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Karin Lienhard

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UNIVERSITE DE NICE SOPHIA ANTIPOLIS

Faculté des Sciences du Sport – Unité de Formation et de Recherche en

Sciences et Techniques des Activités Physiques et Sportives

**ANALYSIS OF WHOLE-BODY VIBRATION EXERCISE EFFECT
ON LOWER LIMB MUSCLE ACTIVITY USING
SURFACE ELECTROMYOGRAPHY
METHODOLOGICAL CONSIDERATIONS
AND PRACTICAL APPLICATIONS**

THESE

En vue de l'obtention du grade de

Docteur d'Université

En Sciences du Mouvement Humain - Ecole Doctorale 463

Présentée et soutenue publiquement par

Karin Lienhard

Le 7 novembre 2014

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ABSTRACT

The aim of this thesis was to analyze the effect of whole-body vibration (WBV) exercise on lower limb muscle activity and to give methodological implications and practical applications. Two methodological studies were conducted that served to evaluate the optimal method to process the surface electromyography (sEMG) signals during WBV exercise and to analyze the influence of the normalization method on the sEMG activity. A third study aimed to gain insight whether the isolated spikes in the sEMG spectrum contain motion artifacts and/or reflex activity. The subsequent three investigations aimed to explore how the muscle activity is affected by WBV exercise, with a particular focus on the vibration frequency, platform amplitude, additional loading, platform type, knee flexion angle, and the fitness status of the WBV user. The final goal was to evaluate the minimal required vertical acceleration to stimulate the muscle activity of the lower limbs. In summary, the research conducted for this thesis provides implication for future investigations on how to delete the excessive spikes in the sEMG spectrum and how to normalize the sEMG during WBV. The outcomes of this thesis add to the current literature in providing practical applications for exercising on a WBV platform.

RÉSUMÉ

L'objectif de cette thèse a été d'analyser l'effet de l'exercice physique réalisé sur plateforme vibrante (whole-body vibration, WBV) sur l'activité musculaire des membres inférieurs, de développer des outils d'analyse méthodologiques et de proposer des recommandations pratiques d'utilisation. Deux études méthodologiques ont été menées pour identifier la méthode optimale permettant de traiter les signaux d'électromyographie de surface (sEMG) recueillis pendant la vibration et d'analyser l'influence de la méthode de normalisation de l'activité sEMG. Une troisième étude visait à mieux comprendre si les pics sEMG observés dans le spectre de puissance du signal contiennent des artéfacts de mouvement et/ou de l'activité musculaire réflexe. Les trois études suivantes avaient pour but de quantifier l'effet de la WBV sur l'activité musculaire en fonction de différents paramètres tels que, la fréquence de vibration, l'amplitude de la plateforme, une charge supplémentaire, le type de plateforme, l'angle articulaire du genou, et la condition physique du sujet. En outre, l'objectif a été de déterminer l'accélération verticale minimale permettant de stimuler au mieux l'activité musculaire des membres inférieurs. En résumé, les recherches menées au cours de cette thèse fournissent des solutions pour de futures études sur : i) comment supprimer les pics dans le spectre du signal sEMG et, ii) comment normaliser l'activité musculaire pendant un exercice WBV. Enfin, les résultats de cette thèse apportent à la littérature scientifique de nouvelles recommandations pratiques liées à l'utilisation des plateformes vibrantes à des fins d'exercice physique.

DECLARATION OF ORIGINALITY AND AUTHORSHIP

I, Karin Lienhard, hereby declare that this thesis and the work reported herein was composed by and originated entirely from me. Information derived from the published and unpublished work of others has been acknowledged in the text and the sources are given in the list of references.

Signed in Nice, France, the 9th of September 2014

A handwritten signature in black ink, appearing to read 'K. Lienhard', written in a cursive style.

Karin Lienhard

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BIBLIOGRAPHY

PREFACE

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LIST OF ABBREVIATIONS

Acc	Acceleration
Amp	Amplitude
ANOVA	Analysis of variance
BF	Biceps Femoris
CV	Coefficient of variation
Disp	Displacement
Freq	Frequency
GL	Gastrocnemius Lateralis
GM	Gastrocnemius Medialis
H-reflex	Hoffmann-reflex
ICC	Intraclass correlation coefficient
ISO	International Organization for Standardization
LOA	Limits of agreement
MUAP	Motor unit action potential
MVC	Maximal voluntary contraction
PSD	Power spectral density
RF	Rectus Femoris
RMS	Root mean square
sEMG	Surface electromyography
sEMG _{RMS}	Root mean square of the surface electromyography
SD	Standard deviation
SOL	Soleus
TA	Tibialis Anterior
TVR	Tonic vibration reflex
VL	Vastus Lateralis
VM	Vastus Medialis
WBV	Whole-body vibration

CHAPTER 1

GETTING TO KNOW WHOLE-BODY VIBRATION

1.1 WHAT IS WHOLE-BODY VIBRATION?



Whole-body vibration (WBV) means transmission of a mechanical repetitive movement to the body. WBV employed as a training device requires a platform that generates sinusoidal oscillations by motors underneath the vibration platform that are transmitted to the person on the machine. The oscillations can be adjusted by manipulating the frequency and/or the displacement of the platform. The resulting acceleration (Acc) is defined by the frequency (Freq) and the amplitude (Amp) of the platform displacement via the equation $Acc = (2 \times \pi \times Freq)^2 \times Amp$. The person using the machine can adapt various positions on the platform such as sitting, standing, squatting, or hovering.



Whole-body vibration exercises have been and are still currently being used amongst professional athletes such as ice hockey players, skiers, golfers or marathon runners as a warm-up or as a method to increase performance. In recent years, WBV training has increased in popularity amongst recreational athletes. As a result, WBV platforms are available in many gyms and are even affordable for individual purchases. The commercialization of WBV platforms has led to numerous companies manufacturing such platforms, leading to a large variety of products available. As the popularity of WBV platforms has grown faster than the scientific background, manufacturers have claimed benefits that are untested and open to debate. Their claims, based on anecdotal evidence, include weight loss, reduced lower back pain, improved balance, increased bone density, enhanced blood flow and increases in muscle strength and power.



These unwarranted claims of positive effects through WBV have made many researchers rightfully frown upon the use of WBV as a proven performance/health stimulant. While the image of WBV has remained dubious to the scientific community, there has been an increase in research performed, and some research has successfully replaced some of the anecdotal evidence with scientific proof. This increase in WBV's scientific understanding is directly correlated with the dramatic uptick in research being completed on its effects.

1.2 WBV AND SCIENCE

WBV has gained more and more popularity amongst the scientific community mainly over the last two decades. To give a general idea of the increase in publications around WBV, an internet search was conducted using the portal of the US National Library of Medicine (<http://www.ncbi.nlm.nih.gov/pubmed>). A first search was launched by using the term **whole-body vibration [All Fields]**. The total number of publications until the end of the year 2013 was 1044, with an exponential increase over the past 20 years (**Figure 1.1**). The highest number of publications per year was reached in 2013 with 130 publications.

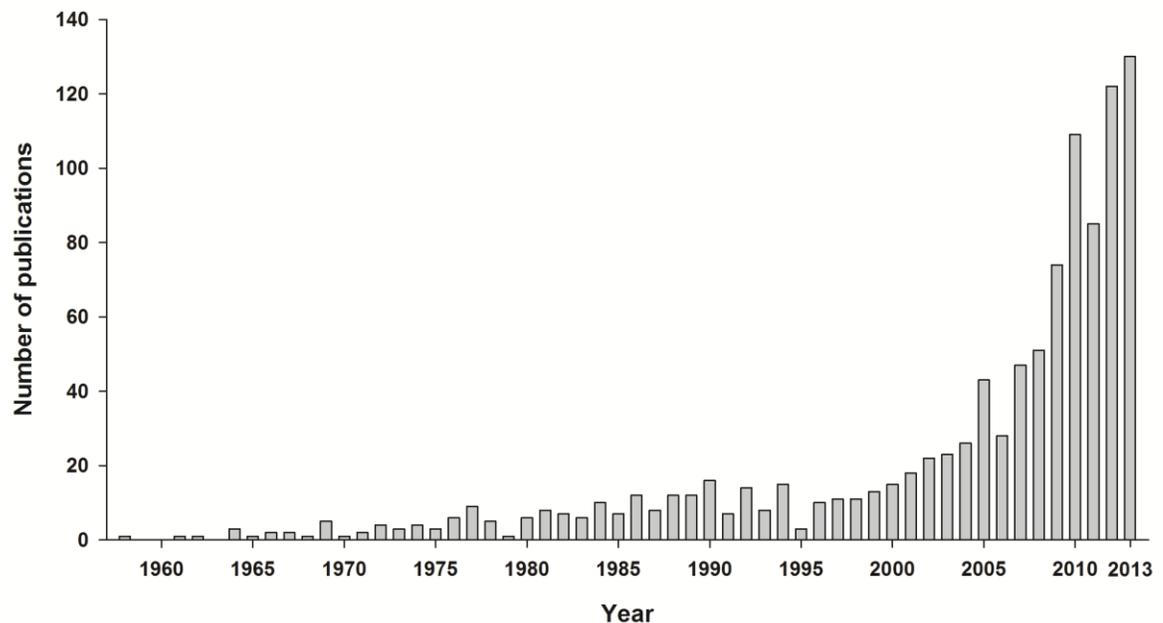


Figure 1.1 The amount of publications sorted by year covering whole-body vibration indexed by the US National Library of Medicine. A total amount of 1044 publications was found by the end of the year 2013.

Out of the claims that have been made by the manufacturers, blood flow and weight loss have received the least attention by the scientific community thus far, with only about 70 and 25 publications respectively. Other fields such as back pain, balance, and bone density have been more in the focus with 100 - 200 publications per topic. By far the most prevalent subject has been muscle strength and power, with over 350 publications.

1.3 WBV AND ITS EFFECTS ON MUSCLE PERFORMANCE

The first scientific studies that have investigated the effect of WBV exercises on muscle strength and power were conducted in the late 90's. Bosco et al. (1999a; 1999b) have tested the immediate effects of WBV exercises on muscle performance, whereas one limb received the vibration treatment and the other limb served as a control entity that received no vibration treatment. Bosco et al. (1999a) found that 5 x 1 min of vibration applied to a contracted arm increased its average power immediately after the treatment compared to the control arm. Similarly for the lower limbs, Bosco et al. (1999b) found that after exposing the leg to 10 x 60 s of WBV, velocity, force and power was increased in the vibration-treated leg but not in the control leg. Bosco et al. (1998) also examined the short-term effects of WBV exercises by testing jumping performance before and after 10 days of a training period. This consisted of WBV exercises in an intervention group and no exercises in a control group. The measurements after the training period showed that only the intervention group improved their power output and their jump height. While Bosco and his group were amongst the first researchers who showed that WBV exercises can positively affect muscle performance, the draw-back of their study set-up was that the control group/limb did not receive the same treatment as the intervention group/limb but without vibration. Hence, the parameter "vibration" was not isolated in their investigations, meaning that it is not clear if the positive adaptations originated from the vibration per se.

Subsequent studies compared the short-term effects of WBV training to the same training without vibration. De Ruyter et al. (2003) found no effect on muscle strength and power that was solely attributable to WBV, as muscle strength of the knee extensor muscles and counter movement jump performance was unaffected after 11 weeks of training regardless of utilizing WBV or not. On the other hand, Delecluse et al. (2003) found a significant effect on muscle strength solely attributable to WBV, as isometric and dynamic knee extensor strength was increased after 12 weeks of WBV training but not after 12 weeks of the same training without WBV. Further, Delecluse et al. (2003) found that muscle strength was positively affected by the WBV training in equal measures as by resistance training, but that an improvement in counter

movement jump performance was only observed after the WBV intervention. Later studies confirmed beneficial immediate (McBride et al. 2010) as well as beneficial short-term effects (Colson et al. 2010; Petit et al. 2010) of WBV exercises on muscle strength of the lower limbs.

The findings from the early 2000's studies have led to the conclusion that WBV triggers something in the muscle that improves performance. Therefore, subsequent investigations were focused on the changes in muscle activity while the body is being vibrated. As a result, with the use of surface electromyography (sEMG) recordings, muscle activity during WBV exercises started to be extensively investigated in the years thereafter.

1.4 MUSCLE ACTIVITY DURING WBV

In order to illustrate that in the early 2000's researchers became more and more interested in examining muscle activity during the exposure to WBV, a more refined search was launched to narrow down the publications that have investigated WBV in relation with muscle activity. The used search term was **whole-body vibration [All Fields] AND (EMG [All Fields] OR sEMG [All Fields] OR muscle activity [All Fields])**. The total amount of publications until the end of 2013 was 111, whereas the most publications were found in the year 2012 (Figure 1.2).

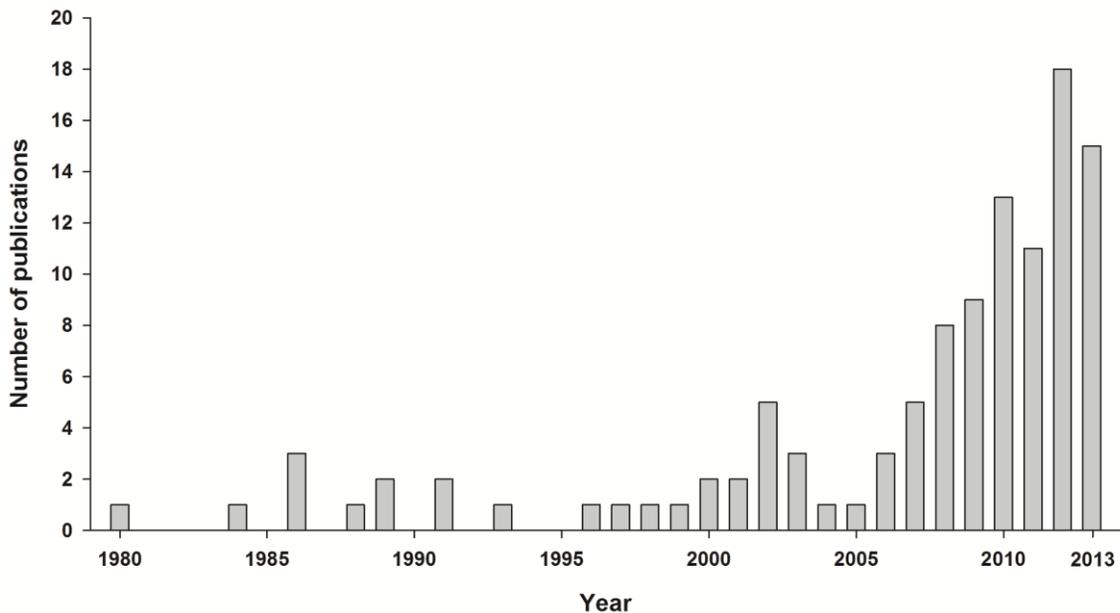


Figure 1.2 The amount of publications sorted by year covering whole-body vibration together with muscle activity or surface electromyography indexed by the US National Library of Medicine. A total amount of 111 publications was found by the end of the year 2013.

1.5 PROBLEMATIC OF WBV STUDIES

Although it was not distinguished whether muscle activity was assessed before, during, or after the WBV exercise, this search shows that the interest in the effect of vibration on muscle activity as measured by sEMG recordings has been increasing over the last decade. Unfortunately, contradictory findings have been reported in these studies. This makes drawing general conclusions about the effectiveness of WBV exercises challenging. This lack of concise conclusions stems from several issues.

First, although directly applied vibration to the muscle belly or the tendon elicits stretch reflex responses, it is unclear if such responses occur during WBV exercises. Second, it is unknown whether the myoelectrical signal during WBV is contaminated by motion artifacts or not. This disagreement has led to unstandardized methods to process and analyze the sEMG signals during WBV. Third, WBV studies have been using different platform types, vibration frequencies and amplitudes, as well as different body positions. Hence, due to the large variety of the used study parameters, it is challenging to explain the conflicting outcomes. Subsequently, it is crucial to develop a sEMG processing method for a standardized use in WBV studies, and to describe the effectiveness of WBV and its optimal parameters leading to the highest positive changes in sEMG activity during the exposure to vibration.

Therefore, **CHAPTER 2** of this thesis aims to describe the potential mechanisms explaining the increased sEMG activity during WBV exercises. **CHAPTER 3** expands on the findings and outcomes of the above mentioned scientific literature, and the problematic of WBV studies are further discussed. **CHAPTER 4** describes the methodology that was used aiming to (1) find the optimal method to process and analyze the sEMG during WBV, (2) to clarify whether the sEMG signal during WBV contains reflex activity and/or motion artifacts, and (3) to scrutinize the effect of WBV and its parameters on muscle activity levels. The detailed studies and their outcomes are each presented in a chapter (**CHAPTER 5-10**). Last, the outcomes of this thesis are discussed in **CHAPTER 11**, and conclusions are drawn in **CHAPTER 12**.

CHAPTER 2

POTENTIAL MECHANISMS DURING WHOLE-BODY VIBRATION

2.1 OVERVIEW OF THE POTENTIAL MECHANISMS

The potential mechanisms leading to increased muscle activity during the exposure to WBV are various, and some of them have only been evidenced with directly applied vibration on the muscle belly or the tendon of interest. The most frequently cited mechanism in relation with WBV exercises is the tonic vibration reflex (TVR), mediated by the primary and secondary endings of the muscle spindles (Cardinale and Bosco 2003; Rittweger 2010). Besides the muscle spindles, other somatosensory receptors provide feedback during direct exposure to vibration, including mechanoreceptors in the skin, tendon organs, joint receptors, nociceptors, and thermal sensors (Brooke and Zehr 2006). While these mechanisms are well documented during directly applied vibration, evidence is lacking for their occurrence during WBV. A more recently identified mechanism with demonstrated occurrence during WBV is vibration damping (Wakeling and Nigg 2001). However, some of these mechanisms have only been evidenced during directly applied vibration to the muscle or the tendon. Nevertheless, there is the potential that all these mechanisms occur during WBV. The most prevalent mechanisms are described in the following.

2.2 TONIC VIBRATION REFLEX

The occurrence of muscle contractions evoked by vibrations has been called the TVR, and has been experimentally established a few decades ago (Hagbarth and Eklund 1965; Matthews 1966). The TVR is based on stretch reflex responses, which are mediated by the muscle spindles. It needs to be emphasized that in these studies and in the ones mentioned in the following, vibration was directly applied to the muscle belly or the tendon of interest.

2.2.1 MUSCLE SPINDLES

Muscle spindles have a fusiform shape and lie in parallel with the skeletal muscle fibers. The muscle spindle fibers are referred to as intrafusal fibers, whereas the skeletal muscle fibers outside the spindle capsule are called extrafusal fibers (**Figure 2.1**). The muscle spindle has an afferent supply, i.e. Ia-afferent and II afferent, to transmit action potentials to the central nervous system. In addition to that, intrafusal fibers receive efferent input, namely from the

alpha (α) and gamma (γ) motor neurons. Alpha motor neurons innervate extrafusal fibers, whereas gamma motor neurons connect to intrafusal fibers. Normal mode of operation involves concurrent activation of the alpha and gamma motor neurons, which is known as alpha-gamma co-activation. The alpha motor neurons activate the extrafusal fibers to produce the force required for the task, and the gamma motor neurons activate the intrafusal fibers to set the desired level of feedback from the muscle spindle. A stretch of the muscle is detected by the primary (Brown et al. 1967; Roll and Vedel 1982) and secondary (Brown et al. 1967) endings of the muscle spindles which results in innervations of the Ia-afferents (De Gail et al. 1966).

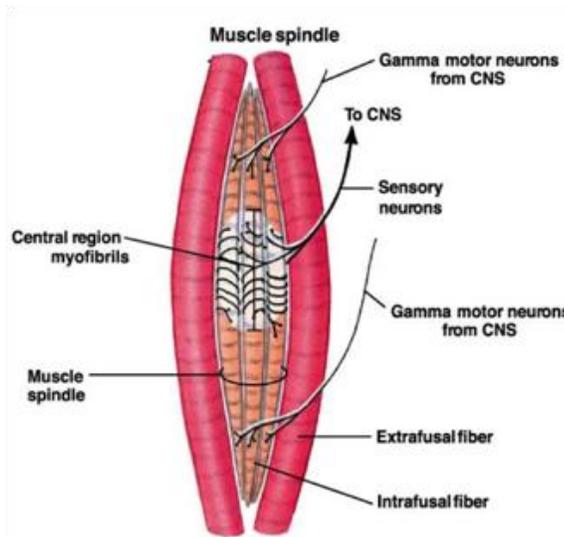


Figure 2.1 Schematic illustration of a muscle spindle.

(Retrieved from <http://www.studyblue.com/notes/n/movement-science-motor-control/deck/1421471>)

2.2.2 STRETCH REFLEX

The TVR is based on stretch reflex responses; rapid muscle contractions as a response to a brief, unexpected increase in length (stretch) of the muscle. The stretch reflex consists of a short-latency response (M1) after 30 ms via homonymous Ia excitation. The medium-latency response (M2) occurs after 50 - 60 ms and is mediated by the transcortical pathway. Occasionally, a long-latency response (M3) is observed occurring after 90 - 100 ms also mediated by the transcortical

pathway (Petersen et al. 1998). The short-latency response is mediated by monosynaptic and polysynaptic pathways. The monosynaptic pathway evokes a homonymous Ia excitation, which means that efferent motor neurons innervate the same muscle from which the afferent signal originated. The function of this pathway is to reverse the stretch by a rapid muscle contraction. The polysynaptic pathway evokes a reciprocal Ia inhibition, which means that Ia inhibitory interneurons evoke inhibitory postsynaptic potentials in the motor neurons that innervate the antagonist muscles (**Figure 2.2**). The function of the polysynaptic pathway is to link the inhibition of an antagonist muscle to the activation of the agonist muscle in order to complete the task (e.g. flexion, extension). Reciprocal inhibition declines during tasks that involve co-activation of agonist and antagonist muscles, and increases during postural activity.

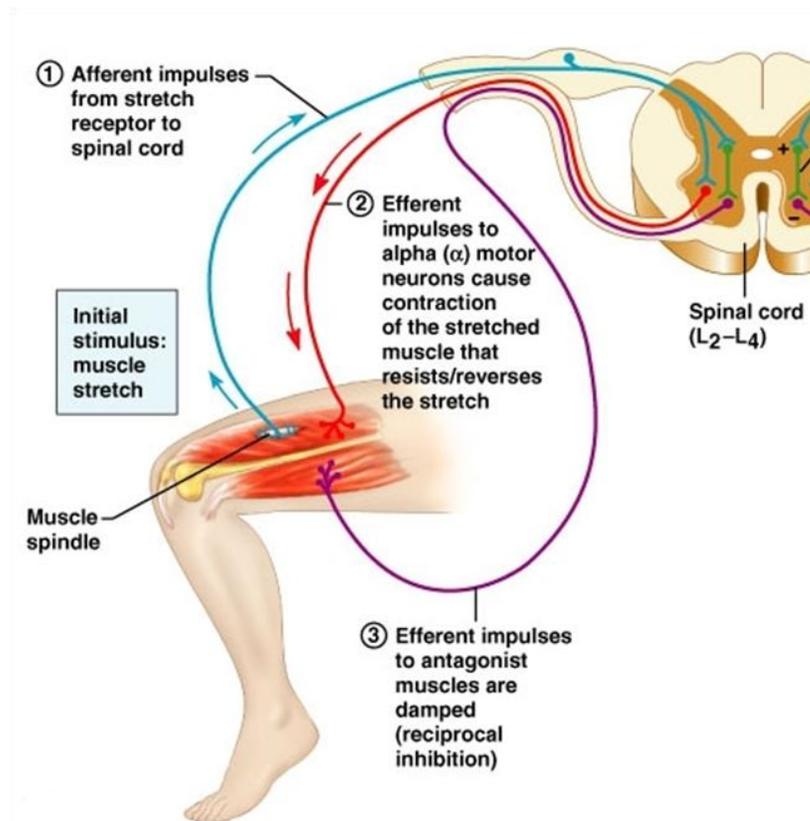


Figure 2.2 Schematic representation of the monosynaptic (1, 2) and polysynaptic (3) stretch reflex pathway.

(Retrieved from

<http://www.studyblue.com/notes/n/flexibility/deck/4090637>)

The TVR evoked during directly applied vibration has been observed in a frequency range of 20 - 200 Hz (Eklund and Hagbarth 1966; Burke et al. 1976a; Burke et al. 1976b; Martin and Park 1997), with increasing motor unit synchronization the higher the frequency up to 100 Hz (Martin and Park 1997). The TVR is also positively correlated to the amplitude of the vibratory input stimulus (Eklund and Hagbarth 1966). The TVR occurs whether the innervated muscle is relaxed or contracted (Bongiovanni and Hagbarth 1990), and the magnitude of the TVR is dependent on the amount of the background muscle activity (Eklund and Hagbarth 1966; Burke et al. 1976a; Gottlieb and Agarwal 1979; Bedingham and Tatton 1984; Ogiso et al. 2002). Background muscle activity increases the magnitude of the TVR response due to its relation to the excitability of the motoneuron pool. On the contrary, the amplitude of the TVR is reduced when the antagonist muscle is contracted due to reciprocal Ia inhibition (Gottlieb and Agarwal 1979). Additionally, the amplitude of the TVR is greater with increasing muscle length or stretch because of an increased sensitivity of the muscle spindles during the stretch (Burke et al. 1976b; Nordin and Hagbarth 1996). The characteristics of the TVR have further been studied by evoking Hoffmann (H)-reflexes. The H-reflex has been studied before and after a vibratory stimulus as its magnitude, like the magnitude of the TVR, represents the excitability of the motoneuron pool. The difference between the H-reflex and the TVR is that the H-reflex bypasses the muscle spindles as the nerve is directly innervated. Before discussing the behavior of the H-reflex during direct vibration, the pathway of the H-reflex is described in the following section.

2.2.3 H-REFLEX

The H-reflex can be evoked by application of a single electrical stimulus to a peripheral nerve, which evokes a twitch response in the muscle, approximately 28 – 35 ms after the stimulus. This response can be measured as the sEMG and/or the force resulting from the twitch. The H-reflex is most commonly studied in the Soleus (SOL), but it can also be evoked in the Quadriceps Femoris, the Hamstrings, and Tibialis Anterior (TA) (Enoka 2008). The pathway underlying the H-reflex involves activation of Ia afferents and subsequent generation of action potentials in the same muscle. Increasing the intensity of the electrical stimulus evokes a short-latency (5 ms) response called the M-wave. While the H-reflex tests the excitability of the motor neuron pool,

the M-wave tests neuromuscular propagation. When the nerve is stimulated, the action potential is propagated both toward the neuromuscular junction and back toward the motor neuron. The action potential (**Figure 2.3**) propagating toward the neuromuscular junction (1) produces an M-wave, whereas the action potential propagating toward the motor neuron (2) produces an H-reflex. As the pathway of the H-reflex involves a spinal loop, the H-reflex is observable shortly after the M-wave response.

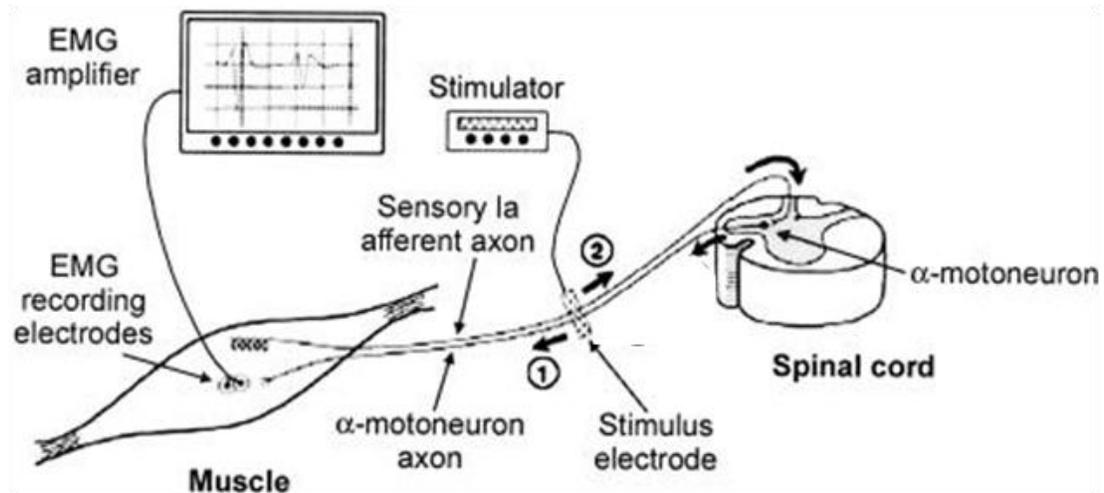


Figure 2.3 Pathway for the Hoffmann (H)-reflex and the M-wave. The electrical stimulus is applied over the peripheral nerve. The action potential that propagates toward the neuromuscular junction (1) produces an M-wave, whereas the action potential that propagates toward the motor neuron (2) produces an H-reflex.

(Retrieved from <http://bmsi.ru/doc/61f60d25-6a9e-426f-a16e-5e1101da0811>)

Studies have shown that the magnitude of the H-reflex increases with increasing baseline muscle activity (Verrier 1985) and with increasing torque levels (Pensini and Martin 2004), and that the H-reflex is linearly correlated to torque levels ranging from 0 to 50% of the maximal voluntary contraction torque (Butler et al. 1993). These observations are in line with the ones for the TVR, and support the following conclusions.

The magnitude of the TVR during direct vibration

- is positively related to the vibration frequency and amplitude of the input stimulus
- is positively related to the background muscle activity
- is positively related to the amount of the stretch of the muscle

2.3 CUTANEOUS MECHANORECEPTORS

Cutaneous mechanoreceptors provide information about the acceleration of the skin and the deeper tissues such as movement of the hair, and displacement and stretch of the skin. This feedback is provided by the Meissner corpuscles, Pacinian corpuscles, Merkel disks, Ruffini endings, and free nerve endings (Figure 2.4).

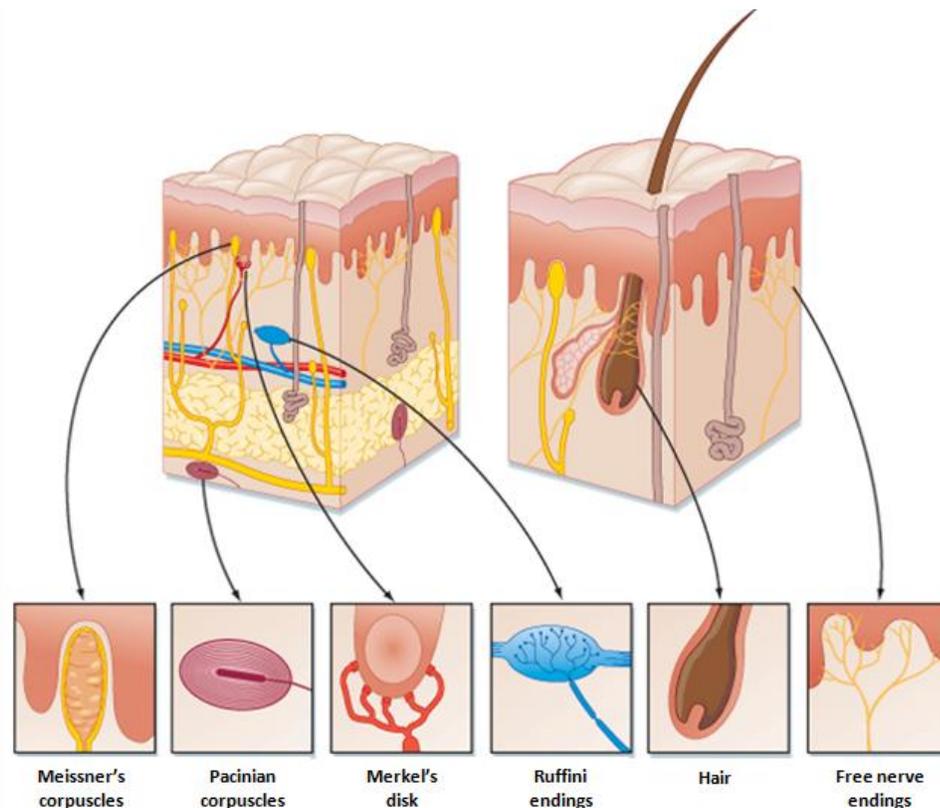


Figure 2.4 Schematic illustration of the cutaneous mechanoreceptors consisting of Meissner corpuscles, Pacinian corpuscles, Merkel disks, Ruffini endings, and free nerve endings.

(Retrieved from <http://users.atw.hu/blp6/BLP6/HTML/C0079780323045827.htm>)

The cutaneous mechanoreceptors can be divided into fast and slow adapting receptors. The fast adapting receptors are the Meissner corpuscles and the Pacinian corpuscles; the Meissner corpuscles are most responsive to a vibratory stimulus around 40 Hz and the Pacinian corpuscles around 100 Hz and above (Rittweger 2010). More specifically, the Meissner corpuscles are sensitive to local, maintained pressure, and their response fades rapidly. The Pacinian corpuscles, which are the largest receptors in the skin, detect rapid changes in pressure and respond mainly to the acceleration component of a stimulus. The slow adapting receptors, the Merkel disks and the Ruffini endings, are sensitive to sustained pressure. Merkel disks are sensitive to local vertical pressure and do not respond to lateral stretch of the skin. The Ruffini endings respond to stretch of the skin and are sensitive to the direction of the stretch. Last, the free nerve endings provide feedback on the movement of the skin and hair.

As the cutaneous mechanoreceptors provide the central nervous system with information about the magnitude, speed and direction of a mechanical stimulus, they are partly responsible for the sensorimotor alterations during a vibratory stimulus. While it has been believed that only the fast adapting mechanoreceptors are responsible for these alterations during vibration, a study by Ribot-Ciscar et al. (Ribot-Ciscar et al. 1989) suggests involvement of both, the slow and the fast cutaneous mechanoreceptor. It needs to be considered that the fast receptors become quiescent after a certain vibration frequency (Ribot-Ciscar et al. 1989), whereas the slow adapting receptors are able to respond one-to-one to vibration frequencies up to 200 Hz (Vedel and Roll 1982). The study by Ribot-Ciscar et al. (Ribot-Ciscar et al. 1989) also showed that the processing of pressure and touch is masked during vibration, and one of their earlier study found such a masking even after the vibration set-off (Ribot-Ciscar et al. 1996).

2.4 TENDON ORGAN

The tendon organ includes one afferent (i.e. Ib afferent) and no efferent connections (**Figure 2.5**). The Ib afferents are innervated when a muscle and its connective tissues are stretched, either through pulling of the muscle or through activation of the skeletal muscle fibers. Hence, the tendon organ provides information about the muscle force. In addition to the spindle afferents, Ib-afferents from the Golgi tendon organs are responsive to vibration (Burke et al. 1976a; Hayward et al. 1986). During the reflex contraction, discharge from the primary and secondary spindle endings decline, whereas discharge from the Golgi tendon organs increase (Burke et al. 1976a).

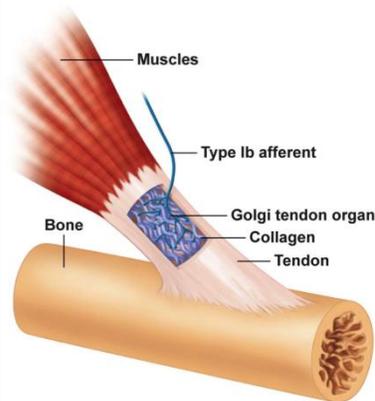


Figure 2.5 Schematic illustration of a tendon organ.

(Retrieved from <http://www.pinterest.com/pin/368732288212709089/>)

2.5 DAMPING

A more recently identified mechanism leading to increased sEMG activity during WBV exercises is damping. During vibration, the muscles and tendons act as a spring-mass system that experiences compression during the upstroke of the vibration platform, and expansion during the down stroke of the platform. The stiffness k and the mass m of the system determine the natural frequency ω_0 and are related with $\omega_0 = \sqrt{k/m}$. As the mass m of the system cannot be altered, adjustment of the natural frequency ω_0 can only be changed through alteration of the stiffness k . When the frequency of the actuator ω_A (i.e., the frequency of the vibration platform) coincides with the frequency of the resonator ω_0 , resonance can occur (Cardinale and Wakeling 2005; Boyer and Nigg 2006). This can lead to accumulation of energy where the vibration amplitude of the resonator is greater than the amplitude of the actuator. This amplitude amplification can lead to destruction – the so-called resonance catastrophe (**Figure 2.6**).

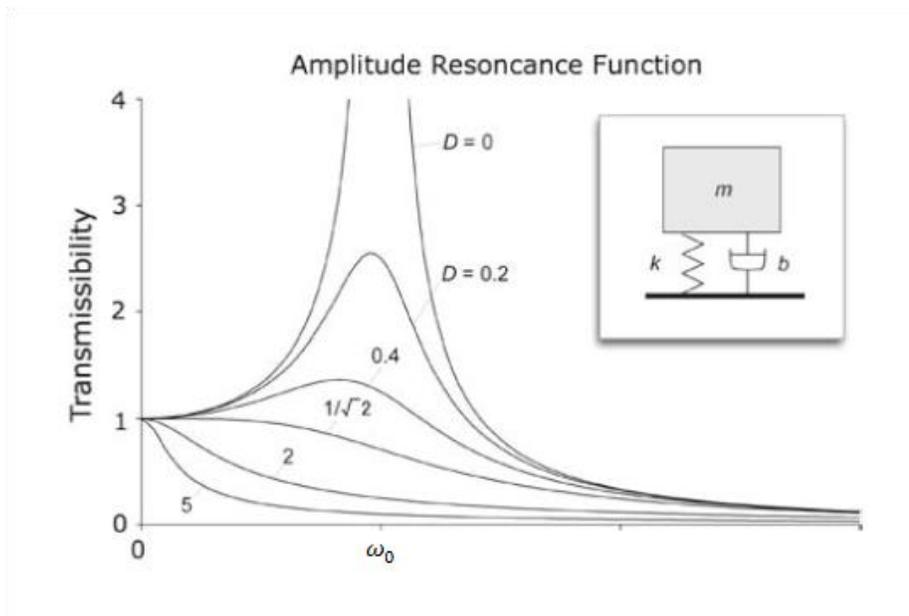


Figure 2.6 Model of the human body acting as a resonator. A mass is linked to a spring system with stiffness k and a dashpot with friction b . Resonance can occur in the system when the frequency of the actuator ω_A is close to the natural frequency ω_0 of the system. Transmissibility is defined as ω_0/ω_A and is amplified when > 1 .
(Adapted from Rittweger et al. 2010)

In order to prevent resonance-related tissue damage, it is speculated that the body aims to minimize vibrations by damping (Nigg 1997). The effects of resonance can be reduced by changing either the stiffness k and thus the natural frequency ω_0 , or the damping characteristics of the tissue, both of which are possible through increased muscle activation (Wakeling and Nigg 2001; Wakeling et al. 2002).

This mechanism has been confirmed during WBV exercises in a study by Wakling et al. (2002). In this study, the subjects were exposed to continuous vibrations and pulsed bursts across a frequency range of 10 - 65 Hz. It was found that muscle activity was elevated together with increased damping of the vibration power when the input frequency was close to the natural frequency of the soft tissue. As the frequency range of commercially available WBV platforms are typically within the range of the natural frequency of the human body, resonance can be expected as well as damping via increased muscle activity.

Besides increased muscle activation, the stiffness of the body can be altered by changing the body position, such as the degree of knee flexion. Effective body stiffness increases with straightening the legs (Lafortune et al. 1996), which increases the natural frequency ω_0 . Therefore, an optimal body position can help to prevent resonance.

In summary, spinal reflexes elicited by e.g. muscle spindles, cutaneous mechanoreceptors, or tendon organs as well as damping via increased muscle activity have been identified as possible mechanisms that could explain the elevated sEMG activity during WBV exercises. However, it is debatable whether spinal reflexes not only occur during directly applied vibration but also during WBV exposure.

CHAPTER 3

LITERATURE REVIEW – DURING WHOLE-BODY VIBRATION

3.1 TONIC VIBRATION REFLEX

Although the TVR is a well established mechanism during directly applied vibration to the tendon and the muscle, it is not clear if the TVR occurs during WBV. In order to clarify this matter, several studies have compared the behavior of the stretch reflex and the H-reflex before and after WBV. The magnitude of the stretch reflex response was found to be enhanced (Rittweger et al. 2003; Melnyk et al. 2008), unchanged (Hopkins et al. 2009; Cochrane et al. 2010; Fernandes et al. 2013; Yeung et al. 2014), and even depressed (Ritzmann et al. 2011) immediately after the WBV exercise. Similarly, the magnitude of the H-reflex was found to be unchanged (Pollock et al. 2012) and lower (Armstrong et al. 2008; Apple et al. 2010; Ritzmann et al. 2011; Games and Sefton 2013; Hortobágyi et al. 2014) immediately after the WBV exercise. Post-WBV enhancement of reflex responses is rather surprising, as one would expect reflex depression as observed during directly applied vibration for the H-reflex (Arcangel et al. 1971; Roll et al. 1980; Van Boxtel 1986) and the stretch reflex (Roll et al. 1980). Acknowledging that some studies found enhancement of the stretch reflex after directly applied vibration (Van Boxtel 1986; Shinohara et al. 2005), it is difficult to pin point as to why such discrepancies have been observed.

A sparse sample of studies has investigated the occurrence of the TVR/reflex activity during WBV by analyzing responses in muscle activity while the subject was exposed to the vibration (Ritzmann et al. 2010; Pollock et al. 2012; Zaidell et al. 2013). The earliest study (Ritzmann et al. 2010) compared the latencies and spectrograms of sEMG recordings during WBV with recordings of stretch reflex responses evoked by passive dorsiflexion with a custom-made ankle ergometer. The stretch reflex latency during dorsiflexion was defined as the time between the onset of the movement and the onset of the first rise in the sEMG signal. During WBV, the latency was defined as the time from the low point of the vibration platform and the onset of the first spike in the sEMG signal. The outcomes showed that the latencies and the spectrograms of the SOL and Gastrocnemius Medialis (GM) during dorsiflexion were extremely similar to those during WBV (**Figure 3.1**). In the same study, a pressure cuff was placed on the belly of the SOL muscle as occlusion has been shown to reduce the short-latency response of

the stretch reflex (Leukel et al. 2009). The influence of the pressure cuff on the amplitude of the stretch reflex was then compared during dorsiflexion and WBV, and it was found that the stretch reflex amplitude was reduced by about 56% during dorsiflexion and by 61% during WBV. Hence, the findings of this study showed that the behavior of the sEMG signal during WBV has similarities to the one of stretch reflex responses.

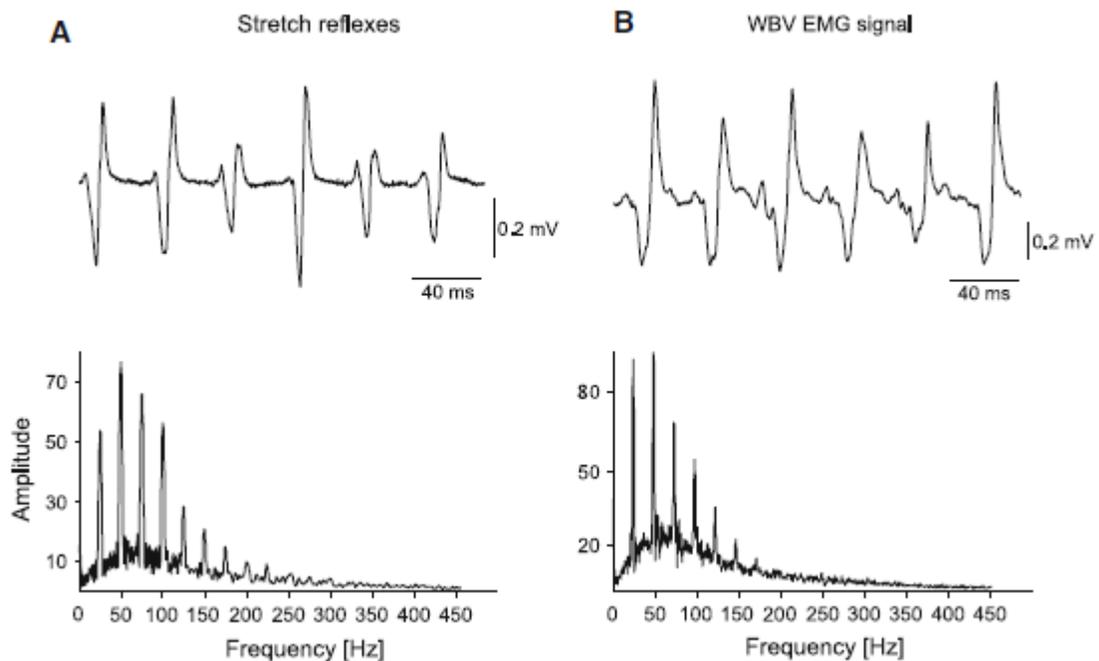


Figure 3.1 (A) Stretch reflex responses of the Soleus evoked by the ankle ergometer at 25 Hz (upper panel) and the mean of the corresponding frequency spectrograms of all participants (lower panel). (B) Surface electromyography signal of the Soleus during whole-body vibration at 25 Hz (upper panel) and the mean of the corresponding frequency spectra of all participants (lower panel).

(Reprinted from Ritzmann et al. 2010.)

In the second study (Pollock et al. 2012), single motor units of the Vastus Lateralis (VL) were recorded during WBV at 30 Hz using intramuscular EMG. Simultaneously, the waveform of the

platform acceleration was measured with the help of a strain gauge (**Figure 3.2**). The results of this study indicated that motor unit firing is phase locked to the vibration cycle as there was a strong relationship between the timing of the motor unit firing and the phase of the WBV cycle. This observation confirms the presence of reflex muscle activity during WBV, yet it is unclear which somatosensory receptors were involved.

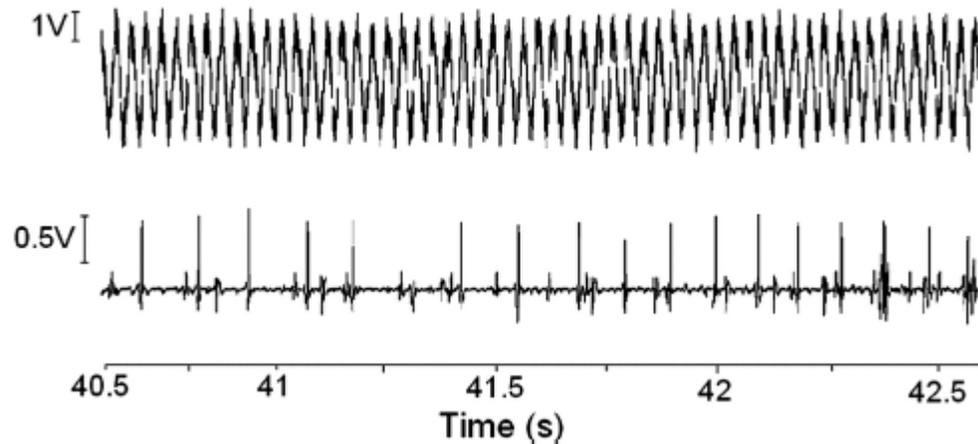


Figure 3.2 Platform acceleration recording (top graph) and intramuscular electromyography of the Vastus Lateralis (bottom graph) during WBV at 30 Hz. The firing of a single motor unit is phase locked to the phase of the vibration cycle. *(Reprinted from Pollock et al. 2012)*

The most recent study (Zaidell et al. 2013) compared sEMG activity of the SOL and TA during WBV and during Achilles tendon vibration. During the procedure, the subjects were seated on a chair with their leg positioned on the platform. WBV and Achilles tendon vibration was induced at 25 and 50 Hz for a time period of 70 s. Further, the effect of background muscle activity on the TVR was tested by passive loading of the leg. The motion artifacts induced by the vibrations of the platform were deleted in the Power Spectral Density (PSD) by subtracting a sine and cosine wave from the sEMG signals. However, the outcomes of this study were that the Root Mean Square (RMS) of the SOL and TA increased over the 70-sec time period during the WBV treatment as well as during the Achilles tendon vibration. This can be explained by increased

muscle spindle sensitivity, which points towards the occurrence of stretch reflex responses during WBV. Also, sEMG activity of the SOL and TA was affected by WBV in equal measures as by Achilles tendon vibration, which further supports the occurrence of the TVR during WBV. According to the findings of these studies, it can be assumed that WBV elicits stretch reflex responses and that the TVR occurs during WBV.

While Ritzmann et al. (2010) highlighted that actual stretch reflex responses display excessive peaks in the frequency spectrum (**Figure 3.1A**), it is not guaranteed that the excessive peaks that are observed during WBV (**Figure 3.1B**) are stretch reflex responses. Recently, it has been suggested that the sEMG signals recorded during WBV are contaminated by motion artifacts, caused by the oscillation of the platform. Since the motion artifacts as well as the stretch reflex responses would be phase-locked to the vibration frequency, it is not clear if these excessive peaks contain stretch reflex responses (or reflex activity), motion artifacts, or both.

3.2 OCCURRENCE OF MOTION ARTIFACTS

A typical WBV platform induces vibrations via sinusoidal oscillations, which are transmitted to the person standing on the platform. The vibrations travel through the entire body of the WBV user, which means that the muscles as well as the sEMG electrodes are oscillating. Percussion of the electrodes can result in recordings of motion artifacts. These artifacts are typically visible in the frequency domain as sharp peaks (from now on referred to as spikes) at the vibration frequency f_0 and at the multiple harmonics of the vibration frequency ($2f_0, 3f_0, \dots$).

The first research group who drew attention to this issue was Abercromby et al. (2007a), by illustrating a sEMG signal that was recorded during WBV in the frequency domain. **Figure 3.3** is a reprint of the original illustration and shows spikes at the vibration frequency $f_0 = 30 \text{ Hz}$ and at a few multiple harmonics ($2f_0 = 60 \text{ Hz}, 3f_0 = 90 \text{ Hz}, \dots$). Nevertheless, the excessive spikes per se are not a proof for the occurrence of motion artifacts, as these spikes could originate from stretch reflex responses. In order to verify the existence of motion artifacts during WBV, several studies have measured the artifacts with so-called dummy electrodes. In the study where this set-up has been introduced (Fratini et al. 2009a), electrodes were placed on the patella assuming that any signal recorded during WBV would be due to the shaking of the electrodes. Unfortunately, even though excessive spikes in the sEMG spectrum were located, it cannot be excluded that these peaks were not sEMG recordings from surrounding muscles. To address this limitation, a similar set-up has been repeated (Ritzmann et al. 2010; Sebik et al. 2013) by positioning electrodes on the muscle with several layers of tape between skin and electrodes to ensure exclusive motion artifact related recordings without any cross-talk from surrounding muscles. While Sebik et al. (2013) found motion artifact-contaminated sEMG signals, Ritzmann et al. (2010) observed only a marginal contribution.

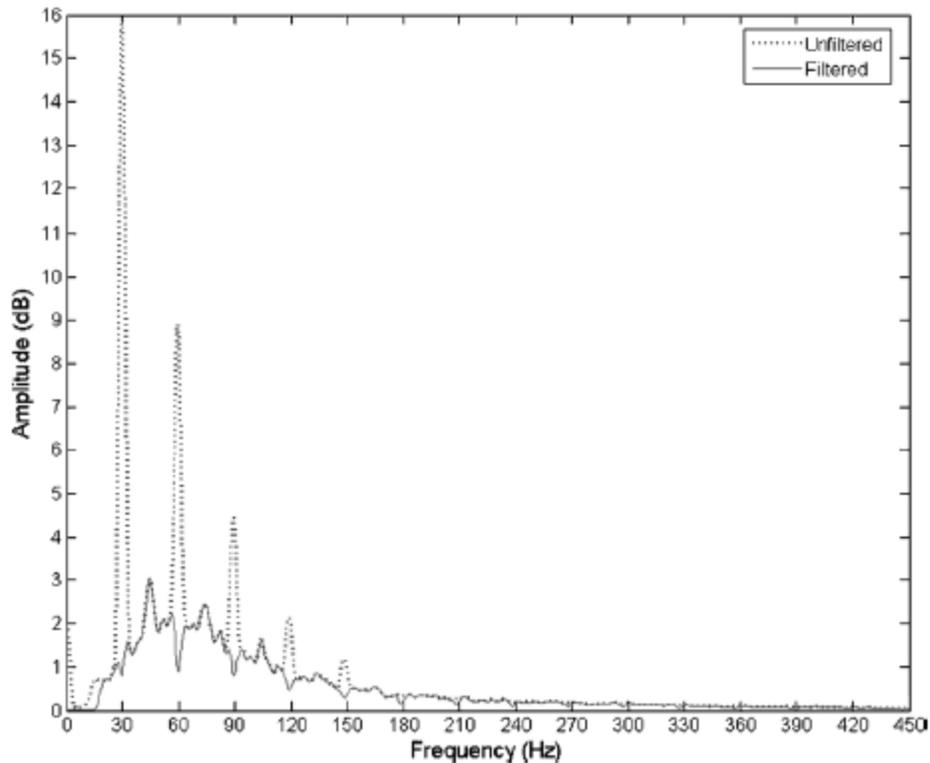


Figure 3.3 Surface electromyography signal recorded during whole-body vibration at 30 Hz in the frequency domain. Excessive peaks are observable at the vibration frequency $f_0 = 30 \text{ Hz}$ and its multiple harmonics $2f_0 = 60 \text{ Hz}$, $3f_0 = 90 \text{ Hz}$, *(Reprinted from Abercromby et al. 2007a)*

Possible reasons for the observed discrepancy between these studies (Fratini et al. 2009a; Ritzmann et al. 2010; Sebik et al. 2013) could be the type of the used equipment such as the electrodes or the cables (Webster 1984). Using conjoined electrodes vs. single electrodes might have the advantage that the conjoined electrodes are moving in phase to each other, which impedes a difference in potential between the electrodes. Additionally, the degree of cable shielding could have an influence, as motion artifacts are reduced by using shielded cables vs. unshielded cables (Clancy et al. 2002). However, it is unclear as to why Ritzmann et al. (2010) did not observe significant spikes in the dummy electrode signal.

3.2.1 sEMG SPIKES - MOTION ARTIFACTS OR REFLEX ACTIVITY?

While it is very likely that WBV elicits stretch reflex responses, it is unknown if such responses are contained in the isolated spikes. Hence it is not surprising that the scientific community has been debating in the past few years whether the spikes are due to stretch reflex responses, motion artifacts, or both. Although the isolated spikes have never been analyzed, a very recent study (Sebik et al. 2013) gave evidence that motion artifacts and stretch reflex responses may contribute to the sEMG spectrum during WBV.

The study by Sebik et al. (2013) used an innovative sEMG filtering regime to demonstrate the occurrence of motor unit synchronization during WBV. First, the sEMG signals were band-pass filtered between 80 - 500 Hz in order to delete the low-frequency sEMG spectrum where motion artifacts might have occurred. Then, the sEMG signals were rectified in the time domain, meaning that all the negative values were multiplied by -1. Last, the sEMG signals were transformed into the frequency domain, where excessive spikes were again observed at the vibration frequency and its multiple harmonics (**Figure 3.4B**). These new spikes could only originate from motor unit synchronization, as motion artifacts, unlike stretch reflex responses, are strictly sinusoidal. Although the method introduced by Sebik et al. (2013) was useful to prove motor unit synchronization during WBV, its feasibility as a method to delete the motion artifacts is questionable, as a band-pass filter between 80 - 500 Hz rejects a large number of myoelectrical information (see **CHAPTER 4**).

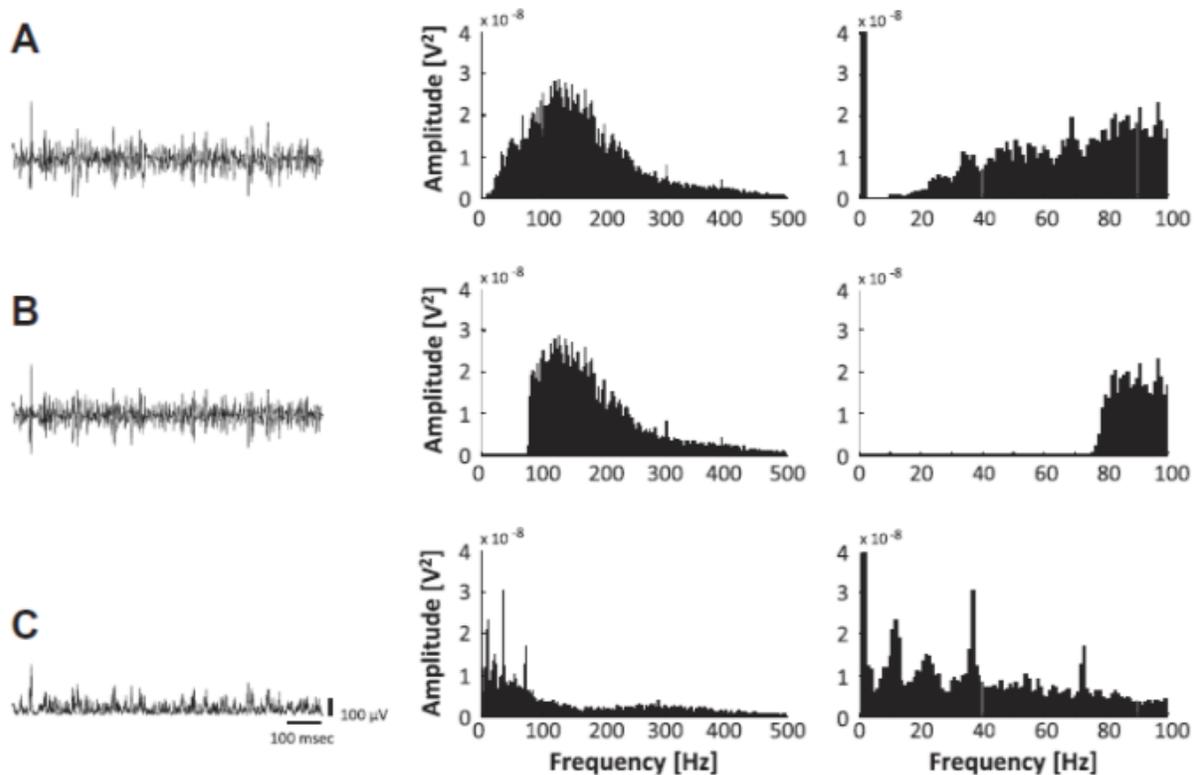


Figure 3.4 Representative sEMG signals and frequency spectrograms recorded from the Gastrocnemius Medialis during whole-body vibration at 35 Hz. Shown in columns: raw sEMG signal; frequency spectrogram; magnified frequency spectrogram in the 0-100 Hz range. (A) Unprocessed sEMG signal; (B) band-pass (80-500 Hz) filtered sEMG signal; (C) the filtered sEMG signal after rectification, showing dominant peaks at around 35 Hz and 70 Hz due to synchronization.

(Reprinted from Sebik et al. 2013)

While it is strongly assumed that stretch reflex responses as well as motion artifacts are contained in the sEMG signal during WBV, it is important to identify their contribution in the spikes. Even more so it is surprising that to date no scientific study has analyzed the content of the isolated spikes.

3.3 METHODOLOGICAL ISSUES IN WBV STUDIES

3.3.1 sEMG PROCESSING

As it is unclear if the spikes observed in the sEMG spectrum contain mostly motion artifacts or stretch reflex responses, different strategies have been used to cope with these spikes. Researchers who have assumed that these spikes are mainly stretch reflex responses have been including these spikes in the calculation of the sEMG magnitude (Cardinale and Lim 2003; Cochrane et al. 2009; Di Giminiani et al. 2013; Krol et al. 2011; Perchthaler et al. 2013; Ritzmann et al. 2013; Roelants et al. 2006). Conversely, researchers who have speculated that these spikes are mainly motion artifacts have been withdrawing these spikes and have been excluding them from the calculation of the sEMG magnitude (Abercromby et al. 2007a; Fratini et al. 2009a; Hazell et al. 2007; 2010; Marín et al. 2012a; 2012b; Pollock et al. 2010). Besides the decision whether these spikes should be deleted or not, it seems that researchers have been given the choice to select the appropriate filtering regime. Hence, different methods have been used including a band-stop filter centered at the vibration frequency and its multiple harmonics (Abercromby et al. 2007a; Fratini et al. 2009a; Marín et al. 2012a; 2012b, Pollock et al. 2010) and a band-pass filter rejecting the entire spectrum where the spikes might have occurred (Hazell et al. 2007; 2010). As a consequence, the WBV research studies that have been published used either no filter, a band-stop filter, or a band-pass filter to delete the excessive spikes. Unfortunately, to date it is unknown if the magnitude of the muscle activity is comparable between studies using different filtering regimes. Neither the band-stop nor the band-pass filter has been validated, which means that their respective effect on the quantification of the sEMG magnitude is unknown. Therefore, it is absolutely necessary to analyze how the method of choice influences the magnitude of the sEMG during WBV.

Another draw-back of never having validated the different sEMG processing methods is that none of the methods might be the optimal choice to filter the excessive spikes. For example, using a band-pass filter with a low cut-off frequency of 100 Hz removes the entire sEMG spectrum below that frequency. Hence, although the motion artifacts might be deleted,

important myoelectrical information is lost over a wide range of the frequency spectrum. On the other hand, using a band-stop filter targets the specific frequencies that are contaminated, yet more sEMG activity is deleted than necessary. This can be seen on page 25 in **Figure 3.3** originally published by Abercromby et al. (2007a). The dotted line represents the raw sEMG signal in the frequency domain, whereas the solid line represents the band-stop filtered sEMG signal. Evidently, the band-stop filter produces notches in the sEMG spectrum which have no physiological background. These notches are a byproduct of using a band-stop filter and might be a severe draw-back. Therefore it would be of great benefit to develop a new processing method that only deletes the excessive spikes without adding notches to the spectrum.

3.3.2 sEMG NORMALIZATION

After filtering the sEMG signal and calculating the RMS, the next step is typically normalization of the RMS of the sEMG ($sEMG_{RMS}$). In WBV studies, the $sEMG_{RMS}$ has been expressed as a percentage of the baseline $sEMG_{RMS}$ during standing (%standing, Marín et al. 2012a; Ritzmann et al. 2013; Tankisheva et al. 2013), or as a percentage of the $sEMG_{RMS}$ during a maximal isometric voluntary contraction (%MVC, Hazell et al. 2007; 2010; Perchthaler et al. 2013; Pollock et al. 2010; Roelants et al. 2006). Surprisingly, although normalization of the $sEMG_{RMS}$ is important for comparison between subjects and muscles (Burden 2010), this step has been skipped in numerous studies (Abercromby et al. 2007a; Avelar et al. 2013; Cardinale and Lim 2003; Cochrane et al. 2009; Di Giminiani et al. 2013; Krol et al. 2011; Marín et al. 2009; 2012b) and the $sEMG_{RMS}$ was reported and compared in Volts (V).

Taken as a whole, measuring the sEMG during WBV is confronted with several methodological issues. First, it is unclear if the excessive spikes in the sEMG spectrum need to be deleted. Second, if they need to be removed, which filter is the most adequate method to do so? Third, it is unknown if the sEMG normalization procedure has an influence of the outcomes of the study. Therefore, before further investigating the instantaneous effects of WBV on muscle activity, these limitations should be addressed and a standardized methodology should be implemented.

3.4 EFFECT OF WBV ON MUSCLE ACTIVITY

The effect of WBV on lower limb muscle activity has been investigated in numerous studies. **Table 3.1** summarizes the most important findings of these studies and the WBV parameters that have been used. Comparing the sEMG activity during WBV to the same exercise without vibration, most studies have found significantly higher sEMG during the vibration. This positive effect of WBV on muscle activity has been reported performing static and dynamic exercises (Abercromby et al. 2007a; Hazell et al. 2007) with various knee flexion angles (Ritzmann et al. 2013; Roelants et al. 2006), using a side-alternating or synchronous vertical vibration platform (Abercromby et al. 2007a; Ritzmann et al. 2013), using a frequency range of 6 to 55 Hz and an amplitude range of 0.5 to 5 mm (Abercromby et al. 2007a; Cardinale and Lim 2003; Cochrane et al. 2010; Di Giminiani et al. 2013; Hazell et al. 2007; 2012; Krol et al. 2011; Marín et al. 2009; 2012a; Ritzmann et al. 2013; Roelants et al. 2006; Tankisheva et al. 2013). However, some authors have found that the effectiveness of WBV can vary between muscles (Cochrane et al. 2009; Hazell et al. 2010), the vibration frequency (Di Giminiani et al. 2013; Hazell et al. 2007), the amplitude of the platform (Marín et al. 2009), or the combination of the two latter parameters (Hazell et al. 2007; Krol et al. 2011). For example, vibration frequencies ranging from 25 to 55 Hz have been found to significantly increase muscle activity during WBV (Abercromby et al. 2007a; Cardinale and Lim 2003; Di Giminiani et al. 2013; Krol et al. 2011; Marín et al. 2009; 2012b; Roelants et al. 2006) in combination with amplitudes ranging from 0.5 to 5 mm. On the contrary, no increases in muscle activity have been reported for vibration frequencies below 25 Hz in combination with amplitudes between 0.5 and 2 mm (Di Giminiani et al. 2013; Krol et al. 2011;). The exception to that is the study by Cochrane et al. (Cochrane et al. 2009), where significant increases in sEMG activity were found with 6 Hz and 3.1 mm amplitude WBV in the VL and GM. Certainly, this is only one example where contradictory findings have been reported. It is difficult to seek out the responsible source, as these studies diverse in a number of parameters. Besides using different frequencies and amplitudes, different vibration platform types, exercise modalities, body postures, and sEMG processing and normalization methods have been used. This makes it difficult to draw general conclusions about the effectiveness of WBV exercises.

Table 3.1 Summary of the scientific studies that have investigated the effects of whole-body vibration (WBV) on lower limb muscle activity. Note that the amplitude (Amp) is the displacement from baseline to peak.

Filter	Authors and year	Subjects	Platform type	Freq	Amp	Exercise modality	Knee angle	Measured Muscles	sEMG magnitude	Results WBV compared to baseline	Results Comparison of parameters	
Band-stop	Abercromby et al. (2007a)	9 males 7 females (32.7 yrs)	RV VV	30 Hz	2 mm	Static	18.5°	VL BF GL TA	sEMG _{RMS} (V)	sEMG VL, BF, GL, TA: Significant increases for both platform types	Platform type: sEMG VL, GL: RV > VV sEMG TA: VV > RV	
						Dynamic	10-35°				sEMG VL, BF, GL, TA: Significant increases (RV) sEMG GL, TA: Significant increases (VV)	Exercise type: sEMG VL, BF, GL, TA: Static > dynamic
												Knee angle: sEMG VL, GL, TA: Highest increases for smallest knee angles
Band-stop	Avelar et al. (2013)	34 males (26.3 yrs)	VV	30 Hz	4 mm	Static	60° 90°	VL	sEMG _{RMS} (V)	sEMG VL: No significant increases for both knee angles	Knee angle: sEMG VL: 90° > 60°	
No filter	Cardinale et al. (2003)	16 females (23.5 yrs)	VV	30 Hz 40 Hz 50 Hz	5 mm	Static	80°	VL	sEMG _{RMS} (V)	sEMG VL: Significant increases for all the measured frequencies	Frequency: sEMG VL: 30 Hz > 50 Hz 40 Hz > 50 Hz	
No filter	Cochrane et al. (2009)	9 males (33.4 yrs)	RV	6 Hz	3.1 mm	Static	18°	VL GM SOL TA	Mean rectified amp (V)	sEMG VL and GM: Significant increases		
No filter	Di Giminiani et al. (2013)	20 males (20 yrs)	VV	20 Hz 25 Hz 30 Hz 35 Hz 40 Hz 45 Hz 50 Hz 55 Hz	0.5 mm	Static	60° 90°	VL GL	sEMG _{RMS} (V)	sEMG VL and GL: Significant increases for frequencies 25 – 55 Hz	Frequency: sEMG VL: 55, 50 Hz > 40, 35, 30, 25, 20 Hz 40 Hz > 25, 20 Hz 35, 30, 35 Hz > 20 Hz sEMG GL: 55, 50, 45, 40, 35, 30, 25 Hz > 20 Hz 55, 50, 45 Hz > 35 Hz 55, 50, 45, 40 Hz > 30 Hz 30 Hz > 25 Hz Knee angle: sEMG VL: 90° > 60°	

Filter	Authors and year	Subjects	Platform type	Freq	Amp	Exercise modality	Knee angle	Measured Muscles	sEMG magnitude	Results WBV compared to baseline	Results Comparison of parameters
Band-pass	Hazell et al. (2007)	10 males (24.4 yrs)	VV	25 Hz 30 Hz 35 Hz 40 Hz 45 Hz	2 mm 4 mm	Static	60°	VL BF	sEMG _{RMS} (%MVC)	<u>sEMG VL:</u> Significant increases at 35, 40, 45 Hz and both amplitudes <u>sEMG BF:</u> Significant increases at 35 Hz (4 mm), 40 and 45 Hz (both amplitudes)	Frequency: <u>sEMG VL:</u> 40, 45 Hz > 25, 30 Hz <u>sEMG BF:</u> 45 Hz > 25 Hz Amplitude: <u>sEMG BF:</u> 4 mm > 2 mm
						Dynamic	0° - 60°			<u>sEMG VL:</u> Significant increases for all conditions <u>sEMG BF:</u> Significant increases at 35, 45 Hz (4 mm)	Frequency: <u>sEMG VL:</u> 35, 40, 45 Hz > 25 Hz Amplitude: <u>sEMG VL, BF:</u> No difference between 2 mm and 4 mm
Band-pass	Hazell et al. (2010)	13 males (23 yrs)	VV	25 Hz 35 Hz 45 Hz	4 mm	Dynamic (without load and with load equal to 30% of body mass)	20° - 90°	VL BF GM TA	sEMG _{RMS} (%MVC)	<u>sEMG VL:</u> Significant increases at 45 Hz (with load) <u>sEMG BF:</u> Significant increases at 45 Hz (both loads) and 35 Hz (with load) <u>sEMG GM:</u> Significant increases at 45 Hz (both loads); 35, 25 Hz (with load) <u>sEMG TA:</u> Significant increases at 45 Hz (with load)	Frequency: <u>sEMG VL:</u> 45 Hz > 25 Hz (with load) <u>sEMG BF:</u> 45 Hz > 25 Hz (both loads) 45 Hz > 35 Hz (with load) <u>sEMG GM:</u> 45 Hz > 25 Hz (no load) <u>sEMG TA:</u> 45 Hz > 35 Hz (no load) Load: <u>sEMG VL, BF, GM:</u> with load > no load (all conditions) <u>sEMG TA:</u> no load > with load (all conditions except 35 Hz)
No filter	Krol et al. (2011)	29 females (21.8 yrs)	VV	20 Hz 40 Hz 60 Hz	2 mm 4 mm	Static	90°	VL VM	sEMG _{RMS} (V)	<u>sEMG VL and VM:</u> Significant increases at 20 Hz (4 mm), 40 and 60 Hz (both amplitudes)	Frequency: <u>sEMG VL:</u> 60 Hz > 40 Hz > 20 Hz (both amplitudes) <u>sEMG VM:</u> 60 Hz > 40 Hz > 20 Hz (2 mm) 60 Hz > 20 Hz (4 mm) Amplitude: <u>sEMG VL, VM:</u> 4 mm > 2 mm (all frequencies)
No filter	Marín et al. (2009)	10 males (28.7 yrs)	VV	30 Hz	2 mm 4 mm	Static	80°	VL GM	sEMG _{RMS} (V)	<u>sEMG VL and GM:</u> Significant increases with the 4 mm amplitude	Amplitude: <u>sEMG VL, GM:</u> No difference between 2 mm and 4 mm
Band-stop	Marín et al. (2012a)	3 males 17 females (79.6 yrs)	VV	25 Hz 35 Hz 45 Hz	1.55 mm 0.5 mm	Static	30°	VM VL BF GM	sEMG _{RMS} (%standing)	Frequency: <u>sEMG of all muscles averaged:</u> 45 Hz > 25 Hz (1.55 mm) Amplitude: <u>sEMG of all muscles averaged:</u> 1.55 mm > 0.5 mm (35, 45 Hz)	

Filter	Authors and year	Subjects	Platform type	Freq	Amp	Exercise modality	Knee angle	Measured Muscles	sEMG magnitude	Results WBV compared to baseline	Results Comparison of parameters
Band-stop	Marín et al. (2012b)	5 males 9 females (74.8 yrs)	VV	30 Hz 46 Hz	1.25 mm 0.55 mm	Static	30°	VL GM	sEMG _{RMS} (V)	<u>sEMG VL and GM:</u> Significant increases for both WBV conditions	
No filter	Perchthaler et al. (2013)	17 males 34 females (55 yrs)	RV	6 Hz 12 Hz 18 Hz 24 Hz 30 Hz	1.3 mm 2.6 mm 3.9 mm	Static	30° 45° 60°	RF VL VM BF ST	sEMG _{RMS} (%MVC)	Frequency: <u>sEMG RF, VL, VM averaged:</u> 30 Hz > 12 and 6 Hz <u>sEMG BF and ST averaged:</u> 30 Hz > 6 Hz Amplitude: <u>sEMG RF, VL, VM averaged:</u> 3.9 and 2.6 mm > 1.3 mm <u>sEMG BF and ST averaged:</u> 3.9 mm > 2.6 mm > 1.3 mm Knee angle: <u>sEMG RF, VL, VM averaged:</u> 60° and 45° > 30° <u>sEMG BF and ST averaged:</u> 60° > 45° > 30°	
Band-stop	Pollock et al. (2010)	12 subjects (31.3 yrs)	RV	5 Hz 10 Hz 15 Hz 20 Hz 25 Hz 30 Hz	5.5 mm 2.5 mm	Static	15°	SOL GL TA RF BF GM	sEMG (%MVC)	Frequency: <u>sEMG SOL, GL, TA, BF:</u> Increasing with increasing frequency Amplitude: <u>sEMG GL, TA:</u> 5.5 mm > 2.5 mm (all frequencies) <u>sEMG SOL, RF, BF:</u> 5.5 mm > 2.5 mm (most frequencies)	
No filter	Ritzmann et al. (2013)	10 males 8 females (25 yrs)	RV VV	30 Hz	2 mm	Static (without load and with load equal to 1/3 of body mass)	10° 30° 60°	SOL GM TA BF VM RF	iEMG (%standing)	<u>sEMG of all muscles:</u> Significant increases averaged for all knee flexion angles and both platform types	Platform type: <u>sEMG of all muscles:</u> RV > VV Frequency: <u>sEMG of all muscles:</u> Increasing with increasing frequency Knee angle: <u>sEMG RF, VM, TA:</u> 60° > 30° > 10° <u>sEMG GM, SOL:</u> 10° > 30° > 60° Load: <u>sEMG SOL, GM, TA, VM, RF:</u> with load > no load

Filter	Authors and year	Subjects	Platform type	Freq	Amp	Exercise modality	Knee angle	Measured Muscles	sEMG magnitude	Results WBV compared to baseline	Results Comparison of parameters
No filter	Roelants et al. (2006)	15 males (21.2 yrs)	VV	35 Hz	2.5 mm	Static	55° 90°	RF VL VM GM	sEMG _{RMS} (%MVC)	<u>sEMG of all muscles:</u> Significant increases for both knee angles	Knee angle: <u>sEMG of all muscles:</u> No difference between 55° and 90°
Band-stop	Tankisheva et al. (2013)	5 males 3 females (28.7 yrs)	VV VV VV	30 Hz 35 Hz 40 Hz 50 Hz	0.010 mm 0.015 mm 0.03 mm 0.10 mm 0.06 mm 0.12 mm	Static	0° 45° 70°	SOL TA VM RF BF GMAX GMED	sEMG _{RMS} (%standing)	<u>sEMG SOL, VM, BF, GMAX:</u> Significant increases in sEMG _{RMS} for most conditions	Knee angle: <u>sEMG of all muscles:</u> No difference between 0°, 45° and 70°

Amp, Amplitude; BF, Biceps Femoris; Freq, Frequency; GL, Gastrocnemius Lateralis; GM, Gastrocnemius Medialis; GMAX, Gluteus Maximum; GMED, Gluteus Medialis; MVC, Maximal voluntary contraction; RMS, Root mean square; RF, Rectus Femoris; RV, side-alternating platform; sEMG, surface electromyography; SOL, Soleus; ST, Semitendinosus; TA, Tibialis Anterior; V, Volts; VL, Vastus Lateralis; VM, Vastus Medialis; VV, vertical synchronous platform; WBV, Whole-body vibration; yrs, years.

3.5 EFFECT OF WBV PARAMETERS ON MUSCLE ACTIVITY

The magnitude of the muscle activity during WBV is dependent on many factors. First, the intensity of the WBV stimulus is determined by the magnitude of the acceleration (Mester et al. 1999). The extent of the acceleration can be adjusted by means of the **frequency** and the **amplitude** of the platform displacement. Second, the three-dimensional direction of the acceleration transmitted to the WBV user varies between different **platform types** such as the vertical synchronous, side-alternating, or horizontally vibrating platform (Pel et al. 2009). Third, the WBV stimulus can be further manipulated by changing the conditions of the WBV user by for example using **additional loads** or changing the **knee flexion angle**. The effect of these parameters on muscle activity of the lower limbs during WBV is discussed in the following.

3.5.1 FREQUENCY

The effect of frequency on muscle activity during WBV is the most investigated parameter to date. Most studies have found that the frequency is positively correlated with the sEMG activity of the lower limb muscles (Di Giminiani et al. 2013; Hazell et al. 2007; Krol et al. 2011; Marín et al. 2012a; Perchthaler et al. 2013; Pollock et al. 2010; Ritzmann et al. 2013). Hazell et al. (2007) found a positive frequency effect in the VL and Biceps Femoris (BF) between frequencies of 25 to 45 Hz. In other studies, frequency-dependent increases in muscle activity were reported for lower frequencies, i.e. in the Rectus Femoris (RF), Vastus Medialis (VM), VL, BF, and Semitendinosus from 6 to 30 Hz (Perchthaler et al. 2013), in the BF, TA, Gastrocnemius Lateralis (GL) and SOL from 5 to 30 Hz (Pollock et al. 2010), and in the SOL, GM, TA, BF, VM and RF from 5 to 30 Hz (Ritzmann et al. 2013). Again other studies found a frequency-dependent effect over a wider range of frequencies, with reports of significant increases in VL activity from 20 to 55 Hz (Di Giminiani et al. 2013) and significant increases in VL and VM activity from 20 to 40 to 60 Hz (Krol et al. 2011). However, there are two studies which have reported contradictory findings. Cardinale et al. (2003) measured sEMG activity of the VL during WBV at 30, 40 and 50 Hz and demonstrated that the sEMG was the highest at 30 Hz and the lowest at 50 Hz. These findings have been confirmed for the GL (Di Giminiani et al. 2013), with higher sEMG at 30 Hz than at 40 to 55 Hz. Interestingly, both studies used a vertical synchronous vibration platform and did not

remove the spikes visible in the sEMG spectrum. Motion artifacts during WBV are related to the acceleration magnitude of the underlying soft tissue (Fratini et al. 2009b). Interestingly, the acceleration measured on the soft tissue of the shank and the thigh is the greatest around 30 Hz (Cook et al. 2011). Therefore, it is possible that Cardinale et al. (2003) and Di Giminiani et al. (2013) found the highest sEMG activity at 30 Hz because of the influence of the motion artifacts. However, another study (Krol et al. 2011) that used the same platform type and the same sEMG processing regime (i.e., no deletion of the spikes, calculation of the RMS using 100 ms moving time window) observed contrary results, i.e. increasing muscle activity with an increasing frequency from 20 to 40 Hz and from 40 to 60 Hz.

3.5.2 AMPLITUDE

Before discussing the effect of the platform amplitude, it needs to be emphasized that the magnitude of the platform displacement has been reported in two different ways. Some researchers have reported the peak-to-peak displacement of the platform, whereas others have referred to the amplitude (displacement from baseline to peak) of the platform (**Figure 3.5**).

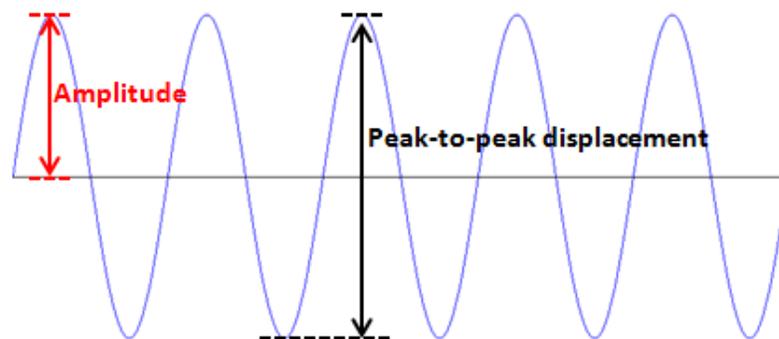


Figure 3.5 Illustration of the two notations that are currently being used to report the displacement of the platform. The amplitude (displacement from baseline to peak) is represented in red; the peak-to-peak displacement is demonstrated in black.

While both methods are valid, problems occur when it is not clear whether the amplitude or the peak-to-peak displacement was reported. Furthermore, numerous studies have reported the platform displacement without actually measuring it, fully trusting in the specifications made by the manufacturers. This can lead to errors. In this thesis, it is referred to the amplitude unless other stated. Hence, for the literature where the peak-to-peak displacement (*Disp*) was reported, the values were transformed to the corresponding amplitude (*Amp*) using the formula

$$Amp = \frac{1}{2} \times Disp.$$

The effect of amplitude on muscle activity has been investigated in numerous studies, reporting conflicting results. Regarding the knee extensor muscles, significant higher muscle activity was found with a 4 mm compared to a 2 mm amplitude in the VL and VM (Krol et al. 2011). Similar results have been reported for the quadriceps muscles in another study (Perchthaler et al. 2013); sEMG activity was significantly increased with increasing amplitude from 1.3 to 2.6 to 3.9 mm. Further, Pollock et al. (2010) found significantly higher muscle activity of the RF with an amplitude of 5.5 mm than 2.5 mm. In contrast, no significant amplitude-dependent effect has been reported in the VL between amplitudes of 2 and 4 mm (Hazell et al. 2007; Marín et al. 2009), and between an amplitude-frequency combination of 30 Hz – 1.25 mm and 46 Hz – 0.55 mm (Marín et al. 2012b). Concerning the knee flexor muscles, consistent results have been demonstrated. Significantly higher sEMG activity was found in the BF using a high amplitude compared to a low amplitude, i.e. 4 mm vs. 2 mm (Hazell et al. 2007) and 5.5 mm vs. 2.5 mm (Pollock et al. 2010). Additionally, Perchthaler et al. (2013) found increasing muscle activity of the hamstrings with increasing amplitude from 1.3 to 2.6 to 3.9 mm. The effect of amplitude on the plantar flexors is controversial, as again inconsistent results have been found. Significant higher sEMG activity for the higher amplitude (5.5 mm vs. 2.5 mm) has been demonstrated in the GM and SOL (Pollock et al. 2010), while other studies found no significant difference between amplitudes in the GM (Marín et al. 2009; 2012b). The effect of amplitude on the TA has been analyzed in only one study thus far (Pollock et al. 2010), with reports of significantly higher sEMG activity for the 5.5 than the 2.5 mm amplitude.

3.5.3 PLATFORM TYPE

To date, there are only two studies investigating the effect of platform type on muscle activity of the lower limbs (Abercromby et al. 2007a; Ritzmann et al. 2013). The platform types that have been compared are the side-alternating type that generates vibrations by rotating along the sagittal axis and the vertical synchronous type that moves synchronously in the vertical direction (Figure 3.6).

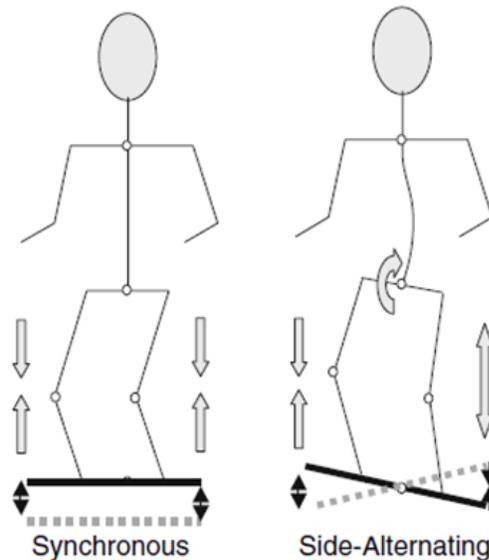


Figure 3.6 Illustration of the vertical synchronous and the side-alternating whole-body vibration platform types. In the synchronous mode, the entire platform moves up and down, with a result that both legs move in phase. In the side-alternating mode, the right and left leg operate in anti-phase.

(Reprinted from Rittweger 2010)

These two platform types have been compared during WBV at 30 Hz and 2 mm amplitude (Abercromby et al. 2007a; Ritzmann et al. 2013). Ritzmann et al. (2013) reported that muscle activity of selected lower limb muscles (i.e. SOL, GM, TA, BF, VM, RF) was significantly higher during side-alternating than vertical synchronous WBV. For the side-alternating platform, the amplitude of the displacement is dependent on the distance between the feet and the central

axis. In order to match the amplitudes between the two platform types, the distance from the center to the feet was chosen corresponding to an amplitude of 2 mm. Nevertheless, Ritzmann et al. (2013) reported that the measured amplitude during synchronous WBV was significantly lower than during side-alternating WBV. As this difference was very small (5%), it can be assumed that their results are valid nonetheless. A more severe limitation to the mentioned study might be the lacking processing method to delete the excessive spikes in the sEMG spectrum. It is well known that sEMG signals during WBV can be contaminated by motion artifacts (Fratini et al. 2009a; Sebik et al. 2013), and that their magnitude is related to the magnitude of the vibrations of the underlying soft tissue (Fratini et al. 2009b). As the accelerations that the soft tissue experiences might vary between the two platform types, it is possible that the observed difference in sEMG magnitude is partly due to differences in motion artifacts. However, another study (Abercromby et al. 2007a) which accounted for these limitations has reported similar results. They found significantly higher muscle activity of the VL and GL during side-alternating WBV, and significantly higher muscle activity of the TA during vertical synchronous WBV. It needs to be considered that the two different platform types may not deliver equivalent acceleration magnitudes in the three different planes (Pel et al. 2009). Thus, each platform type could lead to muscle-specific adaptations depending on the anatomical function of the respective muscle. However, although many different types of WBV platforms are available on the market, it is not clear which type induces the greatest enhancement of muscle activity. As an example, there is no study thus far investigating the effect of a horizontal vibration platforms or dual mode platforms on muscle activity of the lower limbs.

3.5.4 LOAD

The effect of additional loading on muscle activity during WBV has been analyzed in only three studies. Ritzmann et al. (2013) reported that using an additional load equal to 1/3 of the body mass significantly increased muscle activity in the SOL, GM, RF, VM and TA. No changes in muscle activity were observed in the BF. Hazell et al. (2010) found that with a load corresponding to 30% of the body mass, muscle activity in the VL, BF, and GM was significantly

enhanced, and in the TA significantly decreased. The inconsistent findings between the two studies concerning the BF and TA might be attributed to the use of different platform types and exercise modalities, as Ritzmann et al. (2013) assessed static squats on a side-alternating platform type and Hazell et al. (2010) dynamic squats on a synchronous platform type. However, their findings are trivial, as the changes in muscle activity were mainly attributed to the addition of load, and not to an interaction between load and WBV. It would be of higher interest to identify if the use of an additional load has an effect on the increase in muscle activity as an adjustment to WBV. Hazell et al. (2010) investigated the effect between additional loading and muscle activity during WBV, and found no significant interaction effect [load x WBV] in the VL, BF, and GM. A significant interaction effect was found in the TA, which means that the muscles were affected to a higher degree by WBV with the additional load. It is important to mention that although the muscle activity of the TA increased more with the additional load, the background muscle activity of the TA was still lower with the additional load. However, such an effect of additional loading has been observed in another study (Zaidell et al. 2013). In the SOL, an effect of frequency on muscle activity was found when the leg was passively loaded with approximately 45% of the body mass, but no frequency-dependent increases were observed without the load. Contrary findings were reported in the TA, as an effect of frequency on muscle activity was only found without the use of an additional load. While it is unclear how additional loading manipulates the effect of vibration on muscle activity, it might be a successful method to further enhance muscle activity during WBV. As previously described, the TVR is positively correlated with the background muscle activity. Using additional loads likely elevates background muscle activity of some lower limb muscles which might further enhance the TVR during WBV. This could then lead to higher increases in sEMG and maximize the neuromuscular adaptations to this form of exercise.

3.5.5 KNEE FLEXION ANGLE

Before discussing the outcomes of the studies that have measured the effect of knee flexion on muscle activity levels, it is important to point out that the knee flexion angle has been reported in two different manners; (1) the effective angle between the hamstrings and the calf muscles

and (2) the change of flexion with 0° corresponding to full extension. In this document, knee flexion angles are described using the latter method, thus always referring to the change of flexion as compared to full extension. Therefore, degrees of knee flexion angles from studies using method (1) were transformed. In an early study conducted by Roelants et al. (2006), no significant changes in muscle activity of the RF, VM, VL, and GM were observed between knee flexion angles of 55° and 90°. Contrary to that, Di Giminiani et al. (2013) and Avelar et al. (2013) found that increasing the knee flexion angle from 60° to 90° results in increased sEMG activity of the VL. Further, Ritzmann et al. (2013) reported that increasing the knee flexion angle from 10° to 30° to 60° increased muscle activity of the RF, VM, and TA, and decreased muscle activity of the GM and SOL. Additionally, Perchthaler et al. (2013) demonstrated increased quadriceps and hamstring muscle activity by augmenting the knee flexion angle from 30° to 45° to 60°. However, increasing the knee flexion angle amplifies the lever arm and the knee joint torque. According to the muscles' anatomical functions, the muscles of the upper leg and the dorsi flexors need to generate higher forces in order to maintain the posture (Pincivero et al. 2003), and vice versa for the plantar flexors. Therefore, the use of comparing sEMG activity between knee flexion angles during WBV is questionable, as the changes in muscle activity are rather posture-related than an adjustment to vibration. Since the TVR is positively correlated to the stretch of the muscle, it could be more interesting to verify whether changing the length of the muscle, i.e. changing the knee flexion angle, results in higher increases in muscle activity as a response to WBV. Thus far, one study (Abercromby et al. 2007a) has tested this hypothesis with opposite results. The greatest increases in muscle activity were found for the smallest knee flexion angles. As this study was conducted under dynamic squatting conditions, a similar set-up needs to be repeated using static squat exercises.

3.6 ACCELERATION TRANSMISSION DURING WBV

Vibration transmission during WBV is frequently discussed, as high accelerations can be detrimental for the human body (ISO 2631-1 1997). Acceleration transmission is typically analyzed by reporting transmissibility values, which reflect the amount of platform acceleration that travels up the body. Transmissibility is calculated by dividing the acceleration magnitude measured at different locations on the body by the magnitude of the platform acceleration. Transmissibility values greater than one indicate amplification of the vibrations, whereas transmissibility values lower than one indicate vibration damping. The general findings in WBV studies are that transmissibility decreases with increasing distance from the platform (Bressel et al. 2010; Crewther et al. 2004; Kiiski et al. 2008; Pel et al. 2009; Pollock et al. 2010; Tankisheva et al. 2013). This means that the accelerations are the highest at the foot and the lowest at the head. **Figure 3.7** illustrates transmissibility values measured on several bony structures using a wide range of vibration frequencies (Kiiski et al. 2008).

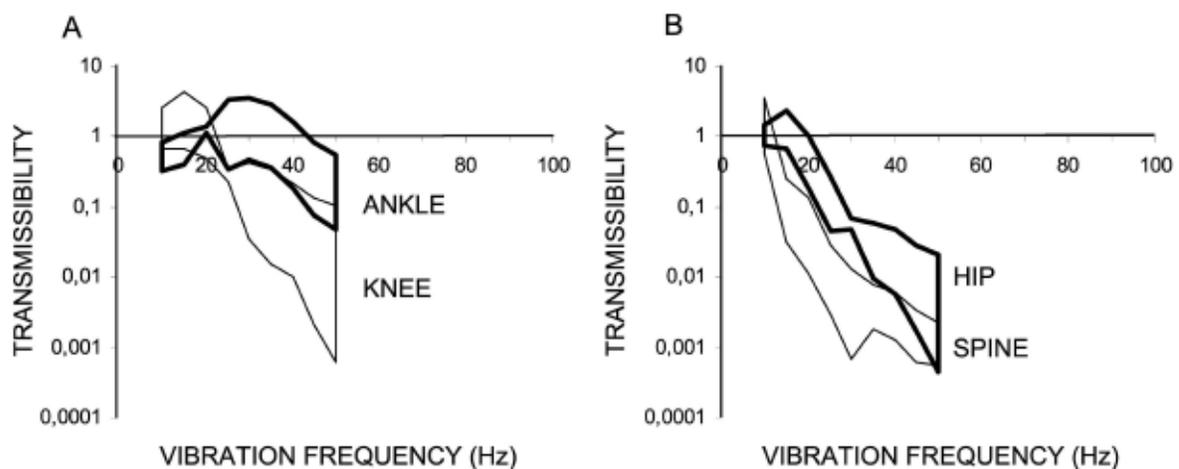


Figure 3.7 Transmissibility during whole-body vibration at various frequencies to (A) the ankle and the knee, and (B) to the hip and the spine. The applied amplitude was 0.5 mm on a synchronous vertical vibration platform.

(Reprinted from Kiiski et al. 2008)

3.6.1 AMPLIFICATION OF VIBRATIONS

In some occasions, amplification of the accelerations has been reported. Amplification of accelerations is possible when resonance occurs, i.e. when the input frequency is close to the natural frequency of the body (Wakeling and Nigg 2001). Harazin and Grzesik (1998) determined the resonance frequency during WBV of several bony structures throughout the body. Their findings are summarized in **Table 3.2**, indicating around which vibration frequencies amplification of the accelerations can be expected. As a matter of fact, amplification around the indicated frequency range in Table 3.2 was found in following studies at the ankle (Bressel et al. 2010; Kiiski et al. 2008; Pollock et al. 2010; Tankisheva et al. 2013), knee (Kiiski et al. 2008; Pollock et al. 2010), and hip (Kiiski et al. 2008; Rubin et al. 2003).

Table 3.2 Resonance frequencies of several bony structures during WBV exercises.

(Reprinted form Harazin and Grzesik 1998).

Structure	Resonance frequency
Metatarsus	4-8 Hz, 12.5 Hz, 31.5 – 125 Hz
Ankle	4-8 Hz, 12.5 Hz, 25 – 63 Hz
Knee	4-8 Hz, 12.5 – 25 Hz
Hip	4 – 10 Hz
Shoulder	4 – 8 Hz, 12.5 Hz
Head	4 – 6.3 Hz, 12.5 Hz

3.6.2 VIBRATION DAMPING

Acceleration damping during WBV has been found in several studies. In summary, occurrence of damping was found on the ankle with a transmissibility of 0.8 (Pel et al. 2009), on the knee with a transmissibility between 0.5 and 0.8 (Muir et al. 2013; Pel et al. 2009), on the hip with a transmissibility between 0.2 and 0.8 (Rubin et al. 2003; Tankisheva et al. 2013), and on the head with a transmissibility lower than 0.3 (Bressel et al. 2010; Muir et al. 2013; Tankisheva et al. 2013).

It is speculated that the body aims to minimize vibrations that could lead to discomfort by damping (Nigg 1997). Vibrations can be induced to the tissues through impacts, such as during running, or in a continuous manner, such as during WBV. As previously mentioned, if the input vibration frequency coincides with the natural frequency of the tissue, the amplitude of the tissue vibration will greatly increase due to resonance (Boyer and Nigg 2006; Cardinale and Wakeling 2005). The effects of resonance can be reduced by changing either the natural frequency or the damping characteristics of the tissue, both of which are possible through increased muscle activation (Wakeling and Nigg 2001b; Wakeling et al. 2002). Furthermore, it has been shown that during WBV, increased muscle activity and increased damping of the vibration power occurred when the input frequency was close to the natural frequency of the muscle (Wakeling et al. 2002). Whether or not this damping mechanism is based on actual stretch reflex responses, increased muscle activity is one of the body's strategies to attenuate the vibrations.

Another mechanism that could influence damping is proprioceptive acuity. Individuals with superior proprioception, such as competitive athletes, are expected to stabilize their ankle and knee joints better than individuals with a relatively poor proprioception (Courtney et al. 2013; Muaidi et al. 2009). Also, it can be assumed that highly trained individuals are able to stabilize their joints more due to greater forces of the muscles crossing these joints as compared to poorly trained or older individuals. Superior proprioceptive acuity and greater muscle forces would presumably both result in better stabilization of the ankle and knee joints. This would

result in a better transmission of the vibrations, as the joints are less able to act as a spring suspension system to damp the vibrations (Nikooyan and Zadpoor 2011).

Furthermore, limb stiffness through passive components such as tendon and connective tissue (Pearson and McMahon 2012) can affect acceleration transmission. It can be expected that the higher the musculotendinous and musculoarticular stiffness, the more vibrations are transmitted through the body. This would mean that populations presenting high limb stiffness would transmit the vibrations more (Rabita et al. 2008). This could be the case for highly trained athletes as regular exercising results in leg stiffening (Pearson and McMahon 2012). Last, soft tissue properties can affect acceleration transmission, i.e., the higher the amount of subcutaneous fat, the better the vibrations are being damped (Wakeling and Nigg 2001).

3.7 EFFECT OF WBV PARAMETERS ON ACCELERATION TRANSMISSION

Table 3.3 provides a summary of the outcomes concerning acceleration transmission during WBV exercises. To which extent the accelerations are amplified or damped depends on many factors such as the vibration platform type, the input frequency, knee flexion angle and possibly the characteristics of the WBV user. The findings in previous studies of how these parameters affect vibration transmission are discussed in the following.

3.7.1 PLATFORM

Transmission of acceleration between different platform types has been investigated in one study thus far (Abercromby et al. 2007b). Comparison between a vertical synchronous and a side-alternating vibration platform during WBV at 30 Hz and 2 mm amplitude showed that the accelerations measured at the head were significantly higher during synchronous than during side-alternating WBV. In fact, the energy to the head was 189% greater during synchronous WBV. This might be attributed to damping the vibrations by rotation of the pelvis during side-alternating WBV, because of the alternating upward forces being applied to the feet (see **Figure 3.6** on page 38).

3.7.2 FREQUENCY

Comparison of acceleration transmission for different vibration frequencies has been conducted in several scientific studies (Caryn et al. 2014; Crewther et al. 2004; Kiiski et al. 2008; Pollock et al. 2010). Caryn et al. (2014) have measured acceleration transmission to the head and found decreasing transmissibility with increasing vibration frequency from 20 to 50 Hz. Similarly, Kiiski et al. (2008) observed the highest transmissibility between 10 and 25 Hz at the knee, hip and spine with decreasing transmissibility until 90 Hz. Crewther et al. (2004) have found that the g-forces on the tibial tuberosity, greater trochanter and the jaw are higher during WBV at 20 Hz than at 10 and 30 Hz. Finally, Pollock et al. (2010) have demonstrated that the acceleration measured on the knee, hip, and head increased from 5 to 15 Hz, peaked around 15 Hz and decreased until 30 Hz. Altogether, it was found that platform acceleration was transmitted the most when the vibration frequency was rather low, i.e. between 10 and 20 Hz, which may be due to the occurrence of resonance (Harazin and Grzesik 1998).

3.7.3 KNEE ANGLE

To date, few studies have investigated the influence of the knee flexion angle on acceleration transmission to the body, reporting contradictory results. Avelar et al. (2013) found no difference in transmissibility to the lower limbs between knee flexion angles of 60° and 90°, whereas Tankisheva et al. (2013) found lower transmissibility with flexed knees (45° and 70°) than with straight knees (0°). Transmissibility outcomes to the upper body are more consistent; a decrease in transmissibility was found from standing with locked knees (0° knee flexion) to standing with bent knees at 20° (Rubin et al. 2003), and to standing with bent knees at 45° and 70° (Tankisheva et al. 2013). Acceleration transmission to the head has been measured in several studies, with most studies reporting that bending the knees decreases head vibrations. Caryn et al. (2014) found lower head accelerations with flexed knees at 40° compared to relaxed standing (20°) and standing with straight legs (0°). Similarly, decreased acceleration transmission to the head has been found with bent knees (45°, 70°) compared to with straight legs (0°) (Harazin and Grzesik 1998; Tankisheva et al. 2013). In contrast, Abercromby et al. (2007b) found that the acceleration measured on the head decreased from a knee flexion angle

of 10-15° until 26-30°, but increased from 26-30° to 31-35°. However, the results by Abercromby et al. (2007b) have been obtained during dynamic squatting, whereas all the above mentioned studies compared transmissibility between knee flexion angles during static squatting. Taken as a whole, it is not completely understood how changing the knee flexion angle alters the vibration transmission to the body during WBV exercises.

3.7.4 POPULATION

Only one study has investigated acceleration transmission between populations, i.e. between children and adults (Bressel et al. 2010). It was found that transmissibility to the body including the head was the same between the two populations, with one exception for the lower body; transmissibility to the ankle and the hip was 42% and 62% higher in the children than the adults for one specific vibration frequency (33 Hz). While it is unclear why the vibrations were more transmitted in the children to the lower body at this specific frequency, it can be concluded that vibration transmission during WBV is similar between children and adults. It remains to be determined whether vibration transmission is also identical between populations with different fitness levels.

Table 3.3 Summary of the outcomes concerning acceleration transmission during whole-body vibration (WBV) exercises.

Author and year	Subjects	Platform type	Freq	Amp	Exercise type	Knee angle	Measured Acceleration	Measuring tool	Magnitude	Results Transmissibility	Results Comparison of parameters
Abercromby et al. (2007b)	9 males 7 females (32.7 yrs)	RV VV	30 Hz	2 mm	Dynamic	10 - 15° 16 - 20° 21 - 25° 26 - 30° 31 - 35°	Platform Head	Skin-mounted accelerometers (3D)	RMS (resultant 3D)		Platform type: Knee angle: Head: VV > RV (all knee angles) Head: decreased from 10-15° until 26-30°, increased from 26-30° to 31-35°
Avelar et al. (2013)	34 males (26.3 yrs)	VV	30 Hz	4 mm	Static	60° 90°	Platform Knee Hip	Skin-mounted accelerometers (vertical, medio-lateral)	RMS (vertical, medio-lateral)		Knee angle: Knee, Hip: No difference between 60° and 90°
Bressel et al. (2010)	4 males 7 females (9.3 yrs) 5 males 5 females (25.9 yrs)	VV	28 Hz 33 Hz 42 Hz	0.5 mm - 0.7 mm	Static	25° 28°	Platform Ankle Knee Hip Sternum Head	Motion analysis	RMS (resultant 3D)	<u>Ankle:</u> Transmissibility > 2 <u>Knee:</u> Transmissibility ~ 0.5 <u>Hip, Sternum, Head:</u> Transmissibility < 0.3	Populations: <u>Transmissibility at the Ankle, Hip:</u> Children > Adults (30 Hz)
Caryn et al. (2014)	12 males (26.3 yrs)	VV	20 Hz 25 Hz 30 Hz 35 Hz 40 Hz 45 Hz 50 Hz	0.5 mm 1 mm	Static	0° 20° 40°	Platform Head	Skin-mounted accelerometers (3D)	RMS (resultant 3D)	<u>Head:</u> Transmissibility > 1 at 20 and 25 Hz and 0° knee angle. Transmissibility < 1 for all other conditions	Knee angle: <u>Transmissibility at the Head:</u> 20° > 40° (20-45 Hz) 0° > 40° (all frequencies)
Crewther et al. (2004)	11 males 12 females (26.1 yrs)	RV	10 Hz 20 Hz 30 Hz	1.25 mm 3 mm 5.25 mm	Static	5° 60°	Platform Tibial tuberosity Greater trochanter Jaw	Skin-mounted accelerometers (vertical)	?	Acceleration significantly decreased with increasing distance from platform	Frequency: <u>Acc averaged on all structures:</u> 20 Hz > 10, 30 Hz Amplitude: <u>Acc averaged on all structures:</u> 5.25 > 3 > 1.25 mm Knee angle: <u>Acc averaged on all structures:</u> 60° > 5°

Author and year	Subjects	Platform type	Freq	Amp	Exercise type	Knee angle	Measured Acceleration	Measuring tool	Magnitude	Results Transmissibility	Results Comparison of parameters
Harazin and Grzesik (1998)	10 males (23.1 yrs)	VV	4 Hz - 300 Hz	?	Static	Locked Relaxed 45° 70°	Platform Metatarsus Ankle Knee Hip Shoulder Head	Accelerometer (vertical)	Spectral power (vertical)	<p><u>On all bony structures:</u> Transmissibility decreased with increasing frequency; <u>Resonance on:</u> <u>Metatarsus:</u> 4-8 Hz, 12.5 Hz, 31.5-125 Hz <u>Ankle:</u> 4-8 Hz, 12.5 Hz, 25-63 Hz <u>Knee:</u> 4-8 Hz, 12.5-25 Hz <u>Hip:</u> 4-10 Hz <u>Shoulder:</u> 4-8 Hz, 12.5 Hz <u>Head:</u> 4-6.3 Hz, 12.5 Hz</p>	Knee angle: <u>Transmissibility at the head for 16 – 63 Hz:</u> 45°, 70° < Locked and Relaxed
Kiiski et al. (2008)	4 males (24 – 47 yrs)	VV	10 Hz - 90 Hz (in steps of 5 Hz)	0.05 mm 0.5 mm 1 mm 3 mm	Static	Relaxed	Platform Ankle Knee Hip Spine	Skin-mounted accelerometers (3D)	RMS (resultant 3D)	<p><u>Transmissibility > 1:</u> <u>Ankle:</u> 10-40 Hz <u>Knee:</u> 10-25 Hz <u>Hip:</u> 10-20 Hz <u>Spine:</u> 10 Hz</p>	
Muir et al. (2013)	4 males 2 females (24.1 yrs)	VV VV CV	34 Hz 33 Hz 23 Hz - 48 Hz	1.7 mm 3.3 mm 0.1 mm 1.4 mm - 0.85 mm	Static	Locked Relaxed 45° 90°	Platform Proximal tibia	Skin-mounted accelerometers (3D)	Peak-to-peak (resultant 3D)	<p><u>Transmissibility at 45°:</u> Tibia: 0.5; Cranium: 0.05 <u>Transmissibility at 90°:</u> Tibia: 0.7; Cranium: 0.08 <u>Transmissibility for Locked:</u> Tibia: 0.9; Cranium: 0.8 <u>Transmissibility at 90°:</u> Tibia: 0.6; Cranium: 0.1 <u>Transmissibility for Locked:</u> Tibia: 0.5; Cranium: 0.1 <u>Transmissibility at 90°:</u> Tibia: 0.3; Cranium: 0.02</p>	

Author and year	Subjects	Platform type	Freq	Amp	Exercise type	Knee angle	Measured Acceleration	Measuring tool	Magnitude	Results Transmissibility	Results Comparison of parameters
Pel et al. (2009)	8 subjects (34 yrs)	RV		1.8 mm						<u>Ankle:</u> Transmissibility = 0.85 <u>Knee:</u> Transmissibility = 0.8 <u>Hip:</u> Transmissibility = 0.02	
		VV	25 Hz	1.1 mm	Static	80°	Platform Ankle Knee Hip	Skin-mounted accelerometers (vertical)	RMS (vertical)	<u>Ankle:</u> Transmissibility = 0.55 <u>Knee:</u> Transmissibility = 0.09 <u>Hip:</u> Transmissibility = 0.03	
		CV		0.6 mm						<u>Ankle:</u> Transmissibility = 1 <u>Knee:</u> Transmissibility = 0.2 <u>Hip:</u> Transmissibility = 0.1	
Pollock et al. (2010)	12 subjects (31.3 yrs)	RV	5 Hz 10 Hz 15 Hz 20 Hz 25 Hz 30 Hz	5.5 mm 2.5 mm	Static	15°	Platform Toe Ankle Knee Hip Head	Motion analysis (3D)	Peak acceleration (vertical)	<u>Toe, Ankle, Knee, Hip:</u> Transmissibility > 1 (2.5 mm)	Frequency: <u>Toe, Ankle, Knee, Hip:</u> Peak transmissibility at 10 Hz Amplitude: <u>Toe, Ankle, Knee, Hip, Head:</u> Transmissibility 2.5 mm > 5.5 mm
Rubin et al. (2003)	5 males 1 female (23-33 yrs)	?	15 Hz - 35 Hz (in steps of 1 Hz)	Adjusted amplitude for 18 N	Static	0° 5° 20°	Platform Greater trochanter L4	Bone-mounted accelerometers (2D)	(resultant 2D)	<u>0° knee angle with frequency < 20 Hz:</u> Transmissibility at hip > 1 <u>0° knee angle with frequency > 25 Hz:</u> Transmissibility at hip, spine = 0.8	Knee angle: <u>Hip, spine:</u> Transmissibility 0° > 5° > 20°
Tankisheva et al. (2013)	5 males 3 females (28.7 yrs)	3 x VV	30 Hz 35 Hz 40 Hz 50 Hz	Low High	Static	0° 45° 70°	Platform Ankle Tibial tuberosity Medial femur Greater trochanter Hip L3, C7 Sternum Head	Skin-mounted accelerometers (3D)	RMS and peak-to-peak (vertical)	<u>Ankle:</u> Transmissibility of 1.3 – 2.9 <u>Hip:</u> Transmissibility of 0.04 – 0.2 <u>Head:</u> Transmissibility of 0.03 – 0.15	Knee angle: <u>L3:</u> 45° > 70° <u>Greater trochanter, trunk, head:</u> 0° > 70°, 45°

Amp, Amplitude; Freq, Frequency; RMS, Root mean square; RV, side-alternating platform; VV, vertical synchronous platform; yrs, years.

3.8 POTENTIAL SIDE-EFFECTS

While most platform types deliver vibrations with an input frequency close to the natural frequency of the body, it is not clarified if such frequencies are detrimental. Resonance might be dangerous as it results in an amplification of the acceleration, which then might overload or damage structures of the body (Mester et al. 1999; Rubin et al. 2003; Wakeling and Nigg 2001b). Furthermore, high accelerations have been associated with back pain (Rubin et al. 2003) and altered visual perception (Pel et al. 2009). The International Organization for Standardization (ISO) has guidelines (ISO 2631-1 1997) related to maximum occupational vibrations. Muir et al. (2013) have compared the daily acceleration threshold limit defined for occupational exposure to vibrations to the acceleration magnitude induced by several WBV platform types. They found that exercising on two out of three of the tested vibration platforms induced vertical peak-to-peak accelerations above 8g or resultant tri-axial peak-to-peak accelerations above 3g. Such acceleration magnitudes are declared as high risk occupational conditions by the ISO and less than one-minute of daily exposure is advised. Another study (Abercromby et al. 2007b) calculated the estimated vibration dose value of side-alternating and synchronous WBV platforms by measuring the platform acceleration and calculating the frequency-weighting coefficients as defined by ISO-2631 (1997). The coefficients are based on the frequency and direction of the applied vibration and are multiplied in a way that the value reflects the health and hazard posed to the human body. Abercromby et al. (2007b) found that the accelerations induced at 30 Hz and 2 mm amplitude are on the upper limit of the ISO recommendations for daily exposure. A more recent study (Tankisheva et al. 2013) performed a similar investigation but instead of using the coefficients defined by ISO 2631-1 (1997), coefficients similar to the ones in norm BS 6841 were applied. Nevertheless, similar results have been reported, with only one out of the three tested platforms inducing accelerations low enough for a safe daily use of 15 min. As the guidelines by the ISO focus on very high vibration frequencies (up to 80 Hz) induced over long periods (4 to 8 hours) rather than on lower frequencies induced for shorter periods by machines such as WBV platforms, it is questionable if those guidelines are entirely applicable to training/medical devices. Further, the coefficients

published by ISO 2631-1 (1997) are specified for persons receiving vibrations through their bottom in a sitting position and not through their feet in a standing position.

Considering that the greatest increase in muscle activity is typically induced by using the highest platform settings (Di Giminiani et al. 2013; Krol et al. 2011; Perchthaler et al. 2013; Pollock et al. 2010; Ritzmann et al. 2013), it is of great interest to establish an acceleration safety threshold for exercising on WBV platforms. Until this is accomplished by the ISO, WBV exercises should be performed at the lowest platform acceleration that still provokes significant increases in muscle activity.

3.9 SUMMARY

The diverse use of sEMG processing and normalization methods in WBV studies has made it extremely difficult to compare outcomes of different studies. The first step to implement standardized sEMG processing during WBV is to sensitize researchers to the present issue by demonstrating the influence of each processing and normalization method on the quantification of the sEMG activity. The next step is to find and communicate the optimal sEMG analysis method and to apply it in a consistent and standardized manner.

Simultaneously, it is important to verify whether the excessive spikes observed in the sEMG spectrum contain motion artifacts and/or reflex responses. Although it is known that WBV elicits stretch reflex responses (Ritzmann et al. 2010), synchronous motor unit activity (Pollock et al. 2012) and tonic vibration reflex-like activity (Zaidell et al. 2013), it is still unclear if the spikes per se contain such responses. Knowing the content of the spikes will contribute in finding the optimal sEMG processing method during WBV.

Another issue that needs to be addressed concerns the previously reported results describing the effects of WBV exercises on muscle activity that are partly contradictory and widely insufficient. It is still unclear which platform settings (i.e. frequency and amplitude) elicit the highest increases in sEMG levels. It would be of great benefit to determine a minimal required platform acceleration to achieve significant changes in muscle activity. Further, more research is needed to clarify the influence of the platform type, additional loading, and knee flexion angle on changes in muscle activity. Examining these parameters will help to determine the optimal WBV condition in which the greatest adaptations can be expected.

Finally, acceleration transmission has been described in numerous research studies, yet it is undetermined how changes in knee flexion angle and subjects' individual characteristics affect transmissibility. Gaining such insight could help to design WBV training protocols imposing the lowest possible accelerations on the head, and to characterize WBV-associated risks for WBV users with a different fitness background.

3.10 HYPOTHESES

Therefore, the overall goal of this thesis was to investigate the changes in muscle activity induced by WBV using the optimal sEMG processing and normalization method. A further goal was to determine the effect of frequency, amplitude, additional loading, knee flexion angle and training status on the increases in sEMG levels. Another goal was to describe transmissibility during WBV in two populations presenting different training levels and for two knee flexion angles.

The research for this thesis was divided into six studies (Table 3.4). **Study I&II** were methodological studies, validating the different sEMG processing methods including a newly developed filter based on spectral linear interpolation and different sEMG normalization methods. **Study III** was an explorative study and served to gain insight whether the isolated spikes observed in the sEMG spectrum contain motion artifacts and/or reflex activity. It is important to mention that the analyses for studies I-III were conducted concomitantly and that their outcomes served for methodological decisions. **Study IV-VI** were physiologically applied studies, where the methodological knowledge acquired from the previous studies was applied investigating the effect of WBV exercises on lower limb muscle activity and acceleration transmission to the head.

Table 3.4 Overview of the studies that have been conducted for this thesis.

Methodological	Study I	What is the best sEMG processing method for WBV exercises?
	Study II	Which normalization method should be used for sEMG data obtained during WBV exercises?
Explorative	Study III	Do the isolated spikes in the sEMG spectrum contain motion artifacts, reflex activity, or both?
Applied	Study IV	How do frequency, amplitude and additional loading affect muscle activity of the lower limbs during WBV exercises?
	Study V	How does the direction of the imposed vibration influence the muscle activity of the lower limbs during WBV exercises?
	Study VI	What is the influence of a person's fitness level and the influence of the knee flexion angle on increases in muscle activity and head transmissibility due to WBV exercises?

CHAPTER 4

METHODS

4.1 PARTICIPANTS

All studies included healthy young adults, aged between 18 and 35 years (**Table 4.1**). The participants were either physically active (exercising ≥ 5 h/week) or inactive (exercising ≤ 2 h/week). The participants were new to WBV exercises, and were free from a recent lower limb injury. The experiments were performed in accordance with the Declaration of Helsinki (1964), and written informed consent was obtained from each participant prior to the assessment.

Study I-IV comprised 18 physically active participants, recruited from the Faculty of Sport Sciences of the University of Nice Sophia Antipolis, Nice, France.

Study V included 30 physically active participants, recruited from the Faculty of Kinesiology of the University of Calgary, Alberta, Canada.

Study VI included 30 physically active and 28 physically inactive participants, recruited from Calgary, Alberta, Canada.

Table 4.1 Characteristics (mean \pm SD) of the subjects who participated in studies I-VI.

	Fitness	Methodological		Explorative	Applied		
		Study I	Study II	Study III	Study IV	Study V	Study VI
N	active	18	18	18	18	30	30
	inactive	-	-	-	-	-	28
Male/female	active	10/8	10/8	10/8	10/8	15/15	15/15
	inactive	-	-	-	-	-	14/14
Age (years)	active	23.8 \pm 3.2	23.8 \pm 3.2	23.8 \pm 3.2	23.8 \pm 3.2	25.9 \pm 4.3	25.9 \pm 4.3
	inactive	-	-	-	-	-	26.9 \pm 3.8
Physical activity (h/week)	active	7.5 \pm 4.2	7.5 \pm 4.2	7.5 \pm 4.2	7.5 \pm 4.2	≥ 5	≥ 5
	inactive	-	-	-	-	-	≤ 2

4.2 DATA ACQUISITION

Table 4.2 summarizes the data that were assessed in each study. In short, sEMG activity of selected lower limb muscles was recorded by using bipolar silver-chloride electrodes (see **4.3 SURFACE ELECTROMYOGRAPHY**). The excessive spikes observed in the power spectrum were withdrawn in all studies with the help of spectral linear interpolation (see **4.4 sEMG PROCESSING DURING WBV**). sEMG data were normalized to the sEMG data recorded during a maximal voluntary contraction (MVC) in all studies, whereas in study II, normalization to the MVC was compared to normalization to the maximal electrophysiological response (Mmax, see **4.5 NORMALIZATION METHODS**). In study III, motion artifacts were quantified by placing so-called dummy electrodes on the patella (see **4.6 MOTION ARTIFACTS RECORDINGS**). Further, the acceleration delivered by the WBV platform was measured with the help of 3D accelerometers in all studies (see **4.7 PLATFORM ACCELERATION**). However, two different platform types have been used. In studies I-IV, a vertical synchronous WBV platform (Power Plate) was applied; in study V, a novel WBV platform (Total Image Fitness) was applied that has a side-alternating mode in the vertical direction, a circular mode in the horizontal direction, and a dual mode that runs the two modes simultaneously; in study VI, the side-alternating mode of the Total Image Fitness platform was used (see **4.8 PLATFORM TYPES**). The amount of the platform acceleration transmitted to the head was quantified in study VI (see **4.9 ACCELERATION TRANSMISSION**). Finally, the participants' body posture was monitored in all studies by measuring the knee flexion angle with the help of an electrical twin-axis goniometer (see **4.10 BODY POSTURE**).

All the signals were analysed and processed with the help of Matlab software (version 7.13 The Mathworks, Inc., Natick, MA, USA). Statistical analysis was conducted using STATISTICA software (Statsoft, version 7.0 Tulsa, OK, USA) or using IBM SPSS software (Version 20, Chicago, IL, USA).

Table 4.2 Overview of the signals that were measured and the devices that were used in studies I-VI.

	Methodological		Explorative	Applied		
	Study I	Study II	Study III	Study IV	Study V	Study VI
sEMG recordings	VL, SOL (Bipolar electrodes, Biopac® Systems Inc.)	VL, SOL	GL, SOL, VL, RF, BF (Bipolar electrodes, Biopac® Systems Inc.)	GL, SOL, VL, VM, RF (Bipolar electrodes, Biopac® Systems Inc.)	TA, GM, SOL, VL, VM, BF (Bipolar electrodes, Biovision)	TA, GM, VL, VM, BF
sEMG processing	Interpolation, Band-stop filter, Band-pass filter	Interpolation	Interpolation	Interpolation	Interpolation	Interpolation
sEMG normalization	MVC (Weight machines)	MVC, Mmax (Weight machines, neurostimulator, Digitimer)	-	MVC (Weight machines)	MVC (Isokinetic dynamometer, Biodex System 3)	MVC
Motion Artifacts	-	-	Dummy electrode on knee	-	-	-
3D Platform Acceleration	(3D Accelerometer, Biopac® Systems Inc)		(3D Accelerometer, Biopac® Systems Inc)	(3D Accelerometer, Biopac® Systems Inc)	(3D Accelerometer, Analog Devices USA)	
Platform type	Vertical Synchronous (Power Plate)	Vertical Synchronous (Power Plate)	Vertical Synchronous (Power Plate)	Vertical Synchronous (Power Plate)	Side-alternating, Circular, Dual mode (Total Image Fitness)	Side-alternating (Total Image Fitness)
Acceleration on Body	-	-	-	-	-	Head (1D Accelerometer, Analog Devices USA)
Body Posture	Knee angle 70° (Electrical goniometer, Biopac® Systems Inc.)	Knee angle 70°	Knee angle 70° (Electrical goniometer, Biopac® Systems Inc.)	Knee angle 70° (Electrical goniometer, Biopac® Systems Inc.)	Knee angle 60° (Electrical goniometer, Biometrics Ltd)	Knee angle 20°, 60°

BF, Biceps Femoris; GL, Gastrocnemius Lateralis; GM, Gastrocnemius Medialis; Mmax, Maximal M-wave response; MVC, Maximal voluntary contraction; RF, Rectus Femoris; sEMG, surface electromyography; SOL, Soleus; TA, Tibialis Anterior; VL, Vastus Lateralis; VM, Vastus Medialis.

4.3 SURFACE ELECTROMYOGRAPHY

Electromyography is defined as “... an experimental technique concerned with the development, recording and analysis of myoelectric signals. Myoelectric signals are formed by physiological variations in the state of muscle fiber membranes.” (Basmajian 1978). The electromyography contains information regarding the onset, duration, and magnitude of the depolarisation stimulus transferred to the muscle fiber. Non-invasive electromyography recordings, i.e., sEMG, are managed by placing electrodes on the skin over the muscle belly of interest. The electrodes are typically bipolar, silver/silver chloride pre-gelled electrodes and measure the difference in action potential between them. The electrode placement on human body muscles is guided by anatomical landmarks and has been well documented in the SENIAM guidelines (Hermens et al. 2000).

LIMITATIONS

sEMG recordings are associated by a number of limitations, which have to be considered for the assessment, analysis and interpretation of the results. These limitations have been well described (Basmajian 1978; De Luca 1997; Konrad 2005), and are summarized in the following. First, the electrical conductivity varies with the temperature of the skin and the thickness of the tissue between skin and muscle. As both can vary substantially from person to person, quantitative comparison of the raw sEMG magnitude between subjects should be avoided. Second, cross-talk from neighbouring muscles can produce a significant amount of sEMG, especially when the muscles are located close to each other within the same muscle group. Third, the electrodes are mounted on the skin and not on the muscle directly, which means that changes in body position (e.g. during dynamic tasks) changes the distance between signal and origin and could alter the sEMG magnitude. Last, sEMG signals are able to pick up signals from an electrical environment (e.g. powerline interference), which can produce a noise in the sEMG signal at 50 or 60 Hz.

4.3.1 ANALYSIS IN THE TIME DOMAIN

sEMG analysis in the time domain provides useful information regarding the magnitude of the signal. In numerous studies, the magnitude has been expressed by calculating the RMS of the signal (Abercromby et al. 2007a; Bosco et al. 1999; Cardinale et Lim 2003; Di Giminiani et al. 2013; Fratini et al. 2009a; Hazell et al. 2007, 2010; Hermens et al. 2000; Krol et al. 2011; Marín et al. 2012a; Perchthaler et al. 2013; Pollock et al. 2010; Roelants et al. 2006; Wirth et al. 2011). Therefore, in all studies of the current thesis, the sEMG_{RMS} was calculated in the time domain for the no-vibration trials via the **equation (1)**

$$sEMG_{RMS} = \frac{1}{N} \sum_{n=0}^{N-1} x(n)^2 \quad (1)$$

with N being the length of the data and $x(n)$ the sampled sEMG signal in the time domain.

4.3.2 ANALYSIS IN THE FREQUENCY DOMAIN

While the RMS of the sEMG signal provides important information, additional information can be extracted in the frequency domain by means of e.g. PSD analysis. The PSD of a signal provides information about the frequency content of the signal, i.e., how the power of a signal is distributed over the different frequencies. The PSD is estimated by simply computing the squared magnitude of the discrete-time Fourier transform (DTFT) of a stationary signal implemented by the Fast Fourier Transform (FFT) algorithm (Sörnmo et Laguna 2005). In other words, the PSD is computed by using the theoretical expression $X(f) = E [|\text{FFT}(x)|^2]$, where E stands for the mathematical expectation. In that case, the RMS can be calculated in the frequency domain by summing the $X(f)$ values. The practical computation of the RMS calculation in the PSD is explained in **4.4.4 SPECTRAL LINEAR INTERPOLATION**.

4.3.3 RELATIONSHIP BETWEEN TIME AND FREQUENCY DOMAIN

In order to better understand the relationship between a signal in the time domain and the frequency domain, different signals were simulated and are illustrated in **Figure 4.1**. The simulation of these signals was computed in a simple manner and served as preliminary results to better understand the sEMG spectrum during WBV.

For example, **Figure 4.1A** shows a sinusoidal wave simulated in the time domain (upper panel) at a given frequency ($f_0 = 40\text{Hz}$). Transformation of this signal into the PSD displays only one sharp peak exactly at the frequency ($f_0 = 40\text{Hz}$). A sinusoidal wave such as illustrated in **Figure 4.1A** can be measured on the surface of an oscillating WBV platform using an accelerometer (Pel et al. 2009) or on the patella using sEMG electrodes (Fratini et al. 2009a).

Figure 4.1B illustrates a simulated periodic stretch reflex train. Stretch reflex responses are believed to occur during WBV, and their shape has been illustrated by e.g. Ritzmann et al. (2010). Therefore, single motor unit action potentials (MUAPs) were computed with identical amplitudes and separated by identical time intervals (t_0) in the time domain. Subsequently, the signal was transformed into the PSD, where sharp peaks are visible at $\frac{1}{t_0} = 40\text{Hz}$ and its first three harmonics. No myoelectrical activity was visible at any other frequencies.

In reality, stretch reflex responses vary in magnitude and in time delay, despite the use of a constant input stimulus (Ritzmann et al. 2010). Therefore, a stretch reflex train was simulated consisting of quasi-periodic MUAPs with variable amplitudes and variable time intervals around (t_0). In other words, randomness was added in the time domain to the signal from **Figure 4.1B**, which slightly changed the amplitude and delay of each individual MUAP (**Figure 4.1C**, upper panel). The PSD of this signal (**Figure 4.1C**, lower panel) still displays sharp peaks at $\frac{1}{t_0} = 40\text{Hz}$ and its first few harmonics, but myoelectrical activity can be found over the entire frequency spectrum of the signal. Interestingly, these simulated non-periodic MUAPs resemble the sEMG signal that is typically recorded during WBV. However, assuming that stretch reflex responses

are recorded in sEMG signals during WBV, deleting the sharp spikes at $\frac{1}{t_0} = 40Hz$ and its harmonics still retains most of the myoelectrical activity originating from the stretch reflex responses.

Last, a typical sEMG signal without WBV was simulated (**Figure 4.1D**). This was accomplished with the help of the script for the computation of the quasi-periodic stretch reflex train (**Figure 4.1C**). During an isometric muscle contraction, the activation rate of motor units can go up to 100 pps. Therefore, a number of non-periodic MUAPs with variable amplitudes and variable time intervals were summed in the time domain which resulted in the signal illustrated in **Figure 4.1D**, upper panel. In contrast to the other simulated signals, the PSD of this signal does not display excessive spikes at a given frequency.

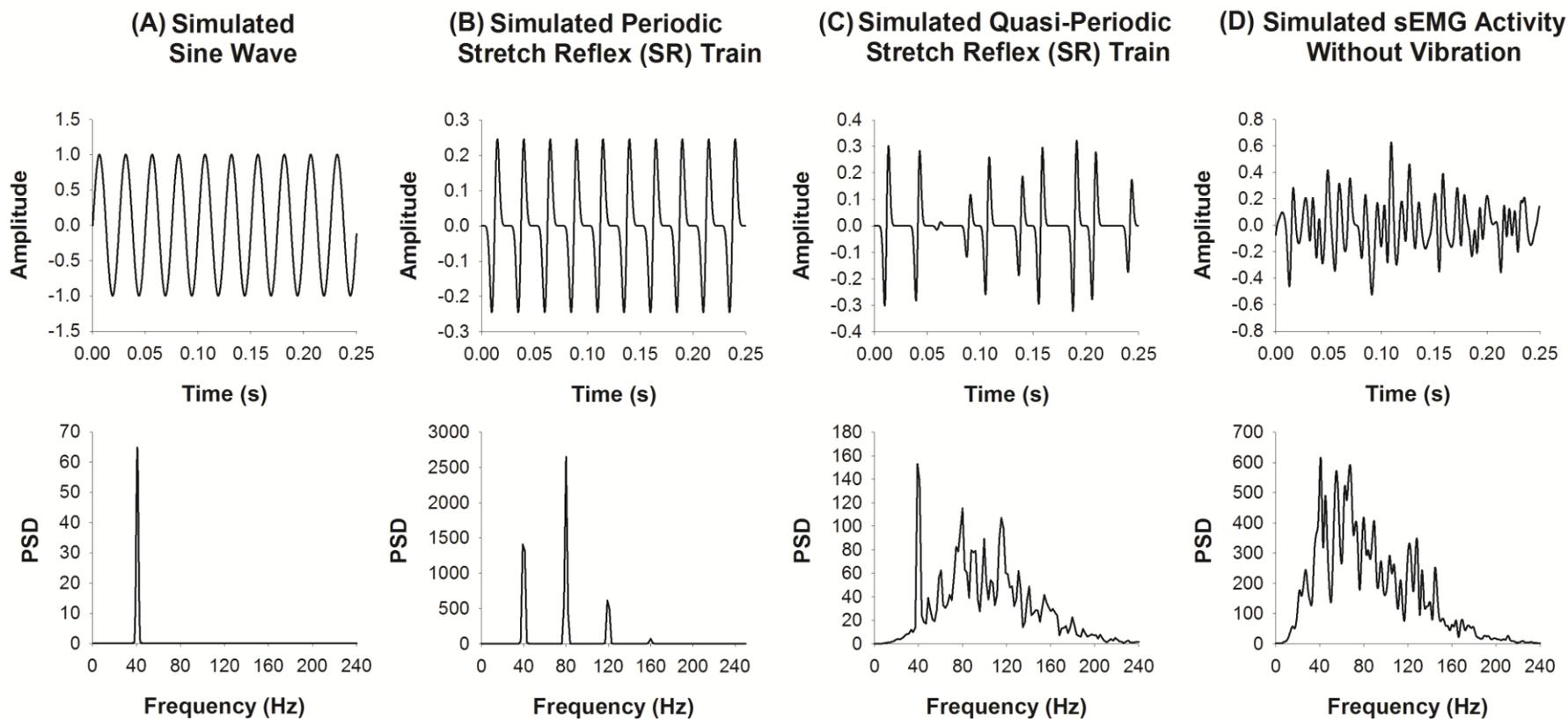


Figure 4.1 Illustration of (A) a simulated sine wave at 40 Hz, (B) a simulated periodic stretch reflex train at 40 Hz, (C) a simulated quasi-periodic stretch reflex train around 40 Hz with variable amplitude, and (D) a simulated surface electromyography (sEMG) signal without vibration. The signals are shown in the time domain (upper panel) and the Power Spectral Density (PSD, lower panel).

4.3.4 sEMG PRE-PROCESSING

Using the amplifier from Biopac® Systems Inc. (MP150 Biopac® Systems Inc., Holliston, MA, USA; CMRR = 110 db, Z input = 1000 M Ω , gain = 1000) for studies I-IV, the sEMG signals were analog filtered with a bandwidth frequency ranging from 10 Hz to 500 Hz. Additionally, noise raising from power line interference was deleted analogically. Digitally, the sEMG signals were processed with a band-pass filter to delete the low- and the high-frequency noise. For studies I-IV, an 18th order band-pass Butterworth filter was applied, whereas a band-pass wavelet filter was used for studies V&VI. Using a wavelet band-pass filter did not change the outcomes as compared to using a Butterworth band-pass filter. The low and the high cut-off frequencies were 10 Hz and 450 Hz for studies I-IV, respectively, and 5 Hz and 300 Hz for studies V&VI. The low cut-off frequency was lower for studies V&VI in order to retain potential stretch reflex responses during WBV at 6 Hz.

BUTTERWORTH FILTER

The ideal band-pass filter would consist of a completely flat pass-band and stop-band. However, this is not the case in reality (**Figure 4.2**). Realizable filters can be described by the ripple factor which specifies the maximum and minimum deviation from the ideal value in the pass-band and the stop-band. Further, the transition zone (i.e., bandwidth signal between pass-band and stop-band) is not a brick-wall response, and is more ideal the narrower. The relative steepness of the filter's transition zone is described by the filter's order; the higher the order the narrower the transition zone. The Butterworth filter is an ideal method to process sEMG signals as its response in the pass-band is maximally flattened with a minimized pass-band ripple.

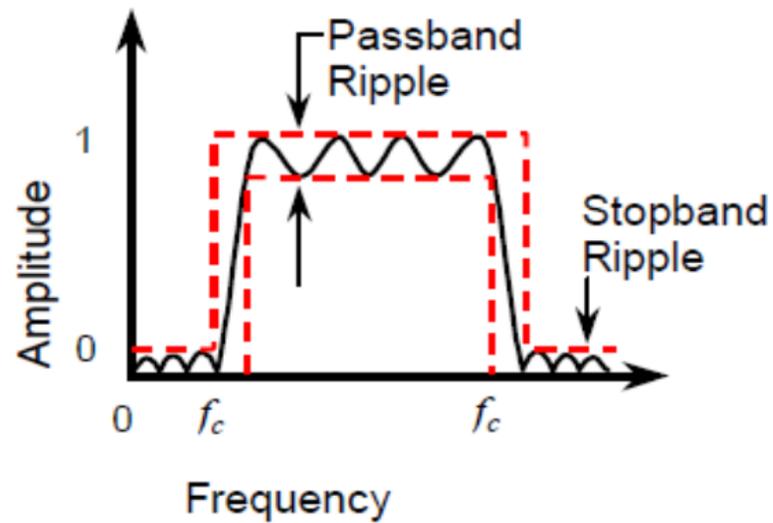


Figure 4.2 Characteristics of a band-pass filter, including the ripple for the pass-band and the stop-band, and the transition zone with the corner frequency f_c .

WAVELET FILTER

Wavelet filters are typically used to provide a time-frequency representation of a signal which can be crucial during dynamic tasks such as running. Wavelet transforms typically use filterbanks to decompose the signal in different bandwidths. However, wavelets are a good method to use for band-pass filtering.

4.4 sEMG PROCESSING DURING WBV

sEMG signals were recorded during WBV for preliminary analyses. Such a signal is illustrated in **Figure 4.3A** in the time domain and in **4.3B** in the frequency domain. The signal was recorded from the VL muscle during WBV at 30 Hz. While the sEMG signal in the time domain does not highlight any specific characteristics to WBV, the sEMG signal in the frequency domain clearly shows excessive spikes at the vibration frequency ($f_0 = 30\text{Hz}$) and at a few multiple harmonics. As a matter of fact, the signal illustrated in **Figure 4.3B** reminds of the PSD of the quasi-periodic stretch reflex train illustrated in **Figure 4.1C**. It is possible, that the excessive spikes visible in the sEMG power spectrum during WBV originate, at least in parts, from stretch reflex responses.

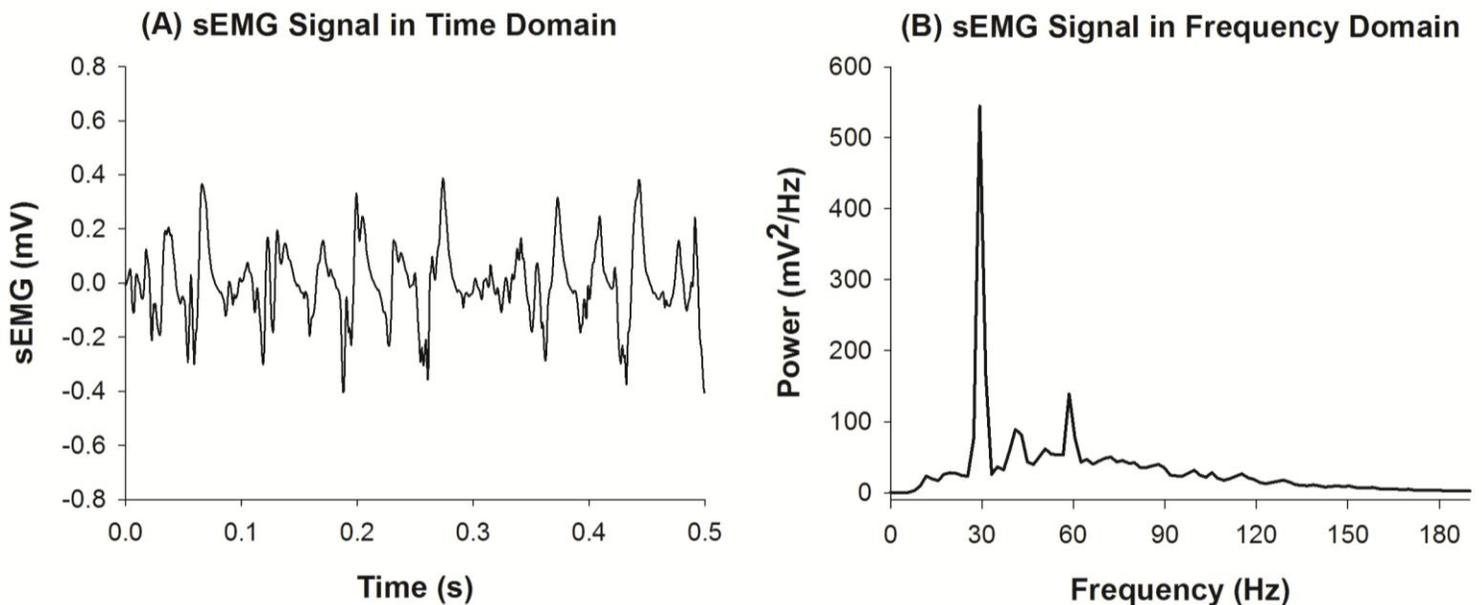


Figure 4.3 Surface electromyography (sEMG) signal of the Vastus Lateralis during whole-body vibration at 30 Hz illustrated in (A) the time domain and (B) the frequency domain. In the frequency domain, excessive spikes are visible at the vibration frequency ($f_0 = 30\text{Hz}$) and its multiple harmonics.

At the same time, and also for preliminary purposes, the electrophysiological signal was recorded from the patella during WBV. Such a signal obtained during WBV at 30 Hz is shown in **Figure 4.4**. In the time domain, the patella signal resembles a sinusoidal wave at 30 Hz. In the

frequency domain, excessive spikes are observed at the vibration frequency and to a lower extent at its multiple harmonics. No myoelectrical activity is shown for all the other frequencies. Therefore, a preliminary conclusion was that using the current study set-up, the sEMG signals are contaminated by motion artifacts caused by the oscillation of the WBV platform.

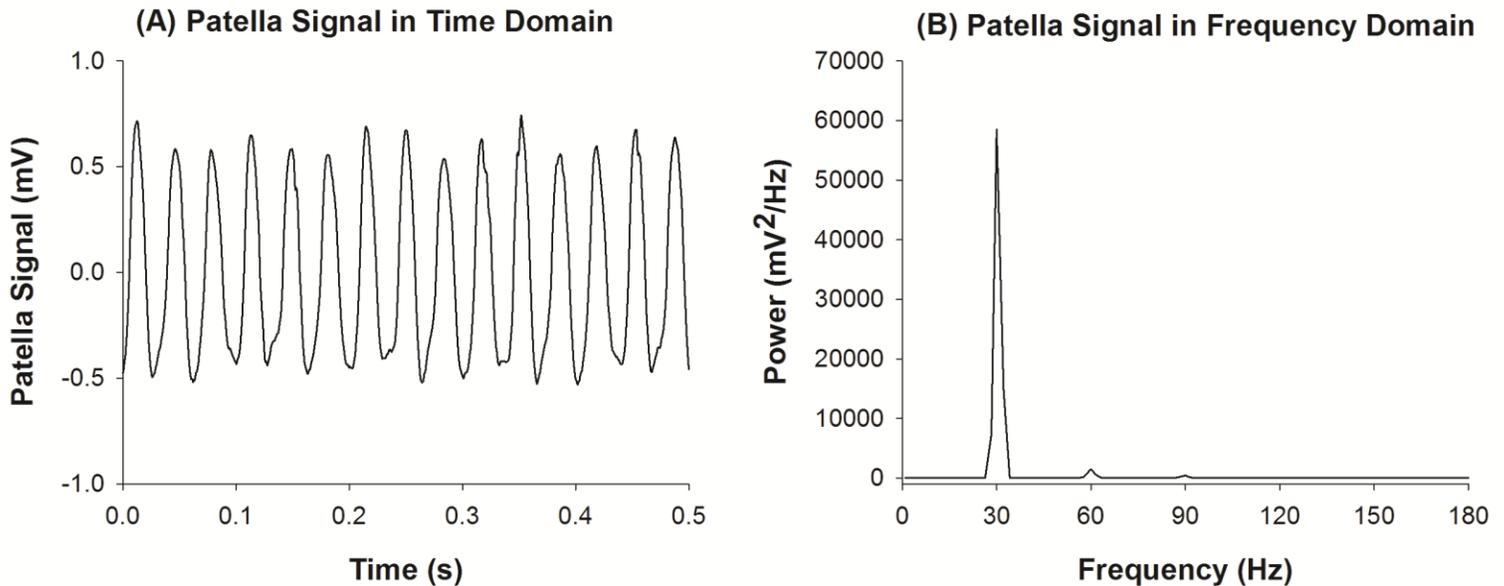


Figure 4.4 Electrophysiological signal recorded from the patella during whole-body vibration at 30 Hz illustrated in (A) the time domain and (B) the frequency domain. In the frequency domain, excessive spikes are visible at the vibration frequency ($f_0 = 30\text{Hz}$) and its multiple harmonics.

While deleting these spikes eliminates the motion artifacts caused by the vibration platform, it is possible that some information originating from the stretch reflex responses is deleted as well. On the other hand, if the spikes are not deleted, sEMG signals are most likely contaminated by motion artifacts and it is possible that the RMS of the muscle activity during WBV is an overestimation of the real muscle activity. Therefore, it was important to compare the influence of the different state-of-the-art sEMG processing methods (no-filter method, band-stop filter, band-pass filter) on the quantification of sEMG activity during WBV. Additionally, a new method based on spectral linear interpolation was developed and compared the already existing methods.

4.4.1 NO FILTER

In numerous studies, the spikes have been considered as mostly stretch reflex responses with negligible contribution of motion artifacts (Di Giminiani et al. 2013; Krol et al. 2011; Perchthaler et al. 2013; Ritzmann et al. 2013; Roelants et al. 2006). Hence, in these studies, no specific filter was applied to delete the spikes in the sEMG spectrum. In order to calculate the amplitude of the unfiltered sEMG in study I, the $sEMG_{RMS}$ was calculated from the unfiltered sEMG in the time domain using *equation (1)*.

4.4.2 BAND-STOP FILTER

Band-stop filters to delete the spikes in the sEMG spectrum have been applied in numerous studies (Abercromby et al. 2007a; Fratini et al. 2009a, 2009b; Marín et al. 2012a, 2012b; Pollock et al. 2010; Wirth et al. 2011). So far, two different band-stop filter methods in relation with WBV have been introduced. Fratini et al. (2009a, 2009b) filtered the sEMG signals using a set of notch filters with a -3dB band of 1.5 Hz, centred at the first three spikes. This method has been reapplied by Wirth et al. (2011). The other band-stop filter method was introduced by Abercromby et al. (2007a), who filtered the sEMG signals using a 17th-order Chebyshev type II. The order of the filter was previously calculated with the choice of a stop-band of ± 1 Hz, a pass-band of ± 1.5 Hz, a minimum stop-band attenuation of 100 dB, and a maximum pass-band ripple of 0.01 dB. The filter was centered at the frequency of the fundamental and the harmonics up to 450 Hz. This method has been reapplied by Pollock et al. (2010) and Marín et al. (2012a, 2012b). The outcomes of the two methods were compared in a preliminary analysis, which showed that the band-stop filter introduced by Fratini et al. (2009a, 2009b) delivered slightly higher sEMG values than the band-stop filter by Abercromby et al. (2007a). This is not surprising considering that the band-stop filter by Fratini et al. (2009a, 2009b) deleted only the first three spikes, whereas the method by Abercromby et al. (2007a) deleted the spikes up to 450 Hz (e.g., eleven spikes were deleted at 40 Hz WBV). However, the difference between the two methods averaged for all trials measured in study I was (mean \pm SD) $12.8 \pm 2.9\%$ for the SOL, $11.9 \pm 2.1\%$ for the GL, $11.8 \pm 1.4\%$ for the VM, $11.0 \pm 1.2\%$ for the RF, and $11.4 \pm 1.4\%$ for the VL. As the band-stop filter introduced by Fratini et al. (2009a, 2009b) has been reapplied less than the

band-stop filter by Abercromby et al. (2007a), the latter method was used in study I. Therefore, sEMG signals were filtered in the time domain using the band-stop filter by Abercromby et al. (2007a) with subsequent calculation of the sEMG_{RMS} using *equation (1)*. The effect of the band-stop filter on the sEMG signal in the PSD is illustrated in **Figure 4.5**, blue signal.

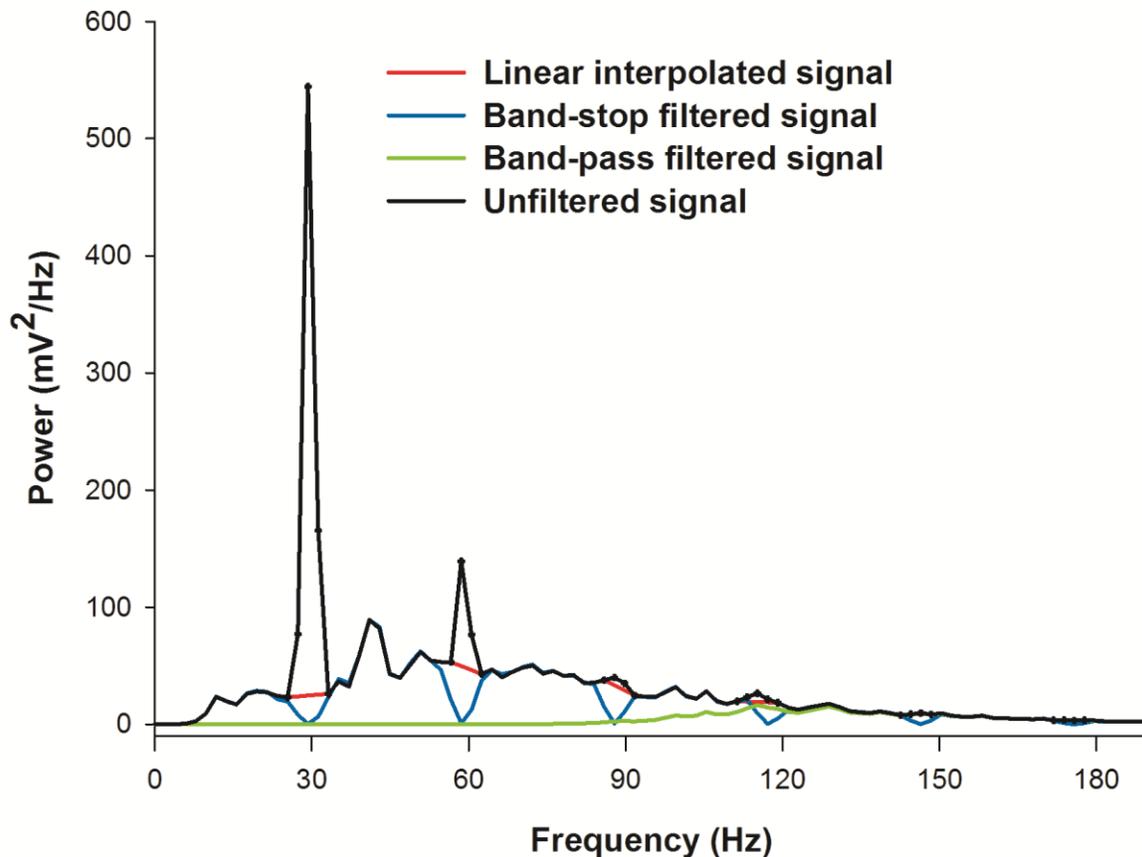


Figure 4.5 Example of a surface electromyography (sEMG) signal during whole-body vibration at 30 Hz. The black signal represents the unfiltered sEMG, the red signal the linear interpolated sEMG, the blue signal the band-stop filtered sEMG, and the green signal the band-pass filtered sEMG.

4.4.3 BAND-PASS FILTER

A band-pass filter for withdrawal of the spikes in the sEMG spectrum has been applied by Hazell et al. (2007, 2010) whereas only high frequency signals (100 - 450 Hz) were retained. This approach was based on a scientific study (Potvin and Brown 2004) showing that high frequency sEMG signals above 140 Hz improve biceps brachii strength estimates compared to common high pass cutoffs in the range of 10 – 30 Hz. Thus, sEMG signals were filtered in the time domain using a dual passed 6th order Butterworth band-pass filter between 100 and 450 Hz with subsequent calculation of the sEMG_{RMS} using **equation (1)**. The effect of the band-pass filter on the sEMG signal in the PSD is illustrated in **Figure 4.5**, green signal.

4.4.4 SPECTRAL LINEAR INTERPOLATION

The aim of the proposed interpolation method was to implement a filtering technique that removes only the spikes from the sEMG signals in the PSD, without adding notches to the signal (band-stop filter) or deleting a wide frequency spectrum (band-pass filter). Unlike the time-domain methods of the state-of-the-art (band-stop filter, band-pass filter), we assumed that only the excessive part of the spikes should be removed in order to preserve the maximum of the sEMG information at the oscillating frequency and its harmonics. As a consequence, the proposed interpolation method does not allow a time-reversal transformation, and the RMS needs to be calculated directly within the PSD (**Figure 4.5**).

sEMG signals recorded during WBV contain sEMG activity that is not related to the spikes ($emg(t)$), and additional activity ($emg_s(t)$) that accounts for the spikes. Consequently, in the time domain, the entire signal ($signal(t)$) can be expressed as: $signal(t) = emg(t) + emg_s(t)$. Transforming $signal(t)$, $emg(t)$, and $emg_s(t)$ into the FD results in $S(f)$, $E(f)$, and $E_s(f)$, which are complex numbers. Thus, direct subtraction of the spikes $E_s(f)$ from the entire signal $S(f)$ is not possible in the FD. However, with the help of the squared modulus (i.e., the PSD), modeling the sum of $E(f)$ and $E_s(f)$ allows this subtraction in the frequency domain using the equation $|E(f)|^2 = |S(f)|^2 - |E_s(f)|^2$ assuming that the data are independent and using the linearity properties of the FD and autocorrelation functions.

The computation of the PSD was accomplished with the help of the Welch method. After subtraction of its mean value, the original sEMG signal was split up into 50% overlapping data segments of the length L ($L = 1024$). The overlapping segments were windowed using a Hamming window. The PSD was then computed as the square magnitude of the discrete Fourier transform of the windowed overlapping segments. Removal of the spikes from the PSD of the sEMG was applied for the first ten spikes (fundamental and 9 harmonics), since no spikes were observed after the 9th harmonic. In order to locate the frequency of the fundamental in the sEMG PSD, the frequency of the fundamental from the vertical acceleration signal was calculated. Therefore, the maximum amplitude within an interval range of ± 4 Hz centered on the frequency provided by the platform setting was searched in the acceleration signal. With the frequency of the acceleration's fundamental, we defined the frequency of the fundamental (F_1) and the 9 harmonics in the sEMG PSD using $F_i = i * F_1$ ($i = 2, \dots, 10$). This procedure was applied for each testing condition and each subject, and the frequency of the acceleration's fundamental measured at the platform level always corresponded to the frequency of the fundamental of the sEMG signals. In a second step, the intervals in $S(f)$ where the spikes spread out were located. In order to account for the spikes' variable width, the following test was performed. If the relative difference was $\frac{|S(f-2\text{ Hz})|^2 - |S(f-4\text{ Hz})|^2}{|S(f-2\text{ Hz})|^2} < 0.1$ for $f = F_1, F_2, \dots, F_{10}$, then $f_{low} = f - 2\text{ Hz}$ was selected as the lower bound of the spike interval, otherwise $f_{low} = f - 4\text{ Hz}$. The corresponding upper bound of the interval was assessed by using the same method with $|S(f + 2\text{ Hz})|^2$ and $|S(f + 4\text{ Hz})|^2$. This procedure guaranteed cancellation of the spikes regardless of their width. The linear interpolation was then computed by replacing the spikes with a straight line between the located intervals and provided the interpolated PSD (**Figure 4.5**, red signal). Finally, the RMS was calculated directly within the PSD using the interpolated PSD $|E(f)|^2$ with the help of the **equation (2)**:

$$sEMG_{RMS} = \sqrt{\frac{1}{L} \sum_{f=0}^{\frac{Fe}{2}} |E(f)|^2} \quad (2)$$

with L being the length of the data to compute the PSD and Fe the sampling frequency.

The RMS calculated in the PSD using **equation (2)** corresponds to the RMS calculated in the time domain using **equation (1)** according to Parseval's theorem. However, to transform the signal into the PSD, Hamming windows were used with an overlap of 50%. This step was crucial, as it smoothed the signal in the PSD, which allowed a more accurate interpolation. Due to the smoothing of the signal, the square root of the power in the PSD resulted in slightly different values than the RMS of the signal in the time domain. In order to verify this difference, the two methods were applied on the unfiltered sEMG signals for all conditions and muscles used in study I. Comparison of the outcomes showed that the discrepancy between the calculation of the RMS in the time domain and the calculation of the square root of the smoothed power in the PSD are minute, with differences (mean \pm SD) of $0.02 \pm 0.10\%$ for the VL, $0.12 \pm 0.14\%$ for the RF, $0.04 \pm 0.16\%$ for the VM, $0.02 \pm 0.48\%$ for the SOL, and $0.10 \pm 0.23\%$ for the GL.

However, besides preserving the maximum of the sEMG information, a further advantage of the linear interpolation method over the other methods is the possibility to calculate the RMS of the isolated excessive spikes $E_s(f)$. This is possible due to the properties of the PSD, since the interpolated sEMG in PSD $E(f)$ can be subtracted from the unfiltered sEMG in the PSD $S(f)$ using the formula $|E_s(f)|^2 = |S(f)|^2 - |E(f)|^2$. Quantifying the magnitude of the spikes and verifying their behavior can be useful to clarify whether these spikes contain stretch reflex responses. For this purpose, the RMS of the isolated spikes $E_s(f)$ was calculated (separately for the fundamental and the harmonics) in study III.

The spectral linear interpolation was used in all the other studies to remove the excessive spikes in the sEMG spectrum. **Figure 4.6** illustrates examples of sEMG signals in the time domain (upper panel) and the PSD (lower panel) for the different platform types that have been used. The unfiltered sEMG signals are illustrated in black, and the interpolated signals in red.

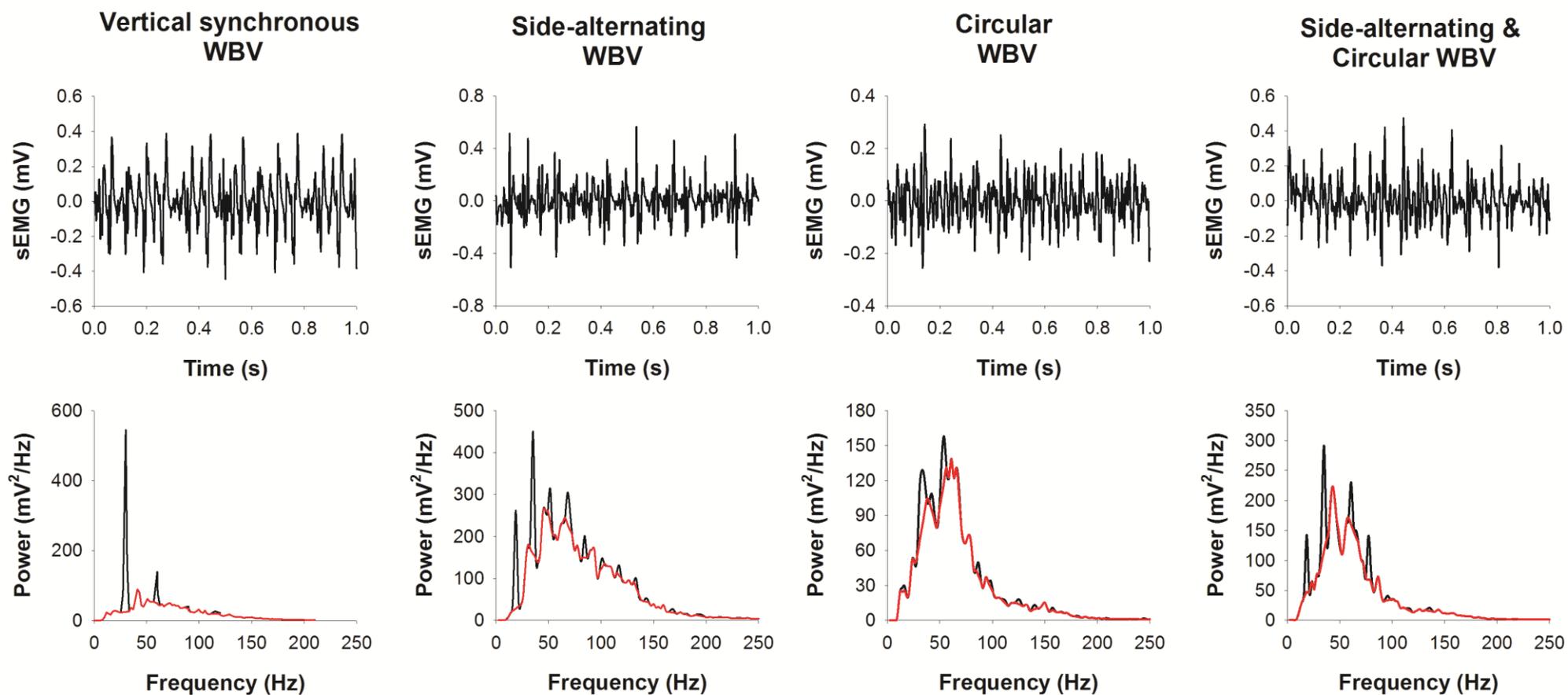


Figure 4.6 Illustration of surface electromyography (sEMG) signals in the time domain (upper panel) and the Power Spectral Density (PSD, lower panel) during vertical synchronous whole-body vibration (WBV, 30 Hz, 1.0 mm amplitude), side-alternating WBV (16 Hz, 4.0 mm amplitude), circular WBV (14 Hz, 0.8 mm amplitude), and the combination of the side-alternating (16 Hz, 4.0 mm amplitude) and the circular WBV (14 Hz, 0.8 mm amplitude). The black and the red signals represent the unfiltered and the interpolated sEMG in the PSD, respectively.

4.5 NORMALIZATION METHODS

As mentioned above, the sEMG magnitude is dependent on many technical, anatomical and physiological factors. Therefore, the magnitude of the raw sEMG should not be compared between subjects and muscles without normalization to an individual value (Burden 2010). In WBV studies, the sEMG_{RMS} is commonly expressed as a percentage of baseline sEMG_{RMS} during standing, or as a percentage of the sEMG_{RMS} during an MVC. While normalization to the sEMG during standing can result in sEMG activity of >200%, it can be more meaningful to use the sEMG recorded from a MVC as the reference value. This indicates the percentage of the maximal capacity that the muscle is working during the task (Allison et al. 1993). One draw-back of normalization to the sEMG recorded from a MVC is that the subjects must be able to generate MVCs that are close to their maximal activation levels. An electrically evoked maximal electrophysiological response (Mmax, or also referred to as M-wave) is an alternative method to express the maximal sEMG activity (Buckthorpe et al. 2012). In study II, normalization to a MVC was compared to the normalization to a Mmax, and in studies I, IV-VI, the sEMG was normalized to the sEMG during a MVC in order to allow sEMG comparison between subjects and muscles.

4.5.1 MAXIMAL VOLUNTARY CONTRACTION

The sEMG during a MVC represents the maximal muscle activity as maximal muscle fibers are recruited in order to produce the highest force possible. It has previously been shown that MVC assessments on the ankle (Webber and Porter 2010) and the knee (Toonstra and Mattacola 2012) are reliable measurements .

In studies I,II, and IV, the MVC torques were assessed on a leg extension machine (Gymstar 900, Carnielli, Italy) and a home-made ankle ergometer for the knee extensors and the plantar flexors, respectively. For studies V&VI, the MVCs were completed for plantarflexion, dorsiflexion, knee extension, and knee flexion on an isokinetic dynamometer (Biodex System 3, Medical Systems Inc., New York, NY, USA).

In order to calculate the $sEMG_{RMS}$ from the MVC trial, the 500 ms period representing the highest torque was first evaluated. Next, the $sEMG_{RMS}$ of the muscle corresponding to the contraction was calculated for the same 500 ms period (**Figure 4.7**). Then, the $sEMG_{RMS}$ during the exercise was expressed as a percentage of the $sEMG_{RMS}$ during the MVC using the following *equation (3)*.

$$MVC \text{ Normalized } sEMG_{RMS} = \frac{sEMG_{RMS} \text{ during WBV}}{sEMG_{RMS} \text{ during MVC}} \times 100 \quad (3)$$

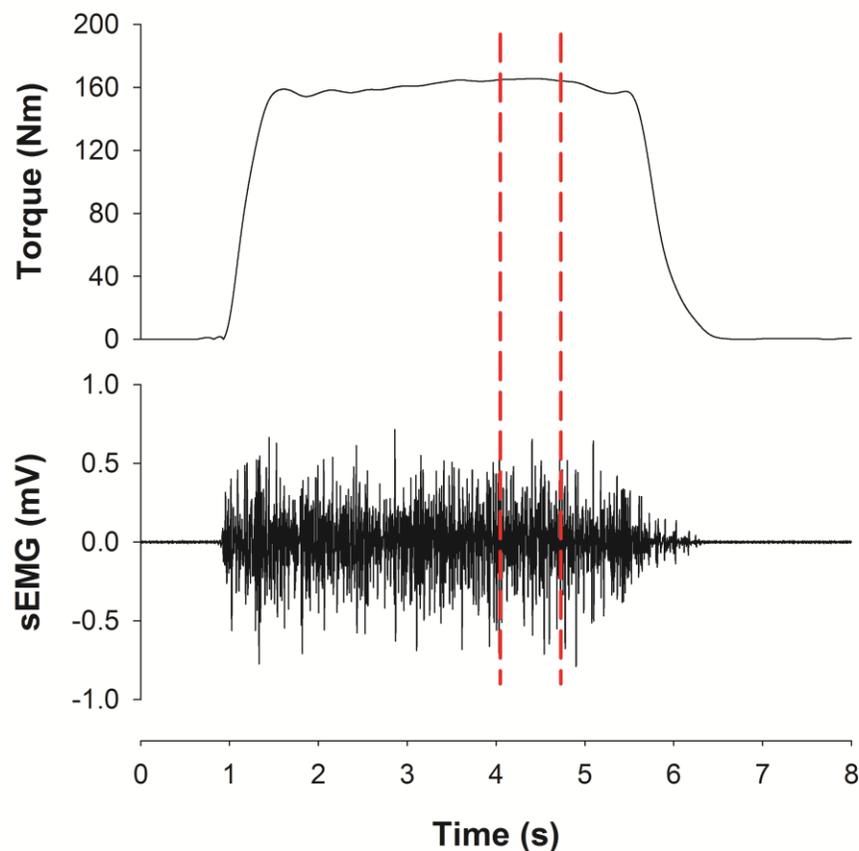


Figure 4.7 Illustration of a maximal isometric voluntary contraction torque during knee extension (upper panel) and the corresponding surface electromyography (sEMG) signal of the Vastus Lateralis (lower panel). Root mean square of the sEMG data was calculated over the 500 ms of the highest torque (red dashed lines).

4.5.2 MMAX

The M-wave is an electrophysiological response caused by action potentials that are evoked in axons of alpha motor neurons in response to an electric (or also magnetic) stimulation applied to the nerve (Enoka 2008). The amplitude of an M-wave comprises the sum of the muscle fiber potentials within the recording volume of the electrodes (Zehr 2002; Tucker et al. 2005; Keenan et al. 2006). The maximal M-wave response (Mmax) has been shown to be reliable (Jenkins et al. 2013).

In study II, M-wave responses of selected lower limb muscles were assessed via percutaneous electrical stimulation of the femoral (for VL, RF, VM, **Figure 4.8**) and tibial nerve (for SOL, GL).

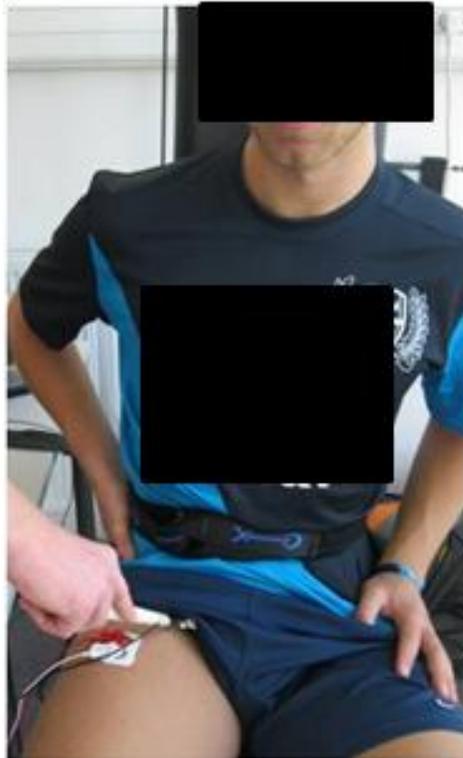


Figure 4.8 Assessment of the M-wave response of the Vastus Lateralis, Rectus Femoris, and Vastus Medialis by percutaneous electrical stimulation of the femoral nerve.

Individual maximal stimulation intensity of each participant was detected during the familiarization trial, by progressively increasing the intensity until consistent peak-to-peak amplitude of the electrophysiological response and unchanged mechanical twitch torque of the knee extensors or plantar flexors were observed. This intensity was further increased by 10% during the assessment to ensure maximal muscle activation (M_{\max}) and was maintained for three continuous single stimulations (1 ms rectangular pulse, 400 V) separated by 4 s. Peak-to-peak amplitude of the three M_{\max} responses (**Figure 4.9**) was calculated) and was averaged for further analyses. Then, the $sEMG_{RMS}$ during the exercise was expressed as a ratio of the averaged M_{\max} response using the following **equation (4)**.

$$M_{\max} \text{ Normalized } sEMG_{RMS} = \frac{sEMG_{RMS} \text{ during } WBV}{M_{\max} \text{ response}} \quad (4)$$

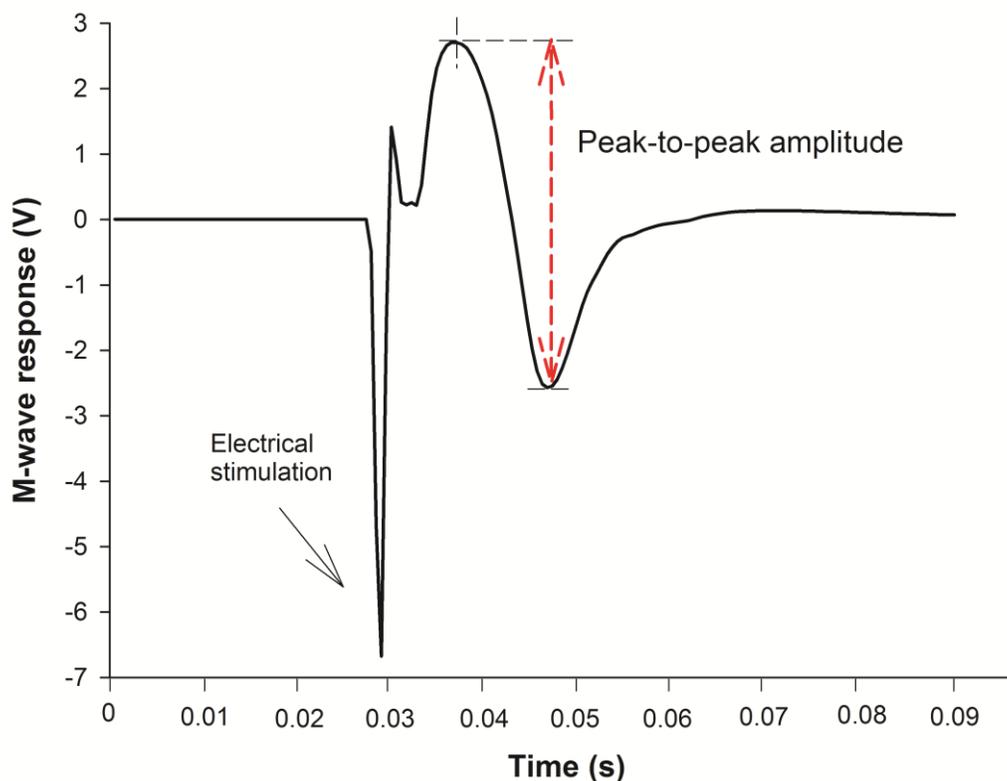


Figure 4.9 M-wave response of the soleus muscle shortly after electrical stimulation of the tibial nerve. Peak-to-peak amplitude of the M-wave response (M_{\max}) was calculated (red dashed lines).

LIMITATIONS

The maximal M-wave amplitude (M_{max}) can change considerably throughout an experiment at different limb positions (Simonsen and Dyhre-Poulsen 1999) or during different levels of muscle contraction (Tucker and Türker 2004). The latter should not have been a limitation to study II, as the M-waves were induced while the subjects were completely relaxed. However, M_{max} can substantially decrease over the period of an experiment even though the subject is at rest (Crone et al. 1999), or from morning to evening sessions (Castaingts et al. 2004). Further, the temperature of the tested muscle (Dewhurst et al. 2005) and reduction of the Na^+/K^+ pump efficiency due to repeated activation of motor units (Nielsen et Clausen 2000) can have an influence on M_{max} .

4.6 MOTION ARTIFACTS RECORDINGS

In study III, motion artifacts were recorded during WBV. Pairs of bipolar silver-electrodes (10-mm diameter, Contrôle Graphique Medical, Brie-Comte-Robert, France) were placed over the left patella in line with the lower limb. Three layers of adhesive tape between the skin and the bipolar electrodes served as isolation of the muscle activity from the electrodes (**Figure 4.10**).



Figure 4.10 Assessment of motion artifacts was accomplished by placing dummy electrodes on the left patella, with several layers of tape between the skin and the electrodes.

In order to assure the validity of measuring artifacts, participants were asked to perform a maximal voluntary knee extension against the static lever arm of a leg-extension machine. During the contraction, the patella signal remained at baseline levels. Hence, we concluded that in case of non-baseline signal recordings on the patella during WBV, the observed signal is due to the shaking of the electrodes, and not due to crosstalk from surrounding muscles. The patella signals were amplified (MP150 Biopac® Systems Inc., Holliston, MA, USA; CMRR = 110 db, Z input = 1000 M Ω , gain = 1000), filtered with a bandwidth frequency ranging from 10 Hz to 500 Hz, and online digitized with a sampling frequency of 2000 Hz.

LIMITATIONS

The insulating tape placed between the electrodes and the skin indeed prevented possible sEMG recordings from surrounding muscles, but at the same time, motion artifacts generated between the electrode and the skin were neglected. The insulating tape raised the resistance between the electrodes to almost infinite, which impeded recordings of electrochemical potentials at the electrode/gel interface, at the gel/skin interface, and between skin layers (Bifulco et al. 2013). However, the cables are another source for the spikes embedded in the sEMG recordings (Webster 1984; Türker 1993; Clancy et al. 2002). Unshielded sEMG cables moved through an ambient magnetic or electric field generate motion artifacts. Shielded sEMG cables can also be a source of error as deformation generates static charges. We acknowledge that in the present study, only the motion artifacts arising from the shielded cables were measured. However, we considered the certainty of excluding crosstalk recordings from surrounding muscles as more important than inclusion of all types of motion artifacts.

4.7 PLATFORM ACCELERATION

Platform acceleration was measured in all studies, as it is important to control the acceleration that is transmitted to the person standing on the platform. The acceleration of the vibration stimuli was assessed with the help of a tri-axial accelerometer (for studies I-IV: 50G, TSD109F, Biopac® Systems Inc., Holliston, MA, USA, sampled at 2000 Hz; for studies V&VI: ADXL 78, Analog Devices USA, encapsulated in a 20/12/5 mm plastic shell, measuring range ± 70 g, frequency response of 0-400 Hz, mass: 2 grams, sampled at 2400 Hz). In studies I-IV, the acceleration signals were filtered using an 18th order band-pass Butterworth filter (Pollock et al. 2010) with a low cut-off frequency of 5 Hz and a high cut-off frequency of 300 Hz. In studies V&VI, the acceleration signals were filtered using a band-pass wavelet filter with a low cut-off frequency of 5 Hz and a high cut-off frequency of 300 Hz. Then, the RMS of each plane was calculated (Abercromby et al. 2007b; Avelar et al. 2013; Cook et al. 2011; Kiiski et al. 2008; Muir et al. 2013; Pel et al. 2009;), as well as the amplitude (displacement from baseline to peak), by averaging the magnitudes of the local maxima and minima within each vibration cycle (Boyer and Nigg 2006; Muir et al. 2013, **Figure 4.11**).

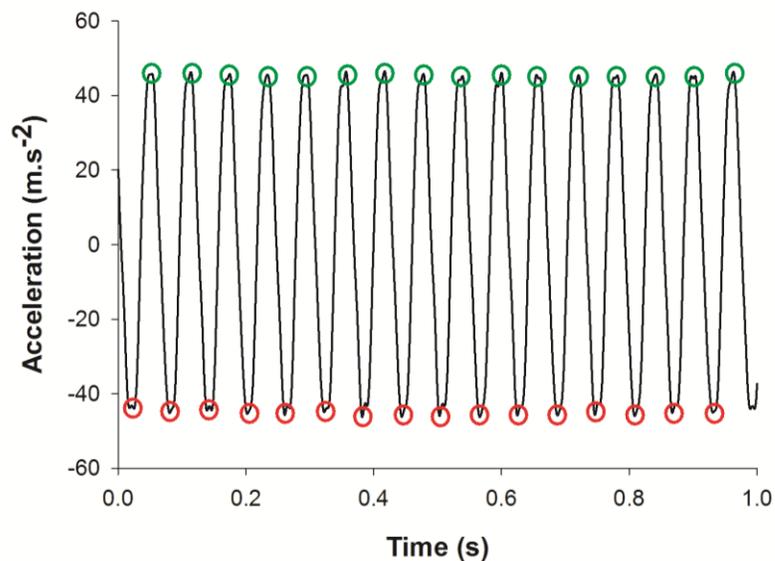


Figure 4.11 Example of a platform acceleration signal. The green and red dots represent the local maxima and minima within each vibration cycle, respectively. The illustrated signal was recorded during side-alternating whole-body vibration at 16 Hz and 4 mm amplitude.

4.7.1 CALCULATION OF PLATFORM DISPLACEMENT

For all studies, the platform acceleration signal was further used to identify the exact displacement of the platform. First, the vertical acceleration signal was double integrated to obtain the displacement signal of the platform. Then, average of the magnitudes of the local maxima and minima within each vibration cycle were computed to calculate the amplitude (displacement from baseline to peak) of the displacement (**Figure 4.12**). The displacement (peak-to-peak amplitude) was reported in studies I-IV, and the amplitude (displacement from baseline to peak) in studies V&VI.

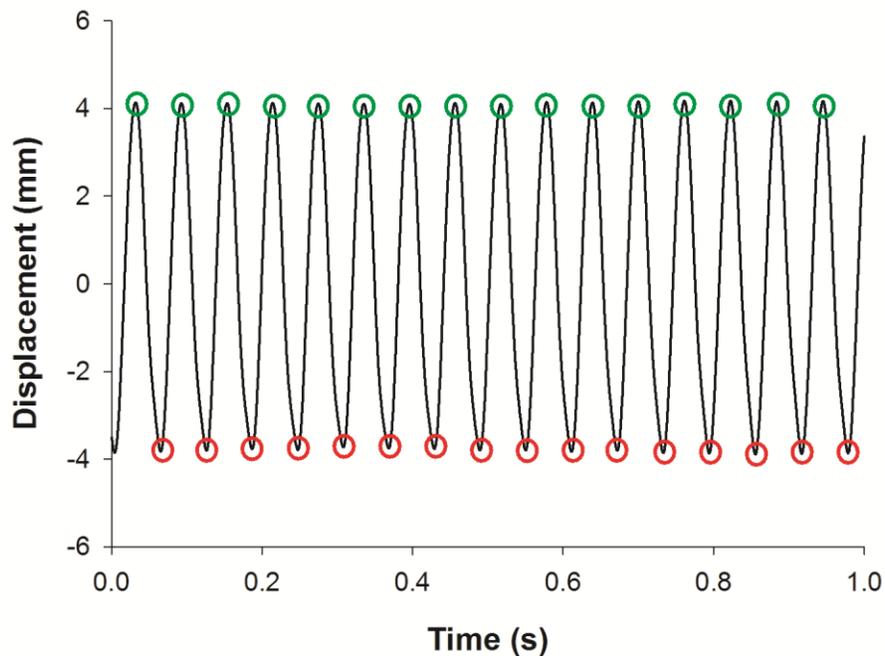


Figure 4.12 Example of a platform displacement signal after double integration of the platform acceleration signal. The green and red dots represent the local maxima and minima within each vibration cycle, respectively. The corresponding acceleration signal was recorded during side-alternating whole-body vibration at 16 Hz and 4 mm amplitude.

4.8 PLATFORM TYPES

In the current thesis, platforms from two different manufacturers were used. For study I-IV, the Power Plate pro6 was applied, whereas for studies V&VI the TBS 100A by Total Image Fitness was utilized (Figure 4.13).



Figure 4.13 The Power Plate pro 6 was used in studies I-IV (left), and the TBS 100A (right) by Total Image Fitness was used in studies V&VI.

The corresponding vibration modes and settings that were applied in the different studies are shown in Table 4.3.

Table 4.3 The platform types, brands, and settings used in studies I-VI.

	<u>Methodological</u>		<u>Explorative</u>	<u>Applied</u>				
	Study I	Study II	Study III	Study IV	Study V		Study VI	
Platform type	Vertical Synchronous	Vertical Synchronous	Vertical Synchronous	Vertical Synchronous	Side-alternating	Circular	Dual mode	Side-alternating
Platform brand	Power Plate	Power Plate	Power Plate	Power Plate	Total Image Fitness	Total Image Fitness	Total Image Fitness	Total Image Fitness
Frequency	30 Hz	25, 30, 35, 40 Hz	30 Hz	25, 30, 35, 40 Hz	6 Hz, 16 Hz	14 Hz, 43 Hz	All combinations	16 Hz
Amplitude	1 mm	1 mm	1 mm	0.6 mm, 1 mm	2.5 mm, 4 mm	0.8 mm		4 mm

The vertical synchronous WBV platform (Power Plate pro6, North America, Northbrook, IL, USA) delivers oscillation in the vertical plane (**Figure 4.14**, left). The frequency setting of this platform ranges between 25 Hz and 40 Hz and is adjustable in steps of 1 Hz. The amplitude (displacement from baseline to peak) settings consist of a low and a high setting, which corresponds to an amplitude of 0.6 mm and 1.0 mm, respectively. Additionally, pressure settings can be manipulated with respect to the WBV user's body mass (setting 1: ≤ 60 kg, setting 2: between 60 and 90 kg, setting 3: ≥ 90 kg). For standardization purposes, the pressure setting 2 was chosen for all subjects, regardless of their body mass.

The second WBV platform (TBS 100A, Total Image Fitness, Calgary, Alberta, Canada) delivers oscillations via single and dual modes. The single modes consist of either side-alternating or circular vibrations, and the dual mode consists of the combination of the two single modes. The side-alternating mode oscillates along the sagittal axis (**Figure 4.14**, middle), with a frequency range of 6 Hz to 16 Hz, adjustable in steps of 1 Hz. The amplitude of the vibration is dictated by the positioning of the feet in relation to the axis of rotation, and was determined prior to the assessments by double integration of the platform acceleration signal. A foot position of 23.3 cm from the central axis corresponded to an amplitude (displacement from baseline to peak) of 4 mm, and a foot position of 14.3 cm corresponded to an amplitude of 2.5 mm.

The circular mode delivers vibrations in the horizontal plane, more specifically in the anterior-posterior and medio-lateral planes (**Figure 4.14**, right). The frequency ranges between 14 Hz and 43 Hz, adjustable in steps of 1 Hz. The amplitude of the circular mode (i.e. the radius of the circular movement) is fixed at 0.8 mm.

The mixed mode is the combination of the side-to-side mode and the circular mode, which are individually adjustable in frequency and amplitude. By using the mixed mode, the WBV platform oscillates along the vertical axis in addition to the circular oscillation in the horizontal plane.

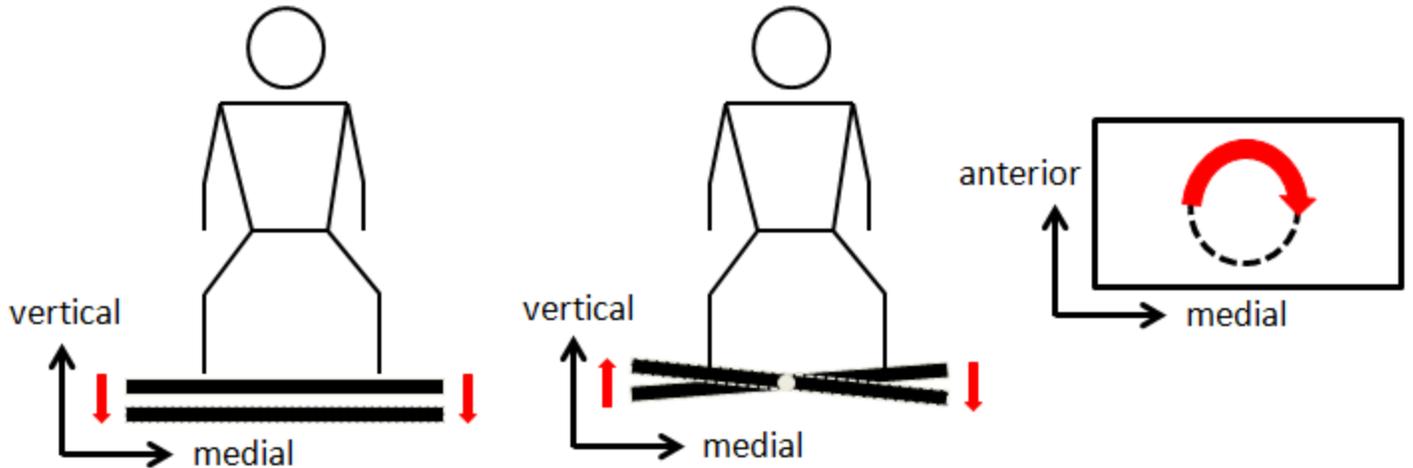


Figure 4.14 Schematic representation of the synchronous vertical vibration mode (left), the side-alternating vibration mode (middle), and the circular vibration mode (right).

The synchronous vertical vibration platform and the side-alternating platform delivered vibrations mainly in the vertical plane, whereas the circular platform delivered accelerations mainly in the horizontal plane (**Figure 4.15**). Dual mode WBV consisting of the side-alternating and the circular mode delivered vibrations in all three planes, with overlapping acceleration phases.

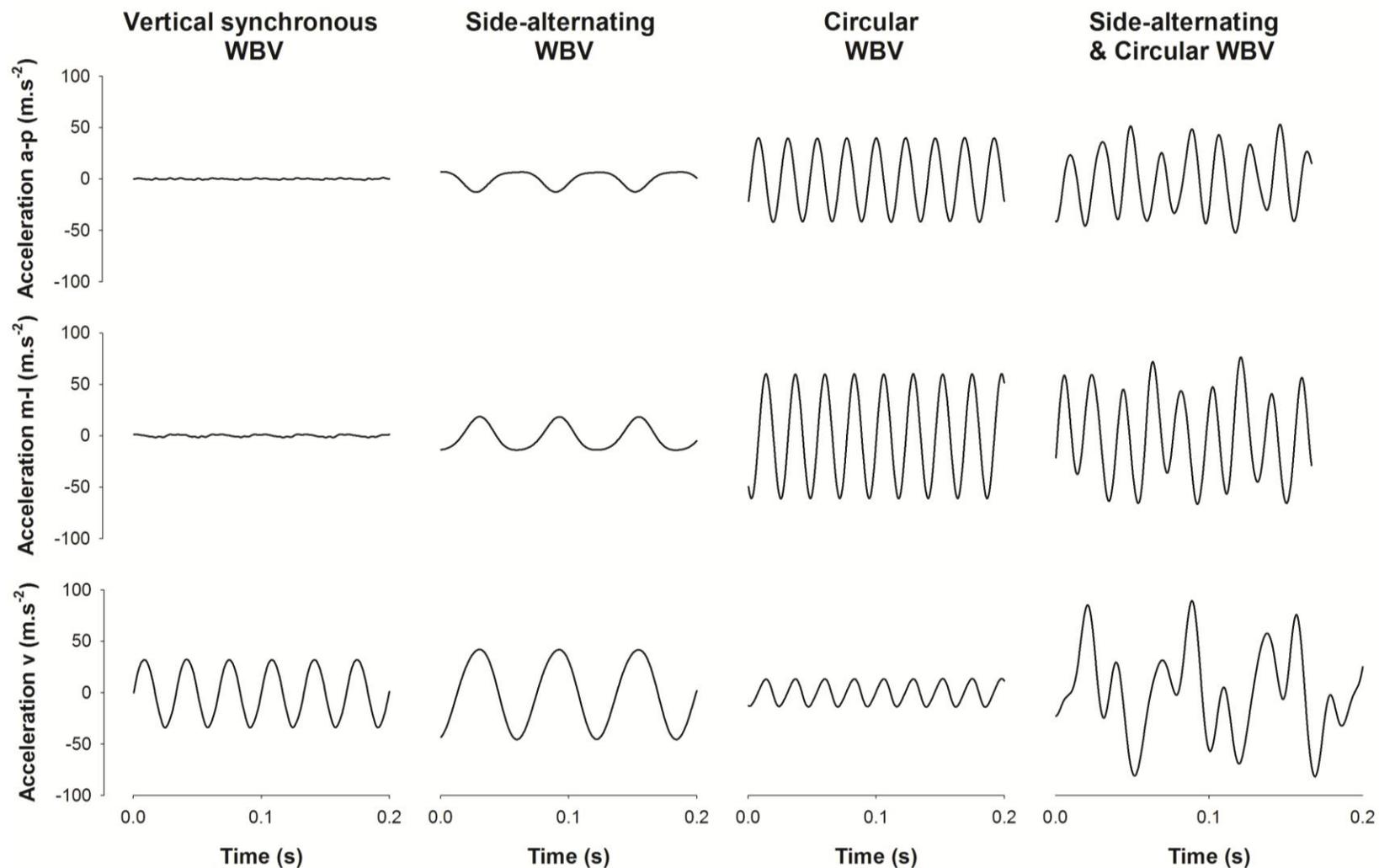


Figure 4.15 Illustration of platform acceleration signals in the anterior-posterior (a-p), medio-lateral (m-l), and vertical (v) plane during vertical synchronous whole-body vibration (WBV, 30 Hz, 1.0 mm amplitude), the side-alternating WBV (16 Hz, 4.0 mm amplitude), circular WBV (43 Hz, 0.8 mm amplitude), and the combination of the side-alternating and the circular WBV.

Table 4.4 illustrates the measured peak accelerations in the anterior-posterior (a-p), medio-lateral (m-l), and vertical (v) plane for the different platform types, frequencies, and amplitudes that have been used.

Table 4.4 Overview of the vibration settings that have been used and their resulting acceleration amplitudes in all three planes (mean \pm SD).

	Vertical Mode		Side-alternating Mode		Circular Mode		Acceleration		
	Freq (Hz)	Amp (mm)	Freq (Hz)	Amp (mm)	Freq (Hz)	Amp (mm)	a-p (m.s ⁻²)	m-l (m.s ⁻²)	v (m.s ⁻²)
Power Plate pro6	25	0.6	-	-	-	-	0.6 \pm 0.2	0.5 \pm 0.4	17.8 \pm 0.8
	30	0.6	-	-	-	-	0.9 \pm 0.3	0.6 \pm 0.5	20.3 \pm 0.7
	35	0.6	-	-	-	-	0.6 \pm 0.3	1.3 \pm 0.7	27.5 \pm 1.3
	40	0.6	-	-	-	-	1.0 \pm 0.5	1.7 \pm 1.2	35.6 \pm 1.7
	25	1.0	-	-	-	-	1.0 \pm 0.3	0.9 \pm 1.1	29.7 \pm 1.2
	30	1.0	-	-	-	-	1.3 \pm 0.5	1.7 \pm 2.1	33.8 \pm 2.0
	35	1.0	-	-	-	-	1.7 \pm 1.0	3.6 \pm 4.8	47.1 \pm 2.9
	40	1.0	-	-	-	-	2.2 \pm 0.7	4.4 \pm 5.1	59.4 \pm 4.4
Total Image Fitness TBS 100A	-	-	6	2.5	-	-	0.5 \pm 0.1	2.7 \pm 0.2	4.2 \pm 0.4
	-	-	16	4.0	-	-	15.6 \pm 4.4	14.6 \pm 2.1	50.5 \pm 5.7
	-	-	-	-	14	2.0	6.6 \pm 0.6	6.7 \pm 0.4	0.9 \pm 0.4
	-	-	-	-	43	2.0	45.4 \pm 6.5	59.6 \pm 3.9	13.1 \pm 4.9
	-	-	6	2.5	14	2.0	6.8 \pm 0.3	8.3 \pm 0.8	4.4 \pm 0.6
	-	-	6	2.5	43	2.0	46.7 \pm 7.2	60.6 \pm 5.7	17.5 \pm 5.9
	-	-	16	4.0	14	2.0	16.6 \pm 3.4	14.6 \pm 1.8	51.9 \pm 4.5
	-	-	16	4.0	43	2.0	57.4 \pm 6.8	71.6 \pm 4.6	71.8 \pm 7.8

Amp, Amplitude; a-p, anterior-posterior; Freq, Frequency; m-l, medio-lateral; v, vertical.

4.9 ACCELERATION TRANSMISSION

In study VI, the vertical acceleration was measured on the head. Therefore, 1D accelerometers (ADXL 78, Analog Devices USA, encapsulated in a 20/12/5 mm plastic shell, measuring range ± 35 g, frequency response of 0-400 Hz, mass = 2 grams) measuring the vertical acceleration were positioned on the middle of the forehead (**Figure 4.16**).



Figure 4.16 Graphic representation of the assessment of the head acceleration. A 1D accelerometer was positioned on the middle of the forehead.

The signals were pre-amplified at the source and sampled at 2400 Hz. The signals were filtered using a band-pass wavelet filter with a low cut-off frequency of 5 Hz and a high cut-off frequency of 300 Hz. Then, the average of the magnitudes of the local maxima and minima within each vibration cycle were computed to calculate the amplitude (displacement from baseline to peak) of each acceleration signal. Examples of acceleration signals measured on the body are shown in **Figure 4.17**. In order to obtain transmissibility values, the acceleration (Acc) amplitude (Amp) of each bony structure was divided by the amplitude of the vertical platform acceleration using the following *equation (5)*.

$$\text{Transmissibility} = \frac{\text{Amp Acc}_{\text{body}}}{\text{Amp Acc}_{\text{platform}}} \quad (5)$$

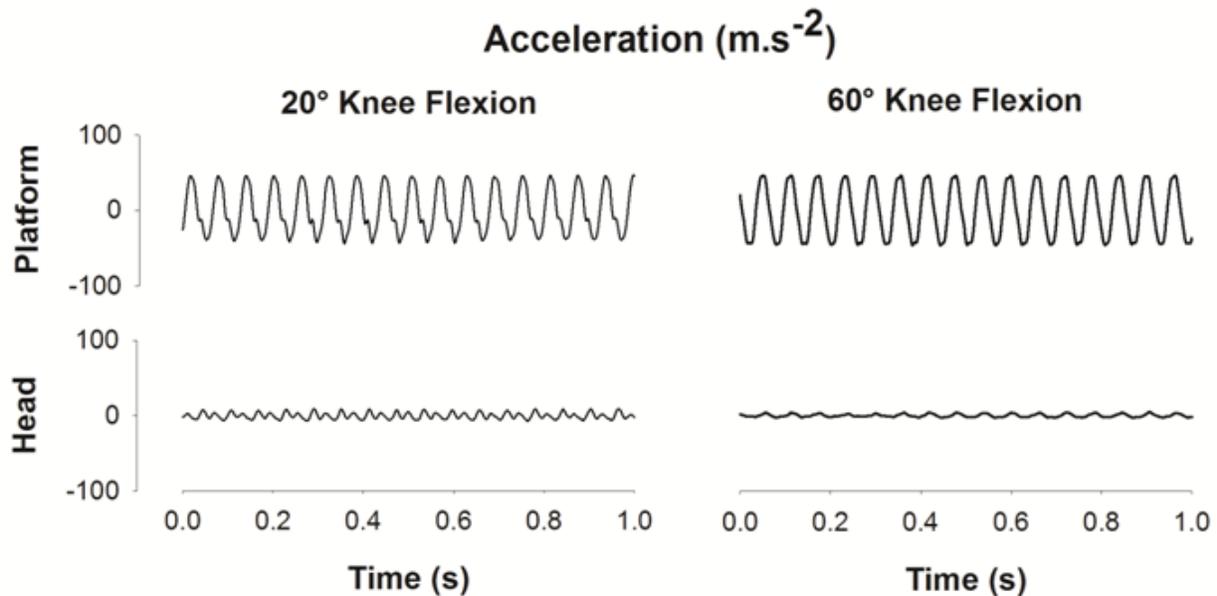


Figure 4.17 Examples of acceleration signals measured on the platform and the head during side-alternating whole-body vibration at 16 Hz and 4 mm amplitude. The left panel illustrates the acceleration signals during a static squat at 20° knee flexion (0° corresponding to fully extended) and the right panel during a static squat at 60° knee flexion.

LIMITATIONS

There are some limitations of the method that was used to assess the acceleration on the bony structures. First, the acceleration signal was only measured in the vertical direction. Although the used platform type (side-alternating) delivers vibrations mainly in the vertical plane (Pel et al. 2009), tri-axial analysis might be needed to more accurately represent the complexities of vibration transmission in the human body (Cook et al. 2011). Second, skin-mounted accelerometers were used, which can allow movements of the soft tissue and skin relative to the bone beneath and may affect the acceleration detected by the accelerometers. Bone-mounted accelerometers are a more accurate method to measure acceleration on bony structures (Rubin et al. 2003), but are associated with the concerns of invasive assessments. Furthermore, it has been shown that there is only minimal overestimation (<10%) of accelerations with the use of skin-mounted accelerometers compared to bone pins (Nokes et al. 1984; Kim et al. 1993).

4.10 BODY POSTURE

For studies I-VI, the knee flexion angle was measured and controlled (**Figure 4.18**) with the help of an electrical twin-axis goniometer (studies I-IV: 150 MM – TSD130B, Biopac® Systems Inc., Holliston, MA, USA; studies V&VI: SG150 Biometrics Ltd, Newport, UK). In study I-IV, the subjects were asked to perform a static squat at a knee angle of 70° (0° corresponding to fully extended). In study V, the static squat was performed at a knee angle of 60°, and in study VI at 20° and 60°. In studies I-IV, the researcher had to ensure that the participant was performing the squat at the desired knee angle. In studies V&VI, the participants were provided with a visual real-time feedback of their knee angle, illustrated on a laptop screen using Matlab software. The trial was repeated if the measured knee angle varied more than $\pm 3^\circ$ of the desired angle.

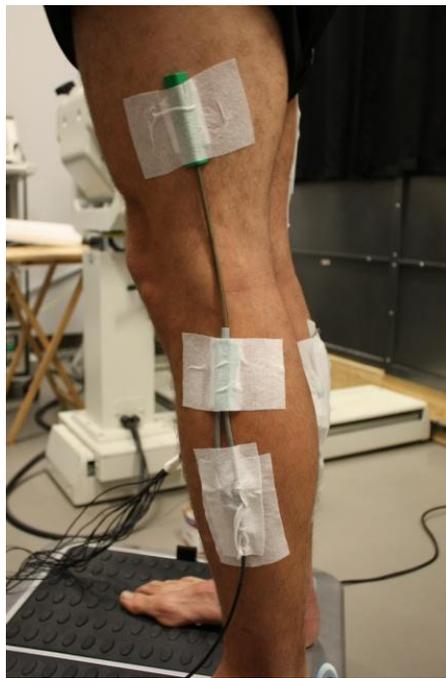


Figure 4.18 The knee flexion angle was measured and controlled using an electrical twin-axis goniometer.

LIMITATIONS

There are some limitations regarding the monitoring of the participants postures. Only the knee flexion angle was measured, although it is well known that leaning forward and backward has an influence on the muscle activity of the lower limb muscles. For example, during knee flexion, the TA acts to shift the centre of gravity forward in order to maintain the posture (Dionisio et al. 2008; Kasai et Kawai 1994). Shifting the line of gravity by leaning forwards or backwards could change the muscle activity required by the TA to maintain balance. As ankle and hip flexion angle and plantar pressure distribution were not monitored in any of the studies, this may have been a confounding factor as the subjects may have used different strategies to maintain their posture during squatting. However, controlling ankle, knee, and hip flexion angle with concomitant visual or audio feedback would overstrain the participants' abilities to control all three joint angles.

CHAPTER 5

STUDY I: sEMG PROCESSING

5.1 PREFACE

The first methodological study was designed to investigate the effect of the sEMG processing method on the quantification of the muscle activity during WBV. This investigation was indispensable as this issue had never been addressed before. The measurements for this study were completed in spring 2012, at the LAMHESS laboratory of the University of Nice Sophia Antipolis. A young healthy cohort was recruited and sEMG activity was measured of two lower limb muscles with and without WBV and with and without an additional load. All sEMG signals were freed from motion artifacts by applying the state-of-the-art methods (band-stop filter, band-pass filter) as well as the newly developed method based on spectral linear interpolation. For comparison reasons, the unfiltered sEMG was also included in the analysis. Although no motion artifacts occur without the use of WBV, the vibration free sEMG signals were filtered nonetheless. Analysis of the no-vibration trial was an important step, as only the unfiltered sEMG during the no-vibration trial depicts the true muscle activity. Hence, the unfiltered sEMG during the no-vibration trial served as a performance measure for the other methods during the same trial, and their RMS magnitudes were compared with the actual muscle activity (i.e. the unfiltered sEMG). The use of an additional load was important to display the properties of the band-stop filter; its systematic bias increased with an increasing background muscle activity. In summary, this study showed that the spectral linear interpolation method provided the closest estimation of the real sEMG activity during the no-vibration trial as well as during WBV exercises, provided that the spikes in the sEMG spectrum are considered as motion artifacts. These results were essential for the subsequent studies, and all following sEMG signals during WBV were processed with the spectral linear interpolation method. This study is currently submitted to the *Journal of Computer Methods in Biomechanics and Biomedical Engineering*.

5.2 THE INFLUENCE OF SEMG PROCESSING ON THE QUANTIFICATION OF NEUROMUSCULAR ACTIVITY DURING WHOLE-BODY VIBRATION

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Abstract

The purpose was to investigate the influence of sEMG processing methods on the quantification of muscle activity during WBV exercises. sEMG activity was recorded while the participants squatted on the platform with and without vibration. The excessive spikes observed in the sEMG spectrum at the vibration frequency and its harmonics were deleted using (1) a band-stop filter, (2) a band-pass filter, and (3) spectral linear interpolation. The same filtering regime was applied on the sEMG during the no-vibration trial. The linear interpolation method showed the highest intraclass correlation coefficients (no vibration: 0.999, WBV: 0.757 - 0.979) with the true muscle activity, followed by the band-stop filter (no vibration: 0.929 - 0.975, WBV: 0.661 - 0.938). While both methods introduced a systematic bias ($P < 0.001$), the error increased with increasing mean values to a higher degree for the band-stop filter. The band-pass filter was in low agreement with the other methods (ICC: 0.207 – 0.697). Spectral linear interpolation should be applied to delete the artifacts in the sEMG signals during WBV, as this method kept more of the myoelectric information and introduced a lower systematic bias than the band-stop filter or the band-pass filter.

Introduction

Muscle activity has been found to be positively affected by WBV, measured by means of sEMG recordings (Abercromby et al. 2007a; Cardinale and Lim 2003). Similarly to the TVR, stretch reflex responses induced by activation of muscle spindles and alpha-motoneurons have been proposed as the underlying mechanism of the observed sEMG increases (Bosco et al. 1999b; Ritzmann et al. 2010). At the same time, sEMG recordings during WBV are contaminated by motion artifacts, displayed as excessive signals (referred to as spikes) in the power spectrum at the oscillation frequency and, to a lesser degree, its multiple harmonics (Abercromby et al. 2007a; Fratini et al. 2009a; Sebik et al. 2013). As there is no consensus on how to process sEMG signals during WBV, different sEMG processing methods have been applied.

In studies where the spikes were considered as mostly stretch reflex responses, no filter was applied (Roelants et al. 2006; Perchthaler et al. 2013; Ritzmann et al. 2013). In contrast, when the spikes were considered as mostly motion-induced artifacts, many studies have used a band-stop filter centred at the vibration frequency and its harmonics to delete the spikes (Abercromby et al. 2007a; Fratini et al. 2009a; Pollock et al. 2010). Another method to withdraw the spikes is to delete the entire signal in the frequency range where the spikes may have occurred using a band-pass filter (Hazell et al. 2007; 2010). As both of these filters delete parts of the signal that account for the actual muscle activity, a new method based on spectral linear interpolation was developed in order to keep the maximum of the myoelectric information. However, to date it is unclear how the sEMG processing method influences the quantification of the sEMG activity measured during WBV exercises.

Therefore, the aim of this study was to test the performance of the different sEMG processing methods (no filter, band-pass filter, band-stop filter, spectral linear interpolation) on the quantification of the muscle activity during WBV. It was hypothesized that (1) the newly developed linear interpolation method would be a valid method to delete the excessive spikes and (2) introduce the lowest systematic bias.

Methods

Participants

Eight female and ten male physically active students (age, 23.8 ± 3.2 years; height, 172.9 ± 8.3 cm; body mass, 67.8 ± 10.9 kg) volunteered to participate in this study. The inclusion criteria consisted of regular participation in sporting activities, and the exclusion criteria included frequent practice of WBV and/or a neurological impairment of the lower extremity. The study was conducted according to the Helsinki Declaration (1964) and the local Ethics Committee gave approval for this study. All participants provided informed consent prior to their participation.

sEMG activity

Pairs of bipolar silver-chloride electrodes (10-mm diameter, Contrôle Graphique Medical, Briec-Comte-Robert, France) were positioned over the muscle belly of the VL and the SOL of the right lower limb to record sEMG activity. The longitudinal axes of the electrodes were placed in line with the presumed direction of the underlying muscle according to the SENIAM recommendations (Hermens et al. 2000), with a centre-to-centre distance of 20 mm. The reference electrode was attached to the left patella. Low interelectrode resistance (< 5 k Ω) was obtained by means of shaving and abrading the skin with emery paper and cleaning the skin with alcohol. sEMG signals were amplified (MP150 Biopac[®] Systems Inc., Holliston, MA, USA; CMRR = 110 db, Z input = 1000 M Ω , gain = 1000), filtered with a bandwidth frequency ranging from 10 Hz to 500 Hz, and online digitized with a sampling frequency of 2000 Hz.

MVC assessment

Isometric MVCs were assessed on a leg extension machine (Gymstar 900, Carnielli, Italy) and a home-made ankle ergometer for the VL and the SOL, respectively. For the MVC measurement of the VL, the participants were seated on the leg extension machine with a hip joint angle of 90° (0°: hip fully extended). The lower leg was attached to the lever arm with a strap 2 to 3 cm proximal to the lateral malleolus with a knee joint angle of 70° (0°: knee fully extended), which was controlled using an electrical twin-axis goniometer (TSD130B, Biopac[®] Systems Inc., Holliston, MA, USA). MVC torques during knee extension and plantar flexion were recorded

using a force transducer (TSD121C, Biopac[®] Systems Inc., Holliston, MA, USA) that was secured to the respective machine. For the MVC measurements of the plantar flexors, the participants were seated on a chair with their right foot attached to the pedal of the ankle ergometer. Knee joint angle was kept at 70°, and the ankle joint at an angle of 0° (neutral position). All participants performed two 5-s MVCs separated by a 45-s rest period. MVC peak torque (i.e., the highest torque plateau over 500 ms) was evaluated using Matlab software (version 7.13 The Mathworks, Inc., Natick, MA, USA). Root mean square values of the sEMG ($sEMG_{RMS}$) were calculated for the trial with the highest MVC torque over the selected period.

WBV trials

A vertical synchronous platform (Power Plate[®] Pro6, Northbrook, IL, USA) was used to deliver the vibrations, with a frequency of 30 Hz and an amplitude (displacement from baseline to peak) of 1.0 mm. The subjects performed 4 x 30-s static squats (no load/no vibration, no load/with WBV, with load/no vibration, with load/with WBV) which were presented in a randomized order. The additional load was applied via a standard weightlifting bar with weight plates resulting in a total weight of 33 kg. The bar was positioned on the participant's shoulders for the trials with additional loading. The participants were asked to flex their knees at a knee angle of 70°. Knee angles during the squats were monitored using an electrical twin-axis goniometer (150 MM - TSD130B, Biopac[®] Systems Inc., Holliston, MA, USA). Subsequent analysis revealed that the participants performed their squats at a knee angle of $68.5 \pm 1.3^\circ$ (mean \pm SD). The acceleration of the vibration stimuli was controlled with the help of a tri-axial accelerometer (50G, TSD109F, Biopac[®] Systems Inc., Holliston, MA, USA) that was placed on the platform aligned with the third toe. The RMS of the vertical acceleration (Pel et al. 2009) over the middle 15 s was 2.4 ± 0.1 g. During the exercises, the participants were barefoot in order to avoid any damping effects due to different footwear.

sEMG signal processing methods

sEMG signals obtained during WBV were processed using different sEMG processing methods (Figure 5.1), followed by the calculation of the RMS and normalization to the sEMG recorded during the MVC. sEMG analysis and the calculation of the RMS were completed in the time domain for the no-filter method, the band-stop and the band-pass filter. For the interpolation method, sEMG analysis and RMS calculation were accomplished in the frequency domain. The sEMG signals during the no-vibration trial were processed according to the same filtering regime as during WBV.

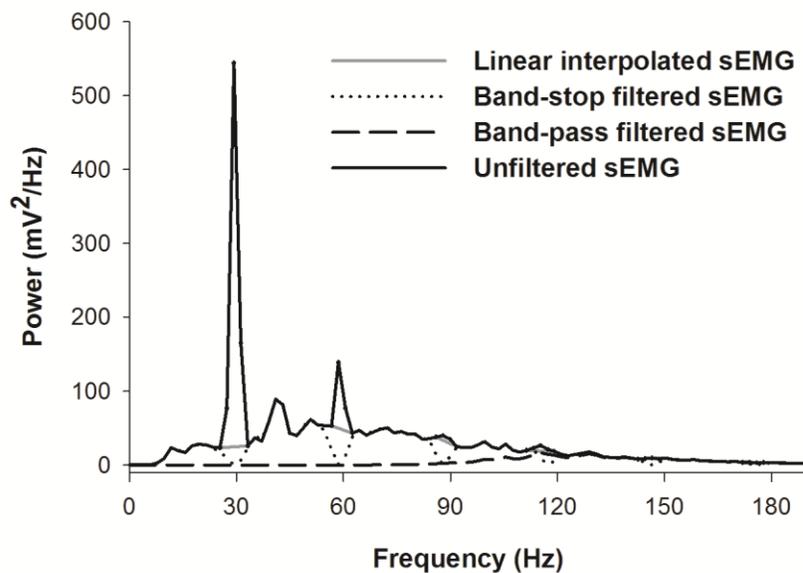


Figure 5.1 Example of a surface electromyography (sEMG) spectrum of the Vastus Lateralis during whole-body vibration at 30 Hz. sEMG signals were processed using the no-filter method (black solid line), linear interpolation (grey solid line), band-stop filter (grey dotted line) and band-pass filter (black dashed line).

Interpolation method

The aim of the proposed method was to implement a filtering technique that removes only the spikes from the sEMG signals the PSD. Unlike other time-domain methods of the state-of-the-art (band-stop filter, band-pass filter), we assumed that only the excessive part of the spikes should be removed in order to preserve the maximum of the sEMG information at the oscillating frequency and its harmonics. As a consequence, the proposed interpolation method does not allow a time-reversal transformation, and the RMS needs to be calculated directly within the PSD. sEMG signals recorded during WBV contain sEMG activity that is not related to the spikes ($emg(t)$), and additional activity ($emg_s(t)$) that accounts for the spikes. Consequently, in the time domain, the entire signal ($signal(t)$) can be expressed as: $signal(t) = emg(t) + emg_s(t)$. Transforming $signal(t)$, $emg(t)$, and $emg_s(t)$ into the FD results in $S(f)$, $E(f)$, and $E_s(f)$, which are complex numbers. Thus, direct subtraction of the spikes $E_s(f)$ from the entire signal $S(f)$ is not possible in the FD. However, with the help of the squared modulus (i.e., the PSD), modeling the sum of $E(f)$ and $E_s(f)$ allows this subtraction in the frequency domain using the equation $|E(f)|^2 = |S(f)|^2 - |E_s(f)|^2$ assuming that the data are independent and using the linearity properties of the FD and autocorrelation functions.

The computation of the PSD was accomplished with the help of the Welch method. After subtraction of its mean value, the original sEMG signal was split up into 50% overlapping data segments of the length L ($L = 1024$). The overlapping segments were windowed using a Hamming window. The PSD was then computed as the square magnitude of the discrete Fourier transform of the windowed overlapping segments. Removal of the spikes from the PSD of the sEMG was applied for the first ten spikes (fundamental and 9 harmonics), since no spikes were observed after the 9th harmonic. In order to locate the frequency of the fundamental in the sEMG PSD, the frequency of the fundamental from the vertical acceleration signal was calculated. Therefore, the maximum amplitude within an interval range of ± 4 Hz centred on the frequency provided by the platform setting was searched in the acceleration signal. With the frequency of the acceleration's fundamental, we defined the frequency of the fundamental (F_1) and the 9 harmonics in the sEMG PSD using $F_i = i * F_1$ ($i = 2, \dots, 10$). This procedure was applied for

each testing condition and each subject, and the frequency of the acceleration's fundamental measured at the platform level always corresponded to the frequency of the fundamental of the sEMG signals. In a second step, the intervals in $S(f)$ where the spikes spread out were located. In order to account for the spikes' variable width, the following test was performed. If the relative difference was $\frac{|S(f-2\text{ Hz})|^2 - |S(f-4\text{ Hz})|^2}{|S(f-2\text{ Hz})|^2} < 0.1$ for $f = F1, F2, \dots, F10$, then $f_{low} = f - 2\text{ Hz}$ was selected as the lower bound of the spike interval, otherwise $f_{low} = f - 4\text{ Hz}$. The corresponding upper bound of the interval was assessed by using the same method with $|S(f + 2\text{ Hz})|^2$ and $|S(f + 4\text{ Hz})|^2$. This procedure guaranteed cancellation of the spikes regardless of their width. The linear interpolation, which was the last step of the procedure, was computed by replacing the spikes with a straight line between the located intervals and provided the interpolated PSD.

RMS values of the filtered sEMG were computed in order to compare the different methods. As the interpolation method does not permit time-reversal transformation to the frequency domain, the interpolated RMS was calculated directly within the PSD. Due to the Parseval's theorem, the RMS calculated in the time domain corresponds to the RMS calculated in the frequency domain. Therefore, after removing the spikes, $sEMG_{RMS}$ values were calculated using

the interpolated PSD $|E(f)|^2$ with the help of the formula: $sEMG_{RMS} = \sqrt{\frac{1}{L} \sum_{f=0}^{\frac{Fe}{2}} |E(f)|^2}$, where L was the length of the data to compute the PSD and Fe was the sampling frequency.

Band-stop filter

The band-stop filtering regime was introduced by Abercromby et al. (Abercromby et al. 2007a), whereas sEMG signals were filtered using a 17th-order Chebyshev type II. The order of the filter was calculated with the choice of a stop-band of ± 1 Hz, a pass-band of ± 1.5 Hz, a minimum stop-band attenuation of 100 dB, and a maximum pass-band ripple of 0.01 dB. The filter was centered at the frequency of the fundamental and the harmonics up to 450 Hz. After processing the sEMG signals with the band-stop filter in the time domain, calculation of the $sEMG_{RMS}$ was

accomplished using the formula $\sqrt{\frac{1}{n} \sum_{t=1}^n signal(t)^2}$, where n was the length of the data.

Band-pass filter

The band-pass filtering sEMG regime employed by Hazell et al. (Hazell et al. 2007; 2010) retains only the high frequency signals (100 - 450 Hz). This approach was based on a scientific study that showed that high frequency sEMG signals above 140 Hz improved biceps brachii strength estimates compared to common high pass cutoffs in the range of 10 – 30 Hz (Potvin and Brown 2004). Thus, sEMG signals were filtered in the time domain using a dual passed 6th order Butterworth band-pass filter between 100 and 450 Hz with subsequent calculation of the sEMG_{RMS} via the same equation as for the band-stop filter.

sEMG normalization

After calculating sEMG_{RMS} during WBV and during the no-vibration trial, the values were normalized to the sEMG_{RMS} obtained during the MVCs according to the formula

$$\frac{\text{sEMG}_{\text{RMS}} \text{ during vibration}}{\text{sEMG}_{\text{RMS}} \text{ during MVC}} \times 100.$$

Statistical analysis

Statistical analyses were conducted in IBM SPSS software (Version 20, Chicago, IL). The Kolmogorov-Smirnov test confirmed the normality of the data. Concurrent validity between the unfiltered, interpolated, band-stop, and band-pass filtered sEMG_{RMS} was analyzed using intraclass correlation coefficient (ICC(2,1)), 95% limits of agreement (%LOA) by Bland and Altman (Bland and Altman 1986), and paired t-test to detect any systematic difference (bias) between the sEMG processing methods. Statistical significance was set at $\alpha = 0.05$.

Results

The $sEMG_{RMS}$ (mean \pm SD) for the different processing methods during the no-vibration trial and during WBV with and without load are shown in **Table 5.1**.

Table 5.1 Root mean square of the surface electromyography ($sEMG_{RMS}$) normalized to the $sEMG_{RMS}$ recorded during a maximal voluntary contraction (MVC) of the Vastus Lateralis (VL) and the Soleus (SOL) with and without whole-body vibration (WBV) and with and without load using different processing methods (mean \pm SD).

		No load		With load	
		No vibration	WBV	No vibration	WBV
VL	No Filter	25.4 \pm 9.9	39.9 \pm 21.9	39.4 \pm 16.3	56.9 \pm 34.8
	Interpolation	25.0 \pm 9.7	27.1 \pm 11.4	38.5 \pm 15.8	40.1 \pm 15.9
	Band-stop	21.9 \pm 8.4	23.8 \pm 9.8	33.8 \pm 13.9	35.5 \pm 13.9
	Band-pass	10.2 \pm 3.9	12.2 \pm 5.1	16.3 \pm 6.5	18.3 \pm 6.7
SOL	No Filter	7.7 \pm 3.6	11.1 \pm 4.8	20.1 \pm 10.1	27.2 \pm 17.5
	Interpolation	7.4 \pm 3.4	9.3 \pm 4.0	19.4 \pm 10.1	23.5 \pm 14.2
	Band-stop	6.4 \pm 2.7	8.2 \pm 3.5	17.4 \pm 8.8	21.2 \pm 12.9
	Band-pass	4.3 \pm 2.0	5.3 \pm 2.2	11.7 \pm 5.5	13.8 \pm 8.3

For the no-vibration trial, the no-filter method showed good concurrent validity with the band-stop filter (ICC \geq 0.929, %LOA \leq 34.1) and the interpolation method (ICC = 0.999, %LOA \leq 3.5), with higher ICC's and lower %LOAs for the latter method (**Table 5.2**). The difference between the no-filter method and the interpolation method (**Figure 5.2A/5.3A**) and the band-stop filter (**Figure 5.2B/5.3B**) increased with increasing respective mean values (heteroscedasticity) to a higher degree for the band-stop filter than the interpolation method. Further, a significant systematic bias ($P < 0.001$, **Figure 5.2**) was observed which was higher between the no-filter method and the band-stop filter (No load: VL: 3.56 \pm 1.80; SOL: 1.27 \pm 1.20; with load: VL: 5.51 \pm

2.45; SOL: 2.71 ± 1.40) than between the no filter method and the interpolation method (No load: VL: 0.31 ± 0.27 ; SOL: 0.10 ± 0.12 ; with load: VL: 0.83 ± 0.58 ; SOL: 0.37 ± 0.29).

During WBV, concurrent validity between the no-filter method and the interpolation method ($ICC \geq 0.758$, $\%LOA \leq 83.7$) and the band-stop filter ($ICC \geq 0.661$, $\%LOA \leq 96.6$) was lower than for the no-vibration trial. A significant systematic bias ($P < 0.001$) was observed which was higher between the no-filter method and the band-stop filter (No load: VL: 16.15 ± 13.26 ; SOL: 2.88 ± 1.64 ; with load: VL: 21.37 ± 22.33 ; SOL: 6.08 ± 4.84) than between the no filter method and the interpolation method (No load: VL: 12.89 ± 11.93 ; SOL: 1.77 ± 1.35 ; with load: VL: 16.91 ± 20.27 ; SOL: 3.74 ± 3.69).

Good concurrent validity was found between the interpolation method and the band-stop filter for the no-vibration trial ($ICC \geq 0.947$, $\%LOA \leq 29.0$) as well as during WBV ($ICC \geq 0.972$, $\%LOA \leq 13.5$). On the contrary, the band-pass filter showed low concurrent validity with the other methods (**Table 5.2**).

Table 5.2 Concurrent validity between the surface electromyography processing methods during the no-vibration trial and during whole-body vibration (WBV) with and without load for the Vastus Lateralis (VL) and the Soleus (SOL).

			No load				With load			
			No vibration		WBV		No vibration		WBV	
			ICC	%LOA	ICC	%LOA	ICC	%LOA	ICC	%LOA
VL	No Filter	Interpolation	0.999	2.1	0.757	71.2	0.999	3.0	0.758	83.7
	No Filter	Band-stop	0.955	15.2	0.653	83.1	0.962	13.4	0.661	96.6
	No Filter	Band-pass	0.341	74.3	0.227	154.9	0.357	80.9	0.207	163.7
	Interpolation	Band-stop	0.965	12.3	0.972	12.5	0.972	10.8	0.973	11.8
	Interpolation	Band-pass	0.350	72.0	0.423	75.6	0.370	78.5	0.379	75.3
	Band-stop	Band-pass	0.429	62.7	0.511	65.6	0.451	71.0	0.460	66.0
SOL	No Filter	Interpolation	0.999	3.5	0.939	26.5	0.999	2.8	0.974	29.1
	No Filter	Band-stop	0.929	34.1	0.858	34.2	0.975	14.6	0.938	40.0
	No Filter	Band-pass	0.660	61.5	0.466	75.6	0.697	62.9	0.679	93.7
	Interpolation	Band-stop	0.947	29.0	0.972	13.5	0.982	17.0	0.990	12.5
	Interpolation	Band-pass	0.678	57.1	0.591	62.4	0.715	65.9	0.778	67.4
	Band-stop	Band-pass	0.796	40.7	0.695	54.7	0.803	52.1	0.843	57.0

ICC, intraclass correlation coefficient; LOA, limits of agreement.

Vastus Lateralis

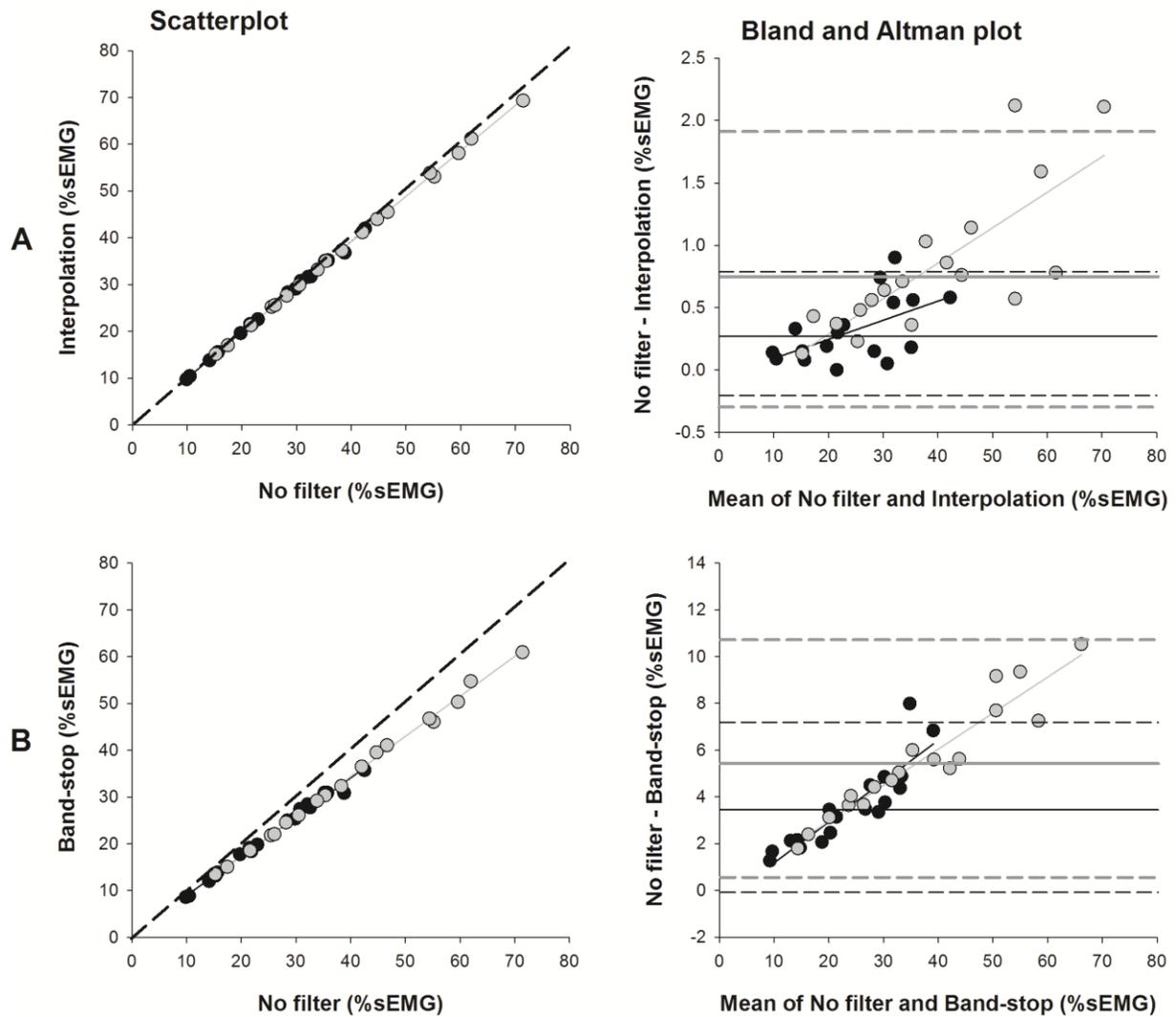


Figure 5.2 Scatterplots and Bland–Altman plots of (A) the no-filter method and the interpolation method and (B) the no-filter method and the band-stop filter of the surface electromyography (sEMG) from the Vastus Lateralis during the no-vibration trial without load (in black) and with load (in grey). On the scatter plots, the dashed line represents the identity line. On the Bland–Altman plots, the solid line represents the bias and the dashed lines represent the upper and lower 95% LOA.

Soleus

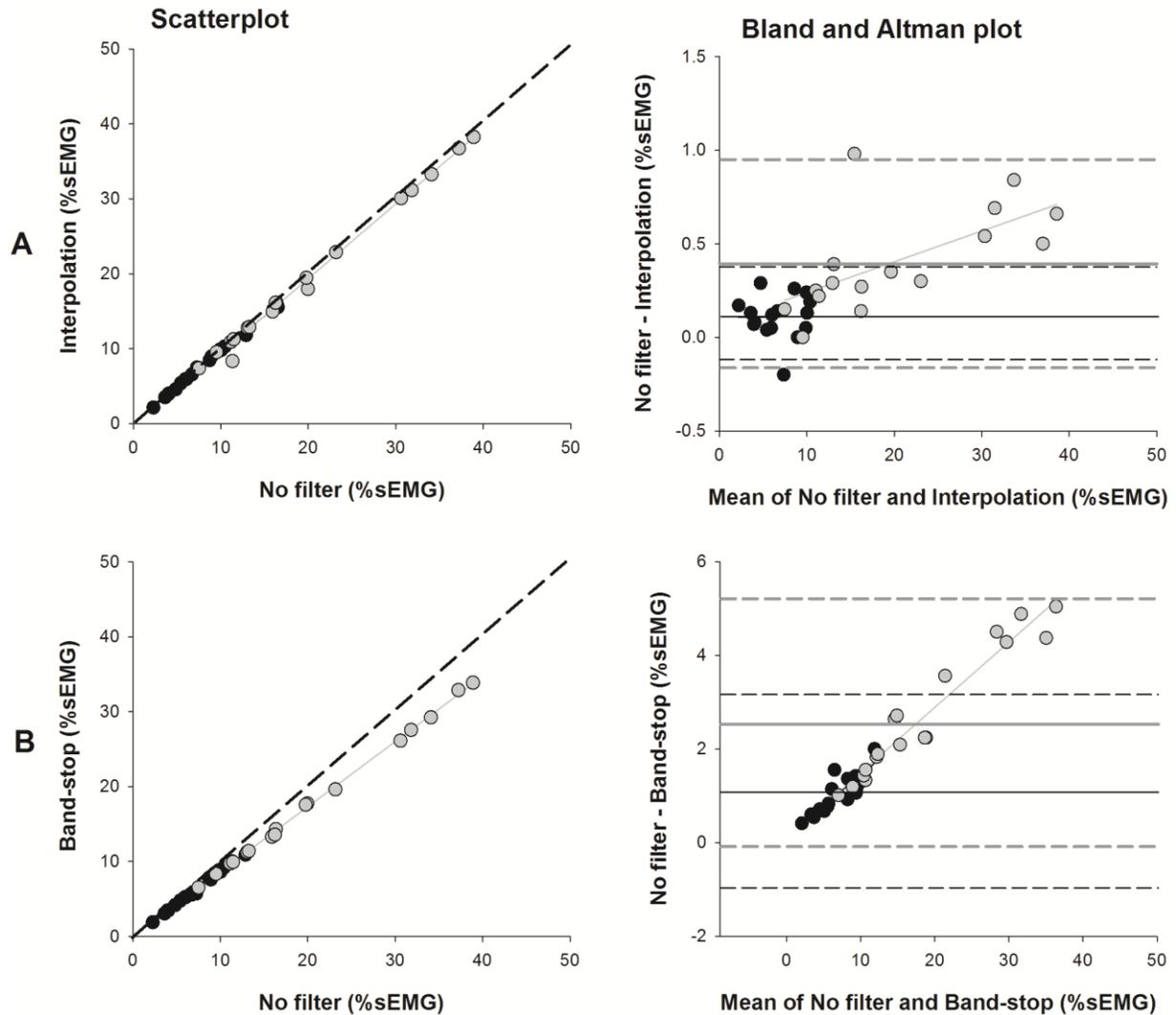


Figure 5.3 Scatterplots and Bland–Altman plots of (A) the no-filter method and the interpolation method and (B) the no-filter method and the band-stop filter of the surface electromyography (sEMG) from the Soleus during the no-vibration trial without load (in black) and with load (in grey). On the scatter plots, the dashed line represents the identity line. On the Bland–Altman plots, the solid line represents the bias and the dashed lines represent the upper and lower 95% LOA.

Discussion

The main findings of this study were that during the no-vibration trial, the no-filter method showed higher concurrent validity with the interpolation method than with the band-stop filter. A significant systematic bias was found for both methods, whereas the bias was higher and increased more with increasing mean values for the band-stop filter than the interpolation method. Comparing the interpolation method with the band-stop filter, it was found that they were in high agreement with and without WBV. Finally, the band-pass filter showed low concurrent validity with the other methods.

During WBV, the unfiltered sEMG is not a good criterion measure, as sEMG signals are contaminated by motion artifacts (Abercromby et al. 2007a; Fratini et al. 2009a; Sebik et al. 2013). In fact, it is unknown to which extent the spikes' magnitude are motion artifacts or reflex activity (Ritzmann et al. 2010; Pollock et al. 2012; Sebik et al. 2013). On the contrary, the unfiltered sEMG during the no-vibration trial is a good criterion measure, as it provides the true amount of the muscle activity. Therefore, comparing the filtering methods to the no-filter method during the no-vibration trial can be used to highlight the individual properties of each processing method. The characteristics of each processing method can then be used to describe their performance during WBV.

Although high ICCs and low %LOAs were found between the no filter method and the interpolation method and between the no filter method and the band-stop filter, a systematic bias was introduced which increased proportionally to the respective mean values. This was the case to a higher extent for the band-stop filter than for the interpolation method, which is not surprising. Using a band-stop filter, the depth of the notches in the frequency spectrum is dependent on the magnitude of the sEMG background signal. Hence, the higher the background sEMG, the greater the error committed by the band-stop filter. This was clearly illustrated with the use of an additional load during the no-vibration trial, as the systematic bias with the band-stop filter was higher with the load (VL: 5.51 ± 2.45 ; SOL: 2.71 ± 1.40) compared to without the load (VL: 3.56 ± 1.80 ; SOL: 1.27 ± 1.20). Although true muscle activity during WBV is unknown, it

can be concluded that the band-stop filter progressively underestimates the sEMG signal during WBV the higher the background muscle activity. Therefore, comparing sEMG signals during WBV over a wide range of sEMG background activity might lead to wrong conclusions. This can be crucial when comparing baseline sEMG activity to the sEMG during WBV with additional loads (Hazell et al. 2010; Ritzmann et al. 2013). Therefore, the use of the linear interpolation method should be favoured as the bias introduced by this method was minute and increased minimally with increasing background sEMG activity.

The band-pass filter was generally in low agreement with the other methods. This finding can be explained by the enormous loss in sEMG activity induced by the band-pass filter, as the entire frequency spectrum below 100 Hz was deleted. Nevertheless, depending on the frequency of the platform, band-pass filtered signals can still include spikes from higher harmonics. However, the band-pass filtering regime with subsequent sEMG rectification has been successful to evidence motor unit synchronization during WBV (Sebik et al. 2013).

Taken as a whole, it was found that the interpolation method and the band-stop filter are in high agreement during WBV, whereas the no-filter method and the band-pass filter are in medium to low agreement with the other methods. Therefore, these methods cannot be used interchangeably, and caution must be taken when comparing outcomes of studies using different sEMG processing methods. For example, in the present study, the VL muscle was working at ~24% of its maximum during WBV using the band-stop filter, whereas by using the no-filter method, sEMG activity of ~40% was found.

Conclusions

In order to delete the motion artifacts contained in the sEMG signals during WBV, it is recommended to use spectral linear interpolation or a band-stop filter centered at the vibration frequency and its multiple harmonics. The use of a band-stop filter can be suboptimal in study set-ups where sEMG activity over a wide range is compared, as the band-stop filter introduced a substantial bias that increased with increasing background muscle activity.

CHAPTER 6

STUDY II: sEMG NORMALIZATION

6.1 PREFACE

The second methodological study was designed to investigate the effect of the sEMG normalization method on the quantification of the muscle activity during WBV. Although it is well established that quantitative comparison of the raw sEMG magnitude between subjects and between muscles should be avoided (De Luca 1997), numerous WBV studies have reported non-normalized sEMG magnitudes (Abercromby et al. 2007a; Avelar et al. 2013; Cardinale and Lim 2003; Di Giminiani et al. 2013; Krol et al. 2011; Marín et al. 2009). Others have normalized the sEMG during WBV to the sEMG during the MVC or to the baseline sEMG during a no-vibration trial (see **CHAPTER 3**). As it was unknown how the normalization method affects the outcomes on the muscle activity levels during WBV exercises, this second methodological study was designed. The assessments for this study were included in the measurements for study I and were completed in spring 2012, at the LAMHESS laboratory of the University of Nice Sophia Antipolis. sEMG activity in eighteen participants was measured of the VL and the Sol during WBV at 25, 30, 35 and 40 Hz and during a no-vibration trial. The motion artifacts that occurred during WBV were deleted in the sEMG spectrum using spectral linear interpolation, as study I identified this method as more suitable than the band-stop or the band-pass filter. Then, the RMS of the sEMG was calculated and normalized to the maximal M-wave response (Mmax), to the MVC, and to the baseline sEMG during the no-vibration trial. The outcomes of this study showed that the sEMG during WBV can be normalized to the Mmax, to the MVC, and to the baseline sEMG, because the coefficients of variation (CV) were similar between the methods and because high and significant correlations were found between them. However, advantage of normalizing the sEMG during WBV to the sEMG during the MVC is the indication of the required muscle activity as a percentage of its maximum. For this reason, the sEMG during WBV was normalized to the sEMG during the MVC in all the other studies, with the exception of study III. The outcomes of this study served as preliminary results and have not been submitted to a Journal.

6.2 DOES THE sEMG NORMALIZATION METHOD HAVE AN INFLUENCE ON THE QUANTIFICATION OF MUSCLE ACTIVITY DURING WHOLE-BODY VIBRATION EXERCISE?

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Abstract

The purpose was to investigate the influence of sEMG normalization methods on the quantification of muscle activity during WBV exercises. First, Mmax responses via percutaneous electrical stimulation as well as MVCs were assessed of the VL and the Sol. Then, sEMG activity of the same muscles was recorded while the participants squatted on the platform without vibration (baseline) and with WBV at four frequencies (25, 30, 35, 40 Hz). The motion artifacts observable during WBV were deleted in the sEMG spectrum using spectral linear interpolation, and the root mean square of the sEMG was calculated and normalized to the Mmax, MVC, and baseline sEMG. The lowest CVs were found for the baseline normalization method (VL: 12-19%; SOL: 57-67%), followed by the MVC (VL: 38-42%; SOL: 44-60%) and the Mmax normalization method (VL: 48-56%; SOL: 46-62%). Further, high and significant correlations were found between the normalization methods (VL: $r = [0.91:0.94]$, $P < 0.001$; SOL: $r = [0.62:0.87]$, $P < 0.001$). This study showed that the sEMG during WBV exercises can be normalized to the Mmax, to the MVC, and to the baseline sEMG. However, advantage of normalizing the sEMG during WBV to the sEMG during the MVC is the indication of the required muscle activity as a percentage of its maximum.

Introduction

The effectiveness of WBV exercises is speculated to be a result of increased muscle activity during the exposure to the vibration. Indeed, many studies have demonstrated higher activity levels during WBV as compared to baseline values by measuring the sEMG (Cardinale and Lim 2003; Cochrane et al. 2009; Di Giminiani et al. 2013; Hazell et al. 2010; Pollock et al. 2010; Ritzmann et al. 2013; Roelants et al. 2006). Despite the numerous studies investigating sEMG activity during WBV, no study thus far has investigated the influence of the sEMG normalization method on the quantification of the muscle activity.

The absolute sEMG signal is influenced by numerous non-physiological factors that are unrelated to the level of muscle activation (Farina et al. 2010). Therefore, normalization of the sEMG signal is crucial to reduce between-subject and intramuscular variability. Nevertheless, several studies reported and compared the sEMG non-normalized in Volts (Abercromby et al. 2007a; Avelar et al. 2013; Cardinale and Lim 2003; Cochrane et al. 2009; Di Giminiani et al. 2013; Krol et al. 2011; Marín et al. 2009; 2012b). In other WBV studies, the sEMG has been expressed as a percentage of the baseline sEMG_{RMS} during standing (%baseline, Marín et al. 2012a; Ritzmann et al. 2013; Tankisheva et al. 2013), or as a percentage of the sEMG during a maximal isometric voluntary contraction (%MVC, Hazell et al. 2007; 2010; Perchthaler et al. 2013; Pollock et al. 2010; Roelants et al. 2006). Although normalization to the MVC is broadly used (Lehman and McGill 1999) and advocated (Burden 2010), normalization to an electrically evoked maximal electrophysiological response (M_{max}), also referred to as M-wave, has been suggested as an alternative method. This normalization method has been recently recommended for sEMG activity recorded during maximal voluntary contractions (Buckthorpe et al. 2012). Nevertheless, to the best of our knowledge, sEMG activity during WBV has never been normalized to M_{max} responses.

Therefore, the aim of this study was to investigate the influence of the sEMG normalization method (M_{max} , MVC, baseline sEMG) on the quantification of sEMG activity during WBV.

Methods

Participants

Eight female and ten male physically active students (age, 23.8 ± 3.2 years; height, 172.9 ± 8.3 cm; body mass, 67.8 ± 10.9 kg) volunteered to participate in this study. The inclusion criteria consisted of regular participation in sporting activities, and the exclusion criteria included frequent practice of WBV and/or a neurological impairment of the lower extremity. The study was conducted according to the Helsinki Declaration (1964) and the local Ethics Committee gave approval for this study. All participants provided informed consent prior to their participation.

Experimental Design

First, the subjects performed a standardized 5-min warm-up on a stationary bike. Second, Mmax responses of the VL and SOL were measured. Third, MVC for knee extension and plantarflexion were assessed while recording the sEMG of the VL and SOL, respectively. Then, baseline sEMG of the VL and SOL was measured while the subjects were squatting at a knee angle of 70° (0° corresponding to full extension). Last, WBV was imposed during the same squatting position at four different frequencies (25, 30, 35, 40 Hz). **Figure 6.1** represents the timeline of the assessments.

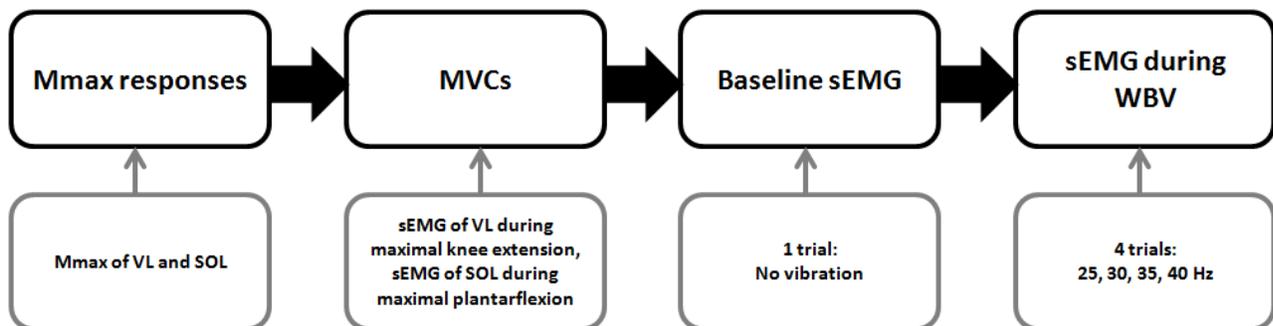


Figure 6.1 Timeline of the assessments that were made in this study. First, Mmax responses were measured of the Vastus Lateralis (VL) and Soleus (SOL), followed by the assessment of maximal voluntary contractions (MVCs) of the same muscles. Surface electromyography (sEMG) was recorded during a no-vibration trial and four whole-body vibration (WBV) trials.

sEMG activity

Pairs of bipolar silver-chloride electrodes (10-mm diameter, Contrôle Graphique Medical, Briec-Comte-Robert, France) were positioned over the muscle bellies of the VL and SOL of the right lower limb to record sEMG activity. The longitudinal axes of the electrodes were placed in line with the presumed direction of the respective underlying muscle according to the SENIAM (Surface Electromyography for the Non-Invasive Assessment of Muscles) recommendations (Hermens et al. 2000), with a centre-to-centre distance of 20 mm. The reference electrode was attached to the left patella. Low interelectrode resistance ($< 5 \text{ k}\Omega$) was obtained by means of shaving and abrading the skin with emery paper and cleaning the skin with alcohol. sEMG signals were amplified (MP150 Biopac[®] Systems Inc., Holliston, MA, USA; CMRR = 110 db, Z input = 1000 M Ω , gain = 1000), filtered with a bandwidth frequency ranging from 10 Hz to 500 Hz, and online digitized with a sampling frequency of 2000 Hz.

MVC Assessment

Isometric MVCs were assessed on a leg extension machine (Gymstar 900, Carnielli, Italy) and a home-made ankle ergometer for the knee extensors and the plantar flexors, respectively. For the MVC measurement of the knee extensor muscles, the participants were seated on the leg extension machine with a hip joint angle of 90° (0°: hip fully extended) and were fixed with a non-elastic belt across their pelvis to minimize body movements during leg extension. The lower leg was attached to the lever arm with a strap 2 to 3 cm proximal to the lateral malleolus with a knee joint angle of 70° (0°: knee fully extended), which was controlled using an electrical twin-axis goniometer (TSD130B, Biopac[®] Systems Inc., Holliston, MA, USA). MVC torques during knee extension and plantar flexion were recorded using a force transducer (TSD121C, Biopac[®] Systems Inc., Holliston, MA, USA) that was secured to the respective machine. For the MVC measurements of the plantar flexors, the participants were seated on a chair with their right foot attached to the pedal of the ankle ergometer. Knee joint angle was kept at 70°, and the ankle joint at an angle of 0° (neutral position). During both, knee extension and plantar flexion MVCs, participants were instructed to fold their arms in front of their chests. All participants performed two 5-s MVCs separated by a 45-s rest period. In the event of an increase or a

decrease in strength of more than 10% between the two MVCs, a third trial was conducted. MVC peak strength (i.e., the highest torque plateau over 500 ms) was evaluated using Matlab software (version 7.13 The Mathworks, Inc., Natick, MA, USA). sEMG_{RMS} were calculated for the trial with the highest MVC strength over the selected period in the time domain.

M_{max} responses

Electrically evoked contractions were induced by a constant current electrical stimulator (DS7AH, Digitimer Ltd, Hertfordshire, England). The femoral and the tibial nerves were stimulated using a monopolar cathode ball (8 mm diameter). A rectangular electrode (Stimex, Wetzlar, Germany) served as the anode and was located, respectively, in the gluteal fold and over the inferior part of the patella. Since the individual maximal stimulation intensity was detected during the familiarization trial, the intensity was rapidly relocated (i.e., the intensity was progressively increased until consistent peak-to-peak amplitude of the electrophysiological response and unchanged mechanical twitch torque of the knee extensors or plantar flexors were observed). This intensity was further increased by 10% to ensure maximal muscle activation (M_{max}) and was maintained for three continuous single stimulations separated by 4 s. The participants were asked to remain relaxed during the stimulations (1 ms rectangular pulse, 400 V). Individual intensities ranged between 60 and 150 mA for the knee extensors and between 60 and 110 mA for the plantar flexors. Peak-to-peak amplitude of the three M_{max} responses was calculated using Matlab software and was averaged for further analyses.

WBV intervention

A vertical synchronous platform (Power Plate[®] Pro6, Northbrook, IL, USA) was used to deliver vibration to the participants. The frequencies provided by the platform ranged from 25 to 40 Hz; frequencies in steps of 5 Hz were selected (25, 30, 35, 40 Hz). The amplitude (displacement from baseline to peak) of this platform was 1.0 mm. The subjects performed 20-s isometric semi-squats while they were asked to flex their knees at a knee angle of 70°. Knee angles during the squats were monitored using an electrical twin-axis goniometer (150 MM - TSD130B, Biopac[®] Systems Inc., Holliston, MA, USA). Subsequent analysis revealed that the participants performed their squats at a knee angle of $68.5 \pm 1.3^\circ$ (mean \pm SD). The acceleration of the vibration stimuli

was controlled with the help of a tri-axial accelerometer (50G, TSD109F, Biopac® Systems Inc., Holliston, MA, USA) that was placed on the platform aligned with the third toe (Lorenzen et al. 2009). RMS of the vertical acceleration (Pel et al. 2009) over the middle 15 seconds of each exercise unit were calculated, identifying values of 2.1 ± 0.1 , 2.4 ± 0.1 , 3.3 ± 0.2 , and 4.3 ± 0.3 g at vibration with 25, 30, 35, and 40 Hz, respectively. During the exercises, the participants were barefoot in order to avoid any damping effects due to different footwear. In total, each participant conducted 5 x 30 s isometric squats which consisted of 4 x 30 s with vibration (frequencies: 25, 30, 35, 40 Hz; peak-to-peak displacement: 2 mm) and 1 x 30 s without vibration (i.e., baseline).

sEMG analysis

sEMG signals during WBV are contaminated by motion artifacts at the vibration frequency and its multiple harmonics. Therefore, the excessive spikes were deleted using linear interpolation in the PSD. Then, the $sEMG_{RMS}$ was calculated directly within the PSD over the middle 15 seconds of each vibration exercise.

sEMG normalization

After calculating $sEMG_{RMS}$ during WBV, the values were normalized to the maximal M_{max} responses, to the $sEMG_{RMS}$ obtained during the MVCs, and to the baseline $sEMG_{RMS}$. Normalization to the maximal M_{max} responses was achieved by dividing WBV $sEMG_{RMS}$ values by the averaged value of the three M_{max} responses on individual basis for each recorded muscle

$sEMG_{RMS_Mmax} = \frac{sEMG_{RMS}}{M_{max}}$. Normalization to the $sEMG_{RMS}$ of the MVC was assessed according

to the formula: $sEMG_{RMS_MVC} = \frac{sEMG_{RMS}}{sEMG_{MVC}} \times 100$. Normalization to the baseline $sEMG_{RMS}$ was

computed using the formula: $sEMG_{RMS_Baseline} = \frac{WBV\ sEMG_{RMS}}{Baseline\ sEMG_{RMS}} \times 100$. In order to calculate

the increase in sEMG, baseline sEMG was subtracted from the sEMG during the WBV trials for the Mmax and MVC normalization methods. For the normalization to baseline, a value of 100 was subtracted to express the %increase due to WBV. **Figure 6.2** illustrates the normalization procedures for Mmax, MVC, and Baseline sEMG.

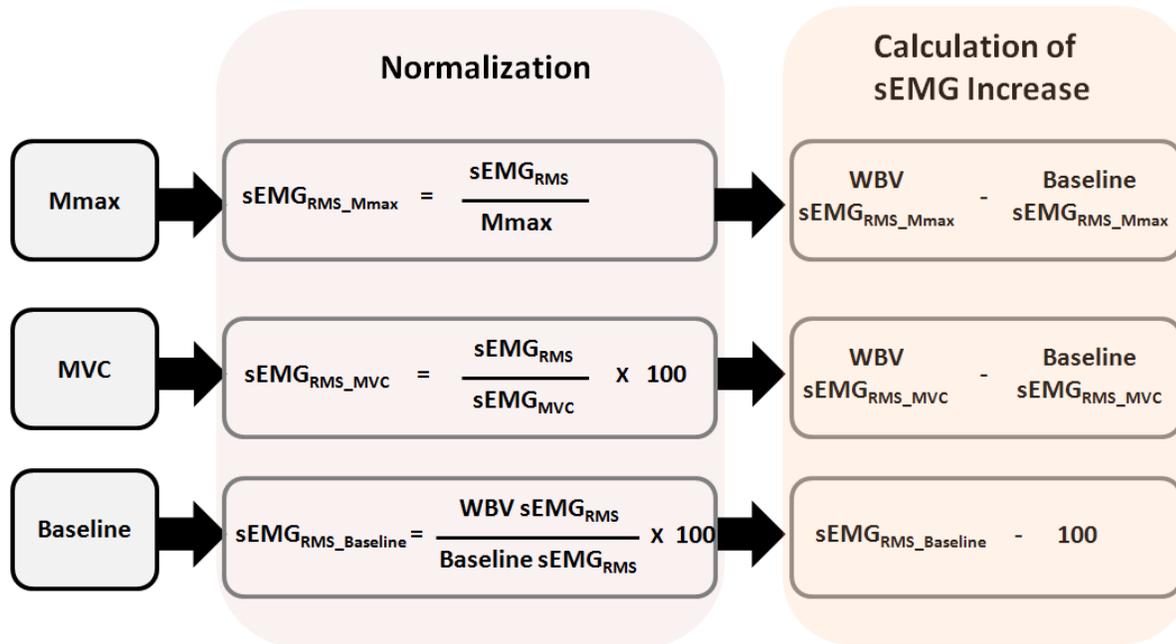


Figure 6.2. Schematic representation of the normalization procedures.

Statistical analysis

Statistical analyses were conducted in Statistica (Statsoft, version 7.0 Tulsa, OK, USA). CVs of the $sEMG_{RMS}$ were calculated for each normalization method and each vibration frequency. Additionally, Pearson's correlation coefficients were calculated for the increase in $sEMG_{RMS}$ between the three normalization methods, with all vibration conditions collapsed to one group. The results are presented as mean \pm SD and the significance level was set at $\alpha = 0.05$.

Results

The sEMG_{RMS} pattern over the four vibration frequencies is illustrated in **Figure 6.3** for each normalization method. Comparing the CV between the normalization methods, the lowest values were found for the baseline normalization method (VL: 12-19%; SOL: 57-67%), followed by the MVC normalization method (VL: 38-42%; SOL: 44-60%) and the Mmax normalization method (VL: 48-56%; SOL: 46-62%). Finally, high and significant correlations were found between the methods. Between the Mmax and the MVC method, the r was 0.94 for the VL ($P < 0.001$) and 0.87 for the SOL ($P < 0.001$). Between the Mmax and the Baseline method, the r was 0.91 for the VL ($P < 0.001$) and 0.62 for the SOL ($P < 0.01$). Between the MVC and the Baseline method, the r was 0.92 for the VL ($P < 0.001$) and 0.85 for the SOL ($P < 0.001$).

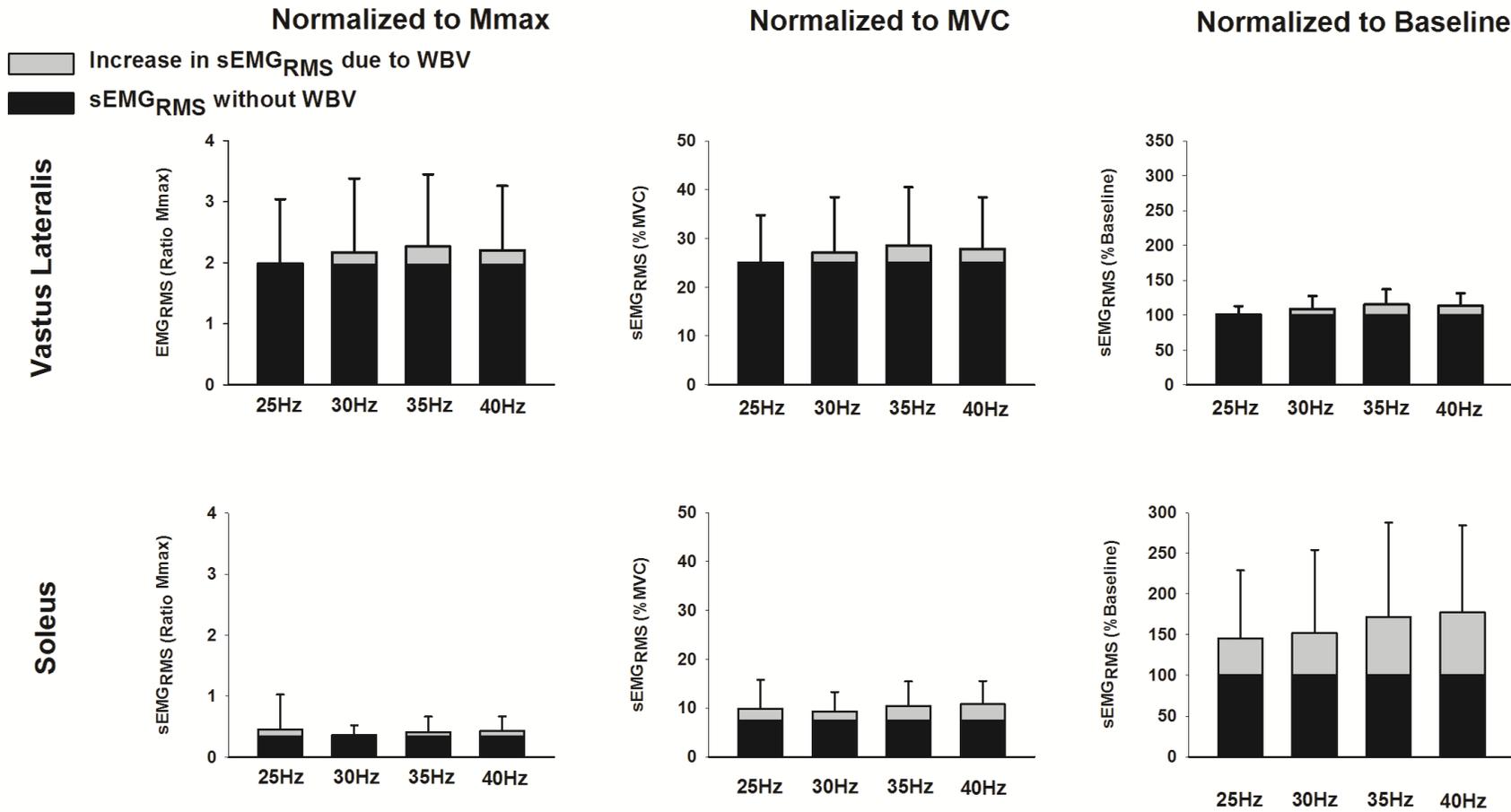


Figure 6.3 Root mean square of the surface electromyography (sEMG_{RMS}) during whole-body vibration (WBV) for the three normalization methods (mean \pm SD). The black bars represent the sEMG_{RMS} without WBV, and the grey bars represent the increase in sEMG_{RMS} due to WBV.

Discussion

The results of this study showed that normalization to the Mmax, to the sEMG during an MVC, as well as to baseline sEMG resulted in a similar sEMG_{RMS} pattern over several vibration frequencies during WBV exercises. Further, the CVs were similar between all normalization methods. Last, high and significant correlations were found between the methods.

The findings of the current study agree with a recent study indicating that normalization procedures did not influence the sEMG signal of the knee extensor muscles during MVCs (Buckthorpe et al. 2012). These authors also compared the use of M_{max} peak-to-peak amplitude and area for sEMG normalization, and recommended the latter method for group comparisons. However, the normalization of sEMG signals to the M_{max} area did not improve within-subject reliability (Buckthorpe et al. 2012). In fact, their CVs (~15.5%) of sEMG_{RMS}/M_{max} peak-to-peak amplitude ratios of the knee extensors were similar to those already reported for healthy subjects (i.e., ranging from 9% to 16%) (Gondin et al. 2005; Place et al. 2007). CVs of about 7.5% were also reported for the SOL (Clark et al. 2007). Although the best normalization method of sEMG signals is still debated (Burden 2010), the sEMG_{RMS}/M_{max} normalization allows a more thorough examination of individual muscles of a muscle group.

Interestingly, high correlation coefficients ($P > 0.87$) were found between the Mmax and MVC normalization methods in the present study. Since all the participants were physically active (i.e., 7.5 ± 4.2 h), it is assumed that they were able to recruit almost all of their motor units during MVCs. Yet this might not be the case for a different population such as untrained subjects, elderly, or patients presenting lower limb impairment, and thus caution should be taken when choosing the optimal normalization procedure for different populations. However, based on the simplicity of collecting sEMG activity during the MVC, and given that percutaneous neurostimulation is not feasible for all the human muscles, sEMG activity during WBV could be normalized to sEMG activity recorded during the MVC, bearing in mind that this procedure presents high between-subject CVs (Buckthorpe et al. 2012).

Although normalization to an individual value should decrease between-subject variability (Burden 2010), normalization to the baseline sEMG showed similar CVs as normalization to the other methods. While it is not clear why similar between-subject variability was observed between the methods, the drawback of normalizing to the baseline sEMG is still exemplified. In **Figure 6.3**, the increases in SOL sEMG_{RMS} due to WBV seem substantial (+50%) when normalized to the baseline sEMG. However, these increases are boosted out of proportion, as the sEMG_{RMS} is expressed as a percentage of the squatting posture. It is well known that the muscle activity of the SOL is minimal during squatting. Therefore, normalization to such baseline values can lead to misleading results. Normalization to the maximal sEMG_{RMS}, i.e. to the sEMG during an MVC, delivers more suitable results. First, it is known to which percentage of the maximum the muscles were contracted during the squatting position without vibration. Second, the increase in muscle activity due to WBV is also expressed as a percentage of the maximum sEMG, which can be useful for training recommendations.

Conclusions

Normalization to the Mmax, to the sEMG during the MVC, and to baseline sEMG are all suitable methods to normalize the sEMG during WBV exercises. However, given the elaborate assessment of Mmax responses, normalization to the MVC and baseline sEMG could be favoured. Further advantage of normalizing the sEMG during WBV to the sEMG during the MVC is the indication of the required muscle activity as a percentage of its maximum.

CHAPTER 7

STUDY III:

MOTION ARTIFACTS VS. REFLEX ACTIVITY

7.1 PREFACE

This explorative study was designed to investigate whether the isolated spikes observed in the sEMG spectrum during WBV contain motion artifacts and/or reflex activity. Insight about the content of the spikes would provide implication on how to process the sEMG signals during WBV exercise. As an example, the gained knowledge from this study was applied in study I, as the properties of the spectral linear interpolation method were designed according to these findings. The measurements for study III were part of the assessments completed in spring 2012, at the LAMHESS laboratory of the University of Nice Sophia Antipolis. The occurrence of motion artifacts was tested by placing electrodes on the patella with layers of tape between the skin and the electrodes. This set-up was used despite its limitations (see **CHAPTER 11**), as no other method is known thus far to measure the artifacts during WBV. In order to test whether the sEMG spikes contain reflex activity, the magnitude of the isolated spikes was analyzed changing the voluntary background muscle activity by means of additional loading. The idea behind this set-up was that if the sEMG spikes contain reflex activity, their magnitude would increase with an increasing background muscle activity, such as it is the case for the TVR induced by direct vibration. Indeed, it was found that the magnitude of the sEMG spikes was correlated to the background muscle activity of the respective muscle. Furthermore, spectral analysis of the patella signal revealed high peaks at the vibration frequency and its multiple harmonics, indicating that the signals were contaminated by motion artifacts. The outcomes of this study suggested that the sEMG spikes during WBV exercises contain reflex activity as well as motion artifacts. This needs to be considered in the interpretation of the results, especially together with the selected filtering regime. This study is currently submitted to the *Journal of Sport Science & Medicine*.

7.2 SEMG DURING WHOLE-BODY VIBRATION CONTAINS MOTION ARTIFACTS AND REFLEX ACTIVITY

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Abstract

The purpose of this study was to determine whether the excessive spikes observed in the sEMG spectrum recorded during WBV exercises contain motion artifacts and/or reflex activity. The occurrence of motion artifacts was tested by electrical recordings of the patella. The involvement of reflex activity was investigated by analyzing the magnitude of the isolated spikes during changes in voluntary background muscle activity. Eighteen physically active volunteers performed static squats while the sEMG was measured of five lower limb muscles during vertical WBV using no load and an additional load of 33 kg. In order to record motion artifacts during WBV, a pair of electrodes was positioned on the patella with several layers of tape between skin and electrodes. Spectral analysis of the patella signal revealed motion artifacts recordings as high peaks at the vibration frequency (fundamental) and marginal peaks at the multiple harmonics were observed. For the sEMG recordings, the RMS of the spikes increased with increasing additional loads ($P < 0.05$), and was significantly correlated to the sEMG signal without the spikes of the respective muscle (r range: 0.54 - 0.92, $P < 0.05$). This finding indicates that reflex activity is contained in the isolated spikes, as identical behavior has been found for stretch reflex responses evoked during direct vibration. In conclusion, the spikes visible in the sEMG spectrum during WBV exercises contain motion artifacts as well as reflex activity.

Introduction

WBV exercises acutely increase muscle activity during the exposure, as shown in numerous studies with the help of sEMG recordings (Cardinale and Lim 2003; Di Giminiani et al. 2013; Krol et al. 2011; Marín et al. 2012; Perchthaler et al. 2013). While the exact underlying mechanisms remain unclear, reflex muscular contractions have been proposed (Cardinale and Bosco 2003). During directly applied vibration, the primary endings of muscle spindles respond up to frequencies of 180 Hz (Roll and Vedel 1982), whereas secondary endings are less sensitive to vibration and respond at lower frequencies (Brown et al. 1967; Burke et al. 1976a). In addition to the spindle afferents, golgi tendon organs are responsive to vibration (Burke et al. 1976a). However, besides reflex activity, motion artifacts are assumed to occur, visible in the sEMG spectrum as excessive spikes at the vibration frequency and its multiple harmonics (Abercromby et al. 2007a). As the reflex activity (Ritzmann et al. 2010) and the motion artifacts (Fratini et al. 2009a) are both phase-locked to the vibration frequency, their influence on sEMG signals is difficult to predict.

Occurrence of motion artifacts during WBV exercises has been evidenced by signal recordings with so-called dummy electrodes. Electrodes were placed on the patella assuming that the recorded signal would be related to motion artifacts only (Fratini et al. 2009a). Although excessive peaks in the sEMG spectrum were located, it cannot be excluded that these peaks were not sEMG recordings from surrounding muscles. To address this limitation, a similar set-up has been repeated (Ritzmann et al. 2010; Sebik et al. 2013) by positioning electrodes on the muscle with several layers of tape between skin and electrodes. While Sebik et al. (2013) found motion artifact-contaminated sEMG signals, Ritzmann et al. (2010) observed only a marginal contribution. As a consequence, there is no consensus whether the spikes in the sEMG spectrum are contaminated by motion artifacts.

The most frequently cited mechanism by which WBV increases muscle activity is the TVR (Cardinale and Bosco 2003; Torvinen et al. 2002). The TVR has been well proven for locally applied vibration on the muscle belly or the tendon (Burke et al. 1976b; Eklund and Hagbarth

1966), but with no conclusive evidence that the TVR is applicable to WBV. However, almost identical latencies between actual stretch reflexes and WBV-induced reflex responses have been reported (Ritzmann et al. 2010). Further, a study by Pollock et al. (Pollock et al. 2012) has shown a strong relation between the timing of motor unit firings and the phase of the vibration cycle using intramuscular sEMG. Additionally, motor unit synchronization has shown to take place during WBV by means of multi motor unit recordings and sEMG signal rectification (Sebik et al. 2013). Interestingly, stretch reflex responses evoked via directly-applied vibration are enhanced by muscle contractions (Burke et al. 1976b; Eklund and Hagbarth 1966). If WBV elicits stretch reflexes, the magnitude of the spikes in the sEMG spectrum is expected to increase with increasing background muscle activity. Surprisingly, this hypothesis has never been examined.

Therefore, the goal of this study was to determine whether sEMG signals during WBV exercises contain motion artifacts and/or reflex activity. The latter was tested by analyzing the magnitude of the isolated spikes in the sEMG power spectrum using different background muscle activity. Contribution of motion artifacts were verified with the help of electrical recordings from the patella during the vibration. We hypothesized that (1) the sEMG measured during WBV contains motion artifacts caused by the vibration of the platform, that (2) motion artifacts are more prevalent in the first spike than in the following spikes, and that (3) the magnitude of the spikes in the sEMG spectrum increases with increasing voluntary background muscle activity.

Methods

Participants

Eighteen (10 men, 8 women) physically active students (mean \pm SD, age: 23.8 ± 3.2 years; height: 172.9 ± 8.3 cm; body mass: 67.8 ± 10.9 kg) volunteered to participate in the study. The exclusion criteria included frequent practice of WBV and/or neurological injuries to the lower extremity. The study was conducted according to the Helsinki Declaration (1964) and the study protocol was approved by the local ethics committee. Participants were informed about potential risks and signed written consent forms prior to their participation.

Experimental design

In order to evaluate if sEMG signals during WBV contain motion artifacts (hypothesis 1) and if they are more pronounced in the first spike than the following spikes (hypothesis 2), electrical signals from the patella were collected during the vibration. To test hypothesis (3), i.e., the magnitude of the spikes in the sEMG spectrum increases with increasing voluntary background muscle activity, sEMG signals of five lower limb muscles were recorded during static squat exercises with and without WBV using no load and an additional load of 33 kg. The additional load was used to increase the voluntary background muscle activity. The WBV device used in this study was a vertical synchronous platform (Power Plate pro6, North America, Northbrook, IL, USA), which delivers vibration in the vertical direction. The vibration frequency was set at 30 Hz, and the amplitude (displacement from baseline to peak) of the platform was 1 mm (preliminarily evaluated via double integration of the platform acceleration signal). The experimental session consisted of four vibration conditions presented in a randomized order, including two WBV trials (no load, 33 kg load) and two control trials without vibration (no load, 33 kg load). WBV trials consisted of 20-s static squats with a 1-min break in between trials. All participants were familiarized with WBV previously to the test session. The additional load was applied via a standard weightlifting bar measuring 150 cm in length with a weight of 5 kg; weight plates were attached to each side to obtain the 33 kg load. For each trial, the bar was positioned on the shoulders of the subjects, and all participants were familiar with loaded squat techniques. The participants were barefoot during the vibration trials in order to avoid damping effects due to different footwear, and foot position was marked on the platform in order to have similar stance between subjects. The participants were advised to flex their knees at an angle of 70° during the WBV exercises, which was monitored and recorded with the help of an electrical twin-axis goniometer (TSD130B, Biopac® Systems Inc., Holliston, MA, USA). When a participant could not maintain the position throughout the exercise, trials were repeated. Subsequent analysis showed that the participants performed their squats at a knee angle of $68.2 \pm 1.3^\circ$ (mean \pm SD). The acceleration of the vibration stimuli was controlled and recorded with the help of a tri-axial accelerometer (50G, TSD109F, Biopac® Systems Inc., Holliston, MA, USA) that was placed on the platform in alignment to the third toe (Lorenzen et al. 2009). RMS

calculation of the vertical platform acceleration proved a RMS (mean \pm SD) of 2.4 ± 0.1 g, with identical values between loads.

sEMG and patella signal recording

Pairs of bipolar silver-chloride electrodes (10-mm diameter, Contrôle Graphique Medical, Brie-Comte-Robert, France) were positioned over the muscle bellies of the VL, RF, BF, SOL, and GL of the right lower limb to record sEMG activity. The longitudinal axes of the electrodes were placed in line with the presumed direction of the respective underlying muscle fibers, with a center-to-center distance of 20 mm. The reference electrode was attached to the left patella. Low interelectrode resistance (< 5 k Ω) was obtained by means of shaving and abrading the skin with emery paper following cleaning the skin with alcohol. In order to record motion artifacts caused by the oscillation of the platform, electrodes were placed over the right patella in line with the lower limb (Miles et al. 1982). Three layers of adhesive tape between the skin and the bipolar electrodes (10-mm diameter, Contrôle Graphique Medical, Brie-Comte-Robert, France) served as isolation of the muscle activity from the electrodes. In order to assure the validity of measuring artifacts, participants were asked to perform a maximal voluntary knee extension against the static lever arm of a leg-extension machine. During the contraction, the patella signal remained at baseline levels. Hence, we concluded that in case of non-baseline signal recordings on the patella during WBV, the observed signal is due to the shaking of the electrodes, and not due to crosstalk from surrounding muscles. The patella and the sEMG signals were amplified (MP150 Biopac® Systems Inc., Holliston, MA, USA; CMRR = 110 db, Z input = 1000 M Ω , gain = 1000), filtered with a bandwidth frequency ranging from 10 Hz to 500 Hz, and online digitized with a sampling frequency of 2000 Hz.

Signal processing

In order to calculate the RMS of the first spike (fundamental) and the following spikes (harmonics), the spikes were separated from the sEMG signal in the frequency domain. This separation of the spikes was performed by using spectral linear interpolation. More specifically, the sEMG signals were first transformed into the PSD. This was accomplished with the help of

the Welch method, using Hamming windows with the length of $L = 1024$. The exact vibration frequency was verified with the help of the acceleration signal from the platform, as peak vibration frequency was identical for the acceleration and the sEMG signals. Subsequently, the magnitudes of the spikes were replaced by a straight line. This filtering regime allowed calculation of the RMS of the interpolated sEMG, the isolated fundamental, as well as the isolated harmonics (**Figure 7.1**). The RMS was calculated via the formula:

$$\text{RMS} = \sqrt{\frac{1}{L} \sum_{f=0}^{\frac{Fe}{2}} |\text{Signal}(f)|^2}$$

where $|\text{Signal}(f)|^2$ was the PSD of the signal, L was the length of the data to compute the PSD, and Fe was the sampling frequency. In order to enable comparison between sEMG signals of the muscles and the electrical signal of the patella, the RMS of the fundamental and the RMS of the harmonics was computed likewise for the electrical signal of the patella. For the patella as well as for the sEMG signals, the ratio between the RMS of the fundamental and the RMS of the harmonics was calculated via the formula: $\gamma = \frac{\text{RMS}_{\text{Fundamental}} - \text{RMS}_{\text{Harmonics}}}{\text{RMS}_{\text{Fundamental}}}$. If $\gamma > 0$, the RMS of the fundamental would be higher than the

RMS of the harmonics. If $\gamma < 0$, the RMS of the harmonics would be higher than the RMS of the fundamental.

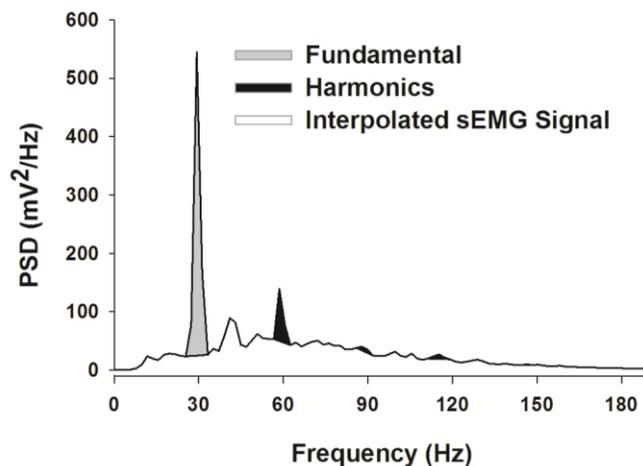


Figure 7.1 Example of a surface electromyography (sEMG) signal of the Vastus Lateralis during whole-body vibration at 30 Hz. The sEMG signals were processed in the Power Spectral Density (PSD), where the first spike (fundamental) and the following spikes (harmonics) were separated from the entire sEMG signal using spectral linear interpolation.

Statistical analysis

All analyses were executed using Statistica software (Statsoft, version 7.0 Tulsa, OK, USA). The Kolmogorov-Smirnov test confirmed the normality of the data, and the level of significance was set to $\alpha = 0.05$. In order to evaluate significant differences between the RMS of the additional loads, repeated analysis of variance (ANOVA) with Bonferroni correction was used. This statistical test was performed for the sEMG of the VL, RF, BF, SOL, and GL, and for the electrical signal of the patella. In order to test if the magnitude of the fundamental and the harmonics increases with increasing sEMG without the spikes, the RMS of the fundamental and the harmonics were correlated to the interpolated RMS of each muscle, calculating Pearson's correlation coefficients. For the patella signal, Pearson's correlation coefficients were computed for the RMS of the fundamental and the harmonics with the interpolated RMS of all muscles. Data of the two load conditions were combined for the correlation analysis. The results are presented as mean \pm SD.

Results

Patella signal

The electrical signal recorded on the patella during WBV consisted of a sinusoidal wave, similarly to the acceleration signal measured on the platform (**Figure 7.2**). In the frequency spectrum, excessive peaks were observed in the patella signal at the vibration frequency and to a lesser extent at the multiple harmonics. No significant difference was found between the RMS of the two loads ($P > 0.05$, **Figure 7.3A**). During WBV, the ratio γ between the RMS of the fundamental and the RMS of the harmonics was 0.7, averaged for both load conditions.

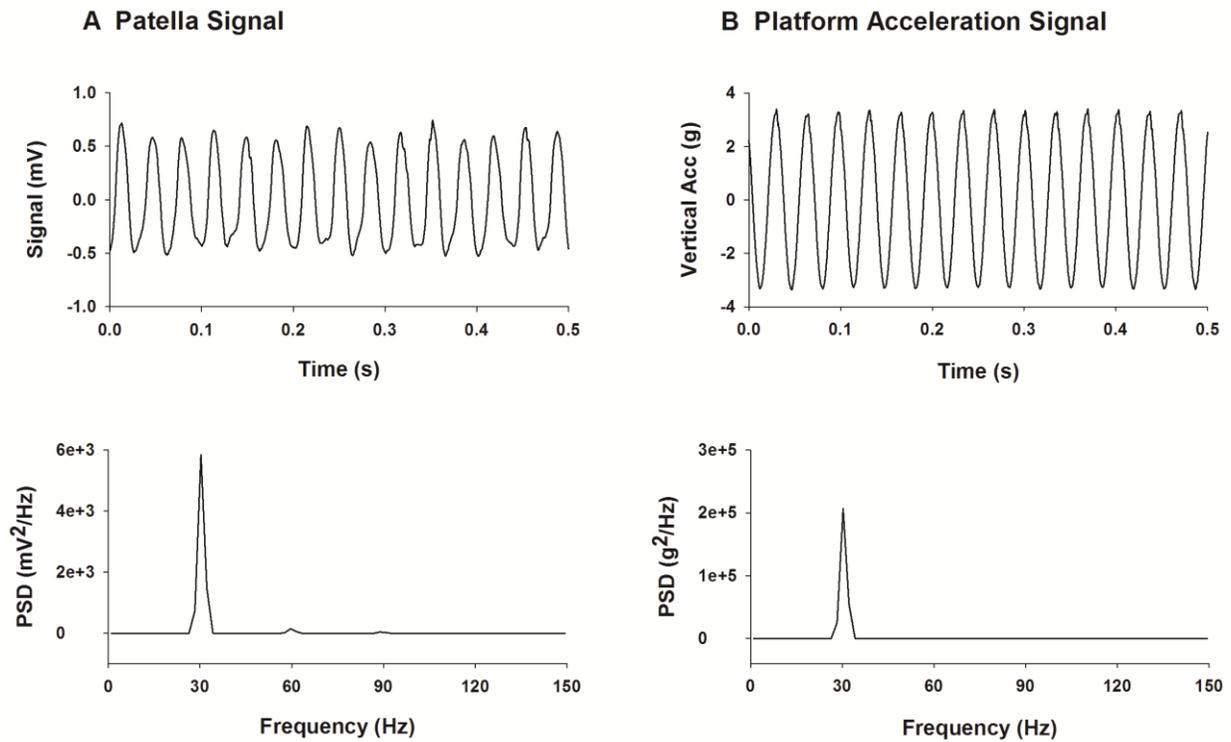


Figure 7.2 Example of (A) an electrical signal recorded on the patella and of (B) a vertical platform acceleration signal illustrated in the time domain (upper panel) and Power Spectral Density (PSD, lower panel) for one representative subject. The ratio $\gamma = \frac{\text{RMS}_{\text{Fundamental}} - \text{RMS}_{\text{Harmonics}}}{\text{RMS}_{\text{Fundamental}}}$ for the patella signal illustrated in (A) was $\gamma = 0.80$.

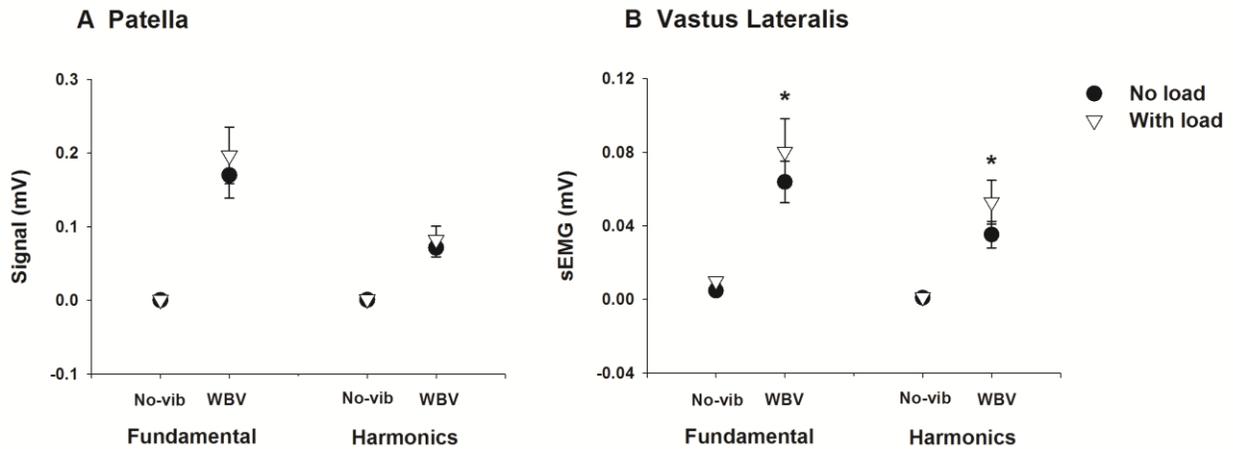


Figure 7.3 Root mean square (mean \pm SD) of (A) the patella signal and (B) the Vastus Lateralis surface electromyography (sEMG) during no vibration (no-vib) and whole-body vibration (WBV) with no additional load, and with an additional load of 33 kg (mean \pm SD). The results are illustrated separately for the first spike (fundamental) and the following spikes (harmonics). *With load $>$ no load ($P < 0.05$).

sEMG signal

During WBV, the RMS of the fundamental as well as the RMS of the harmonics was significantly higher with the additional load as compared without the load in all the measured muscles ($P < 0.05$, **Table 7.1**). **Figure 3B** illustrates the increase in RMS with the additional load for the VL. The ratios between the RMS of the fundamental and the RMS of the harmonics were $\gamma = 0.3$ for the thigh muscles and $\gamma = -0.2$ for the calf muscles (averaged for both loads).

Table 7.1 Root mean square of the fundamental and the harmonics for the surface electromyography of several lower limb muscles during whole-body vibration, without load and with an additional load of 33 kg (mean \pm SD).

		Fundamental (mV)	Harmonics (mV)
Vastus Lateralis	No load	0.064 \pm 0.036	0.035 \pm 0.023
	With load	0.080 \pm 0.053*	0.053 \pm 0.037*
Rectus Femoris	No load	0.026 \pm 0.011	0.017 \pm 0.005
	With load	0.038 \pm 0.015***	0.023 \pm 0.006***
Biceps Femoris	No load	0.013 \pm 0.005	0.007 \pm 0.003
	With load	0.019 \pm 0.008**	0.010 \pm 0.005***
Soleus	No load	0.008 \pm 0.004	0.010 \pm 0.007
	With load	0.018 \pm 0.011***	0.024 \pm 0.017***
Gastrocnemius	No load	0.004 \pm 0.002	0.003 \pm 0.002
Lateralis	With load	0.008 \pm 0.005**	0.009 \pm 0.006***

With load > no load (* $P < 0.05$, ** $P < 0.01$, *** $P < 0.001$).

Correlations

For all muscles, the RMS of the fundamental and the RMS of the harmonics demonstrated significant correlations with the interpolated RMS of the respective muscle ($P < 0.05$). The correlation coefficient (r) between the fundamental and the interpolated sEMG was 0.76 for the VL, 0.72 for the RF, 0.79 for the BF, 0.89 for the SOL, and 0.56 for the GL. For the harmonics, the correlation coefficient (r) was 0.85 for the VL, 0.54 for the RF, 0.81 for the BF, 0.89 for the SOL, and 0.92 for the GL. The RMS of the patella was not related to the RMS of the interpolated sEMG of any muscle that was measured ($P > 0.05$).

Discussion

This study aimed to clarify the content of the isolated spikes that are observed in the sEMG spectrum during WBV. The main findings were that (1) sEMG signals during WBV are contaminated by vibration induced motion artifacts, which was confirmed by motion artifact-recordings by placing electrodes on the patella. In addition, (2) the motion artifacts were more prevalent in the first spike (fundamental) than in the following spikes (harmonics), as the

averaged ratio $\gamma = \frac{\text{RMS}_{\text{Fundamental}} - \text{RMS}_{\text{Harmonics}}}{\text{RMS}_{\text{Fundamental}}}$ for the patella signal was above zero ($\gamma = 0.7$).

Further it was shown that (3) sEMG signals during WBV may include muscle activity phase-locked to the vibration frequency and its multiple harmonics, because the RMS of the spikes increased with increasing background activation of the muscle and was highly correlated to the RMS of the sEMG signal without the spikes.

The high peaks observed in the power spectrum of the patella signal showed that motion artifacts contribute to the sEMG signal during WBV, which confirms the first hypothesis. Furthermore, the RMS of the patella's fundamental was 3.6 times greater than the RMS of the harmonics, which affirms the second hypothesis. Therefore it can be concluded that sEMG signals measured during WBV are contaminated by motion-induced artifacts, and that the artifacts are more distinct in the fundamental than in the harmonics.

Relevance of motion artifacts during WBV using dummy electrodes placed on the patella (Fratini et al. 2009a) or the muscle (Ritzmann et al. 2010; Sebik et al. 2013) has been verified before, reporting conflicting results. Fratini et al. (2009a) and Sebik et al. (2013) proved the existence of motion artifacts by finding excessive peaks at the vibration frequency and its multiple harmonics, whereas Ritzmann et al. (2010) reported a marginal contribution. Possible reasons for the observed discrepancy could be the type of the used equipment such as the electrodes or the cables (Webster 1984). Using conjoined electrodes vs. single electrodes might have the advantage that the conjoined electrodes are moving in phase to each other, which impedes a difference in potential between the electrodes. Additionally, the degree of cable shielding could have an influence, as motion artifacts are reduced by using shielded cables vs. unshielded cables

(Clancy et al. 2002). However, together with the findings of the present study it is challenging to ascertain why Ritzmann et al. (2010) did not observe significant spikes in the dummy electrode signal. However, to minimize motion artifacts during WBV exercises, the electrodes and their cables should be taped to the skin, the amplifier/preamplifier should be placed close to the recording electrodes, and short cables should be used connecting the electrodes with the amplifier (Türker 1993).

Interestingly, it was found that the RMS of the fundamental and the RMS of the harmonics increased with increasing background muscle activity. Identical behavior has been reported for stretch reflex responses evoked by directly-applied vibration to the tendon or the muscle in numerous studies (Bedingham and Tatton 1984; Bongiovanni and Hagbarth 1990; Burke et al. 1978, 1976a, 1976b; Eklund and Hagbarth 1966; Nordin and Hagbarth 1996). Further, there is recent evidence that this mechanism also occurs during WBV exercises (Zaidell et al. 2013). Therefore it is possible that the spikes observed in the sEMG spectrum during WBV consist partially of stretch reflex responses. It is important to mention that we did not measure stretch reflex responses per se. Therefore, other reflex mechanisms induced by e.g. golgi tendon organs and cutaneous mechanoreceptors could explain the findings of the present study. While the underlying mechanisms remain uncertain, the results of this study showed for the first time that the isolated spikes in the sEMG spectrum contain motion artifacts as well as reflex activity.

A higher ratio $\gamma = \frac{\text{RMS}_{\text{Fundamental}} - \text{RMS}_{\text{Harmonics}}}{\text{RMS}_{\text{Fundamental}}}$ for the patella signal ($\gamma = 0.7$) than for the muscles (thigh: $\gamma = 0.3$; calf: $\gamma = -0.2$) was observed. γ ratios higher than zero indicate that the RMS of the fundamental was higher than the RMS of the harmonics and vice versa. As the γ ratios were lower in the muscles than in the patella, we conclude that the magnitude of the harmonics compared to the fundamental was higher in the muscles' sEMG signals than in the patella signal. This observation might be due to contribution of reflex activity in the muscle sEMG. Stretch reflex trains separated by a time interval of t_0 result in a spectrogram with peaks separated by the frequency interval of $\frac{1}{t_0}$, as illustrated by Ritzmann et al. (2010). However, if such a stretch reflex train is contaminated by motion artifacts, the relative contribution of the

fundamental is increased. This effect is shown in **Figure 7.4** with the help of computer simulated signals. **Figure 7.4A** shows the simulated signal of a stretch reflex train in the time domain (upper panel) and PSD (lower panel) with $\frac{1}{t_0} = 30$ Hz. In **Figure 7.4B**, a sinusoidal wave representing motion artifacts with the same frequency as the stretch reflex train was added to the signal from **Figure 7.4A**. Comparing the PSD of the two signals, the ratio γ was found to be higher in the motion artifact contaminated signal ($\gamma = 0.45$) than in the pure stretch reflex train ($\gamma = -0.30$). Hence, it can be concluded that the higher the ratio γ , the higher the contribution of motion artifacts and the lower the involvement of reflex activity. The fact that γ was lower for the muscles than the patella further suggests reflex activity in sEMG recordings during WBV exercises.

One limitation of the present study is the methodological approach of measuring the motion artifacts. The insulating tape placed between the electrodes and the skin indeed prevented possible sEMG recordings from surrounding muscles, but at the same time, the resistance between the electrodes was augmented to almost infinite. This could have resulted in an overestimation of the actual motion artifacts. However, we considered the certainty of excluding crosstalk recordings from surrounding muscles as more important than of the exact estimation of motion artifact contribution.

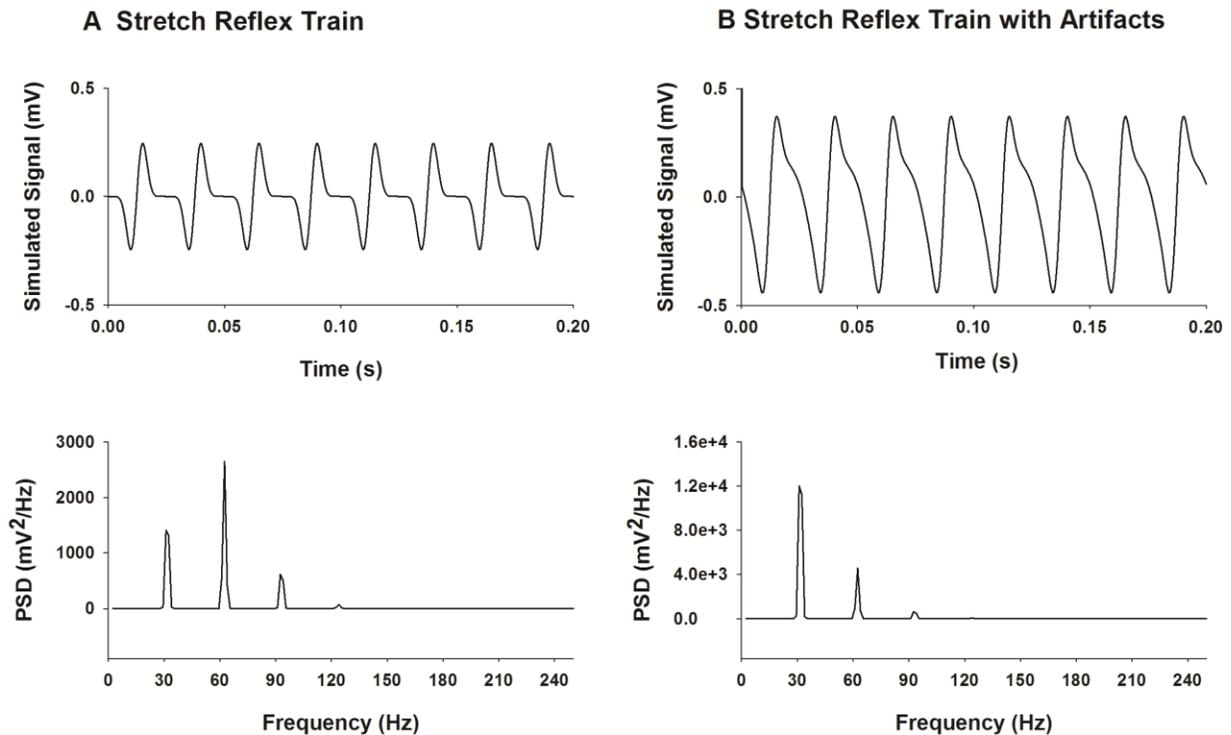


Figure 7.4 Computer simulated signal of (A) a stretch reflex train and (B) a stretch reflex train contaminated with motion artifacts in the time domain (upper panel) and Power Spectral Density (PSD, lower panel) at 30 Hz. The ratio $\gamma = \frac{\text{RMS}_{\text{Fundamental}} - \text{RMS}_{\text{Harmonics}}}{\text{RMS}_{\text{Fundamental}}}$ was $\gamma = -0.30$ for the signal illustrated in (A) and $\gamma = 0.45$ for the signal illustrated in (B).

Conclusions

The isolated spikes that can be observed in the sEMG spectrum during WBV exercises contain reflex activity as well as motion artifacts. This implies that the muscle activity measured during WBV exercises can be overestimated when the excessive spikes are not deleted and vice versa. However, further investigations focusing on this topic are needed with separate analysis for the fundamental and the harmonics as the outcomes of this study showed that their contribution in motion artifacts is unequal.

CHAPTER 8

STUDY IV:

THE OPTIMAL PARAMETERS

8.1 PREFACE

After having clarified important methodological issues, it was time for a physiologically applied study. Besides investigating whether WBV exercise enhances muscle activity compared to the same exercise without vibration, the effect of several parameters was examined. The goal was to find the optimal parameters that would induce the highest increases in muscle activity during WBV exercises. It stands to reason that the vibration frequency and the platform displacement were scrutinized, as they are the two most studied parameters and the only settings that can be adjusted directly on the platform. The WBV platform was provided by Power Plate France, and was ready to use at the LAMHESS laboratory of the University of Nice Sophia Antipolis. The applied frequencies and amplitudes lied within the limitation of the platform. The frequency setting ranged from 25 to 40 Hz, hence four frequencies in steps of 5 Hz were used (25, 30, 35, 40 Hz). The displacement of the platform consisted of a low and a high setting, corresponding to an amplitude (displacement from baseline to peak) of 0.6 mm and 1.0 mm, respectively. For this study, the displacement of the platform was reported using the peak-to-peak notation, thus the peak-to-peak displacement of this platform was 1.2 mm and 2.0 mm. However, a third parameter was studied, namely additional loading. This parameter was of particular interest, as additional loading has the potential to intensify the effect of vibration, via increased TVR responses (as found in study III). Therefore, additional loads of 17 and 33 kg were applied. Although this study was conducted simultaneously as the previous three studies, their methodological suggestions were applied in study IV; the spikes in the sEMG spectrum were deleted using spectral linear interpolation, and the sEMG measured during WBV was normalized to the sEMG during the MVC of the respective muscle. In brief, the outcomes of this study were that the highest increases in muscle activity were achieved with the highest frequency-amplitude combination. However, by using the 33 kg additional load, the muscle activity of the calf muscles was significantly increased even with a low frequency and amplitude. The optimal parameters were found to be a high frequency-amplitude combination together with a high additional load.

8.2 DETERMINATION OF THE OPTIMAL PARAMETERS MAXIMIZING MUSCLE ACTIVITY OF THE LOWER LIMBS DURING VERTICAL SYNCHRONOUS WHOLE-BODY VIBRATION

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ORIGINAL ARTICLE

Determination of the optimal parameters maximizing muscle activity of the lower limbs during vertical synchronous whole-body vibration

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Abstract

Purpose: To describe the most effective parameters maximizing muscle activity during WBV exercises on a vertically vibrating platform. **Methods:** The influence of (1) WBV vs. no vibration, (2) four vibration frequencies (25, 30, 35, 40 Hz), (3) platform peak-to-peak displacement (1.2 mm, 2 mm), and (4) additional loading (no load, 17 kg, 33 kg) on sEMG activity of five lower limb muscles was investigated in eighteen participants. **Results:** (1) Comparing WBV to no vibration, sEMG_{RMS} of the calf muscles was significantly higher with an additional load of 33 kg independently of the displacement and the frequency ($P < 0.05$). During WBV, (2) muscle activity at 40 Hz WBV was significantly higher than at 25 Hz for the GL for all loads, and for the VL and VM using the 33 kg load ($P < 0.05$); (3) sEMG_{RMS} of all lower limb muscles was significantly increased with the 2 mm compared to the 1.2 mm peak-to-peak displacement ($P < 0.05$); (4) an effect of additional load was found in the gastrocnemius lateralis (GL), with significantly higher neuromuscular activation for the 33 kg load than no load ($P < 0.05$). **Conclusions:** On a VV platform, we recommend the use of a high platform displacement in combination with a high vibration frequency to provoke the highest muscle activity enhancement. Without maxing out the acceleration stimuli, calf muscles' sEMG can be enhanced with an additional load of 33 kg which corresponded to 50% of the body mass.

Introduction

The use of WBV training to enhance lower limb strength and power (Bosco et al. 1999; Delecluse et al. 2003; Petit et al. 2010; Roelants et al. 2004) has become increasingly popular over the last two decades. Such chronic effects of WBV exercises are believed to result, amongst others, from augmented neuromuscular activity during the exposure to the vibration. The mechanism leading to increased muscle activity during WBV is believed to result from stretch reflex responses, induced by activation of muscle spindles and alpha-motoneurons, leading to the TVR (Cardinale and Bosco 2003; Pollock et al. 2012; Ritzmann et al. 2010; Torvinen et al. 2002). Numerous studies have indeed observed an increase of sEMG recordings of lower limbs muscles during WBV compared to no vibration (Cardinale and Lim 2003; Cochrane et al. 2009; Di Giminiani et al. 2013; Hazell et al. 2007; Marín et al. 2009; Roelants et al. 2006). Unfortunately, inconsistent observations have been reported depending on the vibration frequency, the peak-to-peak displacement of the platform, and the muscle examined (Cochrane et al. 2009; Hazell et al. 2007; 2010; Marín et al. 2009). Therefore, it is extremely challenging to give recommendations regarding the benefits of WBV exercises.

The intensity of the WBV stimulus is determined by the magnitude of the acceleration (Mester et al. 1999), which is transferred to the subject's body via their feet. The extent of the acceleration can be adjusted by means of the vibration frequency (typically between 10 and 55 Hz), and the peak-to-peak displacement of the platform (ranging from 1 to 12 mm). Additionally, the intensity of WBV exercises can be manipulated for example by using additional loads, by changing the modality of the exercise (static vs. dynamic), and by adapting the position of the person standing on the platform (forefoot stance vs. normal stance, low squat vs. high squat). The heterogeneous use of these parameters contributes to the conflicting results reported in the literature (Cardinale and Lim 2003; Di Giminiani et al. 2013; Hazell et al. 2007), which to date impedes general conclusions. Another factor possibly confounding the outcomes to an even higher extent is the use of different types of WBV platforms. The most prevalent platform types currently available are the side-alternating vibration platform, which oscillates along the sagittal axis, and the vertical synchronous vibration platform, which moves

synchronously in the vertical direction (Pel et al. 2009). Adverse effects on muscle activity between the two platform types have been reported (Abercromby et al. 2007a; Ritzmann et al. 2013). As a consequence, it is imperative to consider what type of platform has been used when comparing outcomes of different studies.

When considering VV platforms, frequency-dependent increases with the greatest sEMG activity at high vibration frequencies (40 – 60 Hz) has been evidenced for the VL (Di Giminiani et al. 2013; Hazell et al. 2007), the VM (Krol et al. 2011), and the SOL (Zaidell et al. 2013), whereas Cardinale and Lim (2003) reported higher VL activity at 30 Hz than at frequencies above 40 Hz. The effect on muscle activity with increasing platform displacement has been investigated scarcely, with reports of greater muscle activity for the higher platform displacement in the BF, but not in the VL (Hazell et al. 2007). Another study using the same platform type found no displacement-dependent effect on muscle activity in the VL and the GM (Marín et al. 2009), whereas Krol et al. (2011) reported a displacement-dependent effect on muscle activity for the VL and VM muscles. To our knowledge, the acute effect of external loading on sEMG activity with vertical synchronous platforms has only been examined in one study, reporting significant effects in the VL, BF, TA, and GM during dynamic squat exercises (Hazell et al. 2010). However, the optimal WBV training parameters using a synchronous platform are widely undetermined.

Therefore, the goal of this study was to examine the effects of vibration frequency, displacement, and additional loads on muscle activity using a vertical synchronous platform. Based on findings of previous studies (Di Giminiani et al. 2013; Hazell et al. 2007; Krol et al. 2011; Marín et al. 2009), the following hypotheses were formulated: We assumed that (1) the effect of vibration on sEMG activity, i.e., higher sEMG during WBV compared to no vibration, is significant for all the tested vibration conditions. Further, during the WBV exercise, we hypothesized that (2) increasing the vibration frequency has a significant increasing effect on muscle activity, that (3) high platform displacement causes higher neuromuscular responses than the low platform displacement, and that (4) increases in sEMG activity as a response to WBV is progressively augmented with augmenting additional loads.

Methods

Participants

Ten male and eight female physically active individuals (mean \pm SD, age: 23.8 ± 3.2 years; height: 172.9 ± 8.3 cm; mass: 67.8 ± 10.9 kg, physical activity: 7.5 ± 4.2 h/week) volunteered to participate in the study. The exclusion criteria consisted of recent injuries to the lower extremity and/or frequent practice of WBV. The study was conducted according to the Helsinki Declaration (1964). The study protocol was approved by the Ethics Committee on Human Experiments in Life and Health Sciences of the University of Nice - Sophia Antipolis, and all participants signed written consent forms prior to their participation.

Experimental design

sEMG signals of five lower limb muscles were collected during static squat exercises with and without WBV with various experimental conditions including (1) two vibration peak-to-peak displacements (1.2 mm, 2 mm), (2) additional load variation (no extra load, 17 kg, and 33 kg), and (3) four progressive increasing vibration frequencies (25 Hz, 30 Hz, 35 Hz, and 40 Hz). All participants completed a standardized warm-up consisting of 5 min cycling on an ergometer (Monark, 818E, Varberg, Sweden) with low resistance. Then, MVCs of the knee extensors and plantar flexors were assessed for sEMG normalization purposes. Subsequently, the participants followed the different WBV conditions, which were presented in a randomized fashion.

WBV exercises

A synchronous vertically oscillating WBV platform was used in this study (Power Plate pro6, North America, Northbrook, IL, USA). The frequency settings of this platform range between 25 Hz and 40 Hz, whereof we chose to examine the effects of vibration frequency in steps of 5 Hz (25 Hz, 30 Hz, 35 Hz, and 40 Hz). The displacement settings include a high and a low setting, whereas the corresponding displacement in mm was not provided by the manufacturer. Therefore, platform displacement was evaluated preliminarily by double integration of the platform acceleration signal, which resulted in peak-to-peak displacement of 1.2 mm and 2 mm. In order to quantify the influence of load variation, trials without load (control trials) were compared to trials with additional loads of 17 kg and 33 kg. The additional loads were applied via a standard weightlifting bar measuring 150 cm in length with a weight of 5 kg. Weight plates were attached to each side to obtain the 17 kg and 33 kg loads. For each trial, the bar was positioned on the participant's shoulders, and all individuals were familiar with loaded squat techniques. The participants were barefoot during the vibration trials in order to avoid damping effects due to different footwear, and foot position was marked on the platform for similar stance purposes. All individuals were asked to flex their knees at an angle of 70° during the squats (0°: knee fully extended), which was monitored and recorded with the help of an electrical twin-axis goniometer (150 MM – TSD130B, Biopac® Systems, Holliston, MA, USA). When a subject could not maintain the knee angle throughout the exercise, trials were repeated. Subsequent analysis showed that the participants performed their squats at a knee angle of $67.3 \pm 1.5^\circ$ (mean \pm SD). Each trial consisted of 30-s isometric squats with a 1-min break in between trials. In total, each participant conducted 27 x 30 s static squats which included 4 frequencies x 2 peak-to-peak displacements x 3 loads and 3 control trials without vibration (no load, 17 kg, and 33 kg). The acceleration of the vibration stimuli was controlled and recorded with the help of a tri-axial accelerometer (50G, TSD109F, Biopac® Systems, Holliston, MA, USA) that was placed on the platform in alignment to the third toe (Lorenzen et al. 2009). Analysis of the acceleration signal revealed that the acceleration stimulus was mainly in the vertical plane (**Figure 8.1**). RMS of the vertical acceleration was calculated, which provided identical values between the three additional loads (**Table 8.1**).

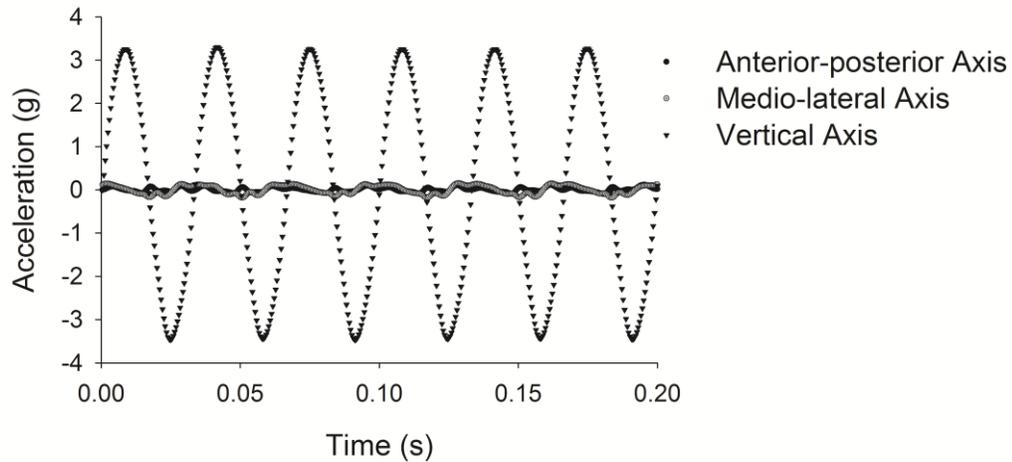


Figure 8.1 Example of an acceleration signal measured on the platform during WBV at 30 Hz, with a peak-to-peak displacement of 2 mm, and no additional load. The platform induced mainly vibrations in the vertical direction.

Assessment of MVCs

Isometric MVCs of the knee extensors were assessed on a leg extension machine (Gymstar 900, Carnielli, Vittorio Veneto, Italy), and MVCs of the plantar flexors were collected on a home-made ankle ergometer. For the MVC assessment of the knee extensor muscles, the participants were seated on the leg extension machine with a hip joint angle of 90° (0° : hip fully extended) and were fixed with a velcro belt across their pelvis to minimize body movements during leg extension. The lower leg was attached to the lever arm with a strap 2 cm to 3 cm proximal to the lateral malleolus with a knee joint angle of 70° (0° : knee fully extended), which was controlled using an electrical twin-axis goniometer (TSD130B, Biopac[®] Systems, Holliston, MA, USA). MVC torques during knee extension and plantar flexion were recorded using a force transducer (TSD121C, Biopac[®] Systems, Holliston, MA, USA) that was secured to the respective machine. For the MVC measurements of the plantar flexors, the participants were seated on a chair with their right foot attached to the pedal of the ankle ergometer. Knee joint angle was kept at 70° , and the ankle joint at an angle of 0° (neutral position). During both, knee extension and plantar flexion MVCs, participants were instructed to fold their arms in front of their chests.

All participants performed two 5-s MVCs separated by a 45-s rest period. In the event of an increase or a decrease in strength of more than 10% between the two MVCs, a third trial was conducted. MVC peak torque over 500 ms was evaluated using Matlab software (version 7.13 The Mathworks, Natick, MA, USA), and the trial with the highest torque was retained. Then, the $sEMG_{RMS}$ was calculated in the corresponding window with the MVC peak torque.

Table 8.1 Root mean square (RMS) of the vertical platform acceleration (in g) for both peak-to-peak platform displacements and the four vibration frequencies. Platform acceleration was averaged for the three loads as no difference was observed.

RMS of Vertical Platform Acceleration in g							
1.2 mm Peak-to-Peak Displacement				2 mm Peak-to-Peak Displacement			
25 Hz	30 Hz	35 Hz	40 Hz	25 Hz	30 Hz	35 Hz	40 Hz
1.3 ± 0.1	1.5 ± 0.1	2.0 ± 0.1	2.6 ± 0.2	2.1 ± 0.1	2.4 ± 0.1	3.3 ± 0.2	4.3 ± 0.3

Mean ± SD.

Muscle sEMG

Pairs of bipolar silver-chloride electrodes (10-mm diameter, Contrôle Graphique Medical, Briec-Comte-Robert, France) were placed over the muscle bellies of the SOL, GL, VM, RF, VL of the right lower limb. The longitudinal axes of the electrodes were aligned with the presumed direction of the respective underlying muscle fibers; center-to-center distance was 20 mm. The reference electrode was attached to the left patella. Low interelectrode resistance (< 5 k Ω) was obtained by means of shaving and abrading the skin with emery paper following cleaning the skin with alcohol. sEMG signals were amplified (MP150 Biopac[®] Systems, Holliston, MA, USA; CMRR = 110 db, Z input = 1000 M Ω , gain = 1000), filtered with a bandwidth frequency ranging from 10 Hz to 500 Hz, and online digitized with a sampling frequency of 2000 Hz. For descriptive purposes, raw sEMG signals during WBV are shown in **Figure 8.2**; illustrating the frequency-, displacement-, and load-dependent increases in sEMG.

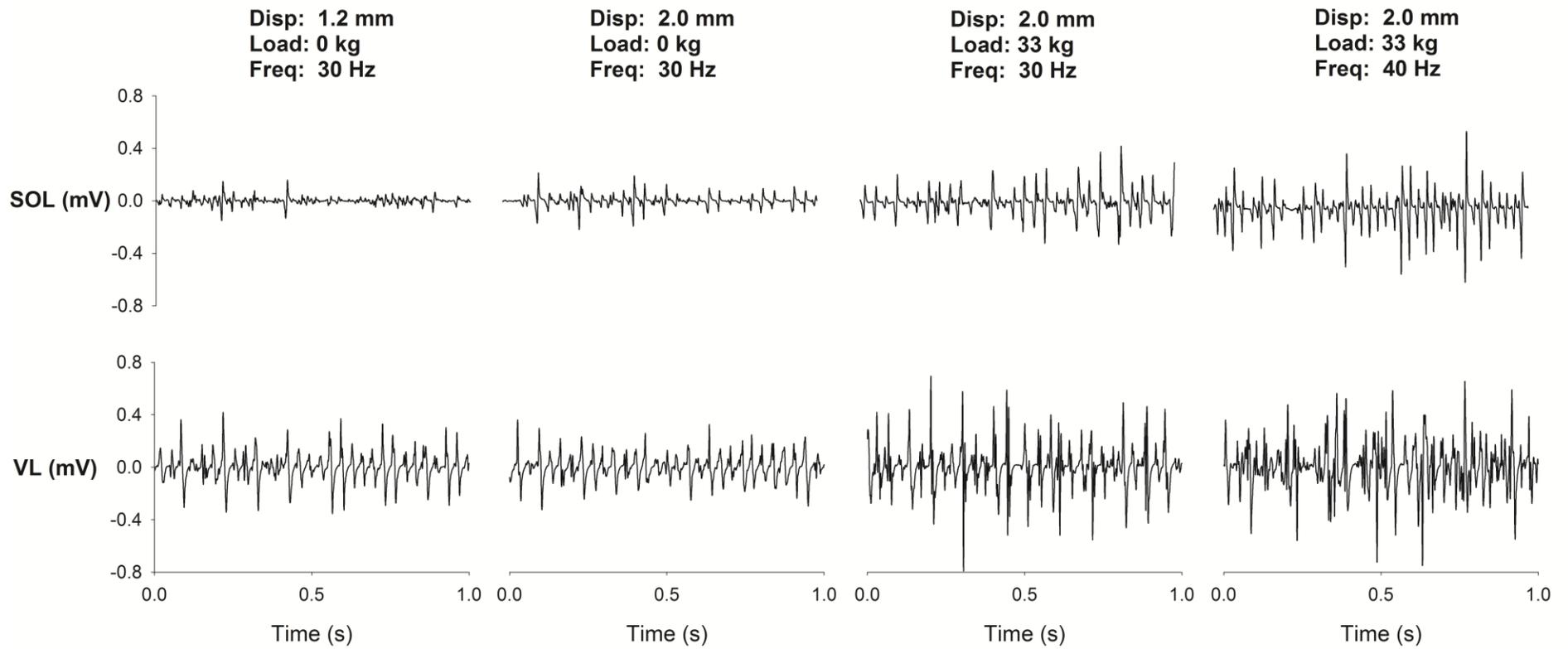


Figure 8.2 Surface electromyography (sEMG) signals of the Soleus (SOL, upper panel) and the Vastus Lateralis (VL, lower panel) for one representative participant during WBV. Note the displacement (Disp)-dependent increase in sEMG activity from the first to the second row, load-dependent increase from the second to the third row, and the frequency (Freq)-dependent increase from the third to the fourth row.

sEMG processing

sEMG signals during WBV contain excessive peaks at the vibration frequency and its multiple harmonics in the frequency domain (referred to as spikes). It is believed that these spikes contain either motion artifacts (Abercromby et al. 2007a; Fratini et al. 2009), stretch reflex responses (Pollock et al. 2012; Ritzmann et al. 2010), or both (Sebik et al. 2013). Therefore, withdrawal of the spikes might result in an underestimation of the true sEMG activity during WBV, whereas keeping the spikes possibly results in an overestimation. However, the authors decided to withdraw the spikes in the sEMG signals, as they suggest that the magnitude of the motion artifacts in the spikes is higher than the magnitude of the stretch reflex responses. Therefore, a new sEMG processing method was developed, based on linear interpolation centered at the vibration frequency and its multiple harmonics. sEMG signals were first transformed into the PSD using hamming windows. Second, linear interpolation was applied 2-4 Hz (depending on the spikes' width) above and below the spikes' peak at the vibration frequency. This procedure was conducted for the first ten spikes. Third, square root of the power was calculated of the sEMG without the spikes, via the formula: $sEMG_{RMS} =$

$\sqrt{\frac{1}{L} \sum_{f=0}^{\frac{Fe}{2}} |sEMG(f)|^2}$, where $|sEMG(f)|^2$ was the interpolated signal in the PSD, L was the length of the data to compute the PSD, and Fe was the sampling frequency. The outcome of this equation is identical to the outcome of the RMS calculation in the time domain via the

formula: $\sqrt{\frac{1}{n} \sum_{t=1}^n sEMG(t)^2}$, where $sEMG(t)$ was the sEMG in the time domain, and n was the

length of the data in the time domain. This procedure was applied for all sEMG signals during vibration, whereas the first 10 s and the last 5 s of the trials were excluded. Last, $sEMG_{RMS}$ was normalized to the sEMG during the MVC via the formula: $\frac{sEMG_{RMS} \text{ during vibration}}{sEMG_{RMS} \text{ during MVC}} \times 100$.

Statistical analysis

The Kolmogorov-Smirnov test confirmed the normality of the data. In order to determine significant effects between $sEMG_{RMS}$ during vibration compared to no vibration (hypothesis 1), repeated measures ANOVAs were performed. $sEMG_{RMS}$ during each WBV condition was compared to the $sEMG_{RMS}$ without vibration of the same load. To evaluate the effects of

platform displacement, additional load, and vibration frequency on sEMG activity during WBV (hypotheses 2, 3, and 4), control trials were subtracted from the WBV trials with the corresponding load to enable sEMG_{RMS} comparison between loads. Then, 3-way repeated measures ANOVAs (displacement (2) x load (3) x frequency (4)) with Bonferroni corrected post-hoc tests were computed. Multiple regression analyses were performed on the mean sEMG_{RMS} of the calf muscles and the thigh muscles to delineate predictive models. In the linear regression model, frequency (25 Hz, 30 Hz, 35 Hz, 40 Hz), peak-to-peak displacement (1.2 mm, 2.0 mm), and load (0 kg, 17 kg, 33 kg) were considered as predictors. For all analyses, the significance levels were set to $P < 0.05$, $P < 0.01$, and $P < 0.001$, and the results are presented as mean \pm SD.

Results

WBV compared to no vibration

sEMG_{RMS} during WBV as compared to no vibration was not systematically affected by the displacement or the vibration frequency in any of the measured muscles. With the additional load of 33 kg, the SOL and the GL showed significantly higher sEMG_{RMS} during WBV for all vibration frequencies and both displacements ($P < 0.05$), with the exception of WBV at 25 Hz with the 1.2 mm peak-to-peak displacement. Tables 8.2 – 8.4 show the sEMG_{RMS} normalized to the MVC of each muscle for the no vibration trial and during WBV at all four frequencies and both displacements (**Table 8.2**), with an additional load of 17 kg (**Table 8.3**), and with an additional load of 33 kg (**Table 8.4**).

Effect of displacement, load, and frequency during WBV

Significant displacement-dependent main effects on sEMG activity was found in all the measured muscles, i.e., SOL ($P < 0.01$), GL ($P < 0.05$), VM ($P < 0.01$), RF ($P < 0.01$), and VL ($P < 0.01$).

A significant load-dependent effect on muscle activity was found in the GL, with higher sEMG_{RMS} for the 33 kg load than no load ($P < 0.05$). Additionally, a significant main effect of frequency (P

< 0.05) was found in the GL, with higher sEMG_{RMS} at 40 Hz vibration compared to 25 Hz vibration ($P < 0.05$).

Significant interactions between load x frequency were observed for the VM ($P < 0.01$) and the VL ($P < 0.01$). Post-hoc tests revealed that in the VM, sEMG_{RMS} at 35 Hz was significantly higher than at 30 Hz vibration with the medium load ($P < 0.01$), whereas with the high load, sEMG_{RMS} at 30, 35, and 40 Hz was significantly higher than at 25 Hz vibration ($P < 0.01$). In the VL, sEMG_{RMS} at 30 and 40 Hz was significantly higher than at 25 Hz WBV with the high load ($P < 0.05$ and $P < 0.01$). Further, a significant interaction between frequency x displacement was found for the VL ($P < 0.05$), with 40 Hz exhibiting higher sEMG_{RMS} increases compared to 25 Hz ($P < 0.01$) and 30 Hz ($P < 0.05$) with the 2 mm displacement. No significant interaction effect between displacement x load x frequency was observed for any of the measured muscles ($P > 0.05$).

The multiple regression analysis results on the average increase in sEMG_{RMS} for the calf muscles and the thigh muscles are shown in **Table 8.5**.

Table 8.2 Root mean square of the surface electromyography (sEMG_{RMS}) normalized to the sEMG_{RMS} during maximal voluntary contractions (MVCs) without the use of an additional load.

sEMG _{RMS} (%MVC) without additional load									
	No Vibration	1.2 mm Peak-to-Peak Displacement				2 mm Peak-to-Peak Displacement			
		25 Hz	30 Hz	35 Hz	40 Hz	25 Hz	30 Hz	35 Hz	40 Hz
SOL	7.4 ± 3.4	8.5 ± 3.6	8.2 ± 4.6	9.5 ± 4.8	10.3 ± 4.2**	9.8 ± 5.9*	9.3 ± 4.0	10.4 ± 5.0	10.8 ± 4.8*
GL	2.9 ± 2.7	2.8 ± 1.9	2.8 ± 1.9	3.2 ± 2.0	3.5 ± 2.1	3.3 ± 2.2	3.1 ± 1.6	3.3 ± 1.7	3.9 ± 2.1
VM	22.7 ± 10.0	24.5 ± 10.8*	24.5 ± 12.7	25.3 ± 12.1*	24.1 ± 11.4	23.3 ± 11.4	26.0 ± 12.2**	27.4 ± 16.2*	24.8 ± 12.4*
RF	14.4 ± 9.3	14.8 ± 10.9	15.9 ± 11.1	15.1 ± 10.6	14.2 ± 10.1	15.2 ± 10.7	16.5 ± 11.5	16.0 ± 10.4	14.6 ± 9.9
VL	25.0 ± 9.7	25.9 ± 10.1	27.3 ± 11.2*	26.8 ± 10.6	26.3 ± 11.0	25.0 ± 9.8	27.1 ± 11.4	28.5 ± 12.0*	27.9 ± 10.6*

Mean ± SD. SOL, Soleus; GL, Gastrocnemius Lateralis; VM, Vastus Medialis; RF, Rectus Femoris; VL, Vastus Lateralis.

With vibration > no vibration (* $P < 0.05$, ** $P < 0.01$).

Table 8.3 Root mean square of the surface electromyography (sEMG_{RMS}) normalized to the sEMG_{RMS} during maximal voluntary contractions (MVCs) with the use of an additional load of 17 kg.

sEMG _{RMS} (%MVC) with additional 17kg load									
	No Vibration	1.2 mm Peak-to-Peak Displacement				2 mm Peak-to-Peak Displacement			
		25 Hz	30 Hz	35 Hz	40 Hz	25 Hz	30 Hz	35 Hz	40 Hz
SOL	12.1 ± 6.2	12.2 ± 6.2	12.6 ± 7.5	13.3 ± 7.7	13.8 ± 7.3	15.4 ± 9.8	16.6 ± 15.7	13.9 ± 7.0	15.0 ± 8.1*
GL	3.6 ± 2.3	3.7 ± 2.4	4.0 ± 2.8	4.0 ± 2.2	4.4 ± 2.6*	4.4 ± 3.1*	4.4 ± 2.7*	4.5 ± 2.2**	5.2 ± 2.8***
VM	29.2 ± 11.7	28.4 ± 11.7	27.8 ± 13.6	31.0 ± 15.8	29.0 ± 14.5	29.8 ± 12.1	28.2 ± 12.0	30.9 ± 15.9	31.4 ± 15.0
RF	17.2 ± 9.5	18.1 ± 9.9	16.9 ± 10.8	18.2 ± 12.6	17.5 ± 11.9	20.4 ± 12.0*	18.1 ± 9.9	18.3 ± 10.7	18.4 ± 10.0
VL	30.8 ± 12.1	30.6 ± 12.7	29.8 ± 11.8	32.3 ± 14.5	30.9 ± 13.1	32.8 ± 14.2	30.3 ± 11.3	33.3 ± 15.0	33.7 ± 14.9*

Mean ± SD. SOL, Soleus; GL, Gastrocnemius Lateralis; VM, Vastus Medialis; RF, Rectus Femoris; VL, Vastus Lateralis.

With vibration > no vibration (* $P < 0.05$, ** $P < 0.01$, *** $P < 0.001$).

Table 8.4 Root mean square of the surface electromyography (sEMG_{RMS}) normalized to the sEMG_{RMS} during maximal voluntary contractions (MVCs) with the use of an additional load of 33 kg.

sEMG _{RMS} (%MVC) with additional 33 kg load									
	No Vibration	1.2 mm Peak-to-Peak Displacement				2 mm Peak-to-Peak Displacement			
		25 Hz	30 Hz	35 Hz	40 Hz	25 Hz	30 Hz	35 Hz	40 Hz
SOL	19.4 ± 10.1	21.3 ± 10.7	24.1 ± 13.1**	23.4 ± 11.8**	22.4 ± 12.0**	24.7 ± 12.9**	23.5 ± 14.2*	23.4 ± 13.2*	24.4 ± 13.7*
GL	6.1 ± 3.8	6.8 ± 4.4**	8.0 ± 5.1***	7.8 ± 4.0**	8.4 ± 5.4*	7.9 ± 4.6**	7.6 ± 4.6**	7.9 ± 4.4**	9.0 ± 5.9**
VM	34.4 ± 14.7	33.6 ± 15.5	38.8 ± 16.5*	36.8 ± 18.9	37.2 ± 17.7	36.9 ± 17.6	38.2 ± 19.1	40.3 ± 20.1*	39.9 ± 19.1*
RF	19.5 ± 11.0	19.9 ± 12.5	20.7 ± 10.0	21.2 ± 13.7	21.6 ± 12.1	22.0 ± 13.5	19.8 ± 11.2	22.4 ± 11.3*	22.3 ± 13.2
VL	38.5 ± 15.8	37.3 ± 15.9	41.3 ± 15.7*	38.4 ± 16.7	41.2 ± 17.3	38.6 ± 16.0	40.0 ± 15.9	42.0 ± 16.5*	43.6 ± 18.2*

Mean ± SD. SOL, Soleus; GL, Gastrocnemius Lateralis; VM, Vastus Medialis; RF, Rectus Femoris; VL, Vastus Lateralis.

With vibration > no vibration (* $P < 0.05$, ** $P < 0.01$, *** $P < 0.001$).

Table 8.5 Multiple regression analysis results for the increase in root mean square of the surface electromyography (sEMG_{RMS}) for the calf and the thigh muscles.

Equation		<i>r</i>	<i>P</i>	
Calf Muscles	$sEMG_{RMS} = -2.209 + (0.049 \times L) + (1.164 \times D) + (0.042 \times FQ)$	0.823	Overall	< 0.001
			Load	< 0.001
			Displacement	< 0.01
			Frequency	NS
Thigh Muscles	$sEMG_{RMS} = -3.881 + (0.014 \times L) + (1.510 \times D) + (0.085 \times FQ)$	0.602	Overall	< 0.05
			Load	NS
			Displacement	< 0.05
			Frequency	NS

L, load; D, displacement; FQ, frequency; *r*, correlation coefficient; NS, not significant.

Discussion

The outcomes of this study were that (1) the sEMG of the lower limb muscles was not systematically positively affected by WBV, with the exception of the calf muscles which showed significantly higher sEMG during WBV compared to no vibration with an additional load of 33 kg. Further, during WBV, we highlighted that (2) only the highest vibration frequency (40 Hz) caused an enhancement in muscle activity compared to the lowest frequency (25 Hz) in the GL averaged for all loads, as well as in the VM and VL for the 33 kg load. Eminently, (3) the high peak-to-peak displacement of the platform (2 mm) increased sEMG_{RMS} in all five selected lower limb muscles to a higher extent than with the low peak-to-peak displacement (1.2 mm). (4) WBV exercises with a high load (33 kg) were found to stimulate muscle activity more than without a load in the GL. The results of the multiple regression analysis computed on the average muscle activity of the thigh and the calf muscles confirmed that the vibration frequency was not a significant predictor of the sEMG increases. Conversely, the displacement was a significant predictor in the calf and thigh muscles, and the load was a significant predictor in the calf muscles.

Numerous studies have investigated the effect of WBV compared to no vibration on lower limb muscle activity using a VV platform and for various platform displacements (1 mm – 4 mm) and

frequencies (20 - 60 Hz) (Cardinale and Lim 2003; Di Giminiani et al. 2013; Krol et al. 2011; Marín et al. 2009; Roelants et al. 2006). Without using an additional load, these authors reported significant effects of WBV on the lower limb muscle activity, which is conflicting with the findings of the present study. Without the additional load, muscle activity was not enhanced over the range of the frequencies (25 – 40 Hz) and the displacements measured (1.2 – 2 mm). The above mentioned studies did not delete the excessive spikes in the sEMG spectrum during WBV, and thus, direct comparison to the current study might be invalid (Fratini et al. 2009). Investigations which have withdrawn the excessive spikes have reported increased VL and GL neuromuscular responses at 30 Hz WBV (Abercromby et al. 2007a) and at 35 – 45 Hz for the VL (Hazell et al. 2007). Except for the GL muscle, these results are in line with the findings of the present study, as muscle activity of the VL was found to be enhanced during vibration at 35 Hz and 40 Hz with the high platform displacement and without load. Finally, only the calf muscles displayed consistent greater sEMG values during WBV with an additional load of 33 kg compared to the corresponding no vibration condition. This observation agrees with a previous study where dynamic squats were performed with an additional load corresponding to 30% of the participants' body mass (Hazell et al. 2010). In summary, our first hypothesis that WBV enhances muscle activity for all the tested conditions is poorly supported as only small sEMG_{RMS} increases were observed. This observation could partly explain why no significant neuromuscular alteration of the knee extensor muscles was observed between a fatiguing protocol (isometric squats) with WBV at 30 Hz and the same protocol without vibration (Maffiuletti et al. 2013). However, it is possible that WBV exercises have a higher effect in untrained and older individuals, which yet needs to be determined.

Our second hypothesis; muscle activity increases with increasing vibration frequency, was not fully supported by the findings of the present study. This result contradicts with an observation indicating that a frequency of 30 Hz induced higher GL activity compared to frequencies above 40 Hz (Di Giminiani et al. 2013). However, as seen from the results of the multiple regression analysis, the frequency did not significantly predict the increases in sEMG_{RMS} activity in the calf and the thigh muscles, although frequency-dependent increases in sEMG_{RMS} were found in the

GL, and the VL and VM with the use of an additional load. Taken as a whole, these results underline that an increase in vibration frequency did not necessarily lead to sEMG enhancements in the calf and thigh muscles. Nevertheless, a high frequency might have an effect on the muscle activity of the thigh muscles in combination with a high platform displacement. As an example, such a combination was successful to enhance jump performance (Petit et al. 2010; Colson and Petit 2013) and knee extensor muscle strength (Petit et al. 2010) after several weeks of training.

In contrast to the frequency, the displacement of the platform positively affected the sEMG of all measured lower limb muscles, which confirms our third hypothesis. Although contradictory results have been found for the VL (Hazell et al. 2007), our observations are in line with previous studies using VV platforms for the VL and VM (Krol et al. 2011), and the GM (Marín et al. 2009). Henceforth, and according to the percentage increases in muscle activity between the displacements (i.e., +59% for the GL, +72% for the SOL, +120% for the VM and RF, and +181% for the VL), we suggest that selecting the highest WBV peak-to-peak displacement is of high importance. Furthermore, considering that the peak-to-peak displacements differed by a very small magnitude (0.8 mm), even greater muscle enhancement can be expected during WBV using higher peak-to-peak displacements. Therefore, increasing the displacement of the platform during WBV was a successful strategy to enhance muscle activity in all lower limb muscles tested. This is particularly valuable when designing WBV protocols. As an example, a recent chronic WBV study has given evidence to believe that the use of a high displacement could also lead to a leaner body mass compared to the low displacement (Martínez-Pardo et al. 2013).

Interestingly, additional loading induced a significant effect in the GL, with higher sEMG increases at 33 kg than without load. Our results are further supported by the multiple regression analysis, as additional loading was identified as a significant predictor of sEMG in the calf muscles but not the thigh muscles. It is important to know that 17 kg corresponded to 25% of the body mass, and 33 kg to 50% of the body mass. The percentage increase in sEMG activity

during WBV with load compared to no load was already higher in the calf than the thigh muscles, ranging from +41% to +145% in the calf muscles and from +18% to +47% in the thigh muscles at 17 kg and 33 kg respectively. This might be a platform type-specific finding, as similar increases for upper and lower leg muscles have been reported using a side-alternating platform (Ritzmann et al. 2013). However, previous studies that used direct vibration on the muscle-tendon complex found that pre-activation of the muscles increased the sensitivity of the spindles with concurrent increase in neuromuscular response to vibration (Bedingham and Tatton 1984; Burke et al. 1976; Eklund and Hagbarth 1966). A study (Zaidell et al. 2013) comparing sEMG during direct vibration with WBV with and without additional loads supported such evidence during WBV, especially for the highest frequency tested (50 Hz). Due to the external loading, the calf muscles were pre-activated to a higher extent than the thigh muscles, thus it can be assumed that they therefore showed a higher response in sEMG activity to the vibration stimuli. Finally, our suggestion can be expanded by considering a recent study showing that an acute WBV exercise performed with a high load (70% of one-repetition maximum) with vascular occlusion could promote endurance-type muscular adaptations in recreationally active participants (Item et al. 2013).

External loading is a particular interesting strategy as the intensity of the WBV stimulus can be boosted without increasing the extent of the acceleration. Chronic exposure to occupational vibration has been linked to many side effects such as spinal degeneration, low back pain, and visual impairment (Cardinale and Pope 2003; Mester et al. 1999; Seidel 1993). Head accelerations while standing or squatting on a WBV platform have been described (Abercromby et al. 2007b) as higher than the recommended daily vibration exposure as defined by ISO 2631-1 (1997). However, boundaries for the daily limit of exercising on a WBV platform remain undetermined (Crewther et al. 2004). Thus, external loading is a better alternative to increase the WBV stimulus than increasing the magnitude of the acceleration. Physically inactive and elderly individuals have the option to use weight vests as an alternative to weight bars, in order to avoid risk of falling and mechanical stress at the spine level.

Conclusions

We have highlighted that WBV did not systematically increase the muscle activity of the lower limb compared to the same exercise without vibration. However, we recommend the use of a high platform displacement in combination with a high vibration frequency to provoke the highest muscle activity enhancement in both the thigh and the calf muscles while engaged in WBV. Alternatively, to train the calf muscles without maximizing the displacement or the frequency, an additional load equal to 50% of the body mass can be applied during WBV training. External loading during WBV is a particularly interesting option considering that the WBV stimulus is enhanced without increasing the extent of the acceleration, as transmission of the acceleration to the head has been associated with health-related concerns.

CHAPTER 9

STUDY V:

ACCELERATION THRESHOLD

9.1 PREFACE

After having determined the optimal frequency, amplitude and loading for exercising on a WBV platform, it was necessary to study the effect of the WBV platform per se. Vertical synchronous and side-alternating platforms have been compared before (Abercromby et al. 2007a; Ritzmann et al. 2013). However, the effect of horizontally vibrating platforms on muscle activity levels has never been investigated. As both the vertical synchronous and the side-alternating platforms induce vibrations mainly in the vertical plane, it suggests itself to question the effectiveness of horizontal vibrations. Therefore, the goal of this study was to examine how the direction of the induced acceleration affects the sEMG activity. This study was conducted in spring 2013, at the Human Performance Laboratory, University of Calgary, Alberta, Canada. Conveniently enough, this laboratory was working together with a Calgary based company, called Total Image Fitness, who designs WBV platforms. At that time, Total Image Fitness had just introduced a new platform type that can deliver vibrations in the horizontal plane via a circular mode, in the vertical plane via a side-alternating mode, and in all three planes via a dual mode (combination of the circular and the side-alternating modes). Hence, this new platform type was used in this study to test the effect of horizontal vs. vertical vibrations on muscle activity levels. Another aim of this study was to determine a minimal required acceleration threshold to induce significant increases in sEMG activity. Such a threshold would be particularly useful considering that high accelerations are associated with health concerns (ISO 2631-1 1997). Instead of describing the threshold using the frequency and amplitude, the resulting acceleration load was utilized. Previous studies have found that it is rather the acceleration that determines the muscular adaptations than the frequency and amplitude itself (Chen et al. 2014; Marín et al. 2012a; Siu et al. 2010). Also, determination of the required load using the acceleration can be more practical than using the frequency and amplitude. In summary, this study showed that the vertical platform acceleration, but not the horizontal platform acceleration, had a significant effect on sEMG activity of the lower limbs. Further, it was found that a vertical platform acceleration of $18 \text{ m}\cdot\text{s}^{-2}$ was sufficient to enhance the muscle activity of the lower limbs during WBV exercises on a side-alternating platform.

9.2 QUANTIFICATION OF VERTICAL ACCELERATION THRESHOLD TO ENHANCE LOWER LIMB MUSCLE ACTIVITY DURING WHOLE-BODY VIBRATION EXERCISES

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Abstract

The purpose of this study was to identify the influence of the platform acceleration direction on sEMG levels during WBV exercises and to determine an acceleration threshold that results in augmentation of sEMG activity. Therefore, a WBV platform was used that delivers vertical vibrations via a side-alternating mode, horizontal vibrations via a circular mode, and vibrations in all three planes via a dual mode. sEMG of selected lower limb muscles was measured in thirty individuals while they performed a static squat on a vibration platform. The WBV trials included two side-alternating trials (Side-L: 6 Hz, 2.5 mm; Side-H: 16 Hz, 4 mm), two circular trials (Circ-L: 14 Hz, 0.8 mm; Circ-H: 43 Hz, 0.8 mm), and four dual mode trials which were the combinations of the single mode trials (Side-L/Circ-L, Side-L/Circ-H, Side-H/Circ-L, Side-H/Circ-H). Further, control trials without vibration were assessed, and 3D platform acceleration was quantified during the vibration. Significant increases in sEMG compared to the control trial were found in most muscles during Side-H (+8 to +117%, $P < 0.05$), Side-L/Circ-H (+19 to +57%, $P < 0.05$), Side-H/Circ-L (+6 to +224%, $P < 0.01$), and Side-H/Circ-H (+18 to +203%, $P < 0.001$). Further, it was found that the vertical platform acceleration was a significant predictor of the sEMG in the lower limb muscles, with a threshold of $18 \text{ m}\cdot\text{s}^{-2}$. The present results emphasize that a vertical platform acceleration of at least $18 \text{ m}\cdot\text{s}^{-2}$ should be induced to enhance muscle activity during WBV exercises.

Introduction

WBV has become an established exercise modality to improve muscle activity, strength and power of the lower limbs in healthy participants, athletes, the elderly, and health compromised individuals (Cochrane 2011). Numerous studies have indeed shown greater muscle activity of the lower limbs during WBV compared to no vibration with the help of sEMG recordings (Cardinale and Lim 2003; Cochrane et al. 2009; Hazell et al. 2010; Marín et al. 2012a; Ritzmann et al. 2013; Roelants et al. 2006). This increased muscle activity observed during WBV is believed to result from stretch reflex responses (Pollock et al. 2012; Ritzmann et al. 2010), induced by changes in length of the muscle spindles due to rapid changes in the length of the muscle-tendon units (Cochrane et al. 2009). To date, the WBV platforms used in these studies induced the vibrations mainly in the vertical direction via side-alternating or synchronous vertical platforms while WBV platforms inducing oscillations in the horizontal planes were newly commercialized. Although side-alternating, synchronous vertical and horizontal platforms deliver distinct magnitudes of acceleration in the different planes (Pel et al. 2009), surprisingly enough, all these platforms have a similar effect on jump performance (Bagheri et al. 2012).

Besides the platform type that delivers vibrations in the horizontal plane, a new model inducing vibrations in all three planes (i.e. dual mode) is available that has never been examined. Dual mode WBV platforms are unique in that two single modes, individually adjustable in frequency and amplitude, are combined. Unlike the commonly used single mode platforms that allow vibrations in either the vertical direction (Cardinale and Wakeling 2005; Cochrane 2011; Pel et al. 2009; Ritzmann et al. 2013) or the anterior-posterior and medio-lateral directions (Bagheri et al. 2012; Muir et al. 2013; Pel et al. 2009), dual mode platforms permit vibrations in all three planes. Acknowledging that maximized sEMG activity during WBV exercises leads to positive acute and chronic effects of neuromuscular performance (Di Giminiani et al. 2009; 2010), using dual mode platforms could be an advantage over the single mode platform types. Nevertheless, the influence of platform type on sEMG activity levels of the lower limb muscles has never been quantified.

In many research studies, increasing responses in sEMG activity were found with an increasing vibration frequency and/or amplitude (Hazell et al. 2007; Krol et al. 2011; Perchthaler et al. 2013; Pollock et al. 2010; Ritzmann et al. 2013). However, this effect might not be due to the frequency and the amplitude per se, as increasing the frequency and amplitude also results in an increase in acceleration. Indeed, the acceleration load imposed on the neuromuscular system during WBV exercises is characterized by the interaction between the vibration frequency and the amplitude of the platform displacement (Mester et al. 1999). Marin et al. (Marín et al. 2012b) tested the effect of frequency and amplitude vs. the effect of the resulting vertical platform acceleration. Two WBV conditions consisting of different frequencies and amplitudes that added up to identical acceleration magnitudes were used. In that study, similar increases in sEMG activity between the two conditions were found, indicating that the main load parameter is rather the vertical acceleration than the frequency and amplitude in itself. Besides the changes in muscle activity during the exposure to WBV, the vertical acceleration load also seems to be the key factor for immediate and chronic benefits. Using different frequencies and amplitudes adding up to the same vertical acceleration load showed identical adaptations following WBV exercises (Chen et al. 2014; Siu et al. 2010;). While many studies have been researching the frequency-amplitude combination that is needed to enhance muscle activity of the lower limbs, it would be more convenient to determine a vertical platform acceleration threshold that leads to significant increases of the sEMG.

Therefore, the goal of this study was to compare muscle activity of selected lower limb muscles during WBV in the vertical plane, the horizontal plane, and the combination of vibrations in all three planes. Vertical vibrations were induced via a side-alternating mode, horizontal vibrations via a circular mode, and the combined vibrations via the dual mode using one single WBV platform. It was hypothesized that (1) the vertical platform acceleration would be a predictor of sEMG activity of the lower limbs and that (2) there would be a vertical acceleration threshold that leads to significant increases in sEMG activity compared to the no-vibration trial.

Methods

Experimental Approach to the Problem

To test the first hypothesis, a WBV platform was used that delivers vibrations in the vertical direction via a side-alternating mode, in the horizontal direction via a circular mode, and in all three planes via a dual mode. To test the second hypothesis, different acceleration thresholds were induced, using the lowest and the highest possible frequency-amplitude settings for each vibration mode. For comparison, control trials without vibration were assessed. The sEMG of six lower limb muscles was measured while the participants performed static squats at a knee angle of 60° (0° corresponding to fully extended). The sEMG measured during the WBV and control trials was normalized to the sEMG measured during isometric MVCs.

Subjects

Thirty physically active volunteers (15 males and 15 females, age: 25.9 ± 4.3 years, height: 171.4 ± 9.5 cm, body mass: 69.5 ± 11.2 kg, mean \pm SD) participated in this study. All participants were new to WBV platforms and free from injuries to the lower extremities. The study was approved by the University of Calgary's conjoint health research ethics board, and all participants provided informed consent prior to their participation.

sEMG Recordings

sEMG electrodes were placed on the muscle bellies of the TA, GM, SOL, VL, VM, and BF of the right lower limb according to the SENIAM guidelines (Hermens et al. 2000) using bipolar surface electrodes (Ag-AgCl) with preamplifiers (Biovision, Wehrheim, Germany). A ground electrode was placed on the right tibial tuberosity. The electrodes had a 10 mm diameter and 22 mm inter electrode spacing. Prior to the placement of the electrodes, the skin was shaved, dead skin cells were removed using abrasive tape, and the area was cleaned with an isopropyl wipe. The electrodes were secured to the skin with Cover-Roll stretch tape (Beiersdorf AG, Hamburg, Germany). Voltage signals were pre-amplified at the source (gain = 1000) and sampled at 2400 Hz.

Assessment of MVCs

After completion of a standardized 5-min warm-up on a stationary bike, two isometric MVCs were completed for plantarflexion, dorsiflexion, knee extension, and knee flexion on an isokinetic dynamometer (Biodex System 3, Medical Systems Inc., New York, NY, USA). sEMG of the SOL and GM were recorded during plantarflexion, TA during dorsiflexion, VL and VM during knee extension, and BF during knee flexion. Plantarflexion and dorsiflexion were performed at an ankle joint angle of 90° (neutral position) and the rotational axis of the dynamometer was visually aligned to the lateral malleolus. Knee extension and flexion were performed at a knee angle of 60° (0° corresponding to fully extended) and the rotational axis of the dynamometer was aligned to the lateral femoral condyle. All participants performed two 5-s MVC trials for each setting for a total of 8 MVCs. During these contractions, the participants were instructed to produce as much force as they could and to maintain this force at a constant level. After each trial, the participants were given a 1-min break. During each 5-s contraction, the peak MVC torque over 500 ms was calculated. As there were two trials for each setting, the higher torque was retained to calculate the maximal sEMG. The sEMG_{RMS} was computed during the 500 ms period of the selected torque output from the muscle corresponding to the contraction.

WBV Platform

The platform used in this study (TBS 100A, Total Image Fitness, Calgary, Alberta, Canada) delivers oscillations via two single modes (i.e., side-alternating and circular mode) and a dual mode, which is the combination of the two single modes. The side-alternating mode generates vibrations mainly in the vertical plane by rotation along the sagittal axis, with a frequency range of 6 to 16 Hz. The amplitude of the vibration is dictated by the positioning of the feet in relation to the axis of rotation, and was determined by double integration of the platform acceleration signal using a 3D accelerometer (ADXL 78, Analog Devices USA, encapsulated in a 20/12/5 mm plastic shell, measuring range ± 70 g, frequency response of 0-400 Hz, mass: 2 grams). A foot position of 23.3 cm from the central axis corresponded to an amplitude (displacement from baseline to peak) of 4 mm, and a foot position of 14.3 cm corresponded to an amplitude of 2.5 mm. The circular mode generates vibrations mainly in the horizontal plane by circular rotation

with a fixed amplitude (radius of the circular movement) of 0.8 mm. The frequency for this mode ranges from 14 to 43 Hz. The dual mode of this platform consists of the combination of the two single modes, whereas the intensity of each mode is individually adjustable. **Figure 9.1** illustrates the acceleration signals measured on the platform level during side-alternating, circular, and dual mode WBV in the anterior-posterior plane (Acc a-p), the medio-lateral plane (Acc m-l), and the vertical plane (Acc v).

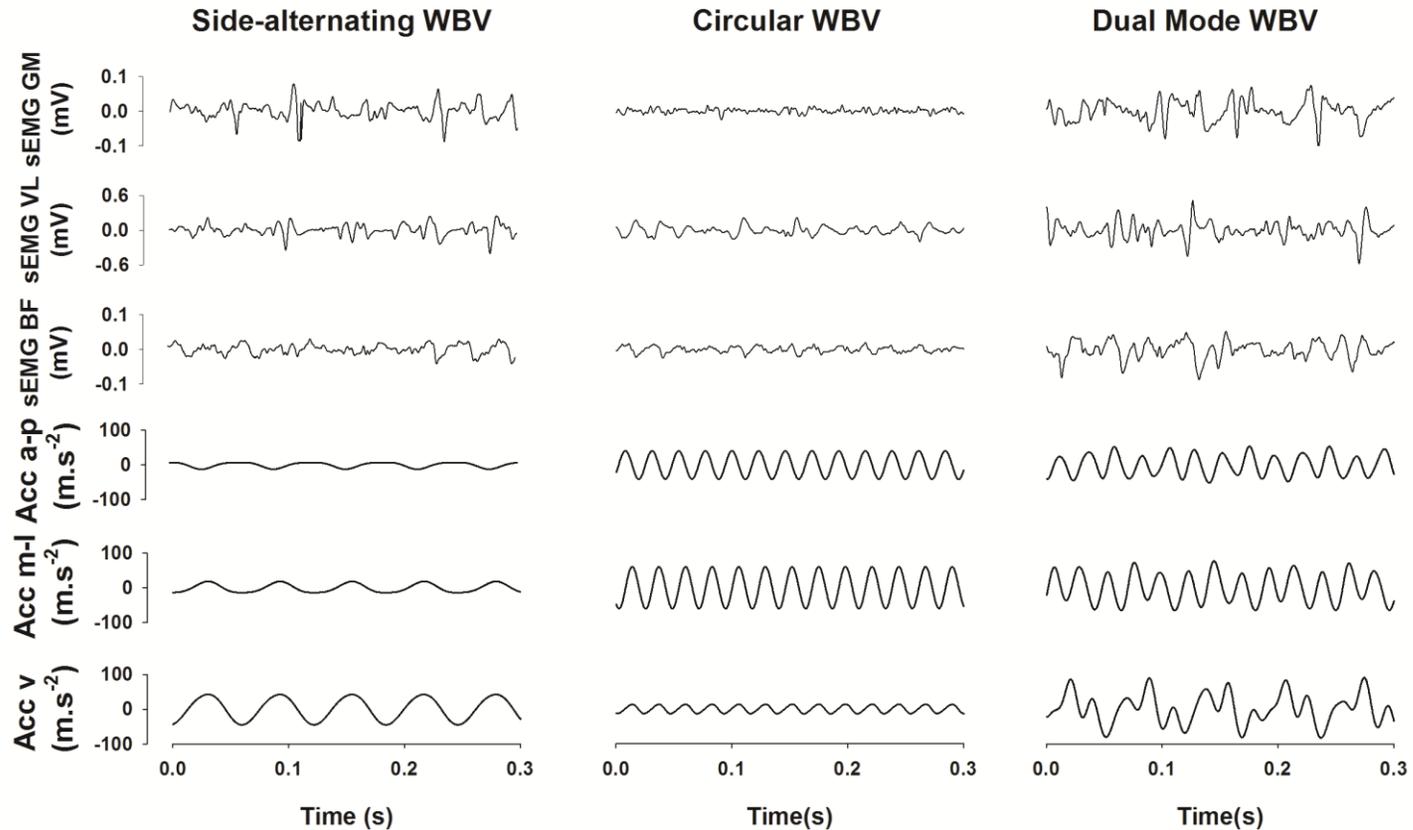


Figure 9.1 Surface electromyography (sEMG) and platform acceleration data of one participant during side-alternating whole-body vibration (WBV) at 16 Hz and 4 mm amplitude (left panel), during circular WBV at 43 Hz and 0.8 mm amplitude (middle), and during dual mode WBV consisting of the combination of the side-alternating (16 Hz, 4 mm) and the circular (43 Hz, 0.8 mm) mode (right panel). The top three rows show sEMG signals of the gastrocnemius medialis (GM), vastus lateralis (VL), and biceps femoris (BF). The bottom three rows illustrate the platform acceleration signals in the anterior-posterior plane (Acc a-p), medio-lateral plane (Acc m-l), and the vertical plane (Acc v).

WBV Protocol

The WBV test protocol consisted for each mode of the lowest and the highest possible intensity settings in order to achieve a wide range of the acceleration magnitudes. For the side-alternating mode, the low intensity setting consisted of the lowest platform setting which was 6 Hz and an amplitude of 2.5 mm (Side-L) corresponding to the narrow foot position. The high intensity setting consisted of the highest platform setting which was 16 Hz and an amplitude of 4 mm (Side-H) corresponding to the wide foot position. For the circular mode, the low and high intensity settings were 14 Hz (Circ-L) and 43 Hz (Circ-H), respectively, with a fixed amplitude of 0.8 mm. Both circular mode trials were performed using the narrow foot position. To create dual mode WBV, both side-alternating settings and both circular settings were combined. This resulted in four dual mode settings (Side-L/Circ-L, Side-L/Circ-H, Side-H/Circ-L, Side-H/Circ-H). Additionally, control trials without vibration were assessed for the wide and the narrow foot positions, which resulted in a total of 10 trials (2 control trials + 2 side-alternating trials + 2 circular trials + 4 dual mode trials), presented in a randomized order (**Table 9.1**). Each trial consisted of 20 s of vibration with a 1-min break between trials. During the exercise, the participants were advised to perform a static squat at a knee angle of 60°. The knee flexion angle was monitored using a digital goniometer (SG150 Twin Axis, Biometrics Ltd, Newport, UK) that was taped to the participant's left knee. During the exercise on the WBV platform, the participants were provided with a visual real-time feedback of their knee angle, illustrated on a laptop screen using MATLAB software (version 7.13 The Mathworks, Inc., Natick, MA, USA). The trial was repeated if the measured knee angle varied more than $\pm 3^\circ$ at any point during the exercise. The participants stood barefoot on the WBV platform in order to avoid damping effects due to different footwear, and they were instructed not to hold onto the handle bar of the platform.

Table 9.1 The whole-body vibration trials assessed in this study and their acceleration amplitudes (mean \pm SD).

	Abbreviation	Foot	Side-alternating		Circular		Acceleration		
		position	vibration		vibration		a - p	m - l	v
		Dis to axis (cm)	Freq (Hz)	Amp (mm)	Freq (Hz)	Amp (mm)	(m.s ⁻²)	(m.s ⁻²)	(m.s ⁻²)
No-vibration trials	NoVib (narrow)	14.3	-	-	-	-	-	-	-
	NoVib (wide)	23.3	-	-	-	-	-	-	-
Side-alternating trials	Side-L	14.3	6.0	2.5	-	-	0.5 \pm 0.1	2.7 \pm 0.2	4.2 \pm 0.4
	Side-H	23.3	16.0	4.0	-	-	15.6 \pm 4.4	14.6 \pm 2.1	50.5 \pm 5.7
Circular trials	Circ-L	14.3	-	-	14.0	0.8	6.6 \pm 0.6	6.7 \pm 0.4	0.9 \pm 0.4
	Circ-H	14.3	-	-	43.0	0.8	45.4 \pm 6.5	59.6 \pm 3.9	13.1 \pm 4.9
Dual mode trials	Side-L / Circ-L	14.3	6.0	2.5	14.0	0.8	6.8 \pm 0.3	8.3 \pm 0.8	4.4 \pm 0.6
	Side-L / Circ-H	14.3	6.0	2.5	43.0	0.8	46.7 \pm 7.2	60.6 \pm 5.7	17.5 \pm 5.9
	Side-H / Circ-L	23.3	16.0	4.0	14.0	0.8	16.6 \pm 3.4	14.6 \pm 1.8	51.9 \pm 4.5
	Side-H / Circ-H	23.3	16.0	4.0	43.0	0.8	57.4 \pm 6.8	71.6 \pm 4.6	71.8 \pm 7.8

Dis, Distance; Freq, Frequency; Amp, Amplitude (displacement from baseline to peak); a-p, anterior-posterior; m-l, medio-lateral; v, vertical.

sEMG Analysis

sEMG analysis was performed using MATLAB software (version 7.13 The Mathworks, Inc., Natick, MA, USA). The sEMG signals were first clipped so that only 10 s remained for further analysis (time: 7-17 s). Then, the sEMG signals were filtered using a band-pass wavelet filter with a low cut-off frequency of 5 Hz and a high cut-off frequency of 300 Hz. Next, the excessive spikes observed in the sEMG spectrum at the vibration frequency and its multiple harmonics were removed via linear interpolation in the PSD. Then, sEMG_{RMS} during WBV and the control trials were calculated and expressed as a percentage value of the maximum sEMG_{RMS} of the respective muscle obtained during the MVC trials.

Statistical Analyses

All statistical procedures were performed using IBM SPSS software (Version 20, Chicago, IL). The Kolmogorov-Smirnov test confirmed the normality of the data. One-way repeated analysis of variance (ANOVA) was used to compare sEMG_{RMS} recorded during each WBV condition (i.e., Side-L, Side-H, Circ-L, Circ-H, Side-L/Circ-L, Side-L/Circ-H, Side-H/Circ-L, Side-H/Circ-H) to the sEMG_{RMS} during the corresponding no-vibration trial. Pearson's correlation coefficients were computed between the sEMG_{RMS} averaged for all muscles and the platform acceleration of each plane. Further, a linear multiple regression analysis was performed on the sEMG_{RMS} averaged for all muscles, including the side-alternating, circular, and dual mode trials. The measured acceleration values (**Table 9.1**) in the anterior-posterior, medio-lateral and vertical directions served as predictors. The level of significance was set at $\alpha = 0.05$, and the results are presented as mean \pm standard deviation (SD).

Results

WBV compared to no-vibration

Figure 9.2 illustrates the $sEMG_{RMS}$ during side-alternating, circular and dual mode WBV compared to the no-vibration trial. Comparing side-alternating WBV to the corresponding no-vibration trial, significant increases in $sEMG_{RMS}$ were found for Side-L in the BF (+18%, $P = 0.020$), and for Side-H in all muscles (TA: +32%, $P = 0.007$; GM: +117%, $P < 0.001$; SOL: +112%, $P < 0.001$; VL: +8%, $P = 0.041$; VM: +8%, $P = 0.049$; BF: +28%, $P = 0.028$).

Comparing circular WBV to the corresponding no-vibration trial, significant increases in $sEMG_{RMS}$ were found for Circ-L in the GM (+23%, $P = 0.017$), and for Circ-H in the TA (+22%, $P = 0.046$), GM (+34%, $P = 0.020$), and BF (+17%, $P = 0.042$).

Dual mode WBV Side-L/Circ-L showed significantly higher $sEMG_{RMS}$ compared to no-vibration in the GM (+47%, $P < 0.001$) and SOL (+26%, $P = 0.019$). The dual mode trial Side-L/Circ-H significantly enhanced muscle activity in the GM (+57%, $P < 0.001$), SOL (+55%, $P = 0.004$), VL (+19%, $P = 0.025$), VM (+23%, $P = 0.002$) and BF (+53%, $P < 0.001$). $sEMG$ activity during dual mode WBV Side-H/Circ-L was significantly higher than the no-vibration trial in the TA (+49%, $P < 0.001$), GM (+224%, $P < 0.001$), SOL (+113%, $P < 0.001$), VL (+9%, $P < 0.001$) and VM (+6%, $P < 0.001$). Finally, the dual mode trial Side-H/Circ-H significantly enhanced $sEMG_{RMS}$ compared to no-vibration in all the measured muscles (TA: +51%, $P < 0.001$; GM: +203%, $P < 0.001$; SOL: +144%, $P < 0.001$; VL: +18%, $P < 0.001$; VM: +21%, $P < 0.001$; BF: +58%, $P < 0.001$).

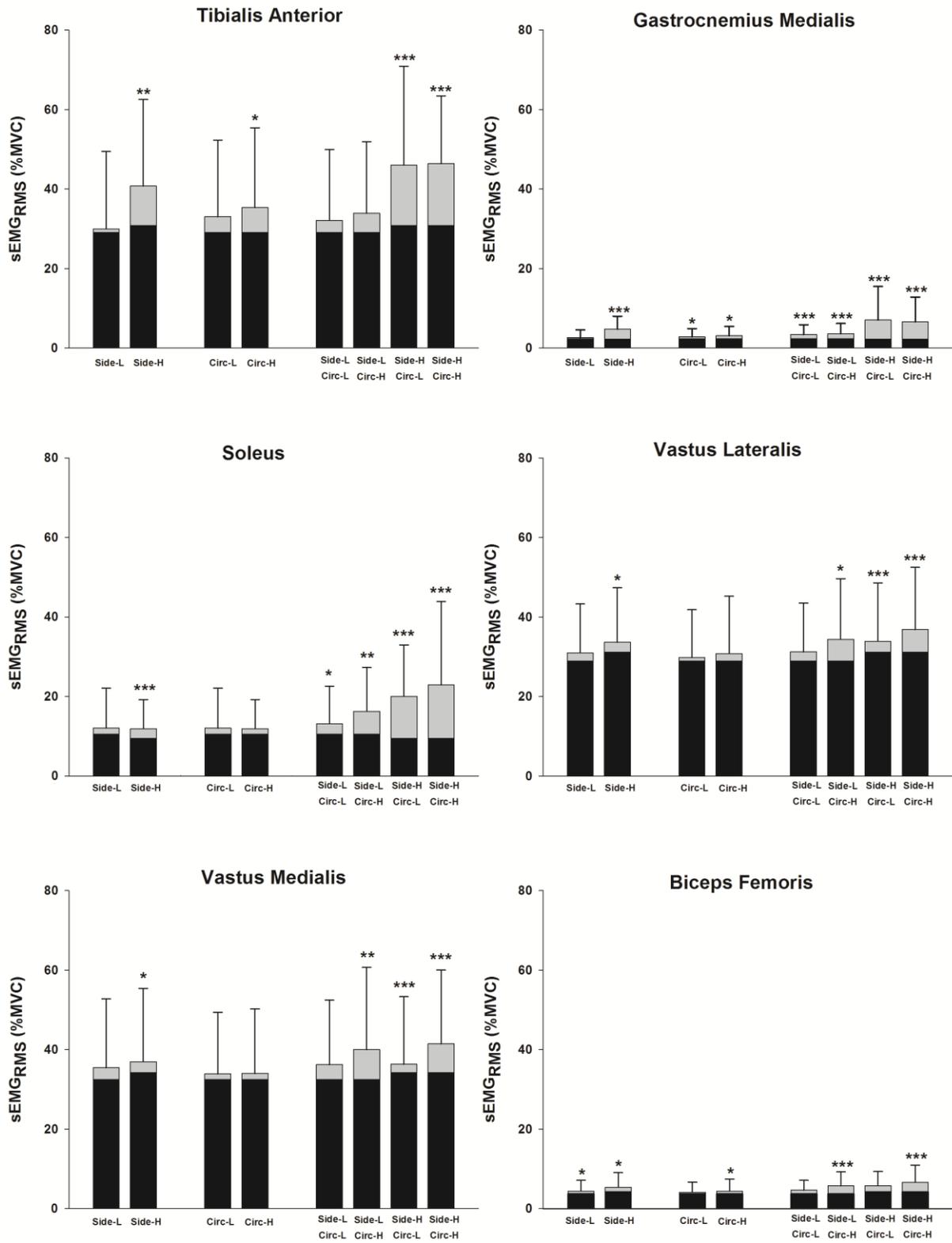


Figure 9.2 Root mean square of the surface electromyography (sEMG_{RMS}) during side-alternating (Side), circular (Circ) and dual mode (Side/Circ) whole-body vibration (WBV) for all the assessed conditions (Side-L: 6 Hz, 2.5 mm; Side-H: 16 Hz, 4 mm; Circ-L: 14 Hz, 0.8 mm; Circ-H: 43 Hz, 0.8 mm). The asterisks indicate significant increases in sEMG_{RMS} during WBV (in grey) compared to the corresponding no-vibration trial (in black, * $P < 0.05$, ** $P < 0.01$, *** $P < 0.001$, mean \pm SD).

Relation between sEMG_{RMS} and Platform Acceleration

The Pearson's correlation coefficient between the averaged lower limb sEMG_{RMS} and the platform acceleration was high and significant for the vertical component ($r = 0.970$, $P < 0.001$), and low and non-significant for the anterior-posterior ($r = 0.543$, $P = 0.164$) and medio-lateral direction ($r = 0.462$, $P = 0.249$, **Figure 9.3**). Similarly, the results of the multiple regression analysis showed that only the vertical direction of the platform acceleration added significantly to the prediction ($P = 0.025$) with the equation $\text{sEMG}_{\text{RMS}} (\% \text{MVC}) = 19.13 + (0.16 \times \text{Acc a-p}) - (0.11 \times \text{Acc m-l}) + (0.08 \times \text{Acc v})$.

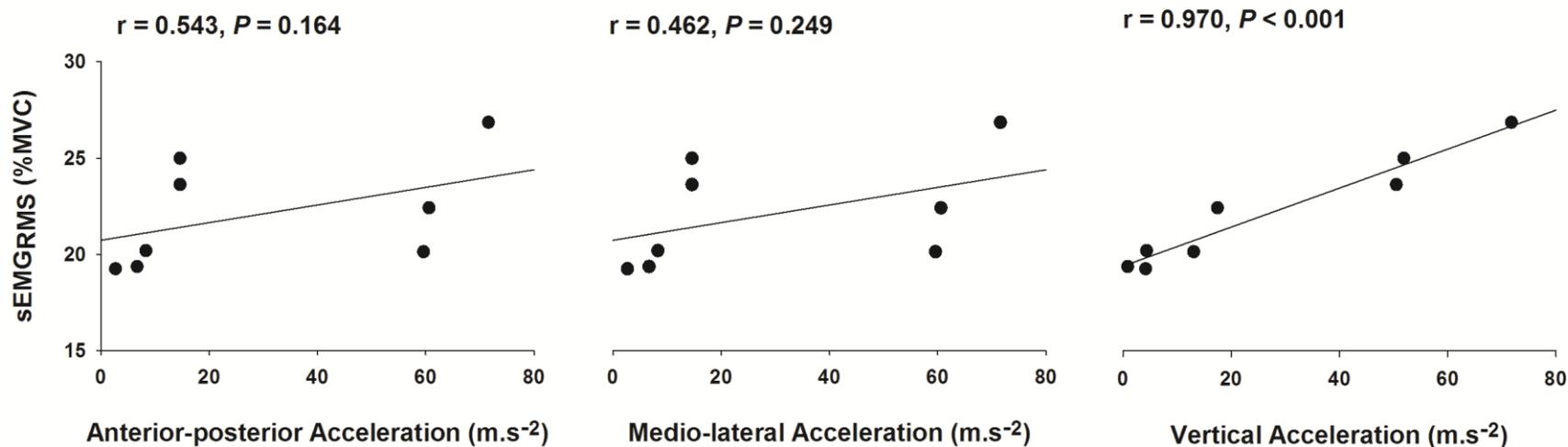


Figure 9.3 Scatterplots of the averaged sEMG_{RMS} for all the measured lower limb muscles and the platform acceleration in the anterior-posterior plane (left), the medio-lateral plane (middle), and the vertical plane (right). Each point represents the averaged sEMG_{RMS} of all muscles and participants for a given WBV trial. The vertical platform acceleration showed a high and significant Pearson's correlation coefficient ($r = 0.970$, $P < 0.001$) with the lower body sEMG_{RMS}.

Discussion

Side-alternating WBV at high intensity (Side-H) and dual mode WBV consisting of the high-intensity side-alternating mode or the high-intensity circular mode (Side-H/Circ-L, Side-L/Circ-H, Side-H/Circ-H) showed significant increases in sEMG activity of most muscles compared to the no-vibration trial. Further, the magnitude of the sEMG activity of the lower limb muscles was positively correlated to the magnitude of the vertical platform acceleration. More specifically, only the vertical component of the platform acceleration was a significant predictor of the muscle activity. Finally, we delineated a vertical acceleration threshold leading to significant improvement of the sEMG activity of the selected lower limb muscles.

To the best of our knowledge, this is the first study quantifying muscle activity during circular WBV in the horizontal plane and during dual mode WBV in all three planes. Numerous studies have investigated the effects of side-alternating WBV on muscle activity and, similarly to the present study, found significant increases in sEMG activity compared to baseline values (Abercromby et al. 2007a; Cochrane et al. 2009; Ritzmann et al. 2013). Further, the measured sEMG_{RMS} values expressed as a percentage of the maximum are comparable to the ones found in previous studies (Pollock et al. 2010; Perchthaler et al. 2013), using the same platform type and acceleration magnitudes measured at the platform level.

Interestingly, we found that the magnitude of the sEMG activity of the lower limb muscles was positively correlated to the magnitude of the vertical platform acceleration. Further, only the vertical component of the platform acceleration was a significant predictor of the muscle activity, which confirms our first hypothesis. This finding seems to be applicable to various platform types, as a previous study performed on a synchronous WBV platform reported that the vertical acceleration of the platform was correlated with the lower body sEMG activity, even though the acceleration was not a predictor of the muscle activity (Marín et al. 2012a). Our results clearly show that the higher the vertical platform acceleration, the higher the responses in sEMG activity. Here, the WBV trials with a vertical acceleration below $4 \text{ m}\cdot\text{s}^{-2}$ showed little to no effect on muscle activity levels. A vertical acceleration of $13 \text{ m}\cdot\text{s}^{-2}$, such as that measured

during circular vibration at 43 Hz and 0.8 mm amplitude, significantly increased muscle activity in the TA, GM, and BF muscles. A minimum vertical acceleration of $18 \text{ m}\cdot\text{s}^{-2}$ was necessary to enhance muscle activity in almost all the measured muscles. Hence, the second hypothesis that muscle activity during WBV would be higher than during the no-vibration trial for a certain vertical acceleration threshold was confirmed. Yet, this conclusion is only applicable to side-alternating WBV platforms. Synchronous WBV platforms have been shown to induce lower responses in sEMG activity compared to side-alternating WBV platforms (Abercromby et al. 2007a; Ritzmann et al. 2013), despite using the same acceleration load. Hence, the marginal threshold of a synchronous WBV platform is likely higher than $18 \text{ m}\cdot\text{s}^{-2}$ and needs to be quantified in future studies.

From a practical point of view, our results indicate that WBV exercises should be performed using the highest frequency-amplitude combination available in the vertical direction, in order to achieve the highest increases in muscle activity. Although this conclusion has been reported elsewhere (Perchthaler et al. 2013; Pollock et al. 2010; Ritzmann et al. 2013), we have delineated for the first time, a vertical acceleration threshold leading to lower body sEMG activity increases compared to a no-vibration trial. It is recommended to measure the vertical acceleration effectively delivered by the platform rather than using the theoretical frequency and amplitude settings provided by the platform manufacturers. Using the platform from the current study, vertical accelerations ranging from $18 \text{ m}\cdot\text{s}^{-2}$ (SideL/Circ H) to $72 \text{ m}\cdot\text{s}^{-2}$ (SideH/Circ H) elicited significant increases in sEMG activity. These values are consistent and within the range of vertical acceleration values previously observed on a synchronous platform (Tankisheva et al. 2013). According to these authors, WBV training within the range of these vertical acceleration values in a flexed knee position, such as the one of the present study, is rather safe. However, as the long-term effects of regular WBV exercises are unclear, and because no safety norms have been specifically developed for WBV devices, it might be preferred to use the lowest vertical acceleration load that is required to enhance the muscle activity of the lower limbs.

The mechanism often proposed for the increases of sEMG activity during WBV compared to no-vibration trials is stretch reflex responses induced by activation of muscle spindles due to changes in length in the muscle-tendon units (Cochrane et al. 2009). Indeed, recent studies have evidenced that synchronous motor unit activity (Pollock et al. 2012) and a tonic vibration reflex-like response (Zaidell et al. 2013) occur during WBV and might account for the augmented sEMG activity. Further, increased muscle activity as a response to vibration has been described as a mechanism to damp the oscillations in order to protect the soft tissue from damage (Wakeling et al. 2002). Although both mechanisms could explain that the vertical acceleration was a predictor of sEMG increases, no measurements were made in the present study to ascertain it. Also, there are no studies that have investigated if these responses could occur during horizontal vibrations.

It needs to be considered that the circular mode induced substantial vibrations in the vertical plane. For example, during circular WBV at high intensity (43 Hz, 0.8 mm), accelerations of $13 \text{ m}\cdot\text{s}^{-2}$ were measured in the vertical plane, even though this mode effectively delivers accelerations of $45 \text{ m}\cdot\text{s}^{-2}$ in the anterior-posterior plane and $60 \text{ m}\cdot\text{s}^{-2}$ in the medio-lateral plane. As only the vertical acceleration was a significant predictor of sEMG activity, it can be speculated that the significant increases in muscle activity during circular WBV were rather due to the vertical than the horizontal vibrations. Even if designed to induce mainly horizontal vibrations, the characteristics of the used platform did not allow to assess the effects of horizontal vibrations without any vertical component, which opposes a limitation to the present study. It is possible that the horizontal vibrations were not identified as a significant predictor of sEMG activity because the vertical accelerations overpowered the effects of the horizontal vibrations. Another limitation of the present study was the relatively low frequency range of the side-alternating mode. Commonly, side-alternating platform vibration frequencies reach up to 30 Hz (Abercromby et al. 2007a; Perchthaler et al. 2013; Pollock et al. 2010; Ritzmann et al. 2013), whereas the maximal frequency of the current platform was 16 Hz. Nevertheless, combined with the amplitude, this frequency setting induced a vertical acceleration platform of $50 \text{ m}\cdot\text{s}^{-2}$, which was sufficient enough to significantly increase muscle activity.

Practical Applications

The results of the present study suggest that trainers and therapists should advise their clients to exercise on WBV platforms using frequencies and amplitudes resulting in a minimal vertical platform acceleration of $18 \text{ m}\cdot\text{s}^{-2}$. Such an acceleration threshold was required to effectively stimulate muscle activity of the lower limbs in a trained population. Thus, it needs to be considered that this threshold might be even lower for their physically less fit, older, and health compromised counterparts, and even higher for highly trained athletes. For progression in training, the vertical acceleration delivered by the platform should be augmented by manipulating the frequency and/or amplitude, as this was found to further enhance lower limb muscle activity during WBV exercises.

CHAPTER 10

STUDY VI:

ACTIVE VS. INACTIVE SUBJECTS

10.1 PREFACE

Once it was clarified how a person should exercise on a WBV platform, it was aimed to clarify who can benefit from WBV exercises. In the previous five investigations, the study population consisted of young, healthy, and physically active men and women. WBV platforms are available in many fitness centers, where they are also used by physically less fit individuals. Hence, the goal of this study was to compare the effect of WBV exercises on sEMG activity between an active and an inactive cohort. Since a squatting position can be challenging for unathletic individuals, a standing position besides the squatting position was included in study VI. As the health concerns related with WBV exercises are more and more discussed, transmissibility to the head was quantified and compared between two knee flexion angles and between the two groups. This study was conducted together with study V in spring 2013, at the Human Performance Laboratory, University of Calgary, Alberta, Canada. The WBV protocol was chosen according to the findings from the preceding studies. WBV was induced in the vertical direction using the side-alternating mode. The induced accelerations were above the established threshold of $18 \text{ m}\cdot\text{s}^{-2}$, and the highest possible frequency-amplitude combination was chosen. The motion artifacts were removed with the help of the spectral linear interpolation method, and the sEMG during WBV was normalized to the sEMG during the MVC of the respective muscle. The outcomes of this study were that active as well as inactive subjects can benefit from WBV exercises by means of enhanced muscle activity of the lower limbs. Inactive subjects are not at higher risk to potential side-effects of vibration than their active counterparts as transmissibility to the head was lower in the inactive group. Finally, while standing on the platform leads to higher increases in muscle activity than the squatting position, head vibrations can be reduced by adapting the latter position.

10.2 THE EFFECT OF WHOLE-BODY VIBRATION ON MUSCLE ACTIVITY IN ACTIVE AND INACTIVE SUBJECTS

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Abstract

The purpose of this study was to compare lower limb muscle activity between physically active and inactive individuals during WBV exercises. Additionally, transmissibility of the vertical acceleration to the head was quantified. Thirty active and twenty-eight inactive participants volunteered to stand relaxed (20°) and squat (60°) on a side-alternating WBV platform that induced vibrations at 16 Hz and 4 mm amplitude. The sEMG was measured of selected lower limb muscles and was normalized to the corresponding sEMG recorded during the MVC. The vertical acceleration on the head was evaluated and divided by the vertical platform acceleration to obtain transmissibility values. Further, control trials without vibration were assessed. The outcomes of this study showed that (1) WBV significantly increased muscle activity in the active (+127%, $P < 0.05$) and inactive participants (+116%, $P < 0.05$), (2) with no differences in sEMG increases between the groups ($P > 0.05$). However, (3), transmissibility to the head was greater in the active (0.080) than the inactive participants (0.065, $P < 0.05$). In conclusion, inactive individuals showed similar responses in sEMG due to WBV as their active counterparts, but were at lower risk to potential side-effects of vibration exposure.

Introduction

The instantaneous effects of WBV have been intensively investigated by focusing on the increases in muscle activity as measured by sEMG recordings (Cardinale and Lim 2003; Marín et al. 2009; Perchthaler et al. 2013). These increases in muscle activity are presumably due to TVR responses (Ritzmann et al. 2010; Zaidell et al. 2013) and increased motor unit recruitment (Pollock et al. 2012), induced by changes in length of the muscle spindles (Cochrane et al. 2009). However, in young healthy people, research studies have been focusing on physically active individuals), and the neuromuscular changes during WBV exposure are widely unknown in their physically inactive counterparts.

There are indications that inactive populations might benefit even more from WBV exercises than active populations. For example, (Rønnestad 2009b) has reported that untrained subjects improved their one-repetition maximum of the leg to a higher extent after exposure to WBV than trained subjects. Also, lower limb power as measured by counter movement jump performance was affected by WBV only in untrained individuals (Rønnestad 2009a). This immediate power improvement in untrained subjects could be attributed to two main mechanisms. First, untrained individuals lack the ability to fully activate their muscles (i.e., mainly the high-threshold motor units) during maximal contractions (Amiridis et al. 1996). Second, WBV exposure leads to reduced recruitment thresholds of high-threshold motor units (Pollock et al. 2012). These observations support that untrained populations might have greater potential to recruit high-threshold motor units during vibration compared to trained populations. Therefore, higher sEMG increases in inactive as compared to active individuals are expected during WBV exercises.

The effect of knee flexion angle on lower limb muscle activity during WBV has been investigated in several studies (Avelar et al. 2013; Di Giminiani et al. 2013; Perchthaler et al. 2013; Ritzmann et al. 2013; Roelants et al. 2006). Most studies reported that the sEMG of the upper leg muscles (Avelar et al. 2013; Di Giminiani et al. 2013; Perchthaler et al. 2013; Ritzmann et al. 2013) and the sEMG of the dorsi flexors (Ritzmann et al. 2013) increased with an increasing knee flexion

angle and vice versa for the plantar flexors (Ritzmann et al. 2013). However, the increases in sEMG activity that were found were rather posture-related than an adjustment to WBV, since the muscles of the upper leg and the dorsi flexors need to generate higher forces in order to maintain the posture when increasing knee flexion (Pincivero et al. 2003). Hence, to effectively investigate the adding effect of WBV between knee angles, the sEMG should be compared to the sEMG activity measured during a respective no-vibration trial (i.e., identical knee angle). In this context, significant increases in sEMG activity due to WBV were observed in the knee extensors and the gastrocnemii muscles, but this augmented sEMG activity was not different between knee angles (55-60° vs. 90°) (Di Giminiani et al. 2013; Roelants et al. 2006). In opposition, Abercromby et al. (Abercromby et al. 2007a) observed that the increases in sEMG due to WBV were the greatest for the smallest knee flexion angle (10°). The increases in sEMG activity decreased as the knee flexion angle increased from 10° to 35°. These discrepant results could be explained with the heterogeneous use of WBV devices, settings, and differences in the experimental procedure. However, it remains to assess how changing the knee flexion angle affects increases in sEMG activity of the lower limb muscles during the exposure to WBV compared to the same no-vibration trial.

Therefore, the goal of this study was to compare lower limb muscle activity between physically active and inactive individuals during WBV exercises. Further, the effect of knee flexion angle was analyzed with respect to the increases in muscle activity due to WBV. Additionally, transmissibility of the vertical acceleration to the head was quantified as head vibrations are associated with health-concerns (ISO-2631-1 1997). It was hypothesized that (1) active as well as inactive individuals would show significant increases in sEMG activity during WBV compared to no-vibration, but that (2) the inactive group would show higher increases in muscle activity than the active group. Finally, according to previous observations (Abercromby et al. 2007a), it was hypothesized that (3) the increases in sEMG activity due to WBV would be higher during relaxed standing (20°) than during squatting (60°).

Methods

Participants

The study included 58 adults aged 18-35 years who were divided into two subgroups of 30 active individuals (15 males, 15 females) and 28 inactive individuals (14 males, 14 females, **Table 10.1**). The inclusion criteria for the active group were five hours or more of physical activity per week and a fitness score of 8.5 or higher in the Baecke et al. questionnaire (Baecke et al. 1982). The inclusion criteria for the inactive group were two hours or less of physical activity per week and a fitness score of 7 or lower in the Baecke et al. questionnaire (Baecke et al. 1982). The exclusion criteria for both groups included a recent lower limb injury, lung conditions, heart problems, high blood pressure, epilepsy, osteoporosis, and diabetes. All participants were new to WBV and they provided informed consent prior to their participation. The study was approved by the University of Calgary's Conjoint Health Research Ethics Board and was conducted according to the ethical standards required by the International Journal of Sports Medicine (Harriss and Atkinson 2013).

Surface Electromyography

sEMG signals were measured of selected lower limb muscles during relaxed standing and squatting with and without WBV, and during maximal isometric MVCs. First, the skin was shaved, dead skin cells were removed using abrasive tape, and the area was cleaned with an isopropyl wipe. Then, bipolar sEMG electrodes (Ag-AgCl) with preamplifiers (Biovision, Wehrheim, Germany) were placed on the muscle bellies of the TA, GM, SOL, VL, VM, and BF of the right leg according to the SENIAM guidelines (Hermens et al. 2000). A reference electrode was positioned on the right tibial tuberosity. The electrodes had a 10 mm diameter and 22 mm inter electrode spacing, and were secured to the skin with Cover-Roll stretch tape (Beiersdorf AG, Hamburg, Germany). Voltage signals were pre-amplified (gain = 1000) at the source and sampled at 2400 Hz.

Assessment of MVCs

After completion of a standardized 5-min warm-up on a stationary bike, MVCs were completed for plantarflexion, dorsiflexion, knee extension, and knee flexion of the right lower limb on an isokinetic dynamometer (Biodex System 3, Medical Systems Inc., New York, NY, USA). Plantar- and dorsiflexion were performed at an ankle joint angle of 90° (neutral position) and the rotational axis of the dynamometer was visually aligned to the lateral malleolus. Knee extension and flexion was performed at a knee angle of 60° (0° corresponding to fully extended) and the rotational axis of the dynamometer was aligned to the lateral femoral condyle. sEMG of the TA was recorded during dorsiflexion, GM and SOL during plantarflexion, VL and VM during knee extension, and BF during knee flexion. All participants performed two 5-s MVC trials at maximal force and were given a 1-min break between trials. Peak MVC torques over 500 ms windows were calculated for both trials, and the higher torque was retained. The sEMG_{RMS} was computed during the 500 ms period of the selected torque output from the muscle corresponding to the contraction.

Acceleration Recordings

After completion of the MVC trials, a 1D accelerometer (ADXL 78, Analog Devices USA, encapsulated in a 20/12/5 mm plastic shell, measuring range ± 35 g, frequency response of 0-400 Hz, mass: 2 grams) measuring the vertical acceleration was positioned on the forehead. The accelerometer was secured to the skin with Cover-Roll stretch tape (Beiersdorf AG, Hamburg, Germany). Additionally, a 3D accelerometer (ADXL 78, Analog Devices USA, encapsulated in a 20/12/5 mm plastic shell, measuring range ± 70 g, frequency response of 0-400 Hz, mass: 2 grams) was placed on the WBV platform aligned with the third toe of the right foot (Lorenzen et al. 2009). The sensors were calibrated by aligning each axis of the accelerometer with gravity to measure the corresponding change in voltage induced by gravity (corresponding to 1 g). Acceleration signals were pre-amplified at the source and sampled at 2400 Hz. The acceleration delivered by the platform was identical between participants and knee flexion angles.

Table 10.1 Demographic, anthropometric and strength characteristics (torque/mass) of the active and inactive participants (mean \pm SD).

	Active (<i>N</i> = 30)	Inactive (<i>N</i> = 28)
Age (years)	25.9 \pm 4.3	26.9 \pm 3.8
Height (cm)	171.4 \pm 9.5	172.8 \pm 8.6
Mass (kg)	69.5 \pm 11.2	74.9 \pm 14.9
BMI (kg/m ²)	23.5 \pm 2.4	25.0 \pm 4.2
Fitness score	10.0 \pm 0.9	6.2 \pm 0.7*
Dorsiflexion (Nm/kg)	0.54 \pm 0.08	0.53 \pm 0.09
Plantarflexion (Nm/kg)	1.87 \pm 0.49	1.59 \pm 0.40*
Knee extension (Nm/kg)	2.52 \pm 0.57	2.17 \pm 0.49*
Knee flexion (Nm/kg)	1.33 \pm 0.32	1.09 \pm 0.29*

BMI, Body Mass Index; *Untrained < Trained ($P < 0.05$).

WBV exercises

WBV was delivered with a side-alternating platform (TBS 100A, Total Image Fitness, Calgary, Alberta, Canada) that generates vibration by rotating along the sagittal axis. The frequency setting of this platform ranges from 6 Hz to 16 Hz, and the amplitude (displacement from baseline to peak) is dependent on the width of the foot position. The widest foot position of 23.3 cm from the central axis of the platform corresponded to an amplitude of 4 mm, as evaluated by double integration of the platform acceleration signal. As responses in sEMG activity are positively correlated to the vertical platform acceleration (Marín et al. 2012a), the highest possible frequency-amplitude combination was chosen within the constraints of the WBV platform, which corresponded to 16 Hz and 4 mm. Although a vibration frequency of 16 Hz is rather low as compared to the ones used in previous studies (Abercromby et al. 2007a; Perchthaler et al. 2013; Pollock et al. 2010; Ritzmann et al. 2013), significant increases in sEMG were reported for the same platform type with a lower frequency-amplitude combination

(Cochrane et al. 2009). After the participants were familiarized with WBV, they performed two trials consisting of a 20-s static squat at a knee angle of 20° (relaxed standing) and 60° (squatting) (0° corresponding to full extension) with and without WBV. The assessment of these knee angles is of importance for inactive people as well as older and health compromised individuals since they might not be able to maintain deeper squat positions. In order to maintain the desired knee flexion angle, the participants were provided with a visual real-time feedback of their knee angle, measured with a digital goniometer (SG150 Twin Axis, Biometrics Ltd, Newport, UK), and illustrated on a laptop screen using MATLAB software (version 7.13 The Mathworks, Inc., Natick, MA, USA). If the measured knee angle varied more than $\pm 3^\circ$ from the desired knee angle at any point during the exercise, the trial was repeated. The participants stood barefoot on the WBV platform in order to avoid damping effects due to different footwear, and they were instructed not to hold onto the handle bar of the platform.

Data Analysis

The sEMG and acceleration signals were first clipped so that only 10 s remained for further analysis (time: 7-17s). Then, the signals were filtered using a band-pass wavelet filter with a low cut-off frequency of 5 Hz and a high cut-off frequency of 300 Hz. Examples of acceleration and sEMG signals are illustrated in **Figure 10.1**. For the sEMG signals, the excessive spikes observed in the sEMG spectrum at the vibration frequency and its multiple harmonics were removed via linear interpolation in the PSD. Then, $sEMG_{RMS}$ during WBV was calculated and divided by the maximum $sEMG_{RMS}$ obtained during the MVC trials and multiplied by 100 to express the $sEMG_{RMS}$ as a percentage of the maximum. For the acceleration signals, the average of the magnitudes of the local maxima and minima within each vibration cycle were computed. The average of the local maxima and minima was used to calculate the averaged peak amplitude (displacement from baseline to peak) of each acceleration signal. Then, the acceleration amplitude measured on the head was divided by the amplitude of the vertical platform acceleration ($50.5 \pm 5.7 \text{ m}\cdot\text{s}^{-2}$) to obtain transmissibility values (Pel et al. 2009).

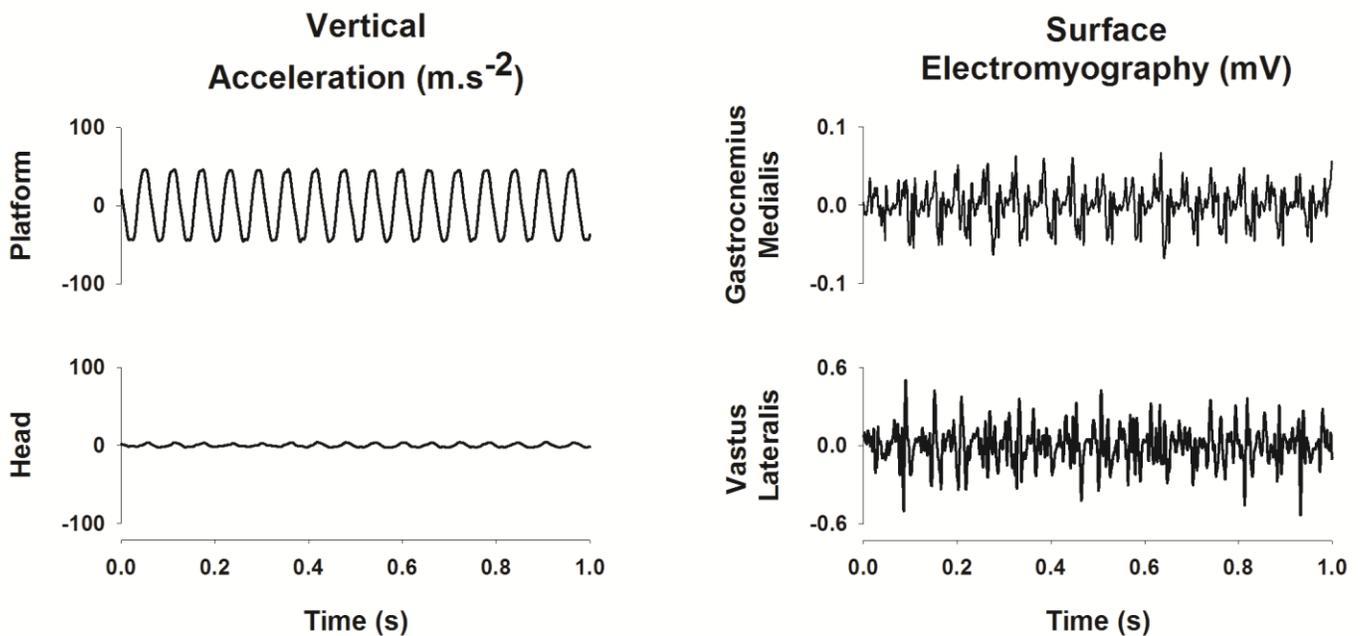


Figure 10.1 Acceleration (left panel) and raw surface electromyography (sEMG, right panel) signals of a representative participant during side-alternating whole-body vibration at 16 Hz and 4 mm amplitude.

Statistical Analysis

All statistical procedures were performed using IBM SPSS software (Version 20, Chicago, IL). The Kolmogorov-Smirnov test was applied to check the normality of the data. A Mann Whitney U test was used to compare the fitness scores between the two groups, and one-way ANOVA's were computed to compare all the other characteristics between them. One-way repeated measures ANOVA was used to compare sEMG_{RMS} recorded during each WBV condition to the sEMG_{RMS} during the corresponding no-vibration trial. One-way repeated measures ANOVAs [knee angle] with "group" as between subject factor were performed to analyze the sEMG_{RMS} during the no-vibration trial, the absolute increase in sEMG_{RMS} during WBV (WBV – no-vibration), and transmissibility to the head. The level of significance was $\alpha = 0.05$, and the data are presented as mean \pm SD.

Results

Subjects' characteristics

Age, height, body mass and BMI were similar between the two groups, whereas the fitness score was significantly higher in the active participants ($P < 0.05$). The maximal torque normalized to the body mass (Nm/kg) was significantly greater in the active than the inactive participants for plantarflexion ($P < 0.05$), knee extension ($P < 0.05$), and knee flexion ($P < 0.05$).

sEMG between groups

Comparing muscle activity between the groups during the no-vibration trial showed significant main effects, indicating that the no-vibration sEMG_{RMS} was significantly higher in the inactive than the active participants for the GM ($P = 0.015$), VL ($P < 0.001$), and VM ($P = 0.016$). Exposure to WBV significantly increased the sEMG_{RMS} in both groups (TA: Active: +234%, Inactive: +293%; GM: Active: +205%, Inactive: +128%; SOL: Active: +81%, Inactive: +49%; VL: Active: +87%, Inactive: +70%; VM: Active: +67%, Inactive: +66%; BF: Active: +87%, Inactive: +88%). **Figure 10.2** illustrates sEMG_{RMS} measured during the no-vibration trials and the significant increases in sEMG_{RMS} due to WBV for both knee angles, both groups, and all the measured muscles. However, these increases in sEMG_{RMS} due to WBV exposure were not significantly different between groups ($P > 0.05$) for any of the measured muscles.

sEMG between knee angles

Comparing the sEMG between the knee angles during the no-vibration trial showed significant main effects, indicating that the no-vibration sEMG_{RMS} was significantly higher at 20° than 60° knee flexion for the GM ($P < 0.001$) and SOL ($P = 0.001$). Conversely, no-vibration sEMG_{RMS} was significantly greater at 60° than at 20° knee flexion for the TA ($P < 0.001$), VL ($P < 0.001$), VM ($P < 0.001$) and BF ($P < 0.001$). The increases in sEMG_{RMS} due to WBV were significantly higher at 20° than at 60° for the GM (20°: +256%, 60°: +77%, $P = 0.005$), VL (20°: +150%, 60°: +7%, $P < 0.001$), VM (20°: +124%, 60°: +9%, $P < 0.001$) and BF (20°: +144%, 60°: +31%, $P = 0.004$).

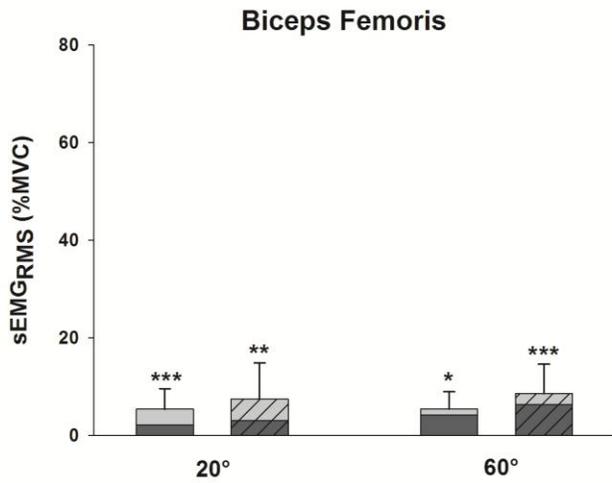
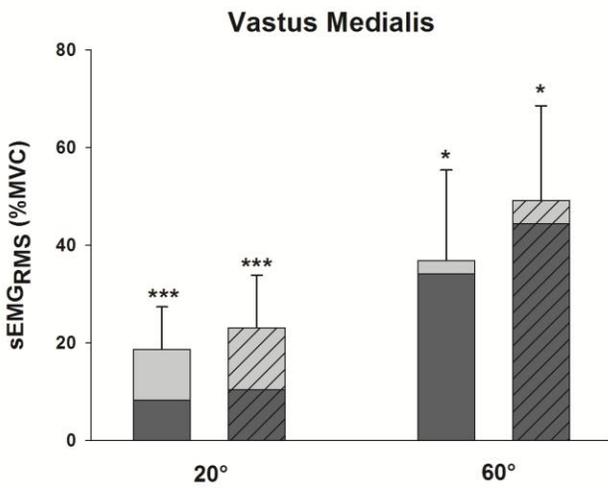
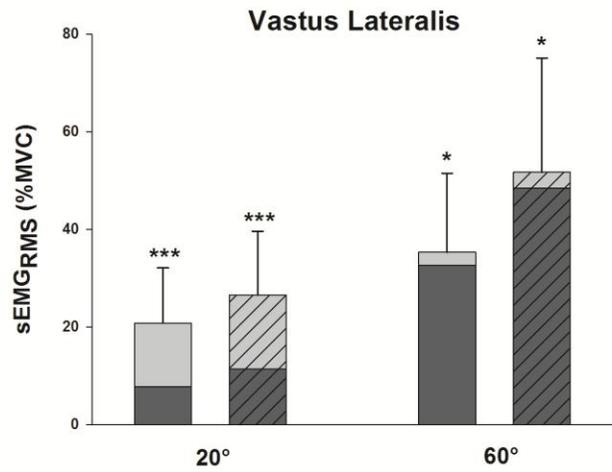
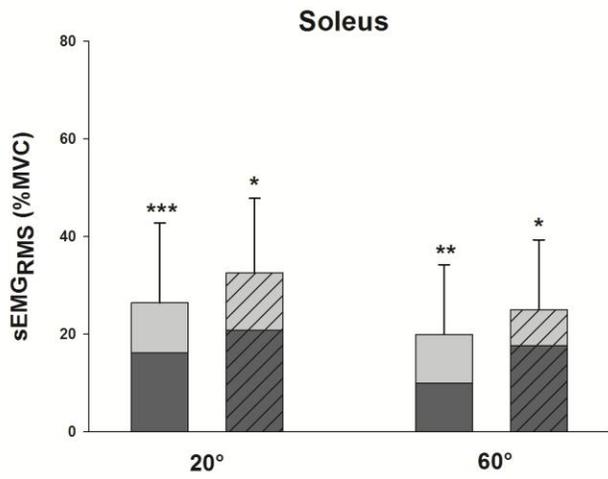
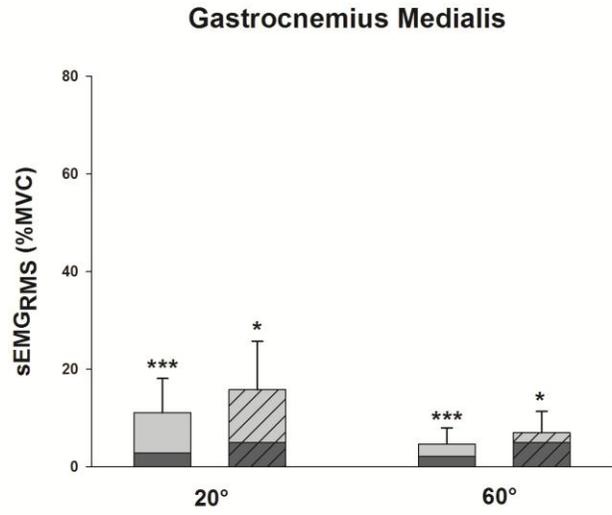
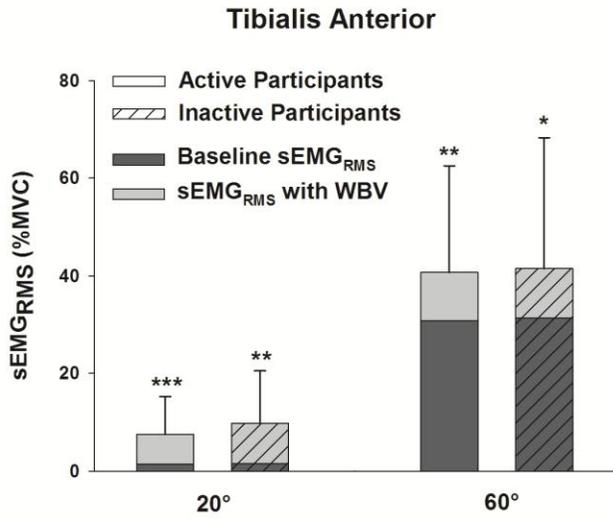


Figure 10.2 Root mean square of the surface electromyography (sEMG_{RMS}, mean \pm SD) during whole-body vibration (WBV) in the active and the inactive groups (dashed lines) during relaxed standing (20°) and squatting (60°) in several lower limb muscles. Significant increases in sEMG_{RMS} (light grey) as compared to no-vibration (dark grey) were found in all the measured muscles (* $P < 0.05$, ** $P < 0.01$, *** $P < 0.001$).

Head transmissibility

Comparing transmissibility to the head between groups and knee angles showed significant main effects for both parameters, with significantly higher values in the active than the inactive participants ($P = 0.033$), and significantly higher values at 20° than at 60° knee flexion ($P < 0.001$). Transmissibility to the head in the active participants was 0.11 ± 0.05 at 20° knee flexion and 0.05 ± 0.02 at 60° knee flexion. In the inactive participants, transmissibility to the head was 0.10 ± 0.05 at 20° knee flexion and 0.03 ± 0.01 at 60° knee flexion.

Discussion

The major findings of this study were that (1) side-alternating WBV at 16 Hz and 4 mm amplitude significantly increased muscle activity of all the measured lower limb muscles in active and inactive individuals during relaxed standing (20°) as well as during squatting (60°) compared to no vibration. Further, (2) baseline sEMG was significantly higher in the inactive than the active participants in the GM, VL, and VM, yet the increases in sEMG due to WBV were identical between the two groups. Comparing the knee flexion angles, it was found that (3) the increases in sEMG due to WBV were significantly greater during relaxed standing than during squatting in the GM, VL, VM, and BF. Last, (4) transmissibility of the vertical acceleration to the head was significantly higher in the active than the inactive participants, and was significantly higher during relaxed standing than during squatting.

Although the used platform frequency of 16 Hz was relatively low, the acceleration stimulus was strong enough ($50.5 \text{ m}\cdot\text{s}^{-2}$) to significantly enhance muscle activity in both groups, which confirms the first hypothesis. As a matter of fact, significant increases in muscle activity have been found for an even lower acceleration input of 6 Hz and 3.1 mm amplitude (Cochrane et al. 2009). However, baseline sEMG without WBV was greater in the inactive group than the active group in the VL, VM, and GM, which is attributed to their differences in lower limb muscle strength. The inactive group produced lower MVC torques during plantarflexion, knee extension and knee flexion than the active group. Consequently, the inactive group activated their lower limb muscles to a greater proportion of their maximum in order to maintain their standing and squatting postures. We initially assumed that WBV would be more beneficial to the inactive group due to their lower ability to fully activate their muscles. However, it cannot be neglected that during the WBV exposure, sEMG activity levels were lower than 60% of the MVC. Considering that the deficit of activation is mainly related to high threshold motor units, the WBV stimulus might not have been sufficient enough to activate them. Also, it has been previously shown that the sEMG activity during the no-vibration trials predicts the increases in sEMG activity due to WBV. Therefore, greater increases in sEMG activity would have been expected in the inactive group, at least for the VL, VM, and GM as they were activated to a greater extent than in the active group. Yet, the increases in muscle activity due to WBV were similar between the two groups, which rejects the second hypothesis. Within the limits of our experimental design, this suggests that the inactive group was not facilitated to recruit more motor units during the vibration exposure in comparison to the active group for the same acceleration stimulus. However for the first time we demonstrated that the effect of vibration on muscle activity is independent of the person's fitness level, and that inactive as well as active individuals can benefit from WBV exercises.

Another interesting result of the current study is that WBV training could be further of particular interest in inactive individuals due to their slight lower acceleration transmission to the head (active: 8.0%, inactive: 6.5%). As a result, inactive populations are possibly at lower risk to the potential side-effects of WBV exercises than their active counterparts. The higher

transmissibility in the active participants could be due to increased lower limb stiffness as an adaptation to regular exercising (Pearson and McMahon 2012). Also, it can be speculated that different soft tissue properties between the two groups was partly responsible for the higher transmissibility in the trained participants (Wakeling and Nigg 2001). However, these parameters were not controlled in the current study, and therefore no conclusions about the underlying mechanisms can be drawn.

As previously reported, it was found that increasing the knee flexion angle from relaxed standing (20°) to squatting (60°) increased muscle activity of the VL, VM, BF, and TA (Avelar et al. 2013; Di Giminiani et al. 2013; Perchthaler et al. 2013; Ritzmann et al. 2013), and decreased muscle activity of the SOL and GM (Ritzmann et al. 2013). However, these adaptations are obviously due to the muscles' anatomical functions to compensate for increased joint torques. According to the third hypothesis, for the first time we showed that the increases in sEMG levels due to WBV were higher during relaxed standing (on average: + 169%) than squatting (+31%) in the GM, VL, VM, and BF. This result is in line with a previous study showing that flexing the knees resulted in lower increases in sEMG activity of the VL, TA and GL due to WBV during dynamic squatting (Abercromby et al. 2007a). Further, although not statistically significant, the same tendency was observed for the VL, VM, GM and rectus femoris muscles during isometric squats performed at 55° and 90° of knee flexion (Roelants et al. 2006). While the standing position favoured the muscular adaptations to vibration, nor muscle length nor muscle pre-activation leading to an increased sensitivity of muscle spindles during direct muscle vibration (Burke et al. 1976a; 1976b) could account for our observations. However, according to the theoretical model of Cardinale and Bosco (Cardinale and Bosco 2003), it can be suggested that the greater sEMG increases in the standing position resulted from muscle stiffness regulation to dampen the WBV stimulus. This suggestion is further supported by the higher acceleration transmission to the head during relaxed standing compared to squatting.

Bending the knees resulted in lower acceleration transmission to the head, which is line with previous findings (Caryn et al. 2014; Rubin et al. 2003a; Tankisheva et al. 2013). WBV exercises

are associated with health concerns because high accelerations can result in back pain (Rubin et al. 2003) and altered visual perception (Ishitake et al. 1998; Pel et al. 2009). The International Organization for Standardization (ISO) has published guidelines (ISO-2631-1 1997) for occupational vibrations, and daily exposure to WBV has been found to overreach the limit determined by the ISO (Abercromby et al. 2007b; Muir et al. 2013; Tankisheva et al. 2013; . Yet it is questionable if those guidelines are entirely applicable to training or medical devices, as the guidelines focus on very high vibration frequencies (up to 80 Hz), induced over long periods of time (4 to 8 hours), and transmitted through the rare and not through the feet. Thus, although less than 11% of the platform acceleration was transmitted to the head, it is unknown if such acceleration magnitudes are detrimental (Caryn et al. 2014).

In summary, it was found that adjusting the knee flexion angle was a trade-off with respect to benefit and safety. The greatest benefits of WBV with regards to increases in muscle activity were found during standing. However, standing on the vibration platform lead to greater safety concerns than the squatting position, as transmissibility to the head was higher during this posture. Thus, both standing and squatting during WBV entail assets and drawbacks which need to be considered when designing a WBV training protocol.

The results of this study have to be interpreted with consideration to the limitations of the current method to assess the acceleration on the head. First, the acceleration signal was only measured in the vertical direction. Although the used platform type (side-alternating) delivers vibrations mainly in the vertical plane (Pel et al. 2009), tri-axial analysis might be needed to more accurately represent the complexities of vibration transmission in the human body (Cook et al. 2011). Second, skin-mounted accelerometers were used, which can allow movements of the soft tissue and skin relative to the bone beneath and may affect the acceleration detected by the accelerometers. Furthermore, it has been shown that there is only minimal overestimation (<10%) of accelerations with the use of skin-mounted accelerometers compared to bone pins (Kim et al. 1993; Nokes et al. 1984).

Conclusions

WBV training is an interesting training method for inactive populations because they are affected by WBV in equal measures as their active counterparts and are at lower risk to potential side-effects of vibration. Individuals who are not able to adapt a squatting position still benefit from substantial increases in muscle activity in a relaxed standing position during WBV exercises. However, caution must be taken for individuals with head or neck sensitivity as transmissibility to the head is higher in this position.

CHAPTER 11

DISCUSSION

11.1 AIMS

The goal of this thesis was to evaluate (I) the optimal method to process sEMG signals during WBV exercises and (II) the influence of the normalization method on sEMG activity. In parallel, it was aimed to (III) gain insight whether the isolated spikes observed in the sEMG spectrum contain motion artifacts and/or reflex activity. Accordingly, the goal was (IV) to explore the effect of WBV exercises on lower limb muscle activity using a vertical synchronous platform and different vibration frequencies, amplitudes, and additional loads. Further, (V) the effect of horizontal and vertical vibrations was assessed on lower limb muscle activity using a platform inducing horizontal vibrations via a circular mode, vertical vibrations via a side-alternating mode, and vibrations in all three planes via the combination of the two modes. The last goal of this thesis was (VI) to compare lower limb muscle activity and acceleration transmission to the head between two knee flexion angles and between populations with different fitness levels using a side-alternating platform.

11.2 SYNOPSIS OF RESULTS

The presented studies addressing these goals orientate from a methodological approach to physiologically applied investigations. The studies build upon each other, as the methodological studies (I&II) as well as the explorative study (III) give the implication on how to process and analyze sEMG signals during WBV exercises. This knowledge then subsequently served as a methodological basis in the applied studies (IV-VI). In summary, study I showed that spectral linear interpolation centred at the vibration frequency and its multiple harmonics should be used to delete the sEMG spikes during WBV exercises as this method provided the closest estimation of the true muscle activity. The band-pass filter was in low agreement with the true muscle activity as this filter removes a wide frequency range. The use of a band-stop filter provided a feasible estimation of the sEMG activity, though it should be considered that this filter can introduce a substantial bias in the quantification of the sEMG activity. In study II, it was found that the sEMG during WBV exercises can be normalized to the Mmax, to the sEMG during the MVC, and to the baseline sEMG. Considering that MVC assessments are relatively unproblematic in comparison to the assessment of Mmax responses, normalization to the MVC

appears to be the most interesting option as it indicates the muscle's relative activation during WBV exposure. In study III, it was found that the isolated spikes in the sEMG spectrum are contaminated by motion artifacts as shown with the help of electrical recordings of the patella during the vibration. At the same time, it was shown that these spikes contain reflex activity, most likely pertaining to stretch reflex responses. Thus, from the first three studies it can be concluded that the spikes observed in the sEMG spectrum are contaminated by motion artifacts, and should therefore be deleted by using spectral linear interpolation. Subsequently, the sEMG should be normalized to the sEMG during the MVC. This exact methodology was applied in the following studies (IV-VI), although different platforms types were used.

The results of study IV showed that a high frequency-amplitude combination provoked the greatest increases in lower limb muscle activity. At the same time it was found that the displacement of the platform has a greater effect on lower limb muscle activity than the frequency as only the displacement was identified as a significant predictor of the sEMG. Further, adding an external load equal to 50% of the body mass enhanced muscle activity of the calf muscles even at a low frequency and amplitude. In study V it was found that the vertical platform acceleration, but not the horizontal acceleration, was positively correlated with the sEMG of lower limb muscles during WBV exercises. More specifically, a minimum vertical acceleration of $18 \text{ m}\cdot\text{s}^{-2}$ was required to significantly increase the sEMG activity of the lower limb muscles. Last, study VI demonstrated for the first time that inactive individuals showed similar increases in sEMG activity during WBV as their active counterparts. Yet, the inactive group showed greater sEMG activity in order to maintain the posture during the exercise. Another finding of study VI was that transmissibility of the vertical acceleration to the head was lower in the inactive than the active group, and that bending the knees from relaxed standing to a half-squat position substantially reduced the transmissibility to the head. Last, it was found in study VI that the sEMG activity was more enhanced due to WBV during relaxed standing than during squatting.

11.3 HOW TO FILTER THE sEMG DURING WBV – AN ONGOING DEBATE

While there is evidence that WBV elicits TVR-like responses (Pollock et al. 2010; Ritzmann et al. 2010; Sebik et al. 2013; Zaidell et al. 2013), the co-existence of motion artifacts is still under debate. Several studies including the present thesis have been dedicated to motion artifact occurrence during WBV exercises (Fratini et al. 2009; Ritzmann et al. 2010; Sebik et al. 2013), yet the methodological limitations (Bifulco et al. 2013) as well as the contradictory findings impede a general conclusion. As a consequence, the content of the isolated spikes in the sEMG spectrum is unknown, which makes it difficult to agree upon a filtering regime. Even more so it is surprising that study III of this thesis was the first study that has scrutinized these isolated spikes. The observation that reflex activity and motion artifacts are present in the spikes seems to further promote the ongoing debate. However, despite the co-existence of reflex activity and motion artifacts, the spikes were withdrawn in this thesis for the following reasons. First, as the motion artifacts are sinusoidal waves, excessive peaks are only expected at the vibration frequency and at a few multiple harmonics. Stretch reflex responses, as seen in the current document as well as in previous studies (Ritzmann et al. 2010), are variable in amplitude and periodicity. Thus, stretch reflex responses display a power spectrum over a wide frequency range and not only at the vibration frequency and its multiple harmonics. Therefore, withdrawal of the spikes leads to deletion of the motion artifacts but allows for the retention of the majority of the stretch reflex-related information. The resulting filtered sEMG spectrum is no longer contaminated by artifacts while maintaining the necessary information about the reflex activity. Second, as seen in study I, deleting the spikes can result in an underestimation of the actual muscle activity, while keeping the spikes can result in an overestimation. In the case that significant increases in muscle activity were found during WBV without deleting the spikes, one cannot assure that these increases would still be significant without the contribution of the motion artifacts, particularly when compared to the artifact-free no-vibration condition. On the contrary, when significant increases were found during WBV after deleting the excessive spikes, one can assume that the actual increases in muscle activity might be even higher. For these reasons it is suggested to delete the excessive spikes during WBV exercises until further insight is gained.

In order to remove the motion artifacts, spectral linear interpolation was applied. This newly developed method withdraws only the excessive spikes in the sEMG spectrum without adding notches to the signal. Notches in the sEMG spectrum have no physiological meaning, which makes the spectral linear interpolation method a more fitting method from a physiological point of view than the band-stop filter. However, besides the fact that the band-stop filter introduces a systematic bias, it was rendered impossible to show the improvement of the linear interpolation method over the band-stop filter in study I. It is likely that the power of the notch was not large enough relative to the power of the entire signal to affect statistical differences between the two methods. Further, the information associated with the reflex responses is not only contained in the spikes but in the entire frequency spectrum (Ritzmann et al. 2010). Therefore, both methods retained most of the reflex activity-related information which could further explain why no differences between the two methods were observed.

11.4 VIBRATION INPUT – THE OPTIMAL PLATFORM TYPE

Strong evidence was found in study V that the vertical and not the horizontal platform accelerations positively affect the sEMG activity of the lower limb muscles. This observation might be due to the following mechanisms. First, the TVR might only be provoked during vertical vibrations. Stretch reflex responses are induced by activation of muscle spindles due to changes in length in the muscle-tendon units (Cochrane et al. 2009). Due to the anatomical function of the muscles that were measured in study V, stretch reflex responses were mainly expected during a stretch in the vertical direction, i.e. during vertical accelerations. The second mechanism that could explain this finding is the damping mechanism. An increase in muscle activity as a response to vibration has been labeled as a mechanism to damp the oscillations in order to protect the soft tissue from damage (Wakeling et al. 2002). It is possible that this mechanism occurs primarily during vertical than horizontal vibrations. However, it is important to mention that no attempts have been made nor in the current thesis nor in the current literature to ascertain the occurrence of these mechanisms during horizontal vibrations.

Although it was found in the present thesis that horizontal vibrations have a lower effect on muscle activity than vertical vibrations, a previous study (Bagheri et al. 2012) found no difference in jump performance immediately after the exposure to WBV between horizontal and vertical oscillations. However, platform types that induce vibrations in the horizontal plane are unsuitable when increases in sEMG levels are targeted. The dual mode platform type that induces vibrations in all three planes is therefore not essential, as the platform types that oscillate only in the vertical direction are sufficient. The two platform types that induce vibrations mainly in the vertical plane are the side-alternating and the vertical synchronous WBV platforms. Both of these platform types were used to deliver vibrations in the current thesis, yet without direct comparison. Previous studies have found that the side-alternating type induces greater sEMG activity of the knee extensors, knee flexors, and plantar flexors (Abercromby et al. 2007a; Ritzmann et al. 2013) than the vertical synchronous type. Due to their different mode of operation, it is possible that they do not completely target the same muscles. Also, there is a lack of studies comparing these platform types controlling the induced accelerations and ascertaining the differences between them. However, besides enhancing muscle activity to a higher extent, side-alternating vibrations transmit less accelerations to the head (Abercromby et al. 2007b). This may be attributed to greater vibration damping by rotation of the pelvis. Thus, side-alternating WBV induces greater sEMG activity and lower acceleration transmission to the head than vertical synchronous WBV, which makes side-alternating platforms an ideal device. However, both platform types were found to significantly increase muscle activity of the lower limbs in the present thesis.

11.5 ACCELERATION THRESHOLD – HOW MUCH IS ENOUGH?

Considering that high accelerations can be detrimental for the human body (ISO 2631-1 1997), determination of the minimal required acceleration amplitude to stimulate muscle activity of the lower limbs is a novel yet integral process. In study V, such an acceleration threshold was successfully established. First, the lower body sEMG_{RMS} (mean of all muscles) was correlated with the platform acceleration. As only the vertical direction showed high and significant correlations ($r = 0.970$, $P < 0.001$) with the lower body sEMG_{RMS}, the threshold was evaluated using the vertical platform acceleration. In the next step, the lowest acceleration amplitude was determined by locating the point where significant increases in sEMG activity were observed in most of the measured muscles. Using side-alternating WBV, the lowest required vertical acceleration was $18 \text{ m}\cdot\text{s}^{-2}$ (**Figure 11.1**). This acceleration amplitude resulted in lower body sEMG activity of 22.4% of the MVC, corresponding to a relative increase of +19% as compared to the corresponding no-vibration trial. As the acceleration threshold may vary between platform types, the same analysis was conducted for the outcomes with the vertical synchronous platform, even though this was not presented in study IV. The correlation coefficient between the lower body sEMG_{RMS} and the vertical platform acceleration was lower than the one found for the side-alternating type, yet relatively high and significant ($r = 0.699$, $P < 0.01$). Without the use of an additional load, only the highest frequency-amplitude combination led to significant increases in most muscles (i.e. three out of five muscles). This condition induced vertical vibrations of $59 \text{ m}\cdot\text{s}^{-2}$, which resulted in lower body sEMG activity of 16% of the MVC. This corresponded to a relative increase of +13% as compared to the corresponding no-vibration trial.

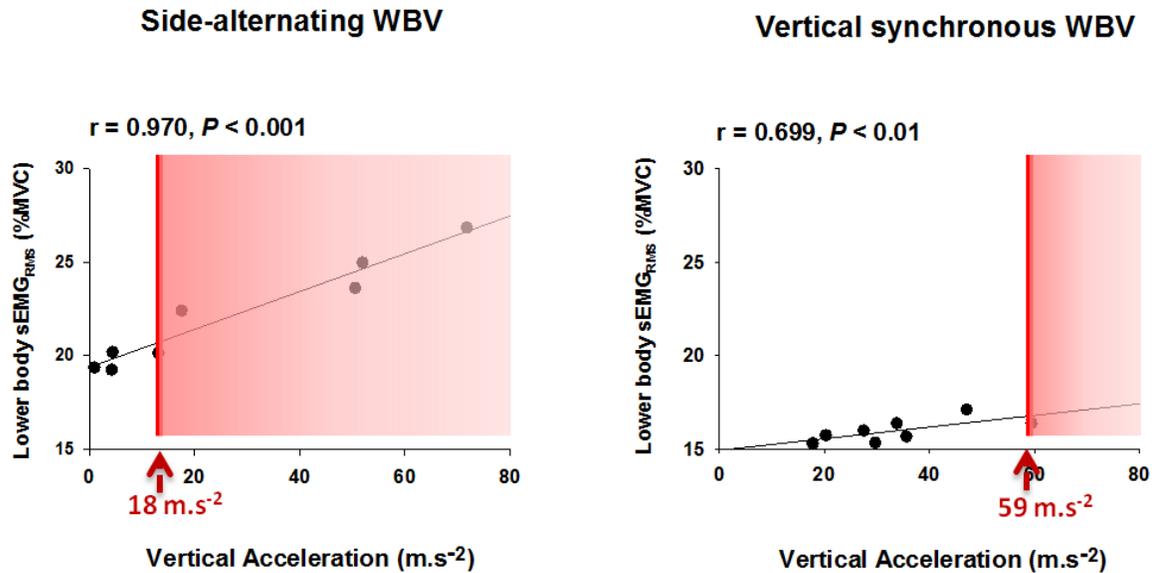


Figure 11.1 Scatterplots of the root mean square of the lower body surface electromyography (sEMG_{RMS}) and the vertical platform acceleration during side-alternating and vertical synchronous whole-body vibration (WBV). Each data point represents the sEMG_{RMS} averaged for all muscles and participants for a given WBV trial. The vertical acceleration threshold that is required to significantly increase muscle activity is lower for side-alternating WBV (18 m.s⁻²) than vertical synchronous WBV (59 m.s⁻²).

Considering that side-alternating WBV elicits greater sEMG responses than vertical synchronous WBV (Abercromby et al. 2007a; Ritzmann et al. 2013) for a given acceleration load, it makes sense that a higher vertical acceleration threshold was found for the latter platform type. As this is the first time that such a threshold has been determined, it is difficult to draw comparison with previous studies. However, side-alternating accelerations of 4.4 m.s⁻² (effectively measured: 5.9 m.s⁻²), which is below the established threshold of 18 m.s⁻², induced significant increases in sEMG activity in only two out of the four measured muscles (Cochrane et al. 2009). Conversely, vertical accelerations of 71 m.s⁻² (calculated using the frequency and amplitude), which is substantially higher than the threshold established here, enhanced sEMG activity of all the measured muscles (Abercromby et al. 2007a; Ritzmann et al. 2013). The acceleration

threshold that we established for vertical synchronous WBV seems to be in agreement with the literature as well. Using vertical accelerations above the established threshold of 59 m.s^{-2} , significant increases in sEMG activity were found in all the measured muscles with accelerations (calculated using the frequency and amplitude) of 71 m.s^{-2} (Ritzmann et al. 2013), 121 m.s^{-2} (Roelants et al. 2006), 126 m.s^{-2} (Krol et al. 2011), 142 m.s^{-2} (Marín et al. 2009), and 177 m.s^{-2} (Cardinale and Lim 2003). In a study (Hazell et al. 2007) where a wide range of frequencies (25 – 45 Hz) and two different amplitudes (2 and 4 mm) were tested, significant increases in sEMG activity were only found once the induced accelerations widely exceeded the threshold of 59 m.s^{-2} (96.7 m.s^{-2} for VL, 126.3 m.s^{-2} for BF). In this study, the motion artifacts were deleted by rejecting the entire frequency content below 100 Hz with the help of a band-pass filter. Hence, this intensive filtering regime could account for having evaluated a higher acceleration threshold. However, a study using a band-stop filter to delete the spikes (Avelar et al. 2013) found no enhancement of VL activity with a calculated platform acceleration of 142 m.s^{-2} . On the contrary, two studies found increased sEMG activity in all the measured muscles using lower accelerations than the established threshold, i.e. with vertical accelerations of 44 m.s^{-2} (Marín et al. 2012a) and 12.3 m.s^{-2} (Di Giminiani et al. 2013). The low acceleration threshold found in the latter study might be present because the spikes in the sEMG spectrum were not deleted. In short, the acceleration thresholds that were established in the current thesis for side-alternating and vertical synchronous WBV are in line with the literature. However, future studies should ascertain the acceleration thresholds that were found in the current thesis. Also, it should be re-investigated how the side-alternating and the synchronous vertical platforms affect lower limb muscle activity, using the same effective (and not calculated) vertical acceleration load. Furthermore, acceleration transmission and the associated risks need to be resumed and compared between the two platform types.

While determination of an acceleration threshold is extremely useful, it entails some limitations. First, when resonance occurs, increased muscle activity is expected to reduce resonance by changing the natural frequency or the damping characteristics of the tissue (Wakeling et al. 2002). Although we showed that the vertical platform acceleration determined by the

frequency and amplitude is linearly related to the lower body sEMG, this relation might deviate for resonance frequencies. Second, the acceleration thresholds were determined for the averaged lower body sEMG for simplicity reasons. It is likely that the acceleration threshold is different for certain muscle groups, especially considering that the vibrations are damped throughout the body. Third, it needs to be considered that the thresholds were determined in a young, healthy, physically active study population. Lower thresholds might be sufficient in older or health-compromised individuals while higher thresholds might be needed in professional athletes. Last, it was found in the present thesis that parameters such as the use of additional loading or changes in knee flexion can enhance the effect of vibration, and therefore change the required acceleration threshold.

11.6 OPTIMAL PARAMETERS – HOW SHOULD WE VIBRATE?

Besides the platform type, other parameters can be manipulated in order to enhance the effect of the vibration. Study IV and V both showed that muscle activity increases with an increasing vibration frequency and amplitude, resulting in the greatest enhancement using the highest frequency-amplitude combination. This finding is in agreement with previous studies that found an effect of the frequency (Di Giminiani et al. 2013; Hazell et al. 2007; Krol et al. 2011; Marín et al. 2012a; Perchthaler et al. 2013; Pollock et al. 2010; Ritzmann et al. 2013) and the amplitude (Hazell et al. 2007; Krol et al. 2011; Perchthaler et al. 2013; Pollock et al. 2010). However, this result is not surprising considering that the TVR, whether evoked via directly applied vibration (Eklund and Hagbarth 1966) or via WBV (Zaidell et al. 2013), is positively correlated to the vibration frequency and/or amplitude.

The use of additional loads is another method to amplify the effect of the vibration on muscle activity levels, as seen in study IV. This enhancement is seemingly originating from the TVR as well, since pre-activation of the muscles increases the sensitivity of the spindles with concurrent increases in neuromuscular responses to vibration (Bedingham and Tatton 1984; Burke et al. 1976a; Eklund and Hagbarth 1966). This observation has been recently confirmed during WBV exercises (Zaidell et al. 2013), especially for the highest frequency tested (50 Hz). The finding

that additional loading increases the effect of WBV has important implications; using additional loads decreases the required acceleration threshold to effectively stimulate the sEMG activity of the lower limbs. This can be seen in study IV, as using an additional load corresponding to 50% of the body mass lowered the acceleration threshold from 59 m.s^{-2} to 47 m.s^{-2} in the thigh muscles and from 59 m.s^{-2} to 20 m.s^{-2} in the calf muscles. Considering that high accelerations can have adverse effects on the human body (Pel et al. 2009; Rubin et al. 2003), lowering the acceleration threshold by means of additional loading is a particular interesting strategy.

The last parameter that was investigated in the present thesis (study VI) was the body position, more specifically the knee flexion angle. During relaxed standing as well as during a squatting position, significant increases in sEMG activity due to WBV were observed for all the measured muscles. Surprisingly, it was found that the sEMG responses to WBV were higher during relaxed standing than during squatting. As previously described, the magnitude of the TVR is positively related to the length of the muscle (Burke et al. 1976b; Nordin and Hagbarth 1996). The quadriceps muscles and the plantar flexors are longer during squatting than during standing, hence it would be expected that the increases in sEMG activity are higher during squatting. Furthermore, the TVR is enhanced the higher the pre-activation of the muscles (Burke et al. 1976a; Eklund and Hagbarth 1966). Pre-activation was higher for the thigh muscles and the TA during the squatting position, yet higher sEMG increases were observed during standing. Although unexpected, these results are in line with the outcomes of a previous study (Abercromby et al. 2007a) - the sole study that has addressed this issue. Although the underlying mechanisms remain dubious, this finding indicates that individuals who are not able to perform squatting exercises still benefit from WBV increases by standing on the platform in a relaxed position.

Taken as a whole, the highest increases in muscle activity during WBV exercises can be expected by using the highest possible frequency-amplitude combination, by using an additional load consisting of the highest weight possible, and by engaging into a relaxed standing position.

11.7 SAFETY ASPECT – HOW MUCH IS TOO MUCH?

Taking the above mentioned measures to induce the highest possible sEMG increases might not be the best decision from a safety point of view. Although no acceleration limit has been determined for WBV exercises, using the highest frequency-amplitude combination might be hazardous. Therefore, it is advisable to use frequencies and amplitudes approximately resulting in the minimal required acceleration load leading to sEMG increases (18 m.s^{-2} for side-alternating and 59 m.s^{-2} for vertical synchronous WBV with the devices used in the present thesis). However, according to the existing guidelines for human exposure to vibration (ISO 2631-1 1997), platform accelerations of 18 m.s^{-2} and 59 m.s^{-2} in combination with vibration frequencies of 16 Hz and 40 Hz, respectively, are defined as high risk accelerations with recommendations for less than one minute of exposure per day. But again, it is questionable if those guidelines are entirely applicable to WBV devices, as the guidelines focus on occupational vibrations induced through the bottom over long periods. Additionally, bending the knees from relaxed standing to squatting has been shown to substantially decrease head vibrations. Therefore, although higher increases in sEMG activity were observed during relaxed standing, a squatting position might be preferable. Last, caution must be taken with populations presenting high muscular-tendon stiffness such as professional athletes or older individuals, since it can be speculated from the present thesis that stiffness facilitates acceleration transmission to the head.

11.8 POPULATIONS – WHO CAN BENEFIT?

Study VI of this thesis was the first investigation to compare the effect of WBV on muscle activity levels and acceleration transmission to the head between populations presenting different fitness levels. Interestingly, the active subjects transmitted more of the vibrations to the head than the inactive subjects. This observation could be because the active participants relied on greater lower limb stiffness, and a better ability to stabilize their ankle and knee joints through a greater proprioceptive acuity. It has been reported that regular exercising results in increased lower limb stiffness through passive components such as the tendon or the connective tissue (Pearson and McMahon 2012). Further, there is recent evidence that

competitive athletes exhibit superior proprioceptive acuity than non-competitive athletes (Courtney et al. 2013; Muaidi et al. 2009). As the active group had higher fitness scores and produced higher torques for plantarflexion, knee extension and knee flexion than the inactive group, it can be speculated that the active participants had higher musculotendinous and musculoarticular stiffness (Rabita et al. 2008), as well as enhanced proprioception resulting both in greater transmission of the acceleration. Although the inactive participants showed higher muscle activity levels relatively to their maximum to maintain the posture, it can be speculated that they stabilized their ankle and knee joints less and, similar to a spring suspension system (Nikooyan and Zadpoor 2011), damped the vibrations more. Finally, different soft tissue properties between the two groups could have been partly responsible for the higher transmissibility in the active participants. Although it was not assessed, it can be speculated that the inactive participants had more subcutaneous fat than the active participants. As subcutaneous fat is known to attenuate vibrations (Wakeling and Nigg 2001), this could be an additional reason why the damping effect was greater in the inactive participants. Nevertheless, it is important to emphasize that no attempts have been made to ascertain these mechanisms. It can be concluded that inactive individuals are not at higher risk of WBV-related health-concerns as the accelerations measured on the head were lower than the ones measured in the active group.

Another interesting result pertains to the effect of WBV on muscle activity between the two groups. It was assumed that inactive individuals would show higher increases in muscle activity than their active counterparts as they might have greater potential to recruit high-threshold motor units during WBV. This assumption was based on the lack of ability in inactive subjects to fully activate their high-threshold motor units during maximal contractions (Amiridis et al. 1996). However, as the sEMG activity in study VI was lower than 60% of the MVC, the WBV stimulus might not have been sufficient enough to activate them. Regardless, higher increases in sEMG activity due to the vibration could have been expected in the inactive group. Based on the pre-activation theory of the TVR, the inactive group should have shown higher increases in sEMG during WBV as they pre-activated their VL, VM, and GM to a greater extent than the

active group. Yet, similar increases in muscle activity were observed between the two groups. This indicates, against our expectations, that the inactive group recruited less motor units in those muscles due to the vibration compared to the active group. This suggests that active individuals would be more affected by WBV exercises than their inactive counterparts. Evidently, it is difficult to ascertain the effect of vibration between populations presenting different fitness levels. Future research should focus on this, keeping in mind that inactive individuals might require higher sEMG levels in certain muscles to maintain the posture.

For the first time it was shown that active as well as inactive individuals benefit from WBV exercises by enduring increased muscle activity levels during the exposure to the vibration. Additionally, it can be assumed that WBV does not impose a higher health threat on inactive individuals. These results should be expanded in future studies to other populations such as the elderly, health-compromised individuals, and professional athletes.

11.9 METHODOLOGICAL CONSIDERATIONS AND LIMITATIONS

The methods employed in the present studies present some limitations. In study III, motion artifacts were measured by placing electrodes on the patella with insulating tape in between. As the insulating tape elevated the resistance between the electrodes to an almost infinite value, recordings of motion artifacts were more likely. In order to address this limitation, the electrodes should have been connected with a resistor as utilized in previous studies (Ritzmann et al. 2010; Sebik et al. 2013). However, even with the use of a resistor, the set-up presents some draw backs. The use of insulating tape impedes recordings of electrochemical potentials at the electrode/gel interface, at the gel/skin interface, and between skin layers (Bifulco et al. 2013). Although the cables are another source for motion artifact occurrence (Clancy et al. 2002; Türker 1993; Webster 1984), it needs to be acknowledged that this set-up was not ideal to record motion artifacts during WBV exercises. As a consequence, the outcomes need to be interpreted with caution.

Another limitation pertains to study IV. For each participant, additional loads of the same weight (17 and 33 kg) were used. As the body mass and strength abilities varied between participants, the used weights imposed different levels of stress. In order to account for the morphological differences between subjects, additional weights corresponding to a proportion of the body mass, the maximal isometric voluntary contraction torque, or the one-repetition maximum (1RM) would have been more suitable, as accomplished in previous studies (Hazell et al. 2010; Ritzmann et al. 2013).

In study V, it was found that the acceleration amplitude and not the frequency and the amplitude per se is the main load parameter during WBV exercises. For the side-alternating mode, the tested frequency-amplitude combinations consisted of a low-low (6 Hz, 2.5 mm) and a high-high (16 Hz, 4 mm) condition. Having included all the possible combinations, i.e. low-low (6 Hz, 2.5 mm), low-high (6 Hz, 4 mm), high-low (16 Hz, 2.5 mm) and high-high (16 Hz, 4 mm) would have allowed a more confident interpretation of the results. However, previous studies have already shown that the same amount of acceleration made up by different frequency-amplitude combinations led to the same outcomes (Chen et al. 2014; Marín et al. 2012b; Siu et al. 2010). Additionally, the assessment of all the conditions was unfeasible as the test protocol was under time constraint.

The last relevant limitation affects study VI. First, active and inactive subjects were tested for comparison reasons. While it was difficult to find inactive individuals willing to participate in an exercise research study, the assessment of their fitness levels could have been more refined. Although strength levels were measured and physical activity habits were assessed using a questionnaire, sport-type specific questions should have been included. Second, only the vertical acceleration on the head was measured, although tri-axial analysis more accurately represents the complexities of vibration transmission in the human body (Cook et al. 2011). Retrospective, it would have been interesting to measure the subjects' body composition using DEXA scan and/or to measure the amount of subcutaneous fat using skinfold calipers. The

knowledge of these parameters could have been helpful to interpret the differences in acceleration transmission between the two study populations.

11.10 PRACTICAL APPLICATIONS

The outcomes of the present thesis allow a number of practical applications that can be useful for trainers and researchers when designing a WBV protocol. The first point concerns the choice of the WBV platform. WBV platforms inducing oscillations mainly in the horizontal plane should be avoided, since it has been found that horizontal vibrations have little effect on lower limb muscle activity. Instead, WBV platforms that induce vibrations in the vertical plane should be used. While both the side-alternating and the vertical synchronous type enhance lower limb muscle activity, side-alternating platforms are the better choice as they induce higher sEMG responses (Abercromby et al. 2007a; Ritzmann et al. 2013) while limiting accelerations to the head (Abercromby et al. 2007b). The second point pertains to the magnitude of the induced acceleration load. Using a side-alternating platform, the minimal induced vertical platform acceleration should be $18 \text{ m}\cdot\text{s}^{-2}$; using a vertical synchronous platform, the minimal vertical platform acceleration should be $59 \text{ m}\cdot\text{s}^{-2}$. While these acceleration magnitudes can be calculated with the help of the frequency and amplitude, it is recommended to measure the effective platform acceleration. As seen in the present thesis, the frequencies and amplitudes communicated by the manufacturer can be different from the actual values. Therefore, it is very likely that some studies have reported a wrong platform displacement, which is devastating for the interpretation and application of their outcomes. However, to increase the stimulus, the acceleration load should be progressively increased by augmenting the frequency and the amplitude. Alternatively or simultaneously, an additional load can be applied to reinforce the effect of vibration on the muscle activity. This load can also be progressively increased to further enhance the effect of vibration. Relaxed standing on the vibration platform leads to higher increases in sEMG activity than a squatting position, but a squatting position is recommended in case it is suspected that the induced head accelerations impose a health threat. The last recommendations concern the analysis of the sEMG signals recorded during WBV exercises. The excessive spikes observed in the sEMG spectrum should be deleted using

spectral linear interpolation. Subsequently, the sEMG measured during the WBV exercise is preferably normalized to the sEMG measured during MVCs.

In summary, the following recommendations can be made from the current thesis to achieve the highest increases in sEMG activity during WBV exercises.

- Use a side-alternating WBV platform or alternatively a vertical synchronous platform.
- Start with an acceleration threshold of 18 m.s^{-2} for side-alternating WBV and 59 m.s^{-2} for vertical synchronous WBV.
- Always measure the platform acceleration, as the theoretical platform acceleration can deviate from the actual values.
- Progressively increase the stimulus by increasing the acceleration load and by using additional loads
- Adapt a relaxed standing position to gain the highest increases in muscle activity. Conversely, adapt a squatting position when the induced head accelerations could impose a health threat.
- For sEMG recordings, delete the excessive spikes visible in the sEMG spectrum using spectral linear interpolation. Subsequently, normalize the sEMG to the sEMG during MVCs.

CHAPTER 12

OUTLOOK & CONCLUSIONS

12.1 OUTLOOK

The current thesis has outlined several issues concerning muscle activity during WBV exercises, yet new questions have emerged that need to be addressed in future studies. The concept of a minimal required platform acceleration (i.e. acceleration threshold) to enhance lower limb muscle activity during WBV exercises has been introduced in this thesis. However, acceleration thresholds were developed for one specific population, performing a certain exercise on a particular WBV platform. More specifically, acceleration thresholds were successfully developed for young, healthy, physically active individuals, who performed static semi-squats on a synchronous vertical or side-alternating WBV platform. Hence, such thresholds need to be expanded for the elderly, health-compromised individuals, as well as for professional athletes. Also, different body positions need to be considered such as relaxed standing or dynamic squatting. Furthermore, the benefits of horizontal vibrations need to be re-addressed. It is possible that vibrations in the horizontal plane affect the sEMG activity of muscle groups that induce movements not only in the saggital plane such as the invertors and evertors of the ankle. Such muscle groups were neglected in the present thesis, which might be accountable for the lacking effect of horizontal vibrations. In addition, further research is required to explore the effect of WBV exercises between different populations, such as between young and older subjects and healthy and health-compromised individuals. On this note, safety regulations are lacking for a safe use of WBV platforms. Such regulations need to be compiled separately for different populations, as it was shown in the present thesis that acceleration transmission can vary between them.

Although the occurrence of motion artifacts during WBV exercises has been addressed in this thesis, it is still unknown whether sEMG recordings are effectively contaminated by motion artifacts. This nescience is linked to the boundaries of recording motion artifacts. Although the limitations of using dummy electrodes are well described, there seems to be no obvious solution on how to address these issues. Another approach, rather simple but original, would be to place dummy electrodes without layers of tape on a piece of synthetic material similar in composition and density to a human subject. As reflex activity can be excluded in this case, any

electrical recordings during the vibration would be related to motion artifacts. The uncertainty about the occurrence of motion artifacts during WBV exercises makes it challenging to settle on a sEMG processing method, which again makes it difficult to compare outcomes between studies. Therefore, future studies should investigate the occurrence of motion artifacts in order to subsequently develop a standardized sEMG processing method.

Another subject that should be explored in the future pertains to the reflex activity during WBV exercises. It is extremely challenging to distinguish the stretch reflex responses from the entire myoelectrical signal. As a consequence, little is known about the properties of stretch reflex responses induced by WBV, such as their amplitude or periodicity. A possible approach to learn more about these reflexes has been accomplished in a preliminary study of the I3S laboratory. sEMG signals from the SOL were measured during WBV at 30 Hz together with the vertical platform acceleration. With the help of the platform acceleration signal, each vibration cycle in the sEMG signal could be determined. Then, all vibration cycles that were recorded during the 4 s of WBV were segmented into one cycle. This was accomplished for the acceleration as well as for the sEMG signal, as illustrated in **Figure 12.1**. The acceleration signal was strictly periodical, as the recordings from each cycle were entirely superimposed (**Figure 12.1A**). On the contrary, the sEMG signal was not strictly periodical, as the recordings were not fully superimposed (**Figure 12.1C**). Nevertheless, computing the average of all the sEMG recordings displayed a signal that resembled a stretch reflex response (**Figure 12.1D**). In the next step, the superimposed sEMG signals were synchronized to match their delays (**Figure 12.1E**), which displayed a relatively high variability of the amplitude of the stretch reflex responses. This method proved to be extremely useful as the stretch reflex responses induced by WBV could not only be extracted, but their amplitudes and periodicity could be studied.

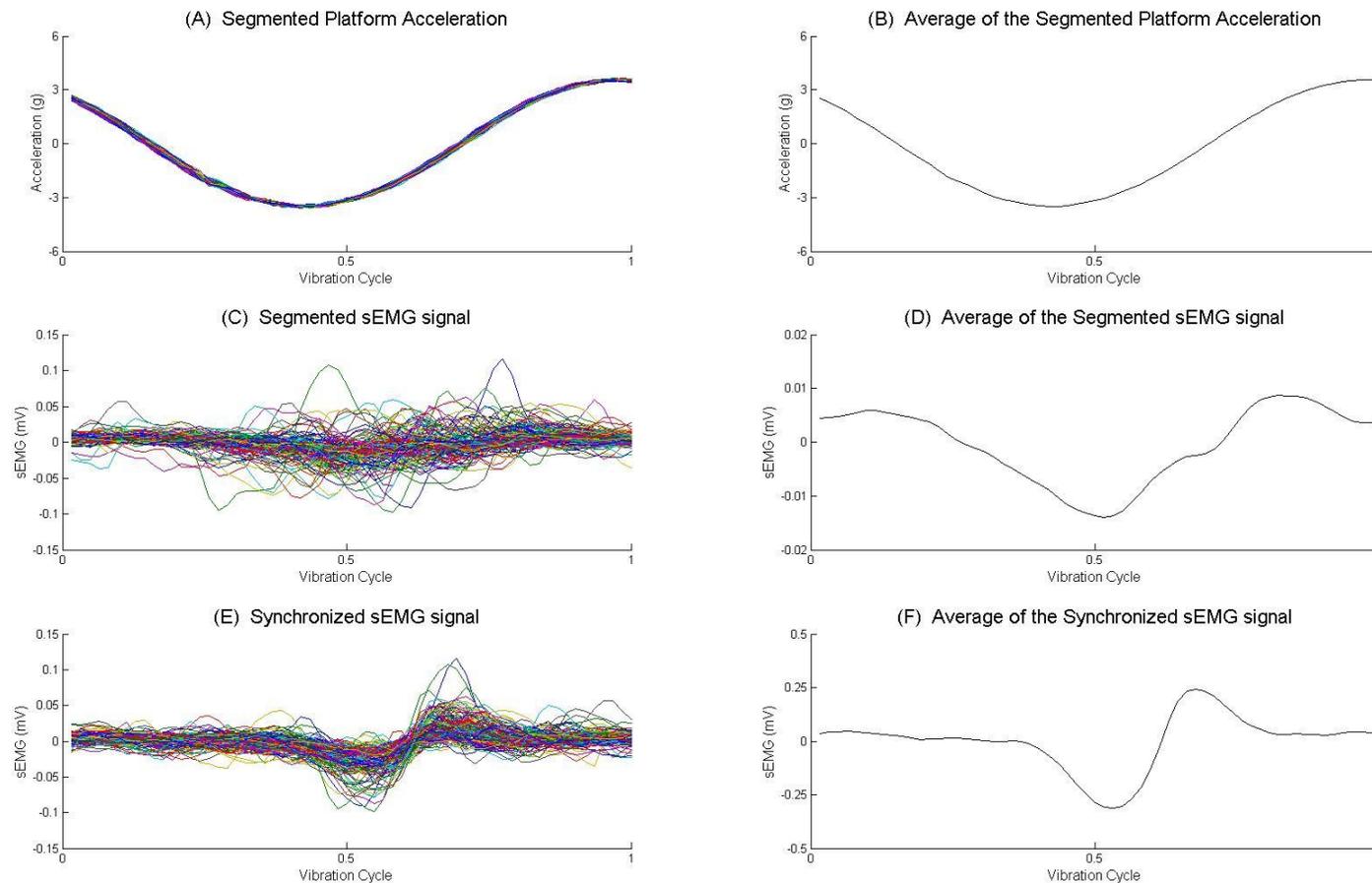


Figure 12.1 Platform acceleration and sEMG signal of the Soleus recorded during whole-body vibration at 30 Hz. The recordings were superimposed for each vibration cycle; the vibration cycles were determined using the platform acceleration signal. The platform acceleration signal is strictly periodical (A), unlike the sEMG recordings (B). Nevertheless, the sEMG recordings reveal stretch reflex responses (D, F), and their variation in periodicity and amplitude can be studied in (C) and (E), respectively.

12.2 CONCLUSIONS

The present thesis has addressed methodological issues that are associated with sEMG recordings during WBV exercises. For the first time it has been shown that the isolated spikes in the sEMG spectrum contain motion artifacts as well as reflex activity. Despite this co-existence, it is recommended to withdraw the excessive spikes in the sEMG spectrum. For this purpose, a new processing method was developed based on spectral linear interpolation which outperformed the commonly used band-stop and band-pass filter. This new method, which was described in detail in this thesis, can serve in future research to filter sEMG signals recorded during WBV exercises. Besides the methodological recommendations, practical applications have been made which can be particularly helpful for WBV research and training protocols with the target of increasing lower limb muscle activity. Using a side-alternating WBV platform, vertical accelerations of a minimum of 18 m.s^{-2} should be induced, whereas with a vertical synchronous WBV platform, vertical accelerations of at least 59 m.s^{-2} are recommended. WBV platforms that vibrate only in the horizontal plane are advised against as horizontal accelerations have little effect on lower limb muscle activity. In order to increase the vibration stimulus, the acceleration load can be progressively increased together with the use of additional loading. The WBV user can stand relaxed on the platform to gain the highest improvement, or can adopt a squatting position to reduce head accelerations. Finally, an important finding of this thesis was that active and inactive individuals can benefit from WBV exercises in equal measures, but that active individuals transmit higher head accelerations. While this thesis provides valuable suggestions for WBV training protocols in young healthy adults, future research is essential to develop WBV training protocols for prevention and rehabilitation purposes.

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