts to the EMG during VE.

Conclusion

In the present study, the EMG signals recorded during VE with different protocols were investigated by spectrum analysis and CV estimation. The results of our experiments support the hypothesis that, during VE, the observed sharp spectral peaks at the vibration frequency in the EMG recordings are more related to individual neuromuscular response to the vibration frequency than to motion artifacts. We may therefore conclude that these responses are induced by stretch reflex. These vibration frequency components may therefore be included in the analysis of EMG signals during VE to obtain more reliable EMG signals to study the neuromuscular response to a vibrating stimulation.

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Conflict of interest The authors declare that they have no conflict of interest.

References

- Abercromby AF, Amonette WE, Layne CS, Mcfarlin BK, Hinman MR, Paloski WH (2007) Variation in neuromuscular responses during acute whole-body vibration exercise. *Med Sci Sports Exerc* 39:1642–1650
- Bendat JS, Piersol AG (1980) Engineering applications of correlation and spectral analysis. Wiley, New York
- Blottner D, Salanova M, Püttmann B, Schiffel G, Felsenberg D, Buehring B, Rittweger J (2006) Human skeletal muscle structure and function preserved by vibration muscle exercise following 55 days of bed rest. *Eur J Appl Physiol* 97:261–271
- Bongiovanni LG, Hagbarth KE, Stjernberg L (1990) Prolonged muscle vibration reducing motor output in maximal voluntary contractions in man. *J Physiol* 423(1):15–26
- Bosco C, Cardinale M, Tarpela O (1999) Influence of vibration on mechanical power and electromyogram activity in human arm flexor muscles. *Eur J Appl Physiol Occup Physiol* 79(4):306–311
- Burke D, Hagbarth KE, Löfstedt L, Wallin BG (1976) The responses of human muscle spindle endings to vibration during isometric contraction. *J Physiol* 261(3):695–711
- Cardinale M, Bosco C (2003) The use of vibration as an exercise intervention. *Exerc Sport Sci Rev* 31(1):3–7
- Cardinale M, Lim J (2003) The acute effects of two different whole body vibration frequencies on vertical jump performance. *Med Sport* 56(4):287–292

- Cochrane D, Stannard S (2005) Acute whole body vibration training increases vertical jump and flexibility performance in elite female field hockey players. *Br J Sports Med* 39:860–865
- Davies SW, Parker P (1987) Estimation of myoelectric conduction velocity distribution. *IEEE Trans Biomed Eng* 34(5):365–374
- Delecluse C, Roelants M, Verschueren S (2003) Strength increase after whole-body vibration compared with resistance training. *Med Sci Sports Exerc* 35(6):1033–1041
- de Ruiter CJ, van der Linden RM, van der Zijden MJ, Hollander AP, de Haan A (2003a) Short-term effects of whole-body vibration on maximal voluntary isometric knee extensor force and rate of force rise. *Eur J Appl Physiol* 88:472–475
- de Ruiter CJ, van Raak SM, Schilperoort JV, Hollander AP, de Haan A (2003b) The effects of 11 weeks whole body vibration training on jump height, contractile properties and activation of human knee extensors. *Eur J Appl Physiol* 90:595–600
- de Talhouet H, Webster JG (1996) The origin of skin-stretch-caused motion artifacts under electrodes. *Physiol Meas* 17(2):81–93
- Eklund G, Hagbarth KE (1966) Normal variability of tonic vibration reflexes in man. *Exp Neurol* 16(1):80–92
- Farina D, Merletti R (2000) Comparison of algorithms for estimation of emg variables during voluntary isometric contractions. *J Electromyogr Kinesiol* 10:337–349
- Farina D, Merletti R (2004) Methods for estimating muscle fibre conduction velocity from surface electromyographic signals. *Med Biol Eng Comput* 42(4):432–445
- Farina D, Negro F (2007) Estimation of muscle fiber conduction velocity with a spectral multidip approach. *IEEE Trans Biomed Eng* 54(9):1583–1589
- Fratini A, Cesarelli M, Bifulco P, Romano M (2009) Relevance of motion artifact in electromyography recordings during vibration treatment. *J Electromyogr Kinesiol* 19:710–718
- Fromm C, Noth J (1976) Reflex responses of gamma motoneurons to vibration of the muscle they innervate. *J Physiol* 256(1):117–136
- Hagbarth KE, Eklund G (1966) Tonic vibration reflexes (tvr) in spasticity. *Brain Res* 2(2):201–203
- Hagbarth KE, Burke D, Wallin BG, Löfstedt L (1976) Single unit spindle responses to muscle vibration in man. *Prog Brain Res* 44:281–289
- Hori Y, Hiraga K, Watanabe S (1989) The effects of thiamylol sodium on the tonic vibration reflex in man. *Brain Res* 497:291–295
- Houtman CJ, Stegeman DF, van Dijk JP, Zwarts MJ (2003) Changes in muscle fiber conduction velocity indicate recruitment of distinct motor unit populations. *J Appl Physiol* 95:1045–1054
- Huhta JC, Webster JG (1973) 60-Hz interference in electrocardiography. *IEEE Trans Biomed Eng* 20(2):91–101
- Hunter IW, Kearney RE, Jones LA (1987) Estimation of the conduction velocity of muscle action potentials using phase and impulse response function techniques. *Med Biol Eng Comput* 25(2):121–126
- Issurin VB, Tenenbaum G (1999) Acute and residual effects of vibratory stimulation on explosive strength in elite and amateur athletes. *J Sports Sci* 17(3):177–182
- Issurin VB, Lieberman DG, Tenenbaum G (1994) Effect of vibratory stimulation training on maximal force and flexibility. *J Sports Sci* 12(6):561–566
- Kahn A (1965) Motion artifacts and streaming potentials in relation to biological electrodes. In: *Proceedings of 6th Int Conf Med Electr Biol Eng*, pp 562–563
- Kinser AM, Ramsey MW, O'bryant HS, Ayres CA, Sands WA, Stone MH (2008) Vibration and stretching effects on flexibility and explosive strength in young gymnasts. *Med Sci Sports Exerc* 40(1):133–140
- Kvornig T, Bagger M, Caserotti P, Madsen K (2006) Effects of vibration and resistance training on neuromuscular and hormonal measures. *Eur J Appl Physiol* 96:615–625
- Li W, Sakamoto K (1996) The influence of location of electrode on muscle fiber conduction velocity and emg power spectrum during voluntary isometric contraction measured with surface array electrodes. *Appl Hum Sci* 15(1):25–32
- Martin BJ, Park HS (1997) Analysis of the tonic vibration reflex: influence of vibration variables on motor unit synchronization and fatigue. *Eur J Appl Physiol Occup Physiol* 75(6):504–511
- Matthews PBC (1966) The reflex excitation of the soleus muscle of the decerebrate cat caused by vibration applied to its tendon. *J Physiol* 184:450–472
- Merletti R, Knaflitz M, de Luca CJ (1990) Myoelectric manifestations of fatigue in voluntary and electrically elicited contractions. *J Appl Physiol* 69(5):1810–1820
- Mischi M, Cardinale M (2009) The effects of a 28-Hz vibration on arm muscle activity during isometric exercise. *Med Sci Sports Exerc* 41(3):645–653
- Mischi M, Rabotti C, Cardinale M (2010) Electromyographic assessment of muscle fatigue during isometric vibration training at varying frequencies. In: *Proceedings of 32th Annu Int Conf IEEE Eng Med Biol Soc*, pp 2338–2341
- Moritani T, Muro M, Nagata A (1986) Intramuscular and surface electromyogram changes during muscle fatigue. *J Appl Physiol* 60(4):1179–1185
- Naeije M, Zorn H (1982) Relation between emg power spectrum shifts and muscle fiber action potential conduction velocity changes during local muscular fatigue in man. *Eur J Appl Physiol* 50:23–33
- Ödman S (1981) Potential and impedance variations following skin deformation. *Med Biol Eng Comput* 19:271–278
- Ödman S, Öberg PA (1982) Movement-induced potentials in surface electrodes. *Med Biol Eng Comput* 20:159–166
- Park HS, Martin BJ (1993) Contribution of the tonic vibration reflex to muscle stress and muscle fatigue. *Scand J Work Environ Health* 19:35–42
- Poston B, Holcomb W, Magg MA, Linn LL (2007) The acute effects of mechanical vibration on power output in the bench press. *J Strength Cond Res* 21(1):199–203
- Rabotti C, Mischi M, Oei SG, Bergmans JW (2010) Noninvasive estimation of the electrohysterographic action-potential conduction velocity. *IEEE Trans Biomed Eng* 57(9):2178–2187
- Rees SS, Murphy AJ, Watsford ML (2008) Effects of whole-body vibration exercise on lower-extremity muscle strength and power in an older population: a randomized clinical trial. *Phys Ther* 88(4):462–470
- Ribot-Ciscar E, Rossi-Durand C, Roll JP (2000) Increase muscle spindle sensitivity to movement during reinforcement manoeuvres in relaxed human subjects. *J Physiol* 523(1):271–282
- Rittweger J, Beller G, Felsenberg D (2000) Acute physiological effects of exhaustive whole-body vibration exercise in man. *Clin Physiol* 20:134–142
- Rittweger J, Mutschelknauss M, Felsenberg D (2003) Acute changes in neuromuscular excitability after exhaustive whole body vibration exercise as compared to exhaustion by squatting exercise. *Clin Physiol Funct Imaging* 23(2):81–86
- Ritzmann R, Kramer A, Gruber M, Gollhofer A, Taube W (2010) Emg activity during whole body vibration: motion artifacts or stretch reflexes? *Eur J Appl Physiol* 110(1):143–151
- Romaiguère P, Vedel JP, Azulay JP, Pagni S (1991) Differential activation of motor units in the wrist extensor muscles during the tonic vibration reflex in man. *J Physiol (Lond)* 444:645–667
- Sands WA, Mcneal JR, Stone MH, Russell EM, Jemni M (2006) Flexibility enhancement with vibration: acute and long-term. *Med Sci Sports Exerc* 38(4):720–725
- Simakov AB, Webster JG (2010) Motion artifact from electrodes and cables. *Iran J Electr Comput Eng* 9(2):139–143

- Staudenmann D, Roeleveld K, Stegeman DF, van Dieën JH (2010) Methodological aspects of semg recordings for force estimation—a tutorial and review. *J Electromyogr Kinesiol* 20(3):375–387
- Tam HW, Webster JG (1977) Minimizing electrode motion artifact by skin abrasion. *IEEE Trans Biomed Eng* 24(2):134–139
- Wakeling JM, Nigg BM (2001) Modification of soft tissue vibrations in the leg by muscular activity. *J Appl Physiol* 90(2):412–420
- Welch PD (1967) The use of fast fourier transform for the estimation of power spectra: a method based on time averaging over short, modified periodograms. *IEEE Trans Audio Electroacoust* 15(2):70–73
- Xu L, Rabotti C, Mischi M (2012) Characterization of a novel instrument for vibration exercise. In: *Proceedings of 34th Annu Int Conf IEEE Eng Med Biol Soc*, pp 2760–2763
- Xu L, Rabotti C, Mischi M (2013) Novel vibration-exercise instrument with dedicated adaptive filtering for electromyographic investigation of neuromuscular activation. *IEEE Trans Neural Syst Rehabil Eng* 21(2):275–282