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Citation for published version (APA):
Xu, Y., Long, X., Luo, Z., Mischi, M., & Xu, L. (2021). Influence of spectral peaks on EMG parameter estimation for vibration-exercise analysis. *IEEE Sensors Journal*, *21*(13), 14141-14147. Article 9262944.
https://doi.org/10.1109/JSEN.2020.3038953

Document license:
TAVERNE

DOI:
10.1109/JSEN.2020.3038953

Document status and date:
Published: 01/07/2021

Document Version:
Publisher's PDF, also known as Version of Record (includes final page, issue and volume numbers)

Please check the document version of this publication:

• A submitted manuscript is the version of the article upon submission and before peer-review. There can be important differences between the submitted version and the official published version of record. People interested in the research are advised to contact the author for the final version of the publication, or visit the DOI to the publisher's website.
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IEEE SENSORS JOURNAL, VOL. 21, NO. 13, JULY 1, 2021

Influence of Spectral Peaks on EMG Parameter Estimation for Vibration-Exercise Analysis

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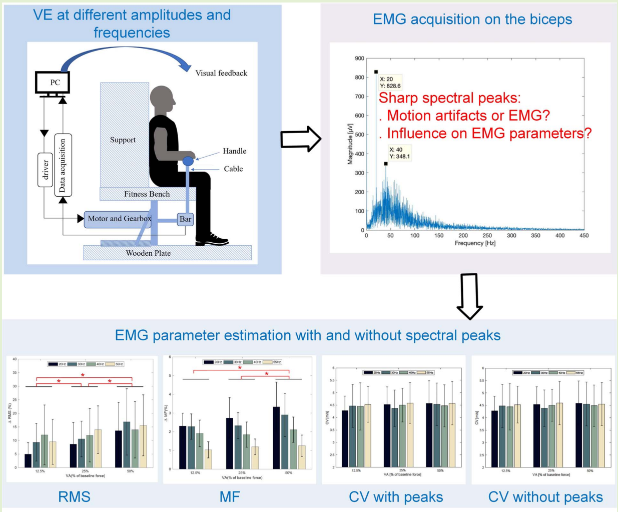

***Abstract*—Many studies have proposed vibration exercise (VE) as a novel training modality for neuromuscular conditioning and rehabilitation. Surface electromyography (sEMG) is widely used for effective measurement of muscle activity. Unfortunately, sharp spectral peaks (SSP) are usually present in the EMG signals recorded during VE. The explanation of these sharp peaks, as muscle activity or motion artifacts, is controversial, complicating EMG parameter extraction for the analysis of VE. The present study aims to quantify the impact of these SSP on the estimation of EMG parameters irrespective of their nature. High-density sEMG was therefore recorded from the biceps brachii muscle during VE with different vibration amplitudes (VA) and frequencies (VF). The power around (±0.5 Hz) VF and its first harmonic was calculated and normalized with the entire EMG power in order to obtain a relative power (P_R) of these peaks. In addition, before and after excluding the SSP, three EMG parameters, i.e., mean frequency (MF), root mean square (RMS), and conduction velocity (CV), were estimated and compared. Our results reveal an average P_R of 21.18±15.68 %. The relative difference in EMG RMS and MF are 12.2±3.8 % and 2.10±1.04 %, respectively. In addition, the impact of these peaks on the MF and RMS seems also to be affected by vibration conditions, such as VA and VF. However, the CV estimation seems not to be significantly influenced by these peaks, indicating these peaks to be primarily reflecting muscle activity and therefore should be included in VE EMG analysis.**

***Index Terms*—Conduction velocity, electromyography, mean frequency, motion artifacts, vibration exercise.**

I. INTRODUCTION

IT HAS been reported that physical training methods integrated with vibration can help attain greater improvements in power performance and muscle strength as compared with conventional resistance training alone [1]. In recent years, vibration exercise (VE) has been proposed with different forms for rehabilitation, recreational and athletic training purposes [2]–[4]. Many studies investigated the short-term [5]–[8] and long-term effects [9]–[11]] of VE. They suggested VE to be able to improve power performance [1], muscle strength [6], balance, and coordination [12], [13]. Besides, VE can increase bone density [14], [15] and muscular metabolism including oxygen uptake [16] and glucose uptake [17].

Tonic vibration reflex (TVR) has been widely adopted to explain the observed benefits of VE [18]. TVR is a specific reflex mechanism originated in vibration-induced deformation of the primary spindle ending, which initiates the reflex loop and causes activation of α-motor neurons [18]–[21]. To dampen the effect caused by vibration, target muscles recruit additional motor units through TVR, leading to an increase in muscle contraction and motor unit synchronization as evidenced by increased electromyography (EMG) [1], [22]–[24].

Surface EMG (sEMG) can noninvasively measure neuromuscular activity and therefore are extensively employed for VE analysis [2], [5]. Several EMG parameters, such as amplitude, conduction velocity (CV), and mean frequency (MF), are widely adopted for VE analysis [24]. In general, the EMG

Manuscript received October 26, 2020; revised November 7, 2020; accepted November 14, 2020. Date of publication November 18, 2020; date of current version June 30, 2021. This work was supported with a start-up funding from the ShanghaiTech University. This article was presented at the IEEE EMBC Conference. The associate editor coordinating the review of this article and approving it for publication was Prof. Rosario Morello. *(Corresponding authors: Lin Xu; Zhe Luo.)*

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Digital Object Identifier 10.1109/JSEN.2020.3038953

amplitude, i.e., root mean square (RMS) value, reflects the intensity of muscle activity and can be used to estimate force production [25]. The EMG MF and CV are frequently employed to analyze myoelectric manifestations of fatigue [24], [26], [27]. Besides, it is well known that high forces are associated with high average muscle fiber CV [28], [29], and changes in CV are correlated with changes in MF [30]–[32].

Unfortunately, the sEMG signals recorded during VE usually exhibits sharp spectral peaks (SSP), which are located at the vibration frequency (VF) and its harmonics. For a couple of decades, the nature of these SSP have been controversially interpreted in the literature. Some authors treated these SSP as vibration-induced motion artifacts and recommended to remove them before estimating EMG parameters [5], [33], [34]. On the contrary, other authors, such as *Martin et al.* and *Ritzmann et al.*, interpreted them as manifestations of synchronized motor unit activation within the vibration cycle trough TVR [35], [36]. More recent studies supported the findings of *Martin* and *Ritzmann* [37]. Based on these results, these SSP were suggested to be considered for VE analysis.

It is clear that the nature of the EMG spectral peaks is essential not only for the understanding of the underlying mechanisms of VE, but also for appropriate estimation of EMG parameters. Due to the controversial interpretation of these SSP, it is unclear whether these SSP should be included or excluded in the extraction of EMG parameters for VE analysis. Therefore, the present study is aiming at quantifying the influence of these SSP on EMG parameter estimation, irrespective of their nature being muscle activity or motion artifacts. Our hypothesis is that such quantitative investigation can provide a guidance for proper EMG analysis during VE and may, in turn, contribute to better understanding the nature of these SSP. For this purpose, isometric VE was performed on the biceps brichii with a dynamic force-modulation VE system and sEMG recorded [38]. The relative power (P_R) of these SSP to the total EMG power was first calculated. Furthermore, three EMG parameters, i.e., RMS, CV, and MF, were calculated with and without SSP and then compared as they are common EMG parameters that have been widely employed for VE analysis and, more in general, characterization of myoelectric activity [24].

II. METHOD

A. Subjects

Eighteen healthy right-handed subjects (age = 28±5 years, 11 males and 7 females) voluntarily participated in the present study. They had no history of neurological irregularities or injuries. Prior to participating in the experiment, each subject was clearly explained the experimental protocol and then signed the written informed consent. The measurement protocol was approved by the Ethical Committee in Máxima Medical Center (MMC, Veldhoven, the Netherlands).

B. Experimental Setup

1) Vibration System: Figure 1 shows the diagram of the adopted VE system that has been realized in our previous studies [38]. In brief, a vibrating force is generated by a motor and applied vertically to the subject's arm through a mechanical interface. The generated vibration force is composed of a baseline and a sinusoidal modulation, as shown in Fig. 2. The mechanical interface consists of an aluminum bar, a cable, and a handle (Fig. 1). A load-cell is integrated in the bar in order to measure the generated force. Besides, a rotary encoder is embedded in the motor, enabling real-time measurement of the angular position of the motor shaft. Therefore, the subject's wrist position can be visually displayed on the PC monitor in order to guide the subjects to perform isometric exercise. The whole system is well controlled and calibrated [38], permitting accurate force generation.

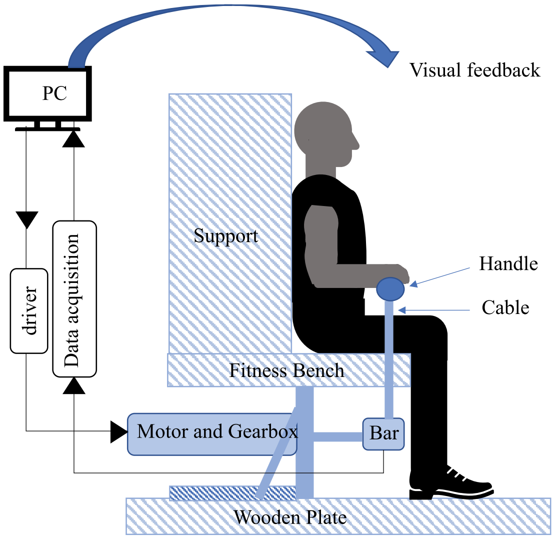


Fig. 1. Diagram of the adopted VE system.

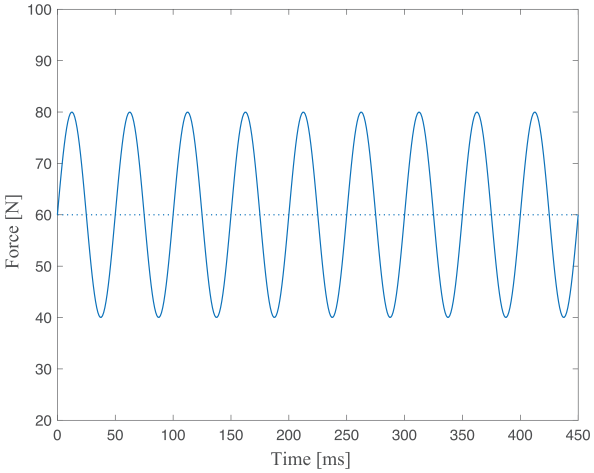


Fig. 2. Example of VE force with 60-N baseline and 20-Hz vibration.

2) Measurement Protocol: We first examined the subject's maximum voluntary contraction (MVC) following the same protocol described in our previous work [39]. Then the subjects performed twelve trials of 30-s isometric contractions under different vibration conditions, including three different vibration amplitudes (VAs), i.e., 12.5%, 25%, and 50% of the baseline force, and four different VFs, i.e., 20, 30, 40, and 55 Hz. The adopted baseline force was 30% of the MVC in order to avoid muscle fatigue. For details of the measurement protocol, please refer to our conference proceedings [39].

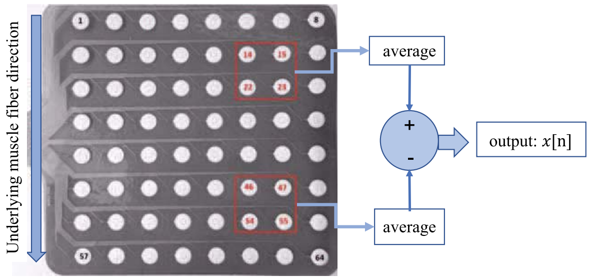


Fig. 3. High density (8 × 8) electrode grid and the two sub-sets for extracting a bipolar signal.

During each experimental trial, sEMG was measured by two 8×8 electrode grids. The diameter and inter-electrode distance of the grid were 2 mm and 8 mm, respectively (Fig. 3). The grids were placed on the biceps brachii of the right hand of the subject with the columns parallel to the direction of the underlying muscle fibers, as shown in Fig. 4. A reference electrode (1 cm, Ag/AgCl) was placed on the right clavicle of the subject. A Refa amplifier with 128 channels (TMS International, Enschede, The Netherlands) was employed to record the detected sEMG signals at 2048 Hz.

C. Signal Processing

1) Preprocessing: In order to avoid transient and fatiguing effects, only data from 3 to 10 seconds were selected for analysis. A channel-quality check was then performed on the extracted data segments. Any channel with constant values or consecutive (> 100) zeros was considered as failed recording and was excluded. If one subject had more than 10 failed channels in any of the 12 trials, the whole subject was excluded from the analysis.

For $\mathrm{P_R}$, MF, and RMS estimation, a bipolar signal was extracted by taking the difference between the average of two sub-sets of four channels, as shown in Fig. 3. For CV estimation, the propagation direction of each EMG recording was first visually checked in order to locate the innervation zone of the muscle. Only the channels located between the tendon and the innervation zone were selected for further analysis. Each selected channel was then normalized in order to have zero mean and unit variance. The normalized monopolar channels were bipolarized by taking the difference of two neighbouring rows in the same column. An example of the obtained bipolar signals from one column is shown in Fig. 5. All the derived bipolar signals were adopted for CV estimation.

In normal conditions, the frequency band of a sEMG signal is considered in the range between 20 to 450 Hz [40]. However, the lowest VF adopted in the present study was 20 Hz. In order to include the components around 20 Hz when extracting EMG parameters, the lower band limit was extended from 20 to 15 Hz in the present study. Since all the four desired parameters can be estimated in the frequency domain, the bandpass filter (15-450 Hz) can be directly implemented in the frequency domain when calculating each EMG parameter. In addition, the power-line interference was also excluded in the frequency domain by setting the amplitude spectrum around (±0.5 Hz) the power-line and its harmonics to zero.

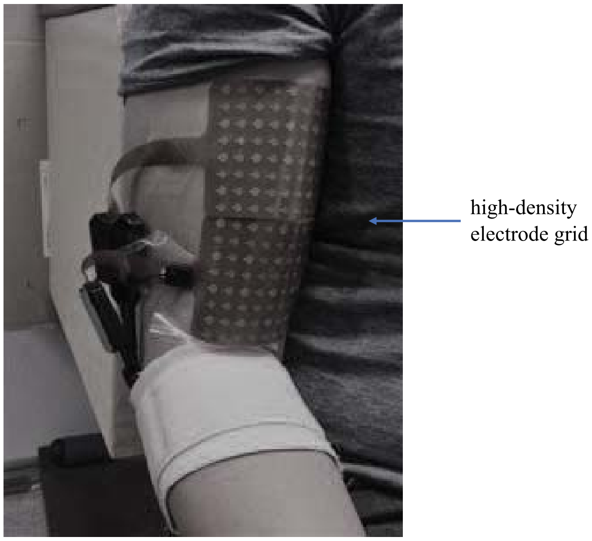


Fig. 4. Position of the electrode grids.

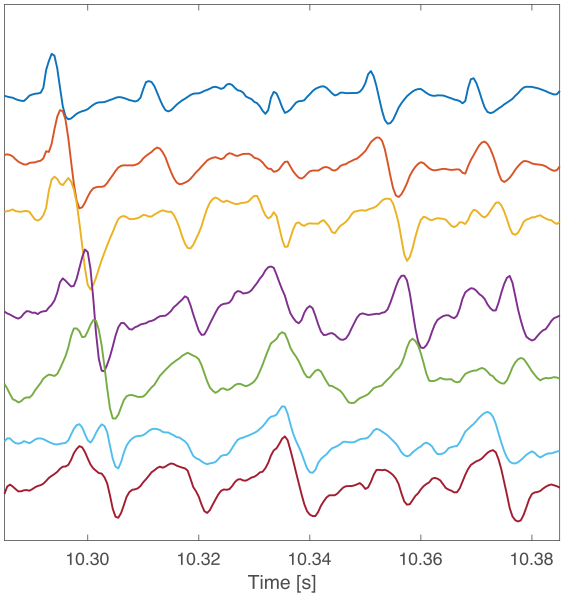


Fig. 5. The derived bipolar EMG in one column.

2) P_R Estimation: The relative power $\mathrm{P_R}$ of the SSP at the VF (f_v) and its first harmonic (FH), i.e., f_h, was calculated as

$$\mathrm{P_R} = \frac{\sum_{n=(f_v-0.5)\times T+1}^{(f_v+0.5)\times T+1} |X[n]|^2 + \sum_{n=(f_h-0.5)\times T+1}^{(f_h+0.5)\times T+1} |X[n]|^2}{\sum_{n=15\times T+1}^{450\times T+1} |X[n]|^2} \times 100\%, \quad (1)$$

where $X[n]$ is the Fourier Transform of the big bipolar signal $x[n]$ (Fig. 3), $T = 8$ is the data duration in seconds, and f_v and f_h are given in Hz.

3) RMS Estimation: The RMS of the EMG signal including the SSP, $\mathrm{RMS_i}$, can be easily calculated in the frequency domain by using the Perseval's equality as

$$\mathrm{RMS_i} = \frac{1}{N}\sqrt{\sum_{n=15\times T+1}^{450\times T+1} |X[n]|^2}, \quad (2)$$

where N is the number of sampling points.

The RMS of the EMG signal excluding the SSP, $\mathrm{RMS_e}$, was calculated in the same way while the amplitude spectrum around (±0.5 Hz) f_v and f_h was set to zero. Then the relative difference in RMS (ΔRMS) was calculated as

$$\Delta\mathrm{RMS} = \frac{\mathrm{RMS_i} - \mathrm{RMS_e}}{\mathrm{RMS_i}} \times 100\%. \tag{3}$$

4) MF Estimation: The MF was estimated as the first statistical moment of the amplitude spectrum of the Fourier Transform,

$$\mathrm{MF} = \frac{\sum_{n=15\times T+1}^{450\times T+1} f[n]\cdot|X[n]|}{\sum_{n=15\times T+1}^{450\times T+1} |X[n]|}, \tag{4}$$

where $f[n]$ is the frequency [Hz] at the sample n.

Similar to RMS, while excluding the SSP, $\mathrm{MF_e}$ was estimated using equation 4 but setting the amplitude spectrum around (±0.5 Hz) f_v and f_h to zero. The relative MF difference (ΔMF) was calculated in a similar way as RMS.

5) CV Estimation: Many algorithms have been proposed for CV estimation in the literature, such as estimation of spectral dips [41], cross-correlation [42], maximum likelihood (ML) [24], [43], [44], and phase-lock-loop [45]. Among them, the ML method described in [24] was adopted in the present study given its good performance on robustness and reliability.

The EMG CV can be computed as

$$\mathrm{CV} = \frac{D \times f_s}{\theta}, \tag{5}$$

where $f_s = 2048$ Hz is the sampling frequency, $D = 8$ mm the distance between adjacent electrodes, and θ the delay of the adjacent EMG waveforms in samples that needs to be estimated.

According to [24], the signal from the ith row and jth column, x_{ij}, can be expressed as

$$x_{ij}(n) = s_j[n - (i-1)\theta] + w_{ij}(n), n = 1, \ldots, N \qquad i = 1, \ldots, R;\ j = 1, \ldots, C, \tag{6}$$

where N is the number of samples, R the number of rows, C the number of columns, w_{ij} a white, Gaussian noise, and s_j the noise-free signal at column j. The ML estimator of θ is obtained in the frequency domain, as given by

$$\hat{\theta} = \arg\max_{\theta}[\sum_{j=1}^{C}\sum_{i=1}^{R}\sum_{k=1}^{R}\frac{1}{2\pi} \times \int_{-\pi}^{\pi} X_{ij}\left(e^{j\omega}\right) X_{kj}^{*}\left(e^{j\omega}\right) e^{j(i-k)\theta\omega} d\omega], \tag{7}$$

where $*$ indicates the complex conjugate, and $X_{ij}(e^{j\omega})$ and $X_{kj}(e^{j\omega})$ denote the Fourier transform of $x_{ij}(n)$ and $x_{kj}(n)$, respectively. Calculating in the frequency domain, θ can be estimated with increased time resolution.

Similar to RMS and MF, CV was calculated both with and without SSP, denoted as $\mathrm{CV_i}$ and $\mathrm{CV_e}$, respectively. The relative difference in CV, ΔCV, was also derived.

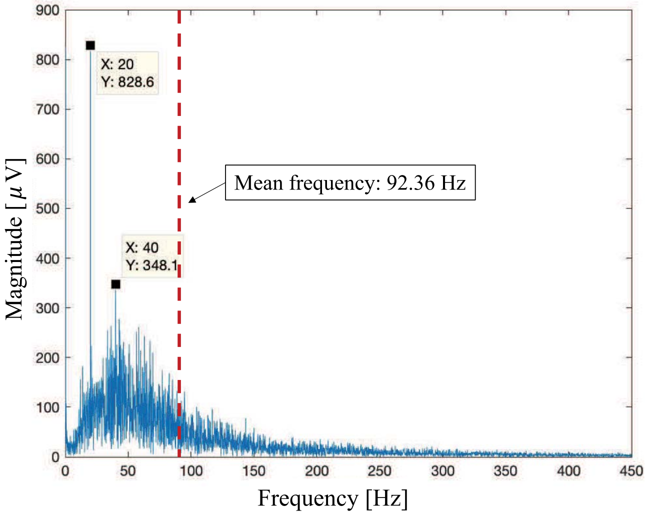


Fig. 6. Example of spectral peaks in the EMG amplitude spectrum at a VF of 20 Hz.

D. Statistical Analysis

The $\mathrm{P_R}$ of the SSP and the three EMG parameters, MF, RMS, and CV, extracted with and without SSP were used for the statistical analysis. Our data are normally distributed as indicated by one-sample Kolmegorov-Smirnov test. For the three EMG parameters, RMS, MF, and CV, Paired Student's t-test was first employed to assess the difference between the values estimated with and without SSP. For each EMG parameter, if the difference between the values estimated with and without SSP was significant, the impact of VA and VF on the relative difference was further analyzed. Since we used a single-group repeated-measure study under different VAs and VFs, our data were dependent. As a consquence, a two-way repeated measure of ANOVA was adopted to test the effects of VF and VA on the relative difference of each EMG parameter. Furthermore, a post-hoc test with Tukey's procedure was employed to assess the differences between two different groups. The same ANOVA and post-hoc approach was applied to the $\mathrm{P_R}$ values of the SSP. The significance level was set to 0.05.

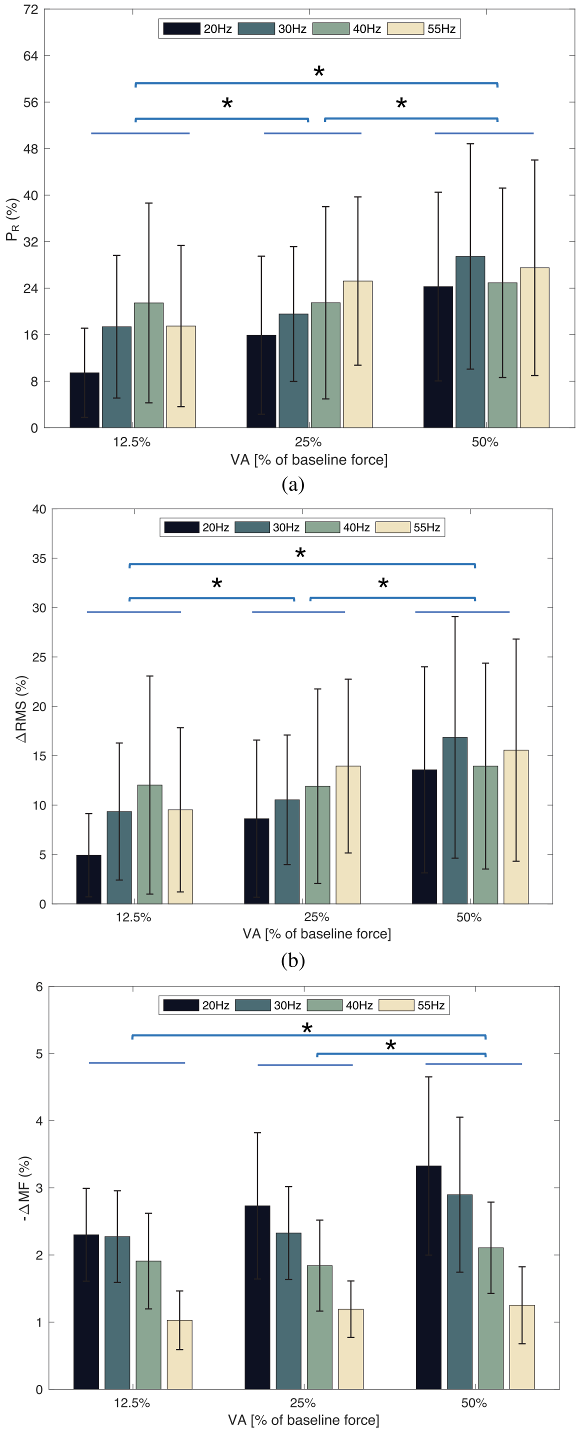
III. RESULTS

A. $\mathrm{P_R}$ Estimation

The EMG spectrum with a VF of 20 Hz is shown in Fig. 6 as an example, in which SSP can be clearly observed at 20 Hz and 40 Hz. The average $\mathrm{P_R}$ of these peaks at different VAs and VFs calculated over all subjects are shown in Fig. 7 (a). On average, the power of the SSP at the VF and FH account for 21.18±15.68 % of the entire EMG power. In addition, the $\mathrm{P_R}$ values increases with increased VAs. The ANOVA analysis reveals that this increase is significant ($p<0.05$). Similarly, the $\mathrm{P_R}$ values increase with increased VFs except for two trials, one with VF at 55 Hz and VA at 12.5 % of the baseline and another with VF at 30 Hz and VA at 50 % of the baseline. But this increase seems not to be significant ($p>0.05$).

B. RMS Estimation

The average RMS calculated with SSP is 501±210 μV, which is significantly larger than the RMS calculated without



(a)

(b)

(c)

Fig. 7. Results: a) P_R; b) ΔRMS; c) $-\Delta$MF. The asterisk (*) indicates a significant difference ($p<0.05$).

SSP (442±192 μV). Figure 7 (b) shows the relative RMS difference at different VAs and VFs. The average ΔRMS over all VAs and VFs is 12.2±3.8 %. Significant ($p<0.05$) increase in ΔRMS is observed with increased VAs. Moreover, similar to P_R, ΔRMS increases with increased VFs except for two trials, one with VF at 55 Hz and VA at 12.5 % of the baseline and another with VF at 30 Hz and VA at 50 % of the baseline. However, the effect of VF on ΔRMS is not significant ($p>0.05$).

C. MF Estimation

The average EMG MF value estimated with SSP is 92.36±11.46 Hz, which is slightly lower than that excluding the SSP (94.30±11.67 Hz). Although small (ΔMF=2.10±1.04 %), this difference is significant, as indicated by the Student's t-test. The average ΔMF over different VAs and VFs is shown in Fig. 7 (c). Similar to ΔRMS, ΔMF increases significantly with increased VAs except for one pair, i.e., 12.5 % vs. 25 % of the baseline. Furthermore, different from P_R and ΔRMS, ΔMF decreases with increased VFs. This decrease is also significant except for one pair between 20 and 30 Hz.

D. CV Estimation

The average value of CV estimated with and without SSP are 4.49±0.78 m/s and 4.49±0.79 m/s, respectively, as shown in Fig. 8. Different from RMS and MF, our statistical analysis reveals no significant difference between CV estimation with and without SSP. Therefore, no ΔCV is calculated nor statistically analyzed.

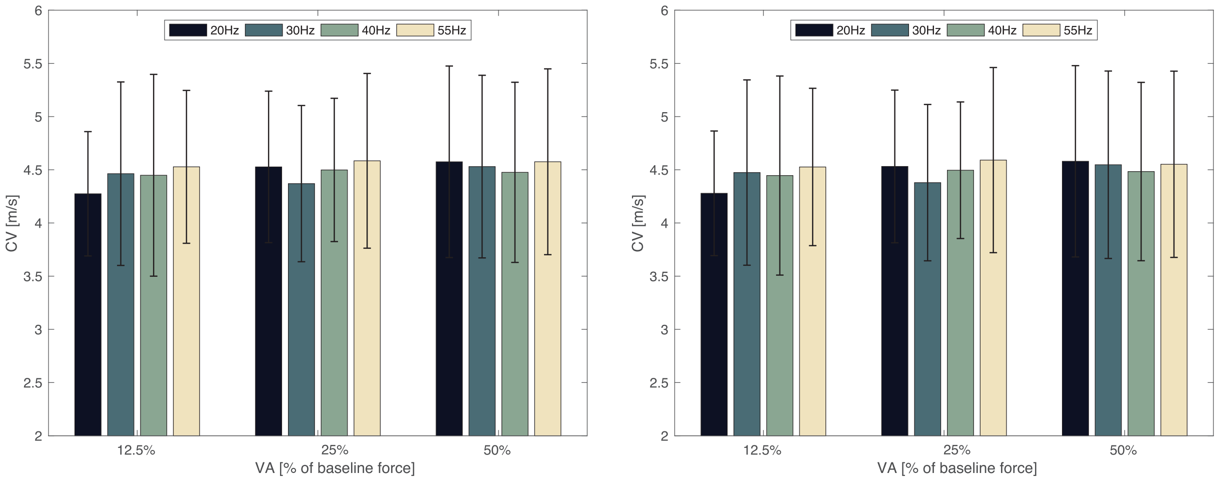
IV. DISCUSSION

The relevance of the sharp peaks in the EMG spectrum recorded during dynamic force modulation is quantitatively studied in our study, irrespective of the controversial interpretation of their nature being muscle activity or motion artifacts. The P_R of the peaks indicating their contribution to the entire EMG power is first calculated. Besides, three widely used EMG parameters, RMS, MF, CV, are estimated with and without SSP and then statistically compared. Our results reveal an average P_R of 21.18±15.68 %, and a significant ΔRMS and ΔMF of 12.2±3.8 % and 2.10±1.04 %, respectively. However, these peaks seem to have no impact on EMG CV.

In the present study, an average power contribution of 21.18±15.68 % is observed for the SSP at the VF and FH, in line with previous studies [33]. Furthermore, the P_R value increases with increased VA and VF, which can be attributed to the fact that the outcomes of VE depends on the initial muscle contraction level and the frequency of the vibratory stimuli, as reported by previous studies [35], [46].

The RMS value has been widely used as an estimate of the EMG amplitude [46]. Our results reveal a significant decrease in RMS when excluding the SSP at the VF and FH, which is expected as less power is included by removing these peaks. Due to the same reason, variation in ΔRMS with changing VA and VF is highly correlated with the variation of P_R, as shown in Fig. 7 (a) and (b).

In our study, the MF of the sEMG signals estimated without SSP are around 92 Hz, in line with previous studies [47]. A significant increase in MF (2.10±1.04 %) is observed after excluding the SSP, which is due to the fact that the SSP are located on the left side of the original MF estimated with SPP (Fig. 6). Removal of those peaks can lead to a shift of the MF towards the right side, thus an increase in MF. In addition, as increased P_R is observed with increased VA, an increase in ΔMF with increased VA is also observed. However, different from P_R and ΔRMS, ΔMF decreases with increased VF. This



(a) (b)

Fig. 8. CV results: a) EMG CV estimation with SSP; b) EMG CV estimation without SSP.

can be explained by the fact that higher VFs (from 15 to 55 Hz) approach the MF of the EMG without SSP (92 Hz), resulting in smaller ΔMF.

The average CV estimated with SSP is 4.49±0.78 m/s, in line with previous studies [31], [48]. However, although accounting for 21.18±15.68 % of the total EMG power, these SSP have no significant influence on the estimated CV. Possible explanation can be that the CV of these SSP is quite similar to that of normal EMG. In fact, previous studies did find these peak components to propagate similar to normal sEMG components [37]. Furthermore, the mechanical vibration that was considered as the source of motion artifacts was also measured by accelerometers in [37], and its CV was much larger (>100 m/s) than normal EMG. All of these findings indicate those SSP to be mainly reflecting muscle activity rather than motion artifacts, which is an evidence of synchronized motor unit activation through TVR [49].

It should be noted that although other two EMG parameters, i.e., RMS and MF, are significantly influenced by the SSP, we may still indicate the SSP to be primarily due to muscle activity based only on the CV results. The reason resides in that the effects of the SSP on the RMS and MF provide no hint on the nature of these peaks due to the estimator of these two parameters. According to equations 2 and 4, removal of any component in the frequency band between 15 and 450 Hz will definitely result in variations in RMS and MF, irrespective of the nature of the removed component being motion artifacts or muscle activity. However, our RMS and MF results provide quantitative indication of the underestimation in RMS and overestimation in MF when excluding these SSP. Taking all into consideration, we suggest to include these SSP when estimating EMG parameters for VE analysis, as neglecting them may lead to overlooking relevant electromyographic activity.

Finally, the findings of the present study are based on but not limited to the upper limb VE with dynamic force-modulation. Fratini et al found a similar power contribution of the SSP when performing lower limb VE with whole-body vibration platform [33]. In addition, Martin and Park observed these peaks by directly applying mechanic vibration to the muscle and suggested them to be vibration-induced muscle activity through the reflex loop [35]. Therefore, our findings seem also to be sound when analysing EMG signals recorded on other limbs or muscle groups with different VE systems, such as whole-body vibration platform.

V. CONCLUSION

In the present study, we quantify the P_R of EMG SSP during VE and their influence on the estimation of three widely used EMG parameters, i.e., RMS, MF, and CV, irrespective of the controversial nature of those peaks. Our results show an average P_R value of 21.18±15.68 %. When excluding these peaks, we observed a significant decrease in RMS (12.2±3.8 %) and an increase in MF (2.10±1.04 %). However, although these SSP account for more than twenty percent of the total EMG power, they have no significant influence on CV estimation, indicating these SSP to be mainly composed of muscle activity rather than motion artifacts. Consequently, there is no need to remove these SSP when analysing VE using sEMG. However, a thorough understanding of their nature requires more further investigation.

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