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Analysis of muscle fatigue induced by isometric vibration exercise at varying frequencies

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Abstract— **An increase in neuromuscular activity, measured by electromyography (EMG), is usually observed during vibration exercise. The underlying mechanisms are however unclear, limiting the possibilities to introduce and exploit vibration training in rehabilitation programs. In this study, a new training device is used to perform vibration exercise at varying frequency and force, therefore enabling the analysis of the relationship between vibration frequency and muscle fatigue. Fatigue is estimated by maximum voluntary contraction measurement, as well as by EMG mean-frequency and conduction-velocity analysis. Seven volunteers performed five isometric contractions of the biceps brachii with a load consisting of a baseline of 80% of their maximum voluntary contraction (MVC), with no vibration and with a superimposed 20, 30, 40, and 50 Hz vibrational force of 40 N. Myoelectric and mechanical fatigue were estimated by EMG analysis and by assessment of the MVC decay, respectively. A dedicated motion artifact canceler, making use of accelerometry, is proposed to enable accurate EMG analysis. Use of this canceler leads to better interpolation of myoelectric fatigue trends and to better correlation between mechanical and myoelectric fatigue. In general, our results suggest vibration at 30 Hz to be the most fatiguing exercise. These results contribute to the analysis of vibration exercise and motivate further research aiming at improved training protocols.**

I. INTRODUCTION

Resistance exercise is a widely accepted training option for improving neuromuscular performance in young and old individuals by increase or maintenance of muscle size and neural drive to the muscles [1]. In fact, although extensive literature covers and proposes various training alternatives for increasing muscle strength and power, resistance exercise is usually preferred over alternative options due to the high neuromuscular demand that imposes [1]. Recent research is even suggesting the inclusion of resistance training in rehabilitation programs for large populations, such as cancer and cardiovascular patients [2], [3].

Vibration training is a relatively new and promising exercise modality that is reported to provide several important advantages over conventional strength training [4]. Whole body vibration platforms are the most widely adopted devices for vibration training and have been suggested to be an effective modality to exercise the lower limbs [5], [6]; alternative devices have also been proposed for the upper limbs [7]. Beneficial effects of vibration exercise are reported in relation to muscle activation, strength, power, joint stability, perfusion, and metabolism [6], [7], [5], [8], [9], [10]. These results seem partly to be ascribed to a neuromuscular phenomenon referred to as tonic vibration reflex (TVR) [11], [12].

Through TVR, muscles try to achieve vibration dampening and joint stiffening by preferential recruitment of faster motor units [11], [13]. In addition, TVR is reported to be modulated by alterations in spindle sensitivity through gamma feedback [11], [14]. In general, although several studies have been carried out, a thorough understanding of the mechanisms and the physiological responses involved in vibration exercise is still lacking, hampering the optimal use of this option in training programs.

In an effort to contribute to a better understanding of the physiological responses to vibration exercise, this paper investigates the fatiguing effect of vibration at different frequencies. To this end, a dedicated prototype is adopted that can apply to a muscle the superimposition of a baseline and a vibrating load. Seven healthy volunteers were asked to perform 20-s sustained isometric contractions of the biceps brachii. Frequencies ranging from 0 (no vibration) to 50 Hz were superimposed to a baseline load corresponding to 80% of the maximum voluntary contraction (MVC).

Mechanical fatigue, defined as the decay in force-production capability, was assessed by MVC tests prior to and after each task. Myoelectric fatigue was assessed by analysis of electromyography (EMG) measured by a 64-channel high-density electrode grid [15]. In particular, the EMG mean frequency (MF) and the conduction velocity (CV) of muscle-fiber action potentials were estimated. Several methods are reported in the literature for CV estimation [16]. Here a modified version of the maximum likelihood (ML) method is adopted [17]. An estimate of the fatiguing effect of exercise can be derived as the angular coefficient of the linear fit of the MF or CV evolutions over time.

EMG measurements performed during vibration exercise are typically affected by severe motion artifacts [7], [18]. These artifacts are most probably determined by variations in the ionic concentration distribution in the dielectric gel induced by vibrations [19]. With the aim of improving the reliability of the estimated myoelectric fatigue, a dedicated motion-artifact removal method is implemented that makes use of accelerometers that are positioned on the electrodes. The measured accelerometric signal is used as reference signal by an adaptive filter (canceler), which assumes the relation between acceleration and EMG artifacts to be modeled by a finite impulse response filter (linear, causal system). Possibly due to the retention of important information at the vibration harmonics, the use of the proposed adaptive

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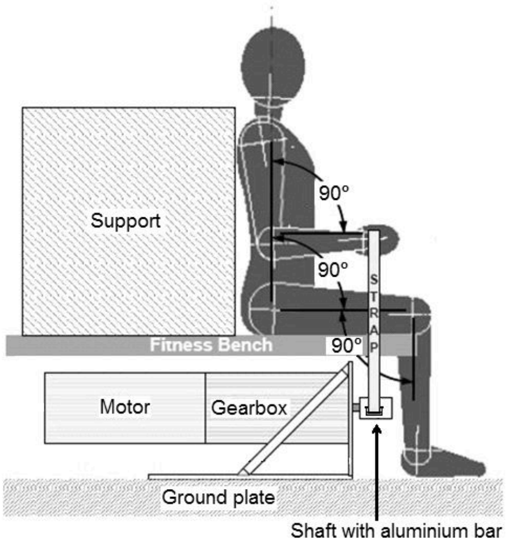

Fig. 1. Scheme of the measurement setup.

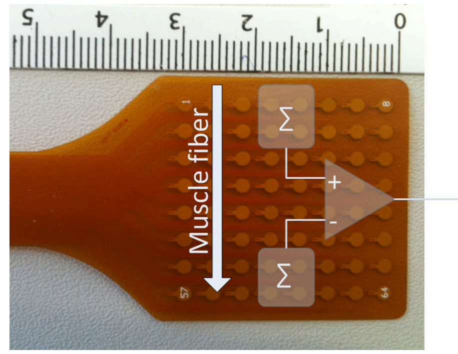

Fig. 2. High-density electrode grid. The muscle fiber orientation is shown together with the two sets of electrodes adopted for MF estimation.

canceler provides better correlation between myoelectric and mechanical estimates of fatigue with respect to removal of all vibration-related harmonics [7], [18].

II. METHODOLOGY

A. Measurement setup

A dedicated electromechanical actuator, whose scheme is shown in Fig. 1, was realized in order to apply an oscillating (vibrating) mechanical load to the biceps brachii. The core of the actuator is an electrical motor (MSK060C Indradyn®, Bosch-Rexroth, the Netherlands) generating a force that can be modulated to produce the required vibrating load at varying frequencies up to over 50 Hz. A planetary gearbox with 10:1 ratio is used to increase the torque at the output shaft. An aluminium bar is connected to this shaft in order to pull a strap that is connected via a handle with the hand of the training subject. The input voltage to the motor drive is generated by a wave generator (PCI 5402, National Instruments, Austin, TX) connected to a PC and controlled by dedicated software implemented in LabView® (National Instruments, Austin, TX).

The actuator was calibrated by means of a LCAE-35kg load cell (Omega Engineering Inc, Stamford, CT) embedded in the aluminium bar. A dynamic calibration was performed in order to determine the transfer function of the complete electromechanical system. This permitted implementing a compensation of the estimated transfer function in order to apply to the muscle a well determined force, the same for all vibration frequencies. This step is essential to assess the physiological effects of different vibration frequencies.

As shown in Fig. 1, the realized actuator enables performing isometric contractions of the biceps brachii from sitting position. The back is supported in order to avoid shoulder movements to influence the contraction of the biceps brachii.

The biceps surface EMG was measured during each experimental trial by a high-density grid (8x8) of contact Ag-AgCl electrodes positioned between the tendon at the elbow side and the muscle belly. This position, distant from the fiber innervation zone, facilitated the CV estimation by detecting a single propagation direction of the action potentials. The adopted high-density grid is shown in Fig. 2. The electrode diameter and the interelectrode distance is 1 mm and 4 mm, respectively. The EMG signals were acquired by a 64-channel Refa® amplifier (TMS International, Enschede, the Netherlands) implementing virtual active grounding and cable shielding for reduction of the electromagnetic interference. The adopted sampling frequency was 2 kHz.

By the adopted EMG recording system, the potential at each electrode is referred to the average potential of all the electrodes. For the CV estimation, all the EMG signals were bipolarized along the fiber direction, leading to 8 parallel columns of 7 bipolar signals. For the MF estimation, one single bipolar signal was extracted by taking the difference of the average EMG in two subsets of four electrodes, forming two squares at the extremities of the electrode grid along the two central columns (parallel to the muscle fibers). The electrode configurations are shown in Fig. 2.

B. Measurement protocol

Four males and three females (mean age $=$ 24.4 ± 2.2 years) took part in the experiment. Prior to each treatment, the MVC for each subject was measured by the embedded load cell. Subjects were asked to pull the handle with 3-s maximal isometric contraction three times, each time followed by 1-min rest. For each 3-s contraction, the maximum force was estimated and the average of the three estimates considered as the MVC.

The treatment protocols consisted then of maintaining a constant tension equal to 80% of the MVC. The experimental conditions were characterized by five sustained (20 s) isometric contractions of the right biceps brachii with no vibration (control), and vibration at 20, 30, 40, and 50 Hz. Each condition was followed by a 15-min resting period. A randomized cross-over design was used. Before and after each trial, the MVC was measured following the same protocol as before treatment. The EMG was measured during each trial. The vibration amplitude was fixed for all frequencies and equal to 40 N.

As shown in Fig. 1, the subjects were instructed to keep the arm attached to the body and the elbow with a fixed a angle of 90° while performing isometric contractions of the biceps brachii, both for the MVC estimation and for the sustained contractions. To this end, the subjects were supported by visual feedback on the elbow flection, implemented by a position encoder embedded in the electromagnetic actuator.

C. EMG feature extraction

First of all, the measured EMG signals are filtered for motion artifact removal. Rather than suppressing or fully removing all the vibration harmonics, as proposed in [18] and [7], respectively, an adaptive filter is designed that estimates adaptively the impulse response between motion signals, measured by 3D accelerometry, and the resulting artifacts in the EMG recording. The impulse response is modeled as a finite impulse response (FIR) filter of 60 taps. A standard normalized least mean squares (NLMS) adaptive scheme is adopted [20]. A cascade of three FIR filters are estimated and applied, one for each direction of motion.

The progression of MF and CV is estimated by time-frequency analysis of the recorded bipolar signals. A sliding window of 2 s with 1.5-s overlap is adopted to calculate the signal Short Time Fourier Transform (STFT). For the analysis in the frequency domain, only frequencies between 20 and 450 Hz are considered [21].

The MF is estimated as the first (normalized) statistical moment of the STFT amplitude spectrum. The bipolar channel extracted from the two subsets of four electrodes defined in Fig. 2 is used for the MF estimation.

The CV is estimated by the modification of the ML method reported in [17]. According to this method, the cost function $\widehat{E}(\tau)$ to be minimized in order to estimate the time delay τ between the EMG recorded at different rows is given as

$$\widehat{E}(\tau) = \frac{2}{N}\sum_{r=1}^{N_r}\sum_{c=1}^{N_c}\sum_{k=1}^{N/2}\Big[a_{rc}\Big(X_{rc}(k) + -\widehat{S}(k)e^{-j2\pi k(r-1)\tau}\Big)\Big]^2, \tag{1}$$

where N is the time-window duration, N_c is the number of columns (7), N_r is the number of rows (8), $S(k)$ is the shape function, and $X_{rc}(k)$ and a_{rc} are the recorded EMG and the adopted weight for channel (c,r), respectively. The shape function $S(k)$ is estimated as the average of all the channels $X_{rc}(k)$ after alignment. The choice for the frequency domain k enables estimating the time delay τ without the resolution limits imposed by the sampling frequency. The weights are chosen to be inversely proportional to the standard deviation of the channel noise and they are defined as

$$a_{rc} = \frac{A}{\frac{2}{N}\sqrt{\sum_{k=1}^{N/2}\Big(X_{rc}(k)\cdot X_{rc}^{*}(k) - \widehat{S}(k)\cdot\widehat{S}^{*}(k)\Big)}}. \tag{2}$$

The Nelder-Mead Simplex search method is used to minimize (1) [22]. The search method is initialized with

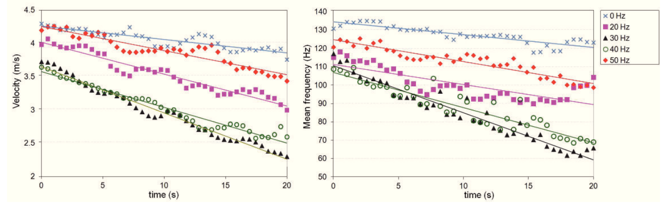

Fig. 3. Example of CV (left) and MF (right) estimates for varying vibration frequencies from 0 (control with no vibration) to 50 Hz.

$\tau = 0.93$ ms. With the given inter-electrode distance of 4 mm, this delay corresponds to CV $= 4.3$ m/s, i.e., the average CV value reported in the literature for the biceps brachii [23]. All the analysis was implemented in Matlab® (MathWorks, Natick, MA).

D. Statistical analysis

The parameters adopted for statistical analysis are myoelectric fatigue and mechanical fatigue. Myoelectric fatigue is estimated as the angle of the regression line to the time evolution of CV or MF, estimated during each task. Mechanical fatigue is estimated as the relative drop in MVC measured before and after each task. The correlation coefficient between mechanical and myoelectric fatigue is investigated for both CV and MF estimates, together with the effect of motion artifact canceling. The effect of motion artifact canceling on the linearity of MF and CV time decay during exercise is also investigated.

For each fatigue estimator, the effect of different vibration frequencies is evaluated with a one-way ANOVA test, being each group represented by fatigue estimates at a given vibration frequency, as well as by paired t-Student tests between control (no vibration) and each vibration frequency.

III. RESULTS

Fig. 3 shows the MF and CV time evolution for a volunteer. Both MF and CV decay over time with a linear fashion. Myoelectric fatigue was therefore estimated as the (negative) angular coefficient of the regression lines. The use of NLMS filtering yielded larger correlation coefficients of these regression lines as compared to full removal all the harmonic frequencies. For MF estimates, the average correlation coefficients over all the measurements were -0.86 ± 0.11 and -0.84 ± 0.13 by NLMS filtering and harmonic removal, respectively. For CV estimates, the average correlation coefficients were -0.82 ± 0.14 and -0.75 ± 0.22 by NLMS filtering and harmonic removal, respectively. All the CV estimates were between 2 and 8 m/s, in line with the values reported in the literature for the biceps brachii [23].

Table I reports the average estimates of fatigue for all the subjects. The reported values were obtained by NLMS filtering. In fact, the use of NLMS filtering produced estimates of myoelectric fatigue that were more correlated with mechanical fatigue estimated by MVC test. An approximately twofold increase was observed in the correlation coefficient between the estimated myoelectric and mechanical fatigue. After NLMS filtering, the resulting correlation coefficients

TABLE I

FATIGUE INDICATORS.

Freq.	Force decay	MF slope	CV slope
0 Hz	-5.32 ± 7.62 %	-0.84 ± 0.57 Hz/s	-63 ± 28 mm/s^2
20 Hz	-9.08 ± 5.83 %	-1.41 ± 0.48 Hz/s	-37 ± 16 mm/s^2
30 Hz	-11.82 ± 6.71 %	-1.11 ± 0.44 Hz/s	-101 ± 81 mm/s^2
40 Hz	-8.71 ± 7.01 %	-1.27 ± 0.64 Hz/s	-54 ± 38 mm/s^2
50 Hz	-6.62 ± 6.73 %	-1.16 ± 0.80 Hz/s	-49 ± 9 mm/s^2

were 0.55 and 0.59 for CV and MF estimates of myoelectric fatigue, respectively.

The estimates in Table I indicate 30-Hz vibration to be the most fatiguing exercise according to CV assessment and MVC test. MF estimates do not show a clear trend. In general, one-way ANOVA on MVC, CV, and MF results suggests the separation between different groups to be not significant. Student t-Test between the control (0-Hz) and 30-Hz groups returns a p-value equal to 0.1 for CV and MVC estimates.

IV. DISCUSSION AND CONCLUSIONS

Although the fatiguing effects of different vibration frequencies could not be clearly distinguished ($p > 0.05$), our results suggest vibration exercise, especially at 30 Hz, to produce a larger degree of fatigue as compared to control condition (0 Hz, no vibration). This is likely to be ascribed to the increased neuromuscular demands placed by vibration, especially when superimposed to high levels of muscle tension ($> 70\%$ MVC) [7]. Previous work using vibration applied to hand muscles has also suggested an effect of vibration frequency on motor unit synchronization causing fatigue [12].

It has been clearly established that the rate of change of spectral variables and CV during a sustained contraction is indicative of muscle fatigue [24]. The results of our study show that possibly different mechanisms of fatigue are evident with vibration, as the agreement between MVC decrease and myoelectric indicators of fatigues was moderate. This suggests that motor unit recruitment patterns may be different during vibration.

The results of our preliminary validation confirm the value of accelerometry-based NLMS filtering for canceling motion artifacts prior to the analysis of vibration training effects.

In the future, a larger population should also be employed with the aim of achieving more significant results. With a larger population, randomization of the task sequence would also become more effective. In general, more studies are needed to elucidate the relative mechanisms of fatigue in vibration exercise.

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