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The Influence of Externally Applied Vibration and Compression on Muscular Performance and Recovery

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The Influence of Externally Applied Vibration and Compression on Muscular Performance and Recovery

by

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Abstract

The impact between heel and ground during walking and running generates a shock wave that travels through the body and induces vibrations to the skeleton and soft tissues. Such vibrations may also be generated by devices such as vibrating tools or whole-body vibration platforms. The implications of vibrations on soft tissues have been studied for many decades, and the effects of vibrations range from induced damage to nerves, muscles and vascular tissue to positive effects such as reduced lower back pain, improved muscular performance and increased bone mechanical properties. The effects of soft tissue vibrations on muscle performance during naturally occurring and artificially induced vibrations are, however, poorly understood. This is indicated by the ongoing debate about the role of soft tissue vibrations during running and the missing guidelines for safe and effective whole-body vibration training. This thesis, therefore, aimed to quantify vibration characteristics during running and whole-body vibration training. In addition, the effects of vibrations on maximal elbow extension torque and the effect of vibration damping by means of compression apparel and the resulting effects on running economy were tested. The results showed that (a) vibrations become more intense and potentially longer during prolonged running, (b) elbow extension torque was reduced during vibration exposure, (c) the local, muscular, oxidative metabolic demand increased with increasing vibration frequency on a vibration platform, (d) the accelerations on soft tissues during vibration training exceeded levels related to tissue injury for short periods of time, and (e) compression apparel did not improve running speeds at the aerobic or anaerobic thresholds and did not improve blood lactate recovery times. Vibrations were shown to affect muscle metabolism and motor performance, while compression apparel may not contribute to running performance. However, small changes in vibration application site, vibration amplitude and frequency, and changes in body position have a large impact on the observed effects and future research is needed to exploit the great potential of safe vibrations and vibration treatment.

Preface

Chapters three to seven are based on the manuscripts listed below:

1. Friesenbichler, B., Stirling, L.M., Federolf, P., Nigg, B.M., 2011. Tissue vibration in prolonged running. Journal of Biomechanics 44, 116-120.

2. Friesenbichler, B., Groves, E.M., Nigg, B.M., 2013. Influence of graduated full-body compression apparel on indicators of running performance and recovery. Journal of Science and Medicine in Sport. (Temporary submission withhold).

3. Friesenbichler, B., Nigg, B.M., 2013. Vibration transmission to lower extremity soft tissues during whole-body vibration. Journal of Science and Medicine in Sport. (Submitted)

4. Friesenbichler, B., Coza, A., Dunn, J.F., Nigg, B.M., 2013. Local metabolic rate during whole body vibration. Journal of Applied Physiology 114, 1421-1425.

5. Friesenbichler, B., Coza, A., Nigg, B.M., 2012. Reduced elbow extension torque during vibrations. Journal of Biomechanics 45, 2203-2207.

Part of this work has been presented at the following conferences:

1. Friesenbichler, B., Stirling, L.M., Federolf, P., Nigg, B.M., 2010. The influence of fatigue on tissue vibration in prolonged running. Oral presentation at the 20th Conference of the Canadian Society of Biomechanics in Kingston, Ontario.

2. Friesenbichler, B., Coza, A., Nigg, B.M., 2011. Elbow extensor torque as a function of vibration frequency. Poster presentation at the 35th Conference of the American Society of Biomechanics in Long Beach, California.

3. Friesenbichler, B., Nigg, B. M., Dunn, J., 2012. Local muscle metabolism during whole body vibration. Poster presentation at the 17th annual Congress of the European College of Sport Science in Bruges, Belgium.

4. Friesenbichler, B., Nigg, B. M., 2013. Vibration Transmission to Lower Extremity Soft Tissues during Whole-Body Vibration. Oral presentation at the 31st International Society of Biomechanics in Sports conference, Taipei, Taiwan.

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Alles Wissen und alles Vermehren unseres Wissens endet nicht mit einem Schlusspunkt, sondern mit einem Fragezeichen

All knowledge and all accumulation of our knowledge does not end with a full stop, but with a question mark

Hermann Hesse

Chapter one: Introduction

The viability of any living organism is strongly dependent on the surrounding environmental conditions. A change in the environment can lead to failure or limited function of certain physiological processes within the organism. Humans and other animals are exposed to changes in their environment and usually either adapt to the changing environment by evolutionary processes (long-term) or by actively altering their direct environment, for example by building a habitat (short-term). Some environmental influences may be very subtle, and the resulting adaptations may therefore be subtle as well but not less important. For example, the impact induced upon ground contact of the heel and the ground during walking or running is an environmental stimulus that humans and other animals have been exposed to for millions of years and they adapted to it. It has been suggested that joint stresses due to walking and running were accounted for in humans by increased articular surfaces, enlargement of the iliac pillar or an enlarged cross-sectional area of the calcaneal tuber compared to australopithecines (Bramble & Lieberman, 2004).

A shock wave upon impact between the heel and the ground travels along the skeletal system upwards through the body and is transmitted to the hard as well as the soft body tissues. The impacts can be seen as a mechanical input to the body characterized by frequency and amplitude. The frequency of this input into the body may sometimes have a frequency spectrum with peaks close to potentially harmful resonance frequencies of soft tissue compartments. It may therefore be speculated, that next to the skeletal system, adaptations in response to impacts in walking and running may occur in soft tissue compartments when the input frequency and natural frequency of the soft tissue compartment are close (resonance). It was speculated for horses that the muscles actively change their mechanical characteristics (e.g., stiffness and damping) in order to avoid resonance (Wilson et al., 2001) and in humans through the *muscle tuning* concept (Nigg et al., 1995; Nigg, 1997; Nigg and Wakeling, 2001).

It was proposed that muscle tuning may affect comfort, performance, and fatigue, although the exact mechanisms related to those effects have not yet been elaborated. It may be speculated that excessive vibrations of soft tissues may induce strains, which, over time, could be perceived as uncomfortable. Performance may be affected because muscle activation is needed to perform the tuning task. Fatigue may be induced earlier, due to the additional expenditure of energy by the muscle tuning process over longer distances.

Although the muscle tuning concept was tested in numerous studies, many unanswered questions remain. If muscle tuning would in fact be related to muscle fatigue, one may ask whether a fatigued muscle also affects the functionality of muscle tuning.

One part of this thesis elaborates on this question and tries to identify the changes in muscle vibrations during prolonged running.

If muscles are actively used to reduce soft tissue vibrations, then an intervention that supports vibration damping may help to reduce the energy consumption of a muscle during exercise. It was shown that compression apparel reduced soft tissue vibrations upon landing from a jump (Doan et al., 2003). In addition, it was shown that externally applied compression improved blood flow in the lower (Lawrence and Kakkar, 1980) and upper (Bochmann et al., 2005) extremities. Based on those findings, it can be speculated that compression apparel may

(a) Reduce soft tissue vibrations during running
(b) Reduce muscle activity
(c) Improve running performance and recovery.

However, the “optimal” levels of compression needed to reduce tissue vibrations and improve blood flow during high intensity exercise are unknown.

One approach to circumvent this limitation is to test multiple levels of compression on the same participants, which is an approach taken in a part of this thesis. The main focus was laid on the third proposed benefit of compression apparel, running performance and recovery.

Vibrations do not only occur during exercise but also in occupational settings where workers operate vibrating machinery. The effects of vibrations on the body were studied in those situations since back pain (Fishbein and Salter, 1950) and vibration induced white finger (a digital vasospastic disorder; Bovenzi, 1998) were related to prolonged vibration exposure.

With the introduction of whole-body vibration platforms as a training device, some studies tested the transmission of vibrations from the platform to the skeleton to establish guidelines for safe usage (Pel et al., 2009; Kiiski et al., 2008; Muir et al., 2013). Although negative effects of vibrations on soft tissues are known (Takeuchi et al., 1986; Necking et al., 1992, Necking et al., 1996; Govinaraju et al., 2006), the vibration transmission to soft tissues has only been reported once on one specific (synchronous) vibration platform type (Cook et al., 2011). In order to assess potential adverse effects on soft tissues when using whole-body vibration training as a long term intervention, the transmissibility needs to be studied in greater detail.

Therefore, one part of this thesis quantified the transmissibility of vibrations to two lower extremity soft tissue compartments, in order to provide essential data for future risk assessments.

Beside safety of a new training regimen, the efficacy of it is probably the most important feature.

Numerous studies have tried to determine at which frequency, amplitude and body position combinations the highest gains in muscle performance can be expected (Ritzmann et al., 2013; Rittweger, 2010; Abercromby et al., 2007). Surface electromyography (EMG) is usually the means by which the immediate effects of a certain amplitude, frequency, and body position combination are quantified. The vibrations on a platform induce significant peaks in the raw EMG signal and one can either treat them as motion induced artifact (noise) and remove them by sophisticated filters (Abercromby et al., 2007) or one can treat them as a neuromuscular response due to the vibration stimulus and not remove them (Ritzmann et al., 2010). Depending on the used (or unused) filter, the results are strongly influenced by this procedure. As an alternative, the activity of a muscle may be quantified by means of near-infrared spectroscopy (NIRS), which noninvasively allows to quantify the aerobic metabolic demand of a muscle. To quantify the metabolic demand of the muscle when using NIRS, the blood flow to the muscle of interest needs to be occluded (Praagman et al., 2003). Only two studies used the occlusion maneuver in conjunction with whole-body vibration training and NIRS so far (Zange et al., 2009; Coza et al., 2011). However, one study did not control for the quality of occlusion, which affects the measurements by the NIRS device (Zange et al., 2009), and the other study tested the effects of vibrations during calf rises (Coza et al., 2011), which is a superposition of voluntary contractions (calf raise) and involuntary contractions (vibration). An isolated measurement of oxidative muscle metabolism during whole-body vibration training has not yet been studied.

One project in this thesis therefore quantified the oxidative metabolic demand at different vibration frequencies while standing on a vibration platform.

Vibrations have not only been studied in terms of applied sciences, the basic effects of vibrations on the neuromuscular system have fascinated researchers for a long time. It is known that vibrations can induce non-voluntary contractions due to the imposed rapid stretches of the muscles or tendons that trigger muscle spindle or Golgi tendon organ reactions. This interaction between vibrations and involuntary neuromuscular activity was named *tonic vibration reflex* (TVR) and was described in detail in 1966 (Eklund and Hagbarth, 1966). Most studies on this topic were conducted by means of direct stimulation by placing small vibratory units on an individual tendon or a muscle belly and it was shown that the TVR increases with increasing vibration frequency (Eklund and Hagbarth, 1966; Gottlieb and Agarwal., 1979) and increasing initial lengths of the receptor bearing muscle (Burke et al., 1976). The potential of vibrations to change neuromuscular activity seemed promising to elicit improvements in muscular performance. A very basic question would be whether or not vibration could increase the maximal voluntary force production. This question was already addressed in part by Eklund and Hagbarth themselves, and no improvements in maximum voluntary finger flexion force were related to vibration stimulation in healthy subjects (Eklund and Hagbarth, 1966). Only a few recent studies have attempted to study the effects on maximal force production during vibration exposure but one showed an increase in maximal elbow flexion strength (+6.4 %) when applying vibrations by means of a belt and in a direction normal to the orientation of the biceps brachii muscle (Kin-Isler et al., 2006). However, vibrations applied in the direction parallel to the muscle’s contraction direction (longitudinal) have not yet been tested, although this scenario may be the most relevant in recreational activities and occupational settings. A custom made, pneumatic vibration device was built within our laboratory which is capable of applying vibrations in a longitudinal manner while simultaneously collecting force data during a maximal voluntary contraction. One study in this thesis therefore tested the effects of longitudinal vibrations on maximal voluntary isometric elbow extension torque at different vibration frequencies and elbow angles.

In summary, this thesis attempts to address the following questions:

(a) Does muscular fatigue affect the vibration characteristics of soft tissues during prolonged running?

This question was addressed in chapter three. Intensity and timing of the triceps surae muscle compartment vibrations were quantified throughout a run to fatigue and comparisons between the beginning and the end of the run were conducted.

(b) Does compression apparel positively affect running performance and recovery?

This question was addressed in chapter four. Four different levels of full-body compression apparel were tested during a run at incremental speeds on a treadmill. Blood lactate and heart rate were quantified throughout the run and during the recovery phase.

(c) To which extent are vibrations transmitted to lower extremity soft tissues during whole-body vibration training?

This question was addressed in chapter five. Vibration frequency and amplitude was quantified at the quadriceps femoris and triceps surae muscle compartments during whole-body vibrations at three different platform frequencies. The tissue vibrations were compared to the platform vibrations to calculate vibration transmissibility.

(d) What is the oxidative metabolic demand of individual lower extremity muscles during whole-body vibration training at different vibration frequencies?

This question was addressed in chapter six. Local oxygen demand of the vastus lateralis and gastrocnemius medialis were quantified during vibration training at three different vibration frequencies.

(e) Can longitudinally applied vibrations increase maximal elbow extension torque?

This question was addressed in chapter seven. A custom-made pneumatic vibrator was designed to apply vibrations to the wrist in order apply longitudinal vibrations to the triceps brachii muscle. Maximal torque and muscle activity of agonistic and antagonistic muscles were quantified at different vibration frequencies and elbow extension angles.

Chapter two: Literature review

2.1 Vibrations

2.1.1 Origins of vibratory stimulation

Rubbing and anointing were said to have energizing effects already in about 1000 B. C., as mentioned in Homer's "Odyssey". Massage was used as a luxury, to speed up recovery from sickness and to increase agility and endurance by the ancient Greeks and Romans. It was also used to treat pain after gymnastic exercise and was recognized as therapeutic agent. It probably was also the Greeks that introduced the use of vibratory stimuli as part of the massages. To generate a vibratory stimulus, one end of a saw was wrapped in cotton fabric and was applied to the part to be treated. The other part was left uncovered and sawed a piece of wood so that mechanical vibrations were transmitted to the treated part. Later, in the sixteenth century, the use of percussion, vibration and passive motion was described in a Japanese book ("Sau-Tsai-Tou-Hoei") to treat rigid muscles, spasmodic contractions or for the relief of rheumatic pains (Snow, 1912).

2.1.2 Definition of vibrations and vibration damping

Vibrations of mechanical systems are highly common phenomena in nature. Sinusoidal vibrations occur due to restoring forces that are proportional to the displacement from an equilibrium position. This dynamic balance between the forces acting on the system can be described by Newton's Law as shown in Equation 2.1.1.

$$\boldsymbol{m\ddot{x} = -sx} \qquad \text{(Equation 2.1.1)}$$

where m is the mass of the oscillator, $\ddot{x}$ is the acceleration, s is the restoring force per unit distance (e.g., spring constant), and x is the displacement. The displacement x from equilibrium in a simple harmonic motion can be defined by the basic Equation 2.1.2.

$$\boldsymbol{x = A \sin(\omega t + \varphi)} \qquad \text{(Equation 2.1.2)}$$

where *x* is the displacement from equilibrium, *A* is the initial amplitude of the motion, *ω* is the angular frequency in rad/s, *t* is the time in seconds and φ is the phase angle.

The period *T* (in seconds), at which the system will oscillate once can be described by Equation 2.1.3.

$$\boldsymbol{T = \frac{2\pi}{\omega}} \qquad \text{(Equation 2.1.3)}$$

A simple harmonic motion is shown in Figure 2.1.1.

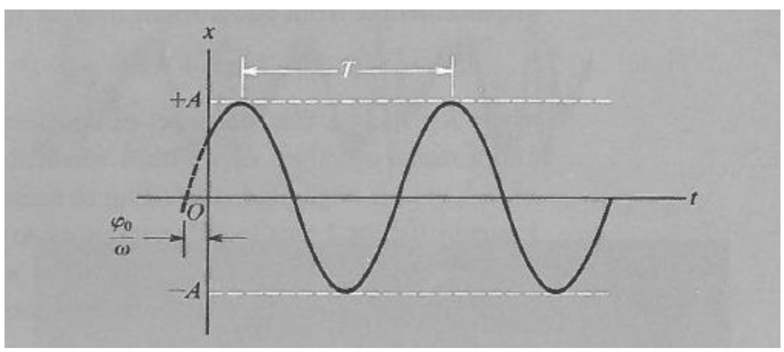

Figure 2.1.1: Simple harmonic oscillation of period *T* and amplitude *A* (French, 1971).

Assuming a simple spring system, the angular frequency *ω* can be described by Equation 2.1.4.

$$\boldsymbol{\omega = \sqrt{\frac{s}{m}}} \qquad \text{(Equation 2.1.4)}$$

where *s* is the spring constant (i.e., stiffness of the spring) and *m* is the mass attached to the spring.

Typically, the free vibrations of any real physical system will dissipate with time due to loss of mechanical energy induced by some form of friction that the vibrating system is experiencing. This reduction of vibration amplitude over time is called damping of vibrations. This resistive force is proportional to the velocity of vibration and it acts in the direction opposite to that of the velocity, which allows to use Newton's Second law, which is shown in Equation 2.1.5.

$$\boldsymbol{m\ddot{x} = -sx - r\dot{x}} \qquad \text{(Equation 2.1.5)}$$

where m is the mass of the oscillator, $\ddot{x}$ is the acceleration, s is the restoring force per unit distance (e.g., spring constant), x is the displacement and r is the constant of proportionality as force per unit of velocity. This system can be illustrated by Figure 2.1.2.

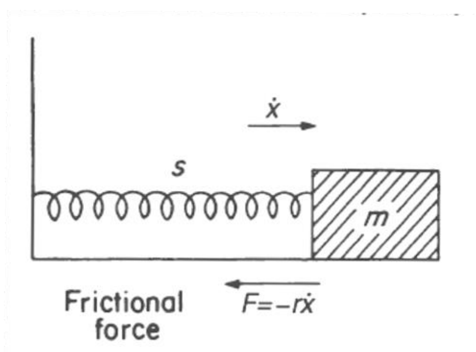

Figure 2.1.2: Simple harmonic system with a damping or frictional force r$\dot{x}$ acting against the direction of motion (Pain, 1999).

In essence, the force by the mass of the oscillator is equal to the force induced by the spring plus the force by friction (or damping), but in opposite direction (towards the equilibrium point). The displacement of a damped simple harmonic motion can be described by Equation 2.1.6.

$$\boldsymbol{x = Ae^{-\gamma t/2}\cos(\omega t + \varphi)} \qquad \text{(Equation 2.1.6)}$$

where x is the displacement from equilibrium, A is the amplitude of the motion, γ is the damping coefficient, t is the time in seconds, ω is the angular frequency in rad/s, and φ is the phase angle.

The angular frequency of a damped oscillator is slightly less than the angular frequency of the freely oscillating system and can be described by Equation 2.1.7.

$$\boldsymbol{\omega^2 = \omega_0^2 - \frac{\gamma^2}{4}} \qquad \text{(Equation 2.1.7)}$$

where ω is the angular frequency of the damped system, ω_0 is the angular frequency of undamped oscillations and γ is the damping coefficient.

The magnitude of the damping coefficient γ will define how the damped oscillator will behave upon excitation. The type of oscillator is defined by:

$$\boldsymbol{\frac{\gamma^2}{4} < \omega_0^2} \qquad \textbf{Underdamped oscillator}$$

$$\boldsymbol{\frac{\gamma^2}{4} = \omega_0^2} \qquad \textbf{Critically damped oscillator}$$

$$\boldsymbol{\frac{\gamma^2}{4} > \omega_0^2} \qquad \textbf{Overdamped oscillator}$$

The three types of a damped oscillator can be illustrated as shown in Figure 2.1.3.

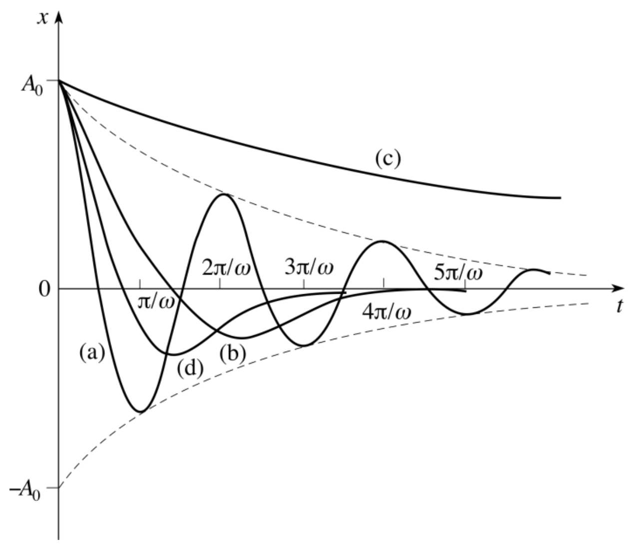

Figure 2.1.3: The effect of (a) underdamping; (b) overdamping; (c) heavily overdamping; and (d) critically damping on the movement of a one-dimensional oscillator (FLAP Open University, University of Reading).

In an underdamped oscillator, the mass will overshoot the zero point and oscillate about the equilibrium point. Critical damping is the quickest approach to zero amplitude for a damped oscillator. Although an underdamped system reaches zero amplitude even quicker, the critically damped system will remain at the zero position earlier than the underdamped system. An overdamped oscillator may or may not cross the equilibrium position, however, the time needed to reach a stable equilibrium position is higher than that of the critically damped oscillator.

The described scenarios are important concepts to consider if one is interested in oscillations of soft tissue packages that are initiated to oscillate once but are not further disturbed anymore by external forces. This may be the case during walking and running, where the heel-strike represents a single inpact that initates the tissue to vibrate once.

The three types of oscillators are characterized by two parameters, ω_0 and γ. The smaller the damping coefficient γ becomes the better the quality of an oscillatory system. This relation can be described by the quality factor Q, which is shown in equation 2.1.8.

$$Q = \frac{\omega_0}{\gamma} \qquad \text{(Equation 2.1.8)}$$

where Q is the quality factor, ω_0 is the angular frequency of the undamped oscillator and γ is the damping coefficient.

However, in certain situations, there is a periodic external driving force that appears in such a short distance that the system does not reach a stable equilibrium before the next stimulus arrives. This situation is referred to as a forced vibration which can lead to the well known "resonance" phenomena. If the driving frequency is close to the natural frequency of the oscillator, then the amplitude of the oscillation can increase to very large amounts using only small forces (French, 1971).

The mechanical equation of motion that describes a forced oscillator applied to a damped mechanical circuit is shown in Equation 2.1.9.

$$m\ddot{x} + r\dot{x} + sx = F_0 cos\omega t \qquad \text{(Equation 2.1.9)}$$

where F_0 is the amplitude of the force. This system can be illustrated by Figure 2.1.4.

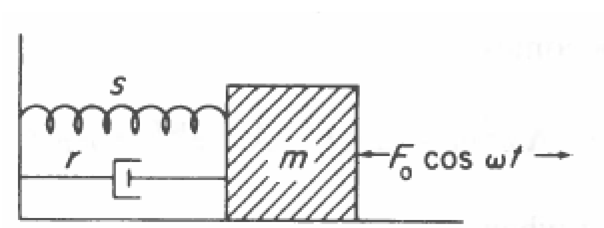

Figure 2.1.4: Mechanical forced oscillator with force $F_0cos\omega t$ applied to a damped oscillator (Pain 1999).

The amplitude at which a forced mechanical oscillator is moving depends on the angular frequency of the undamped system (ω_0), the damping coefficient γ and the angular frequency of the driving force (ω). If ω_0 (also known as natural or fundamental frequency), and ω are close to each other, then we can observe the "resonance" phenomena, which can lead to high amplitude oscillations of the system, which is illustrated in Figure 2.1.5 for different qualities of oscillators (the lower the quality, the higher the damping coefficient, see Equation 2.1.8).

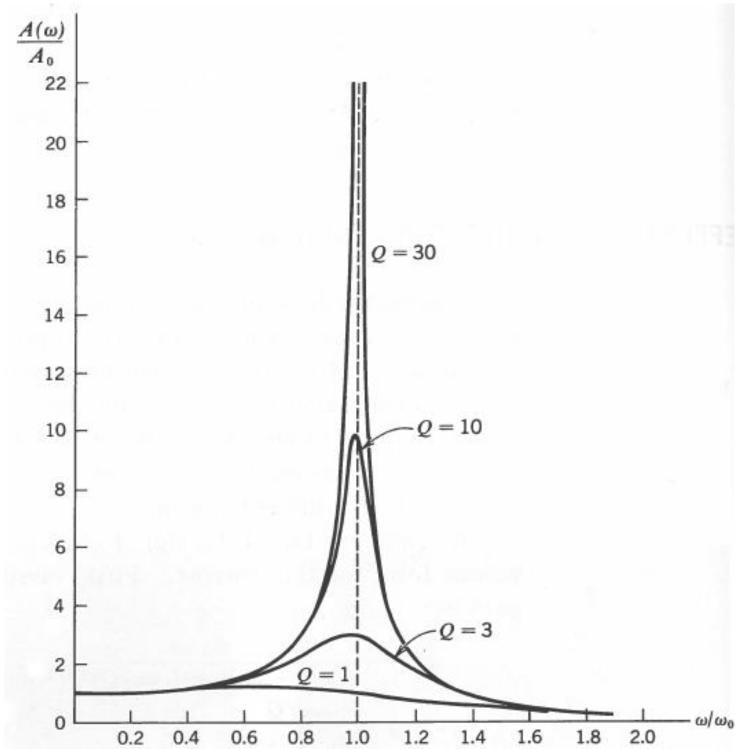

Figure 2.1.5: Oscillation amplitude as a function of driving frequency for different values of Q, under the assumption that the driving force is of constant magnitude but has variable frequencies (French, 1971).

Resonance is an important aspect of forced vibrations and is particularly relevant for situations where forced vibrations are acting upon the body, for example by vibrating tools or devices, such as whole body vibration platforms. The resonance frequency for various lower extremity muscles was quantified during quasi-static conditions (Wakeling and Nigg, 2001). However, it was recently stated that the muscles may vibrate at more than a single frequency during running and sprinting and more sophisticated tools may be used to analyse the muscle vibration damping or resonance frequencies of muscles during locomotion (Enders et al., 2012).

2.1.3 Effects of vibrations on soft tissues

The effects of vibrations on the body have been extensively studied in numerous scenarios over the last 60 years (Rittweger, 2010). This review will primarily focus on literature related to nerve, vascular and muscle tissue, as those areas are the most relevant and applicable for the work within this thesis.

In particular, the following topics are discussed:

(a) Soft tissue vibrations during exercise
(b) Negative effects of vibrations
(c) Vibratory effects on muscle energetic demands and myoelectric activity (EMG)
(d) Vibrations and maximal force production

2.1.4 Soft tissue vibrations during exercise

Running injuries were typically related to excessive impact forces upon ground contact of the heel, excessive mileage, previous injuries, and excessive pronation (van Mechelen, 1992; Clement et al., 1981; Cook et al., 1990; James et al., 1978; Robbins & Gouw, 1990).

The idea of high impact forces and its relation to injuries was challenged and it was speculated that runners with high impact peaks or loading rates did not show a higher incidence of running injuries than those with low ones (Bahlsen, 1989; Stefanyshyn et al., 2001). However, it was seen that soft tissue compartments of the human body are exposed to shocks and potential vibrations resulting from such shocks. Therefore, a new paradigm about the effects of impact loading in relation to EMG activity was proposed and named *muscle tuning* (Nigg et al., 1995; Nigg, 1997; Nigg and Wakeling, 2001).

Muscle tuning was described as a model that considers a signal, such as a heel strike, as an input with amplitude and frequency, which produces vibrations of the soft tissues. The tuning aspect of the paradigm considers the apparent vibrations that are damped by pre-tuning (activating) the muscles before heel strike to create a damped mechanical system and to avoid resonance effects. Muscle activity due to “muscle tuning” is expected to be high when the input frequency is close to the natural frequency of the soft tissue compartment. It has been discussed that muscle tuning may have implications on comfort, performance and fatigue (Nigg and Wakeling, 2001).

A study involving a pendulum apparatus, which allowed the simulation of impact forces present during ground contact, tested different shoe materials in order to change the loading rate. It was shown that the EMG 50 ms before impact and 50 ms after impact in multiple lower extremity muscles was adjusted depending on the loading rate of the impact force, and indicated an adaptation (tuning) effect of muscles before the impact occurred (Wakeling et al., 2001).

In a different experiment, the influence of different vibration frequencies on EMG activity of lower leg muscles while standing on a vibration platform was tested. The frequencies ranged between 10 Hz and 65 Hz and the vibrational power and EMG activity at various lower extremity muscles was quantified using skin mounted accelerometers and surface EMG electrodes. The results showed that different muscles had distinct frequencies at which the power dissipation (the vibration power measured at the platform relative to the one at the soft tissues) was the highest, and the frequencies usually corresponded to the natural frequency of the soft tissue compartments. In addition, the muscle activity showed also the highest EMG activity at frequencies close to the natural soft tissue frequency. It was suggested that soft tissue damping, induced by muscle activity, may be the mechanism by which resonance is minimized at heel strike during running (Wakeling et al., 2002a).

During walking, it was shown that shoes with different midsole hardness affected relative frequency relationships between the input (ground) and the soft tissue compartments. The loading rate (-15.9 %) and the input frequency by the ground (-9.4 %) were lower and closer to the resonance frequency of the soft tissues when using the soft shoes. Interestingly, the force transfer (magnitude of the impact peak) from the ground to the soft tissues was not increased in the softer shoes but the EMG activity in the biceps femoris and lateral gastrocnemius was increased. This result was interpreted as evidence for increased damping of soft tissue resonance at heel strike by muscle activity (Wakeling et al., 2003).

Similarly, a study used different shoe hardnesses and running speeds in heel-toe running and found a strong correlation between EMG pre-activation intensity and impact loading rate, without a change in peak soft tissue acceleration following impact. It was concluded that this result supports the proposed paradigm of muscle tuning and that muscle activity tunes soft tissues to control vibrations (Boyer & Nigg, 2004).

A further test of the muscle tuning paradigm was conducted in an experiment that blinded the subjects to the underground hardness. It was shown that for an unexpected hard landing, the input frequency induced by the impact force shifted closer to the natural frequency of the soft tissue compartments. Within this scenario, the muscles did not show an EMG adaptation prior to landing and soft tissue vibrations increased, in support of the muscle tuning theory (Boyer & Nigg, 2006). Yet the authors expected to find adaptation of EMG activity or vibration intensity at the next ground contact following the "untuned" landing but the results indicated otherwise.

A further experiment that varied the loading rate upon impact by using different shoe midsole hardnesses was conducted during running. Muscle activity was analyzed for a window of 150 ms before heel strike and the results showed significant differences in EMG intensity and the ratio between high and low EMG frequency components. The results indicated that muscle activity was used to tune the muscles for the impact task and that altering the material hardness of the shoe affects pre-activation. Specifically, the EMG frequency components were shifted, suggesting also a different fiber type recruitment. Interestingly, the authors mentioned changes in muscle activity over the duration of the continuous 30 min running experiment, without going into detail which particular changes occurred (Wakeling et al., 2002b).

2.1.5 Negative effects of vibrations related to vibration exposure

Numerous animal studies were conducted to identify which amplitude, frequency, and exposure duration combinations elicited degenerative processes in the soft tissues.

A rat-tail model was used to investigate the effects of 4 hours of vibration at 60 Hz at 49 ms^{-2}. This exposure dose was sufficient to cause myelin decompaction, enlarged separation areas of myelin membranes and increased interstitial space. Those observations are the first signs of nerve damage and may over time lead to numbness and axon loss. Furthermore, persistent reductions in vascular lumen size indicate remodeling processes due to prolonged vibration exposure which may hinder tissue perfusion (Govindaraju et al., 2006).

Rat hind legs were exposed to 82 Hz and 0.21 mm (peak to peak) vibrations for 4 hours on 5 consecutive days in a study by Lundborg and colleagues (1990). The plantar nerves showed clear signs of nerve damage when analyzed by electron microscopy; however, the effects seemed to be reversible within a four week recovery period.

Rat hind limb muscles were studied after vibration exposure for 5 hours per day for 5 consecutive days in two groups that were exposed to two different vibration amplitudes. The analysis revealed that the number of internally located nuclei in muscle fibres was significantly higher in the condition with larger vibration amplitude, and was interpreted as a sign of muscle damage (Necking et al., 1996). A few years earlier, necrosis of muscle fibres was shown in rat hind limbs after vibration exposure for 5 hours per day for 5 days, indicating severe damage of muscle tissue by vibrations (Necking et al., 1992).

The studies involving human subjects originated from vibration induced medical concerns in certain working environments and, for example, the effects of vibrations on workers using vibrating tools have been studied extensively. Common problems are complex vascular, neurologic and osteoarticular disorders in the upper limbs, which are referred to as hand-arm vibration syndrome (Bovenzi, 1998). The occurrence of the vibration induced white finger, a digital vasospastic disorder, is accompanied by sensory impairment, degenerative changes of joints and bones and an increased risk of upper-limb muscle and tendon disorders (Bovenzi, 1998).

Other effects of this vibration-induced disease are intense thickening of muscle layers around small arteries, demyelinating neuropathy in the peripheral nerves and a marked loss of nerve fibers. Also collagenous connective tissues replaced previously elastic fibers (Takeuchi et al., 1986).

Vibrations were traditionally used as a therapeutic intervention but recently became popular as a training tool in form of whole-body vibration platforms. Those platforms usually oscillate in a side-altering motion about an axis normal to the frontal plane of the person standing on it. Another popular way to transmit the vibrations through the platform is by lifting and lowering the entire platform synchronously. From the time of introduction of those vibration platforms, scientists were concerned about potential negative effects of the vibrations that were shown in workers exposed to vibrations.

A few acute adverse reactions were reported in a study involving a whole-body vibration platform. Those reactions included itching of the skin, erythema and edema in about half of the 37 tested subjects and predominantly in women. The skin irritations however resolved after walking for a few minutes (Rittweger et al., 2000). A larger concern may be the long term effects of vibration training, since vibration induced injuries to soft tissues may only be noticed after years of exposure (Takeuchi et al., 1986).

To assess the potential risk of whole-body vibration training, the transmission of vibrations from the vibration platform to the body (transmissibility) was studied. It was shown that different types of vibration platforms generate largely different accelerations. Synchronous vertical vibration platforms generated about 3.8 units of g (9.81 ms^{-2}), side-altering vibration platforms about 14.7 g and a third type of platform, vibrating in circular movements in the transverse plane, about 1.6 g in vertical direction. It was also shown that accelerations are damped predominantly at the ankles and decrease by a factor of about 10 when comparing accelerations at the ankle versus the knee in all platform types (Pel et al, 2009). A limitation of this study was that the vibration amplitude was different for all platform types, which limits the comparability between the results from different platforms and that transmissibility of vibrations from the platform to the body was assessed in vertical vibration direction only.

A different study tested the transmissibility of synchronous vertical vibrations to the ankle, hip and spine at amplitudes between 0.05 mm and 3 mm and at frequencies between 10 and 90 Hz. The study demonstrated a possible amplification of acceleration amplitudes at different frequencies, depending on the body part. Amplifications occurred between 10 and 40 Hz at the ankle, 10 and 25 Hz at the knee, 10 and 20 Hz at the hip, and at about 10 Hz at the spine (Kiiski et al, 2008). This study demonstrated the importance of transmissibility analysis when using whole-body vibration platforms as training devices, since the accelerations at different body parts may exceed the acceleration produced by the devices close to 10-fold.

Transmissibility was also assessed in a study using bone pins in the spine and greater trochanter of the femur. The accelerations at the hip exceeded those of the platform by 30 % at frequencies lower than 20 Hz and in an erect position, but transmissibility decreased substantially at frequencies higher than 25 Hz and as the knee flexion angle increased (Rubin et al., 2003). Transmissibility to the spine was always < 1, meaning vibrations at the spine were always lower than at the platform, indicating a damping effect.

The potential risks of whole-body vibration training were furthermore assessed in a study comparing vibrations between the vibration platform and the cranium to guidelines defined by the International Standards Organization ISO-2631 (Evaluation of human exposure to whole body vibration, 1997). It was shown that some vibration platforms (Vibratfit and Power Plate) exceeded vibration levels deemed to be unsafe even for exposures less than one minute and that the transmissibility to the cranium decreased substantially with increasing knee flexion (Muir et al., 2013).

This result was also shown in a study comparing the transmissibility using two different platforms (side-altering and synchronous) and different knee angles (Abercromby et al., 2007b). An inverse relation between knee flexion angle and transmissibility was shown for both platform types and accelerations transmitted to the head were always grater in synchronous than in side-altering vibration modes and it was shown again that the acceleration amplitudes exceeded the recommendations as defined by ISO 2631.

The applicability of ISO 2631 was questioned and the main point of criticism arose from the means by which the exposure limits within ISO 2631 were established. The exposure limits were based on data of pilots and drivers for vibrations transmitted through the buttocks (Rittweger, 2010). As it was shown that vibrations are damped substantially before reaching the hip (Pel et al., 2009; Kiiski et al., 2008; Muir et al., 2013), a direct comparison of vibrations measured directly at the platform to ISO 2631 may not be valid and standards have to be modified (Rittweger, 2010).

Although adverse effects of vibrations for soft tissues were known (see above), the transmissibility of vibrations to those tissues was hardly tested.

Only one study tested the vibration transmissibility to the shank and the thigh when using a synchronous vibration platform. Two knee flexion angles (30 and 40 degrees) and vibration amplitudes (1.5 and 3 mm), as well as five vibration frequencies (between 20 and 40 Hz) were tested. It was shown that the shank muscle compartment was exposed to the highest acceleration when the vibration frequency was low, amplitude high and the squatting position a deeper one. The same results were found for the thigh, albeit the squat depth did not have an influence. The transmissibility from the platform to the soft tissues was below 1 for both compartments and each tested condition, suggesting no amplification effects of vibrations for soft tissues during whole-body vibration training (Cook et al., 2011). Although the study by Cook and colleagues involved numerous conditions, scenarios like a side-altering vibration platform or more upright standing positions, have not yet been tested.

2.1.6 Effects of vibrations on muscle activity and metabolism

The effect of vibrations on muscle activity and metabolism can be separated into

(a) Effects related to a training response (induced over weeks and months of training)

(b) Effects on muscle activity and metabolism while being exposed to vibrations (immediate effects on the body)

Those immediate effects of vibrations on muscle activity and metabolism are the focus of this review.

Muscle activity of the vastus lateralis muscle was recorded at various vibration frequencies and a 100 degree knee bending angle. It was shown that EMG activity increased with vibration compared to control and the increase in activity at 30 Hz was higher than the one at 40 and 50 Hz (Cardinale & Lim, 2003).

In contrast, no change in vastus lateralis, vastus medialis and biceps femoris activity during whole-body vibration was found in one study (Cormie et al., 2006), which was surprising since the same platform and posture was used as in the work of Cardinale and Lim (2003).

The probably most comprehensive study about the effects of vibration platforms on immediate EMG activity to date was conducted by Ritzmann and colleages (2013). In this study, side-altering and synchronous vibration platforms were tested at various vibration frequencies, knee flexion angles, stance conditions and added loads. The main findings included higher muscle activity in side-altering versus synchronous vertical vibration platforms, higher activity with higher vibration frequency, higher activity with increasing knee flexion angle, higher activity when in forefoot stance for plantar flexor muscles and higher activity with additional weight on the shoulders of the subjects (Ritzmann et al, 2013).

A different study compared the EMG activity of four lower extremity muscles. Specifically, side-altering and synchronous vertical vibration platforms during static and dynamic unloaded squats at different knee angles and contraction types were compared. The results revealed significantly increased EMG activity during vibration exposure for both vibration types and all muscles. Knee extensor muscles were significantly more active using the side-altering platform, while the tibialis anterior was more active using the synchronous platform. Static squatting induced higher or equal muscle activity compared to dynamic squatting and eccentric contractions showed the lowest response, however, exceptions occurred for some muscles. Knee angle influenced all tested lower extremity muscles, with exception of the biceps femoris (Abercromby et al., 2007).

A major difference between the studies of Ritzmann and colleagues (2013) and Abercromby and colleagues (2007) is the means by which the muscle activity data was filtered. Spikes in the raw EMG signal, which occur with the same frequency as the vibration platform, are not removed by the former group, while the latter group completely removes those spikes. This is because the former group believes that the spikes are vibration induced reflexes and an important aspect of vibration induced muscle activity, while the latter group is convinced that those spikes are vibration induced motion artefacts. Until now, there is no consensus on how to treat EMG data when collected during whole-body vibration exposure and new publications on EMG during whole-body vibrations appear with either one of the data treatment methods.

There are at least two alternative methodologies to assess the activity of muscles during whole-body vibration training.

Firstly, a relatively gross estimate of muscle activity can be acquired by quantifying respiratory gas exchange. This allows for a global measure of aerobic effort for the entire body (including oxygen consumption of all other organs of the body) during vibration training. This approach was taken in a series of studies by Rittweger and colleagues (2000; 2001; 2002). It was shown that the oxygen consumption increased with higher vibration amplitudes, frequencies and added external load (Rittweger et al., 2002). The vibrations caused an increase in oxygen uptake to about 6 and about 8 $ml \cdot min^{-1} \cdot kg^{-1}$, which corresponds to an oxygen uptake typically experienced during moderate walking speeds (Rittweger et al., 2001).

The quantification of respiratory oxygen uptake yields information about the total metabolic cost of the body during whole-body vibration training. More detail about individual muscles requires the use of a different approach. The second known alternative to quantifying EMG is the use of near-infrared spectroscopy (NIRS), which is a non-invasive technique to quantify changes in oxygenized and deoxygenized hemoglobin in soft tissues (Jobsis, 1977; Boushel et al., 2001).

The NIRS optodes emit light with a wavelength of approximately 700-950 nm, which is a wavelength that passes well through body tissue (soft and hard tissues) and gets absorbed depending on the level of oxygenation of hemoglobin. This allows the measurement of the concentration change in the oxygenated hemoglobin (ΔO_2Hb) and deoxygenated hemoglobin (ΔHHb) in the muscle tissue.

The oxygenation of the vastus lateralis while performing a squatting exercise on a whole-body vibration platform was tested using NIRS and it was shown that the oxygenation was lower when vibrations were superimposed to the squat exercise. Also more blood was entering the muscle upon cessation of the exercise, thus indicating an increased oxygen demand of the muscle when vibrations are present (Yamada et al., 2005).

A different study tested the oxygenation of the vastus lateralis and the gastrocnemius medialis at different vibration frequencies, which were 30, 40, and 50 Hz. In contrast to the study of Yamada and colleages (2005), the study did not find any significant differences between the vibration and non-vibration (control) condition, independent of the vibration frequency (Cardinale et al., 2007).

The reason why Cardinale and colleagues (2007) did not find any difference in oxygenation may be related to the used measurement techniques (Coza et al., 2011). The measurements by Yamada and colleagues (2005) and Cardinale and colleagues (2007) were conducted without any circulatory occlusion of the tested muscle. Therefore, oxygenated blood can enter the measurement site of the muscle immediately, and the demand of the muscle itself cannot be quantified. In order to exclusively quantify the oxidative demand of the muscle during vibration exposure, the blood supply needs to be suppressed. Using an occlusion technique, a significant increase of 15 % in oxygen utilization rate when vibrations were superimposed to a dynamic calf raise exercise were shown and also increased recovery rate of oxygenation when vibrations were present upon cessation of the dynamic exercise (Coza et al., 2011).

In an earlier study, the effect of occluding the blood flow to the muscle was demonstrated during whole-body vibration training. It was shown that NIRS did not reveal significant changes in muscle oxygenation without occlusion, but a trend to increased consumption when blood perfusion was halted by occlusion during an isometric plantar flexion exercise. The quality of occlusion was not controlled in this study, and blood may have entered the measurement site and prevented the finding of significant changes (Zange et al., 2009).

The effects of changing vibration frequency and amplitude on muscle oxygen demands have not yet been studied using this technique but would provide more detailed insights into the metabolic demands of individual muscles during whole-body vibration training.

2.1.7 Effects of vibrations on muscle force development

The influence of vibrations on muscle strength and power has been studied as a long-term intervention (years and months) and short-term intervention (weeks, days, seconds) (Cochrane, 2011). The most direct form of short-term vibration effects are the immediate effects of vibrations on force production. The immediate effects, with respect to strength and muscular performance, have hardly been studied.

One study investigated the effects of vibrations directly applied to the thigh on EMG and maximal isometric knee extension strength. The results showed that vibrations did not affect root-mean squared EMG, maximal extension strength or the rate of force development (Humphries et al., 2004). However, it seems noteworthy that the maximal extension strength of all 16 subjects was on average 17.8 % higher during vibration exposure but high data variability did not allow to detect significant differences.

Already two years earlier, Warman and Humphries (2002) were able to show significantly improved maximal isotonic strength when vibrations were superimposed on the thigh using the same device (as described in Humphries et al., 2004).

In another study, a standard knee extension machine was modified to provide a 10 Hz vibratory stimulus directly to the lower legs. The results showed significantly increased dynamic strength and power when vibrations were superimposed while completing a set of 8 repetitions at 35 % of the one repetition maximum. Yet, maximum force, or a one-repetition maximum while vibrations are superimposed, was not tested (Mileva et al., 2006).

Only one study on the upper extremity was found which investigated maximal strength during vibration exposure. A Turkish group developed an apparatus to test MVC force at different arm bending angles in vivo, while mechanical vibrations were superimposed to the muscle. The vibrations were transmitted using a leather belt which crossed the biceps perpendicular to humeral long axis and elicited transversal vibrations. Force production increased significantly (+6.4 %) with superimposed vibrations between 6 and 24 Hz. The authors suggested that increased maximal isometric strength was caused by increased muscle spindle activity and motor unit firing and discharge rates resulting from the vibratory stimulus (Kin-Isler et al., 2006).

Vibrations elicited in this study occured purely in transversal direction (i.e., perpendicular to the biceps), however vibrations from vibration platforms, occupational tools or sporting exercise mostly occur in longitudinal direction (i.e., in the sagittal plane). If one thinks about performing tight turns on skis on an icy slope, one can feel the repetitive impacts transmitted through the feet and the small repetitive changes (vibrations) in the knee flexion angle. This is a practical example of how vibrations are transmitted in the sagittal plane where the quadriceps and gluteal muscles are repetitively stretched (eccentric contraction) and contracted (concentric contraction). This scenario is therefore different compared to the way Kin-Isler and colleagues (2006) applied vibrations. To date, it remains unclear how the muscle is influenced by vibrations that are similar to those occurring in a more realistic environment.

2.1.8 Vibration summary

Vibrations have been used for millennia to stimulate the human body, and modern scientific approaches have revealed a surprisingly large spectrum of effects. The effects range from elicitation of back pain and degenerative processes for muscles, nerves and vascular tissue to beneficial effects in muscle training, sports performance and recovery. Considering the effects of vibrations with a focus on the dicsussed topics, it can be summarized that:

1) Muscle activity appears to be (relatively) more elevated before heel strike in walking and running when the impact force rate resembles an input frequency close to the natural frequency of lower extremity soft tissue compartments ("Muscle tuning").
2) Under circumstances with high amplitude, frequency and exposure duration, vibrations can induce damage to nerve, muscle and vascular tissue and the transmissibility during whole-body vibration training seems tolerable for the skeletal system.
3) Whole-body vibration training appears to elevate muscle activity and the methodology used for data analysis is in debate. Energetic demand of the entire body increased to a level comparable to slow walking but the metabolic demand of individual muscles is largely unknown.
4) Immediate effects of vibrations on force development have hardly been explored, and no changes as well as increased isotonic knee extension strength and increased elbow flexion strength were reported.

Based on the current knowledge about vibrations, each of the above discussed topics has unexplored aspects:

Ad 1) Tissue vibrations during running seem to be damped by muscular activity. Using this assumption, one may speculate that during situations of muscular fatigue, such as during a prolonged run, the ability to damp vibrations may be reduced.

Ad 2) Transmissibility of vibrations during whole-body vibration training was studied for the skeleton, but is largely unknown for soft tissues, although negative effects on nerves, muscle and vascular tissue (soft tissues) were related to vibration exposure.

Ad 3) Quantification of the energetic demands of individual muscles during whole-body vibration training has critical shortcomings when quantified by means of surface electromyography and near-infrared spectroscopy can now be used as an alternative to gain a deeper understanding for this form of training.

Ad 4) A custom made vibration device allows the quantification of the effects of longitudinally applied vibrations on maximal force of the upper extremity, which is a way of vibration transmission that is close to many real-life situations and has never been tested before.

2.2 Compression Apparel

2.2.1 Origin of compression treatment in medicine and sports

Externally applied compression has been used as a medical treatment for many decades. The idea of using compression as a medical treatment was related with the desire to understand the coagulation process of blood. Virchow introduced his idea of a triad ("Virchow's Triad"), and claimed that there are three factors that need to be present for blood to coagulate. The factors were (1) blood stasis, (2) vascular damage, and (3) increased clotting factors (Virchow, 1856), which is a concept that is still valid in modern medicine (Janssen et al., 1993). The prevention of excessive coagulation is needed in patients suffering from insufficient venous function or after surgery, in order to prevent deep venous thrombosis or pulmonary emboli. In many patient groups, the use of drugs and related side effects are unacceptable to reduce coagulation (Clagett et al., 1989).

Externally applied compression was originally related to reductions in blood flow volume using high levels of compression (exceeding 50 mmHg) (Friedland et al., 1943) and low to moderate levels of compression (10 – 50 mmHg) (Halperin et al., 1948). Excessively high amounts of compression are in fact used to reduce blood flow to prevent swelling and hematoma after acute injury (Thorsson et al., 1987).

However, the group around Halperin and Stanton showed that, although volume decreased with compression (Halperin et al., 1948), the flow velocity increased (Stanton et al., 1949). Considering that slow blood flow (blood stasis) is one of the causes of blood coagulation, the potential of externally applied compression as a non-medical treatment without drug related side effects was investigated. The same researches later observed that elastic stockings, that were routinely applied to hospital patients, significantly reduced the incidence of expected fatal pulmonary embolism (Wilkins and Stanton, 1953).

The transition of externally applied compression from a purely medical intervention to the use in healthy athletes was rather slow. It probably started with the work of Starkey, who was one of the first to use compression to treat an acute sports injury (Starkey, 1976). He used intermittent compression, by means of a massage unit that applied compression in 15 second cycles, in combination with ice packs to treat ankle sprains and reported significant healing benefits of the used method.

The first researches to use compression for reasons other than treating an injury but rather on healthy subjects with the goal to potentially improve athletic performance were Michael Berry and Robert McMurray in 1987. Based on their knowledge about the flow enhancing properties of compression apparel, they hypothesized that "… increased venous blood flow provided for by wearing the GCS (graduated compression stocking) would be beneficial during exercise. An increased venous blood flow could aid in the removal of metabolites produced by the working muscle", (Berry and McMurray, 1987). They found that blood lactate was lower 15 min after high-intensity running exercise on a treadmill when wearing compression stockings compared to a control condition. The public and scientific interest in this topic increased substantially after this publication and the applications and current knowledge about compression apparel in sports is summarized below.

2.2.2 Compression apparel: Translation of medical knowledge to applications in sports and other activities

2.2.2.1 Effects on blood flow

As mentioned above, one proposed benefit of compression apparel is that it potentially improves blood flow, and the first ones to report an improved blood flow velocity were Stanton and colleagues (1949). Compression was applied to the lower limb and a mean increase of about 45 % in deep vein velocity was observed. It was concluded that pressure as little as 20 mmHg may be sufficient to increase the flow velocity significantly, while reducing flow volume by only a small amount. The assessment of flow velocity and volume was limited to the deep veins in this study. A substantial advancement in understanding blood flow dynamics when using compression was made by the study of Lawrence & Kakkar (1980). Deep venous flow velocity of the lower limb was investigated next to muscle blood flow and subcutaneous blood flow. Probably the most important finding was that the flow velocity in the deep veins was improved at pressures below 20 mmHg, while muscle blood flow and subcutaneous blood flow were not negatively affected.

A later study focussed on the upper extremity and arterial flow volume rather than venous flow. It was found that arterial inflow was doubled when compression between 13 and 23 mmHg was applied to the forearm (Bochmann et al., 2005).

A drawback of many studies on compression and blood flow is that they were conducted at rest, and the beneficial effects during exercise may be questioned. Only a few studies showed that compression apparel/stockings improved leg blood flow during exercise. It was shown that compression improved flow volume in the legs during walking (Ibegbuna et al., 2003) and during stationary walking (Somerville et al., 1974). Furthermore, some studies investigated blood flow during quasi-dynamic exercise, and showed improved venous return with compression during stationary stepping (Husni et al., 1970) and tiptoeing (Christopoulos et al., 1987).

In summary, compression apparel/stockings have the capability to improve blood flow volume and velocity in the upper and lower extremity and are a commonly used medical treatment. The effects were shown in static conditions and low-intensity dynamic exercises.

2.2.3 Effects of compression on soft tissue vibration

Soft tissue vibrations can be induced by the impact between the ground and the heel during various exercises and were proposed to be related to comfort, performance and fatigue (Nigg, 2001). Therefore, it was suggested that muscles have a strategy to reduce soft tissue vibrations by pre-activating the muscles (`muscle tuning`) before heel-strike, in order to increase its vibration damping properties (Nigg et al., 1995; Nigg, 1997; Nigg 2001, Wakeling and Nigg, 2001). Compression apparel was thought as a modality to mechanically reduce soft tissue vibrations after impact.

Only one study showed that anterior-posterior and longitudinal muscle oscillations (as measured by tissue displacement) of the thigh were significantly reduced after landing from a vertical jump in men and women when using compressive shorts (Doan et al, 2003).

A substantial amount of work on this topic has been done within the Human Performance Laboratory at the University of Calgary, where internal studies showed that compression apparel significantly improved the damping coefficient of lower extremity muscles during running (Coza and Nigg, 2008), however, peak acceleration after heel-strike was not significantly reduced with compression (Osis et al., 2006).

Compression apparel reduced tissue vibrations after landing from a jump, and experiments during walking and running, at different speeds and different applied levels of compression were not yet conducted.

2.2.4 Effects of compression apparel on sports performance and muscle perception

The apparent effect of increased blood flow, promoted venous return and accelerated metabolic waste product removal during exercise when using lower limb compression triggered the interest of scientists working on sports performance analysis.

A study done on cyclists, which were wearing lower body compression apparel, showed a likely increase in power output when cycling at the anaerobic threshold and an improvement in muscle oxygenation economy (defined as the power output divided by the percentage of O_2 used within the muscle) (Scanlan et al., 2008). Yet, the same study did not find any improvements in performance during a one hour time trial with compression.

A different study required the participants to complete two 40-km cycling time trials with a break of 24 hours in between. The group wearing compression tights during the 24 hour recovery period was able to improve the cycling performance by decreasing the time needed for the second trial by 1.2 ± 0.4%. The authors concluded that wearing compression apparel during the recovery period may be worthwhile for endurance athletes (de Glanville and Hamlin, 2012).

A study conducted with trained elderly cyclists (63 ± 3 y) showed that cycling power was maintained better between two maximal cycling efforts for 5 min. When using compression stockings, the decrease in power in the second maximal effort was on average 2.1 ± 1.4 % lower and lactate was significantly lower during the 80 min recovery period between trials (Chatard et al., 2004).

Jumping performance of volleyball players, as quantified by mechanical power exerted on a force platform, was not improved in terms of a single jumping power effort when wearing compression, but the mean jumping power over 10 jumps was significantly higher in the compression condition and the finding was proposed to represent a benefit during repeated jump performances (Kraemer et al., 1996).

Scientists expanded their fields of research and tested compression for various activities related to tissue perfusion concerns. Compression tights were, for example, tested during flights over 5 hours or more and wearing the garment improved subjective ratings of leg feeling, pain and discomfort together with increased energy levels, alertness and concentration (Hagan and Lambert, 2008). In addition, ankle swelling decreased significantly when wearing the compression tights in the same study.

One study used an isokinetic dynamometer to assess the effects of compression on muscle activity (EMG), peak torque and power during isometric and isokinetic knee extensions. While torque and power were unaffected by the compression, root mean squared EMG was significantly lower and it was suggested that this might indicate an increase in muscle contraction efficiency (Fu et al., 2012).

A recent study used lower limb compression garments for 24 hours after a simulated rugby game. The group wearing the compressive garment was able to complete a run over 3 km in less time (-2.0% ± 1.9%) as well as repeated 30 m sprint time (-1.2% ± 1.5%) compared to the group wearing a placebo suit (Hamlin et al., 2012). In addition, perceived soreness was reduced by 15.8% ± 26.1% in the compression group which supports the view that compression garment may aid recovery if worn over several hours.

The effects on recovery were also tested in a study involving rugby players. Two 80 min high intensity exercise circuits were completed with a recovery time of twenty-four hours between them. Before and after each circuit the 20 m sprint performance and maximal power in a scrum effort were tested. Wearing compression during exercise and recovery periods had no benefit compared to not wearing compressive garment in this study (Duffield et al., 2008).

Compression stockings were used in a study that tested the effects of compression on running performance as measured by time to exhaustion and on gastrocnemius medialis muscle oxygenation. The stockings significantly increased tissue oxygenation before the running test and also during the passive recovery phase after the run, however, time to exhaustion was not significantly affected (Menetrier et al., 2011).

One study evaluated the effects of various compression apparel designs (stockings, tights, whole body suits) on sub-maximal and maximal running performance as measured by lactate concentration, oxygen saturation in the blood, pH, respiratory oxygen uptake, ratings of perceived soreness before, during and after the test. Furthermore a time to exhaustion test was conducted. None of the compressive apparel designs significantly affected any of the measured performance variables and it was concluded that compression apparel was unable to elicit benefits in running performance (Sperlich et al., 2010).

The influence of ambient temperature on running performance while wearing lower body compression garments was tested at 10 °C and at 32 °C. Time to exhaustion at VO_{2max} and heart rate were not statistically different between compression and non-compression conditions at any temperature. Also, rectal temperature was not different for any of the conditions but skin temperature was higher when wearing compression tights at the 10 °C condition. A small effect size ($d = 0.48$) for increased time to exhaustion at 32 °C with compression was found in 7 out of 10 subjects (Goh et al., 2011).

Two studies were conducted by Ali and colleagues who tested the effect of compression apparel on running performance. Firstly, they tested the effects of three different levels of compression using compression stockings on oxygen uptake, heart rate and blood lactate during a submaximal 40 min run on a treadmill. Also perceived comfort, tightness and pain were assessed and furthermore signs of muscle damage by quantifying creatine kinase, myoglobin, counter-movement jump height and pressure sensitivity. No effects on oxygen uptake, heart rate or lactate were found. High levels of compression were perceived to be tighter and more pain inducing and no differences in creatine kinase, myoglobin and jump height were found after 24 hours (Ali et al., 2010).

The second study was designed to test the effects on a direct indicator of running performance and time needed to complete a distance of 10 km was recorded when wearing three different level of compression stockings or a control stocking. Also, leg power was assessed by measuring countermovement jump height before and after the run. The time to complete the running distance was not significantly different between the conditions and the stockings were not able to improve running performance but low and medium levels of compression were able to maintain jump height better than no or high levels of compression (Ali et al., 2011).

Few studies were able to demonstrate a benefit in running performance when using compression apparel. One example of demonstrated benefits was a study done on moderately trained athletes using compression stockings. Running performance was defined as time under load, work, and aerobic capacity. When using compression stockings, time under load and total work were significantly higher compared to not wearing the stockings although the effect sizes were small (0.4 and 0.3, respectively) (Kemmler et al., 2009). Aerobic capacity (VO_{2max}) was not different between conditions. Furthermore, the authors found a significantly increased running speed at the aerobic and anaerobic thresholds with very small effect sizes ($d = 0.22$ and $d = 0.28$, respectively).

A study on 8 male team-sport athletes investigated the effects of whole-body compression apparel on running performance. In this study, performance was defined by total distance completed, velocity-specific distance, and high-intensity self-paced distance covered during a prolonged high-intensity intermittent running exercise on a treadmill. The results showed a likely improvement in total distance covered and low-intensity distance compared to control when using magnitude-based inferential statistics (Sear et al., 2010). Distance covered during high-intensity running was unchanged as well as blood lactate which was sampled throughout the test. The improvements in covered distance were explained by measured increases in muscle oxygenation.

Reduced oxygen consumption per minute was shown in runners in a study by Bringard and colleagues (2006). When wearing elastic tights, the difference in oxygen consumed between min 2 and min 15 when running on a track at 80 % of VO_{2max} was significantly lower when compared to non-compressive shorts and elastic tights.

In a time to fatigue test, conducted at 105% of the pace of the subjects' best recent 10 km run, maximal heart rate was significantly lower when wearing elastic compression stockings compared to when not wearing them. A trend to increased time to fatigue was found (effect size:

d = 0.32) also when using compression stockings, but the difference was not significant. Lactate at the end of the test did not significantly differ between conditions (Varela-Sanz et al., 2011).

2.2.5 Compression summary

Compression has long been used as a medical treatment for vascular diseases and post-surgery to improve blood flow. The treatment of sports injuries introduced compression into sports and lead to a substantial number of studies conducted with the goal to improve recovery or athletic performance. The state of the art and open questions of compression apparel in sports can be summarized as follows:

1) Compression has repeatedly been shown to accelerate long-term recovery processes of study participants when the apparel was worn for multiple hours post exercise.
2) Compression apparel was shown to positively influence some physiological variables related to athletic performance (heart rate, lactate, muscle oxygenation, etc.) during exercise.
3) Direct measures of performance (time to exhaustion, time trails) were usually not found to be significantly improved, with very few exceptions.
4) It is unknown whether compression apparel can increase blood flow during high-intensity exercises

The literature indicates that the effects of compression are small, if at all existing, and can often not be distinguished from natural variation of the human body or the error induced by the measurement techniques and equipment. Since the amount of compression needed to improve blood flow is a highly sensitive matter and is unknown during high-intensity exercises, one approach to bypass this issue is to test multiple levels of compression on the same subjects and to apply compression to all extremities to maximize the potential effect. This approach was taken into consideration in one project of this thesis.

Chapter three: Tissue vibration in prolonged running

3.1 Introduction

Repeated impacts between the foot and the ground during heel-toe running produce impact shocks which are transferred to the skeletal system and the soft tissue compartments, which will vibrate with respect to the skeleton.

Prolonged exposure to vibrations can have negative effects including reductions in motor unit firing rate and muscle contraction force (Bongiovanni et al., 1990), decreased nerve conduction velocity and reduced peripheral circulation (Dupuis and Jansen, 1981; Gilioli et al., 1981). Short term vibration exposure has been shown to increase the number of muscle fibers with internally located nuclei in rats (Necking et al., 1996), indicating a common non-specific feature of neuromuscular disorders (Dubowitz, 1985). The same study also reported swelling of the muscle fibers due to vibration, probably reflecting the initial stages of fiber damage. During running, the muscle tissue is exposed to vibrations for a relatively short duration compared to the afore-mentioned studies. However, they occur in a highly repetitive manner in prolonged running and at an intensity that is up to five times higher (Boyer and Nigg, 2006) than the intensity used in the study from Necking and colleagues.

It has been suggested that soft tissues are damped to minimize vibrations and that muscles are activated to control the frequency and damping of those vibrations, a process referred to as *muscle tuning* (Nigg, 1997). In support of the muscle tuning paradigm, it has been suggested that the body responds to the impact at heel-strike by muscle activation (Wakeling et al., 2002a) and that lower extremity muscle activity adapts to altered impact forces during running (Wakeling et al., 2001, 2002b). The resulting muscle contractions in turn influence the mechanical characteristics of soft tissues such that vibrations are minimized (Wakeling and Nigg, 2001). This has been shown to occur in non-fatigued muscles. However, it could be speculated that the protective minimization of vibrations is influenced by muscle fatigue. In that case one would expect that the damping effect of *muscle tuning* would be negatively affected by prolonged running.

A second mechanism that would lead to a change in soft tissue vibrations due to fatigue is a possible change in running kinematics. It has been shown that during running the foot lands more flat at heel-strike when runners are in a fatigued state, than when they are non-fatigued (Natrup, 1997). This alteration of ankle kinematics could possibly lead to a reduction of the tension in the Achilles tendon with fatigue (Natrup, 1997). A corresponding reduction of the tension in the triceps surae muscle group is also likely. If the triceps surae is assumed to be a “vibrating string”, a decrease in tension would correspond to a decrease in its natural frequency. A main factor that describes how detrimental the effects of vibration exposure are for the body is the peak acceleration of the muscle compartment during vibration. This is influenced by the amplitude and frequency of the vibration. A decrease in vibration frequency corresponds to slower oscillations, which would lead to a decreased peak acceleration and therefore have a less detrimental effect on the tissue. Thus, the question is whether the triceps surae changes its vibration characteristics with increasing fatigue. The purpose of this study was to identify fatigue dependent changes in soft tissue vibration characteristics during prolonged running.

The hypotheses to be tested were:

H1 The vibration intensity of the triceps surae increases with increasing fatigue

H2 The vibration frequency of the triceps surae decreases with increasing fatigue

3.2 Methods

3.2.1 Subjects

A total of ten healthy, physically active volunteers participated in this study: seven female (age 31.7 ± 7.3 yr; mass 60.1 ± 6.4 kg; height 165.5 ± 4.3 cm; mean ± SD) and three male (age 26.7 ± 2.3 yr; mass 65.3 ± 3.3 kg; height 173.8 ± 3.8 cm). The subjects were recreational runners and were rearfoot strikers. All subjects had a minimum training frequency of three times per week over the last four months and no injuries over the last three months. All subjects gave their informed, written, consent to participate and the institution’s health research ethics board had approved the study.

3.2.2 Protocol

The subjects were asked to complete as many laps as possible of a 230 m outdoor course while maintaining a constant running pace. Each single lap split was provided immediately to the subject to ensure the pace was held constant throughout the run. The appropriate constant running pace was determined for each subject based on their self-reported best 10 km race time within the last year (females: 3.1 ± 0.15 ms^{-1}; males: 3.8 ± 0.26 ms^{-1}; mean ± SD). An additional three seconds per lap was added to account for the course difficulty (slight elevation gain on each lap) and the fact that the psychological aspects of racing were not present. The run was ended when a) the subject was no longer able to maintain the required lap split for three consecutive laps or b) they reached exhaustion and could not continue to run. For this study, fatigue was categorized by the time into an exhaustive run. The degree of exhaustion achieved during the run was quantified by the rate of perceived exertion (RPE) (Borg, 1971), which was shown to serve as a linearly related marker of the time left to exhaustion without threatening homeostasis (Noakes, 2004). Each subject reached a RPE of at least 19 out of the 20 point scale, corresponding to a perceived exertion verbally described as “very, very hard”, before the end of the run. This procedure ensured that the subjects reached a high degree of fatigue by the end of the test. The first five laps of the run were termed the “non-fatigued” state, while the last five laps were termed the “fatigued” state.

3.2.3 Data Collection

All subjects were instrumented with two accelerometers (ADXL 78, range ± 35 g, nominal frequency response 0-400 Hz, Analog Devices USA). A single-axis accelerometer was secured with a vertical orientation to the heel-cup of the left shoe to detect heel-strike. A tri-axial accelerometer was secured to the muscle bulk of the triceps surae to measure the vibrations of that soft-tissue compartment. All accelerometer data were recorded at 2400 Hz using a 12-bit data acquisition system (National Instruments, USA). Sensor calibration was done by aligning each axis of the accelerometer with gravity to measure the voltage value for 1 g. When affixed to the body, the three axes of the tri-axial accelerometer were oriented to be (1) axial (parallel to the long axis of the tibia), (2) medio-lateral relative to the tibia and (3) normal to the skin. The accelerometer on the triceps surae was fixed to the skin using stretch medical adhesive tape (Cover-Roll stretch, BSN Medical, Hamburg, Germany).

3.2.4 Data Analysis

Data collected during the non-fatigued and fatigued states were analyzed and compared. All analysis was performed using custom MATLAB software (Version 7.6.0.324, The MathWorks Inc., Natick, MA, USA). A total of 100 steps were extracted for each subject: 50 steps taken during the first five laps (non-fatigued state), and 50 steps taken during the final five laps (fatigued state). Heel-strike was determined using the heel mounted accelerometer. Individual steps comprised 190 ms of data starting at heel-strike. This duration captured the impact shock related vibrations entirely and was determined by observation of the raw signals. Individual steps were zero-padded to a total length of 2048 data points and a Hamming window function was used to minimize the potential negative influence on the subsequent analysis due to zero-padding. The frequency content of the comprised data for each axial direction was calculated using Fast Fourier transformation and was represented as power spectra (1.2 Hz resolution). A total of six mean spectra were generated for each subject including data corresponding to the three accelerometer axes during the non-fatigued and fatigued conditions.

Local intensity maxima were identified in each of the mean power spectra for the non-fatigued condition. Those values were then compared to the intensities in the spectra from the fatigued condition at the same frequency. The inverse was also examined (i.e, comparison of peaks occurring during the fatigued state relative to the non-fatigued values). This was repeated for each accelerometer axis. Paired Student’s t-tests were used to compare corresponding spectral intensity values and differences were considered as significant at a $p < 0.05$ level.

To account for the second hypothesis, a frequency shift was defined as a difference of at least 2.4 Hz (i.e., twice the frequency resolution of the Fourier transformed signal) between a local maximum in the non-fatigued condition to the corresponding local maximum in the fatigued condition. For example, a shift of a local maximum occurring at 15 Hz in the non-fatigued condition to an analogous local maximum at 17.4 Hz in the fatigued condition was considered as an increase of vibration frequency with fatigue. Relative changes in frequency were calculated as the difference between the fatigued and the non-fatigued condition.

A wavelet transform was used to determine the timing of the maximum vibration intensity relative to heel-strike. Based on a previous publication (von Tscharner, 2000) a filter bank of eight non-linearly scaled wavelets *W(f)* was defined as the following function of frequency, *f*:

$$W(f) = \left(\frac{f}{f_c}\right)^{f_c s} e^{-f_c s (f/f_c + 1)}, \qquad \text{(Equation 3.2.1)}$$

where f_c is the center frequency of the wavelet and *s* is a scaling factor. Wavelets were scaled to resolve the frequencies present in soft tissue vibrations, which have a frequency content ranging from ~10 Hz to 50 Hz (Wakeling and Nigg, 2001). Parameters describing the wavelet bands 1-8 are shown in Table 3.2.1.

Table 3.2.1: Center frequencies and the corresponding time resolutions of wavelet bands 1-8.

Wavelet Band #	1	2	3	4	5	6	7	8
Center frequency [Hz]	2.96	8.27	16.16	26.61	39.58	55.06	73.02	93.46
Time resolution [ms]	146	103	73	60	47	40	37	30

Firgure 3.2.1 shows the raw acceleration signal (axial direction) of one runner early into the run and Figure 3.2.2 provides an example of a wavelet transformed signal in which the time from heel-strike to maximum vibration intensity (Δt) is indicated for band #2. Relative changes of Δt were calculated as the difference between the fatigued and the non-fatigued condition for each subject and accelerometer axis.

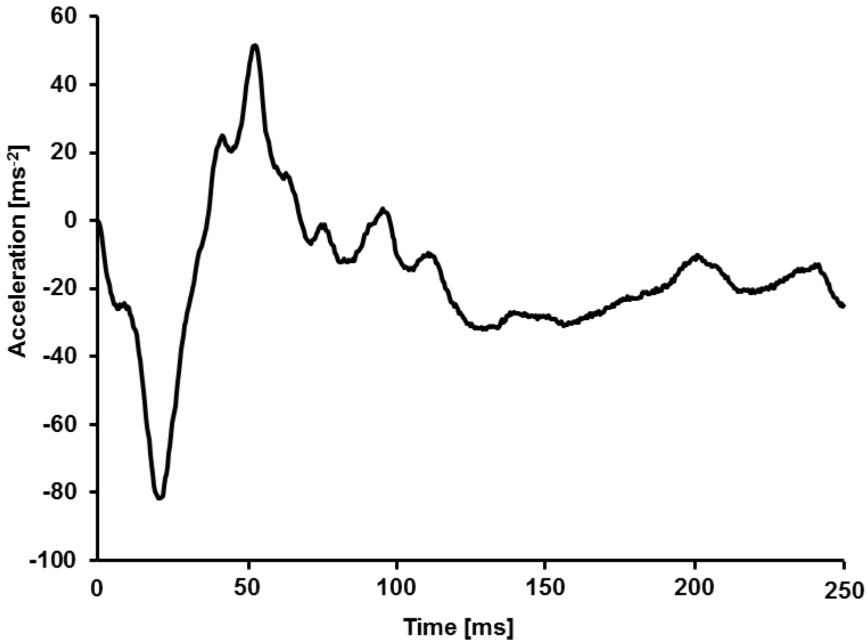

Figure 3.2.1: Raw acceleration signal in axial direction measured from the time of heel-strike until 250 ms after heel strike.

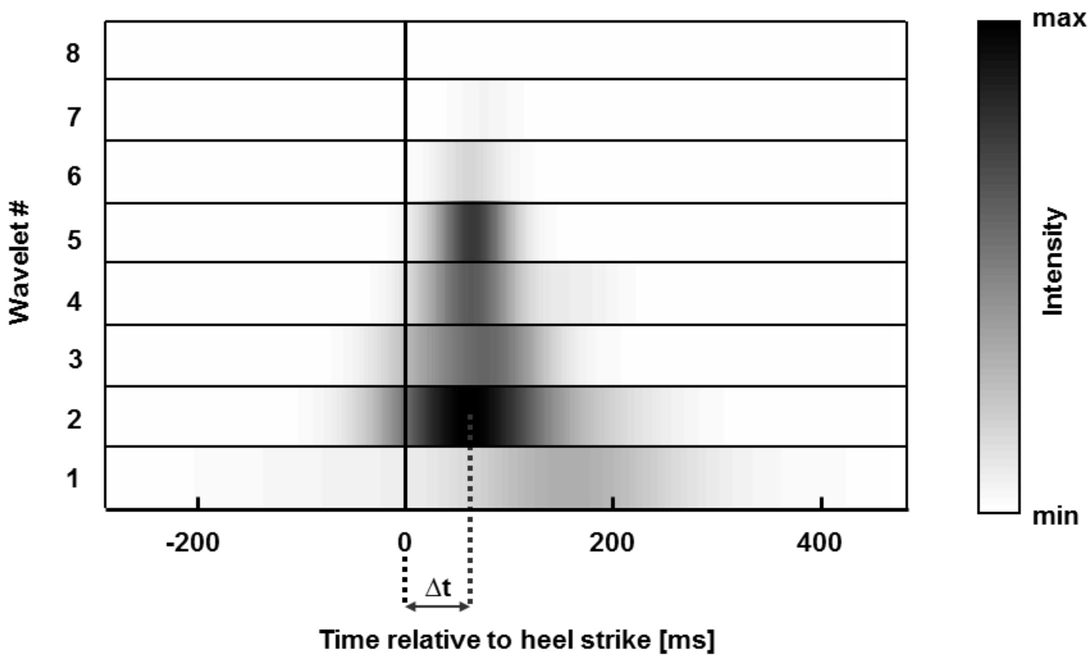

Figure 3.2.2: Wavelet transformed signal for axial acceleration data. The time from heel-strike to maximum intensity in wavelet band #2 is indicated with Δt. Intensity is scaled using shades of gray; the darkest areas represent highest intensity.

3.3 Results

On average, the subjects ran 10.4 ± 2.4 km (mean ± SD) before the run was terminated in a fatigued state (based on the criteria listed previously).

3.3.1 Power Spectra

An example of the power spectra extracted from axial accelerometer data in the fatigued and non-fatigued states is shown in Figure 3.3.1. In this example, local maxima are determined based on the fatigued data and are indicated by arrows.

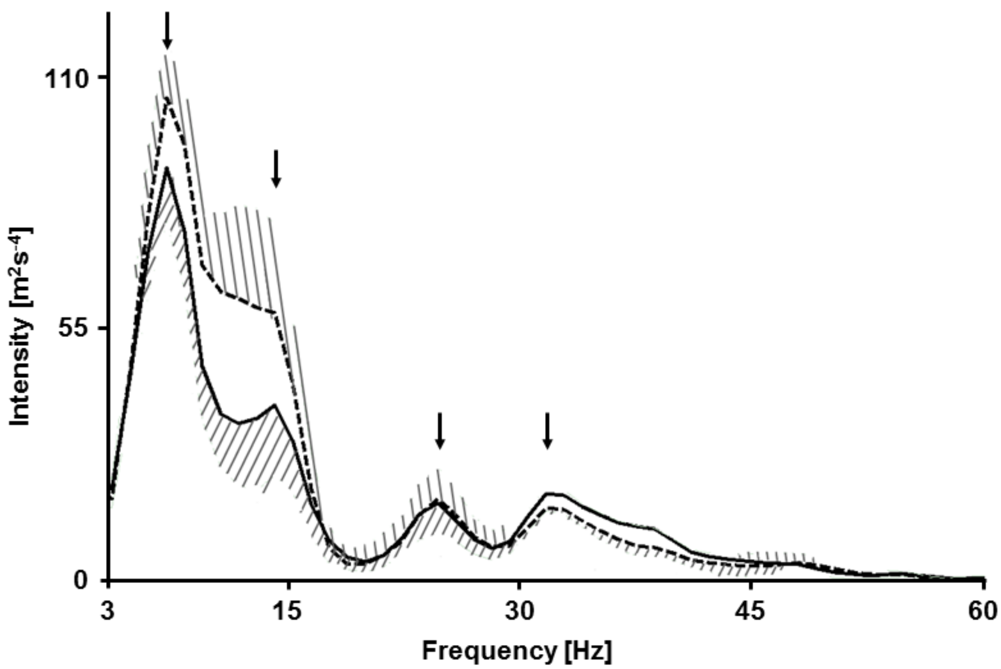

Figure 3.3.1: Example of a power spectrum from accelerometer data in the axial direction. The solid line represents the non-fatigued condition (mean - SD), the dashed line the fatigued condition (mean + SD). The arrows are pointing at local maxima.

Across the ten subjects, a total of 42, 42, and 57 maxima were detected in the axial, medio-lateral, and anterior-posterior directions, respectively. Out of the 42 maxima in axial direction, 21 maxima showed increased vibration intensity and eight showed a decreased vibration intensity. In medio-lateral direction, 21 maxima increased and ten showed decreased vibration intensity. For the anterior-posterior direction, 24 increases and 19 decreases were identified. All of these reported changes were statistically significant ($p < 0.05$).

In the axial direction, no shift to higher frequencies and one shift to lower frequencies occurred. In the medio-lateral direction, the frequency of local maxima increased once and decreased once with fatigue. In the anterior-posterior direction, the frequency increased one time and decreased three times. The mean change in vibration frequency with fatigue (Δf) relative to the non-fatigued state can be seen in Figure 3.3.2

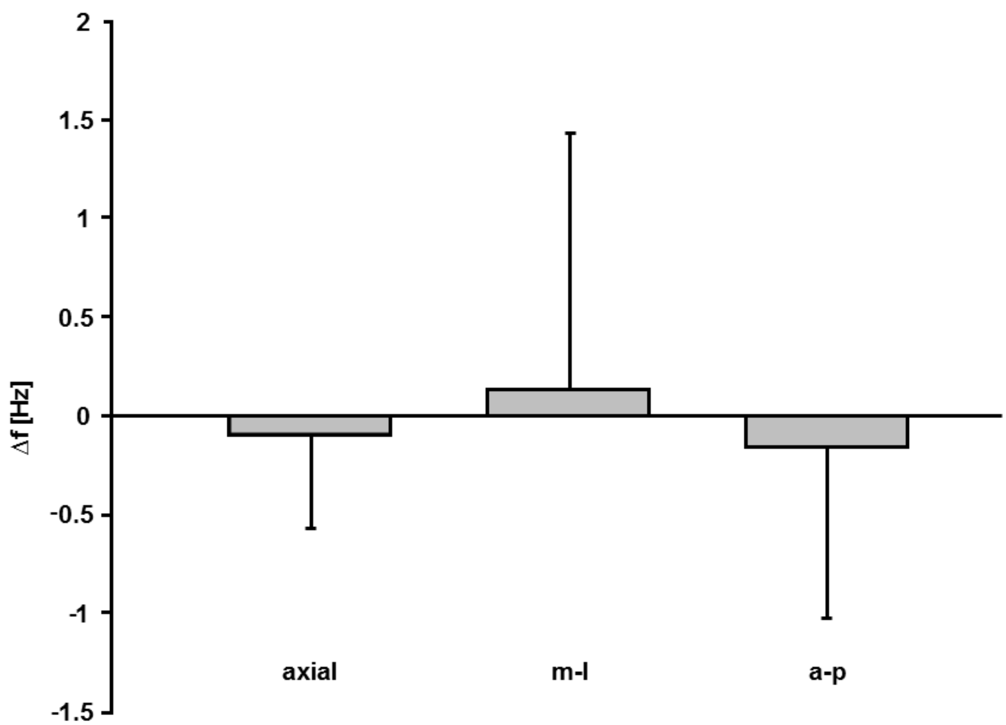

Figure 3.3.2: Mean change in vibration frequency of local maxima between the fatigued and the non-fatigued state for the axial, medio-lateral (m-l) and anterior-posterior (a-p) directions. Bars show the mean change, error bars show the SD from ten subjects.

Identified shifts occurred in a range between 2.4 Hz and 4.8 Hz. Individual and total results are listed in Table 3.3.1

Table 3.3.1 Number of local peaks, changes of intensity and frequency shifts for each subject and direction (m-l stands for medio-lateral and a-p for anterior-posterior direction). Numbers in brackets indicate significant differences ($p < 0.05$). The abbreviations *hi* and *lo* in brackets indicate a shift to lower (lo) or to higher (hi) frequencies with fatigue (F).

Subject #	Direction	Intensity increase in F	Intensity decrease in F	Frequency shifts
1	axial	4 (2)	0	0
	m-l	2 (1)	3 (0)	0
	a-p	5 (4)	1 (0)	1 (lo)
2	axial	3 (2)	1 (1)	0
	m-l	2 (2)	2 (2)	0
	a-p	2 (1)	3 (2)	0
3	axial	3 (2)	1 (1)	0
	m-l	4 (4)	2 (2)	0
	a-p	2 (2)	3 (2)	0
4	axial	2 (1)	1 (0)	0
	m-l	1 (1)	1 (0)	0
	a-p	1 (1)	4 (4)	0
5	axial	7 (5)	0	0
	m-l	2 (2)	0	0
	a-p	2 (1)	4 (2)	0

6	axial m-l a-p	7 (5) 4 (4) 7 (6)	0 1 (0) 0	0 1 (lo) 1 (lo)
7	axial m-l a-p	1 (1) 1 (1) 7 (5)	0 3 (2) 1 (0)	0 0 1 (hi)
8	axial m-l a-p	0 5 (4) 1 (1)	5 (4) 0 2 (2)	1 (lo) 0 0
9	axial m-l a-p	3 (3) 2 (1) 2 (1)	2 (2) 3 (3) 5 (4)	0 1 (hi) 1 (lo)
10	axial m-l a-p	1 (0) 1 (1) 2 (2)	1 (0) 3 (1) 3 (3)	0 0 0
Total	axial m-l a-p	31 (21) 24 (21) 31 (24)	11 (8) 18 (10) 26 (19)	1 (lo) 1 (hi), 1 (lo) 1 (hi), 3 (lo)

3.3.2 Time - Frequency Analysis

A majority of the wavelet bands showed an increased amount of time until the maximal vibration intensity occurred in the fatigued condition (Figure 3.3.3). For the fatigued condition, wavelet bands #2 and #4 showed a significant increase ($p < 0.05$) in the time delay for the axial direction. Additionally, a significant increase in the time delay with respect to fatigue occurred in band #8 ($p < 0.01$) for the medio-lateral direction, as well as in bands #2 - #5 for the anterior-posterior direction ($p < 0.05$). A significant decrease in time to peak signal intensity never occurred with fatigue.

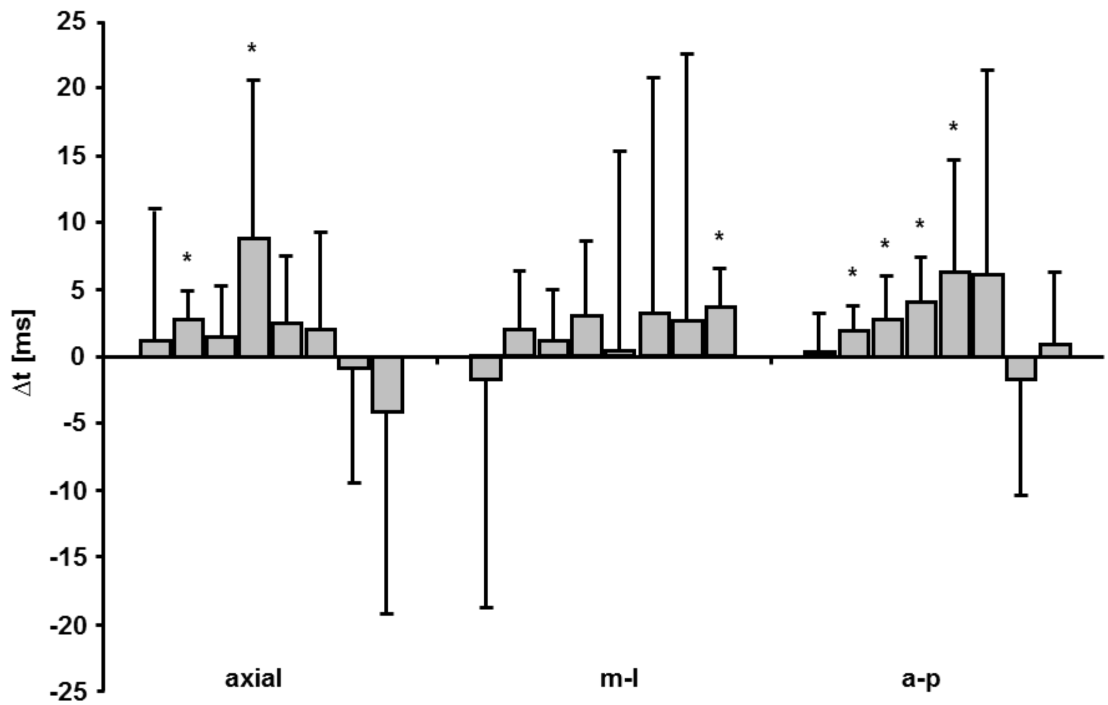

Figure 3.3.3: Mean change of Δt between the fatigued and the non-fatigued state for accelerometer data in the axial, medio-lateral (m-l) and anterior-posterior (a-p) directions for wavelet bands #1 - #8. Bars show the mean change, error bars show the SD from ten subjects. Asterisks denote significant effects with $p < 0.05$.

3.4 Discussion

3.4.1 Intensity changes and time-frequency analysis

The first hypothesis tested in this study was that the intensity of soft-tissue compartment vibrations increases with fatigue. It was shown that the intensity of most vibration peaks increased in axial and medio-lateral direction while anterior-posterior direction intensities did not show systematic changes. The time from heel-strike to maximum signal intensity was significantly longer ($p < 0.05$) in seven wavelet bands across the three vibration directions but was never significantly shortened. This finding was supported by the fact that the time was shortened in only four ($p > 0.05$) of 24 possible wavelet bands (Figure 3.3.3). According to the paradigm of muscle tuning, vibrations are damped by activating the muscle and it was proposed that a fatigued muscle might not be able to dampen the tissues as well as a non-fatigued muscle. It has been shown that fast twitch fibers in human gastrocnemius muscle can develop their maximum tension within 85 ms compared to slow twitch fibers which take more than 99 ms (Garnett et al., 1979). Since in 21 out of 24 wavelet bands maximum vibration intensities occurred within 80 ms after heel-strike, fast twitch fibers were assumed to play an essential role in vibration damping (Wakeling et al., 2002a). It is generally acknowledged that type II (fast/glycolytic) muscle fibers fatigue more rapidly than type I (slow/oxidative) fibers (e.g. Szent-Gyorgyi, 1953; Edström and Kugelberg, 1968; Burke et al., 1973). By taking the considerable fraction of fast twitch fibers in the soleus (30 %) and gastrocnemius muscles (50 %) into account (Edgerton et al., 1975), the earlier fatigue of this fiber type might reduce the capability of the muscles to actively contribute to the process of vibration damping.

The discussed mechanism might also explain the generally increased time span from heel-strike until the maximum vibration intensity was reached (Figure 3.3.3). It is likely that also the total time needed to dampen the vibrations and therefore the duration of time during which the tissue is exposed to possibly damaging vibrations is increased with fatigue. Increases and decreases in vibration intensities in the anterior-posterior direction, however, occurred alike with fatigue which suggests the influence of other variables related to vibration intensity (internal and external factors, described below).

3.4.2 Frequency shifts

The second hypothesis was that the vibration frequency decreases with fatigue. Frequency shifts occurred sparsely compared to the total number of local maxima. However, the identified frequency shifts did not show a systematic pattern (Figure 3.3.2), leading to the rejection of a general frequency shift with fatigue. A decrease in frequency was hypothesized because of a likely tension release of the triceps surae muscle. However, even in the presence of the proposed tension release by changed ankle kinematics (Natrup, 1997), the muscle can influence its stiffness by shortening and/or altering its activation level, which potentially could increase the natural frequency. Moreover, an earlier occurrence of the impact peak of the vertical ground reaction force was previously shown (Christina et al., 2001). This impact peak represents the main input frequency to the skeletal system just after heel-strike. An earlier impact peak corresponds to an increased input frequency and would suggest an increase of the soft tissue vibration frequency.

The mechanical properties of the muscle depend on subject specific muscle fat, connective tissue, vascular components, coupling between tissues, muscle length and mass, muscle shortening velocity and/or relative contribution of different motor units (Epstein and Herzog, 1998; Van den Bogert et al., 1998). The roles of those internal factors on tissue vibration characteristics cannot be isolated and limit the analysis together with external limitation factors such as unknown changes in kinematic and kinetic variables which represent changes in running technique. It was assumed that the subjects' lower limb muscles were fatigued by the end of the run; however this was not directly monitored.

The results of this study showed that with fatigue the soft tissue compartment of the triceps surae vibrated at increased amplitudes without changing the frequency. Additionally, the results suggest that the vibrations after heel-strike lasted longer. Vibrations applied to human tissue can have positive or negative effects. Negative effects include reduction of motor unit firing rate, decrease in maximal muscle contraction force, decrease in nerve conduction velocity and reduction of peripheral blood circulation (Bongiovanni et al., 1990; Dupuis and Jansen, 1981; Gilioli et al., 1981). Positive effects include increased muscle strength and increased bone density (Verschueren et al., 2004). The results of this study showed that the vibrations due to fatigue do change and how they change. However, the results of this study do not provide an answer to the question whether or not these changes are positive or negative for the human locomotor system. The authors speculate that the increased vibration intensities with fatigue are related to a reduced protection of the soft tissue compartment. However, further studies should address this question specifically.

3.5 Conclusions

The findings of this study were that with fatigue a) vibration intensity in the axial and medio-lateral directions increased, b) maximum vibration intensity occurred later after heel-strike and c) no general frequency change was observed. It is suggested that increased vibration magnitudes act on the tissue at extended duration and that the vibration damping mechanism of muscle tuning may be reduced by fatigue. Selected vibration frequency and intensity analysis could be used in the future to identify runners that may be sensitive to vibration-induced injuries.

Chapter four: Influence of graduated full-body compression apparel on indicators of running performance and recovery

4.1 Introduction

Compression has long been used as a treatment for chronic vein insufficiency and as a post-surgical measure to prevent deep vein thrombosis (Agu et al., 1999). The therapeutic effect is based on improvements in blood flow volume and velocity, which aid venous return and prevent venous congestion at injury sites (Lawrence and Kakkar, 1980; Bochmann et al., 2005; Starkey, 1976; Agu et al., 1999). It is not surprising that this treatment, originally medical, was implemented to support athletes while recovering from post-exercise soreness, which may be considered as a mild form of injury (Kraemer et al., 2001).

It has been shown that compression sleeves promote the recovery from soreness after eccentric exercise (Kraemer et al., 2001) and additional studies provided compelling evidence for the recovery benefits induced by compression apparel (Kraemer et al., 2010, Jakeman et al., 2010; Duffield et al., 2008) if they were worn for at least several hours after exercise. Increased blood flow is closely related to oxygen demand (Andersen and Saltin, 1985) and to the removal of blood metabolites during exercise (Berry and McMurray, 1987), and may therefore positively affect exercise performance and recovery immediately after exercise. It was shown that compression apparel reduced blood lactate after high intensity exercise (Berry and McMurray, 1987), reduced oxygen uptake during submaximal exercise (Bringard et al., 2006) and improved muscle oxygenation during exercise (Agu et al., 2004; Scanlan et al., 2008). However, statistically significant improvements in aerobic/anaerobic thresholds, heart rate or performance time were not found (Scanlan et al., 2008; Ali et al., 2011; Sperlich et al., 2010; Berry and McMurray, 1987), which suggests that the shown physiological benefits of compression apparel during exercise may have little, if any, impact on direct measures of performance.

The potential benefits induced by compression apparel are assumed to depend on its capability to improve blood flow. Therefore, one way to further enhance the physiological benefits, and eventually direct performance measures (e.g. time trial), is to maximize the blood flow enhancing effect of compression apparel. The level of compression needed to maximize the blood flow (“optimal compression”) was studied extensively in static situations (Lawrence and

Kakkar, 1980, Bochmann et al., 2005) but only sporadically during low intensity exercises such as stationary stepping (Husni et al., 1970), tiptoeing (Christopoulos et al., 1987) and walking (Ibegbuna et al., 2003). Since the optimal compression levels during high-intensity dynamic exercises (such as running) are yet unknown, testing multiple levels of compression represents one way to circumvent this limitation.

Therefore, the goal of this study was to assess the effects of three different levels of compression on running speeds at estimated lactate thresholds during running. Specifically, the hypotheses to be tested were that compression apparel

H1 improves running speed at aerobic/anaerobic thresholds

H2 reduces maximal heart rate

H3 improves lactate recovery time

4.2 Methods

4.2.1 Subjects

Sixteen healthy, competitive male runners (age: 24.7 ± 4.3 yr; body mass: 74.4 ± 6.1 kg; height: 179.0 ± 5.6 cm; mean ± SD), each with a 10 km race time below 40 min (37.5 ± 2.2 min) were recruited for this study. All subjects completed and signed health screening questionnaires and gave their written informed consent to participate and the University of Calgary's Conjoint Health Research Ethics Board approved the study.

4.2.2 Compression apparel

Prototype whole-body compression suits (adidas AG, Herzogenaurach, Germany), were used for this study. Four different levels of compression, which were Control (Con), Low, Medium (Med) and High, were designed for the study. The different amounts of compression were achieved by changes in sleeve and tights circumference. The level of compression was highest at the distal endings of the upper and lower extremities and decreased with proximity to the trunk (graduated compression). The actual amount of compression applied to the body is affected by the individual limb circumferences of the subjects. The average applied compression for each limb and compression level was estimated for each subject, according to the adapted equation of hoop stress by a thin cylinder (Equation 4.2.1; Rao, 1999).

$$\boldsymbol{p = \frac{F}{w_0 \cdot t_0} \cdot \frac{2 \cdot \pi \cdot t_0}{l} \cdot 0.0075} \qquad \text{(Equation 4.2.1)}$$

where p is the applied pressure [mmHg], F is the force generated by the fabric upon stretch [N], w_0 is the width and t_0 the thickness of the fabric during the stretch [m], l is the circumference of the relaxed fabric at each given body part, and 0.0075 is a constant used to convert from Pascal to millimeters mercury.

Based on the above equation, the average compression exerted by the suits is shown in Table 4.2.1.

Table 4.2.1: Average (± SD) compression applied to the subjects' limbs for each tested suit condition. The applied compression was calculated at the level of maximal circumference of each limb.

	Applied compression [mmHg]			
	Control	Low	Medium	High
Upper arm	0.29 (0.53)	2.15 (1.11)	3.72 (1.06)	7.28 (1.37)
Forearm	0.08 (0.31)	5.47 (0.93)	7.53 (1.1)	9.25 (1.22)
Thigh	0.14 (0.26)	1.84 (0.41)	2.59 (0.44)	3.96 (0.52)
Calf	3.06 (1.03)	8.40 (1.59)	13.70 (3.06)	35.06 (10.73)

4.2.3 Protocol

The subjects were instructed to run on a treadmill (Desmo Pro Evo, Woodway, USA) at incremental speeds and 1 % inclination to simulate level outdoor running (Jones and Doust, 1996). After a 5 min warm-up at 8.9 km/h, the treadmill speed was increased to 9.7 km/h. The treadmill speed was further increased by 0.8 km/h in intervals of three minutes. Blood lactate (La^-) was measured by certified exercise physiologists by means of a lactate analyzer (Lactate

Pro, Arkray Inc. Kyoto, Japan) using capillary finger samples taken within the last 30 sec of each three min stage. Running speed was increased until the subject was not able to complete another full three min stage. After completion of the final stage, the subjects recovered by walking on the treadmill at 5.6 km/h for 15 min and La^- was sampled every three min. An additional sample was taken after 1.5 min after completion of the final stage. Heart rate was measured throughout the test using a heart rate monitor (S610i, Polar Electro, Finland). In total, the protocol was completed five times by every subject. The first session was always done in the control condition (Con-1) and the following four sessions (Con-2, Low, Med, High) in randomized order. Con-1 was used to assess final speed stage for each subject, which was kept constant for the following sessions, and to familiarize the subjects to the suit and the protocol. At least 24 h of rest was given between the tests which were usually completed at the same time of the day and within 2.5 weeks after starting the first testing session. Subjects wore their same personal shoes and were instructed to have the same nutrient intake between each testing session.

4.2.4 Data analysis

The lactate trace during the incremental running test was used to determine the running speed at estimates of the aerobic threshold (S_{AeT}), the anaerobic threshold (S_{AnT}), and the absolute maximal lactate at the final stage (La^-_{max}) (Skinner and Mclellan, 1980). Recovery after running was determined by measuring the duration needed for blood lactate to fall below 50 % of the maximum lactate level ($La^-_{50\%}$) reached during the final running stage. Maximal heart rate (HR_{max}) was measured at the final running stage. Lactate and heart rate data of each suit condition was normalized to each individual's data measured during the Con-1 session.

4.2.5 Statistical analysis

Differences in mean lactate and heart rate variables between the four conditions (Con-2, Low, Med, High) were determined by a one-way repeated measures analysis of variance (ANOVA). Mauchly's tests were used to satisfy the assumptions of sphericity, and if violated, the degrees of freedom for the F-test were adjusted using a Greenhouse-Geisser correction. If significant main effects were observed by the ANOVA (level of significance $\alpha = 0.05$), paired student's t-tests were used to identify which of the four conditions within a dataset were significantly different. The level of significance for the post-hoc student's t-tests was adjusted using a Holm-Bonferroni correction (Holm, 1979). Paired student's t-tests were used to test for a difference between Con-1 and Con-2. Statistical analyses were conducted using SPSS (version 19.0.0, IBM SPSS, USA) and Matlab (version 7.10.0.499, The MathWorks Inc., USA).

4.3 Results

Typical La^- response curves during the incremental running test for each session of one subject are shown in Figure 4.3.1. The average (± SD) running speeds reached during the familiarization session (Con-1) were 12.54 ± 0.84 km/h at S_{AeT} and 14.91 ± 0.79 km/h at S_{AnT}. Maximal lactate (La^-_{max}) was 7.71 ± 1.53 mmol/l and HR_{max} was 187.1 ± 5.02 bpm at the final stage. Time needed to clear 50 % of the lactate measured at the final stage ($La^-_{50\%}$) was 9.37 ± 1.49 min. Percentage changes relative to the Con-1 session are shown for each variable in Table 4.3.1. There were no significant differences between any of the four testing conditions for any of the measured variables. However, there were significant differences between Con-1 and Con-2 for La^-_{max} and HR_{max} and a trend for S_{AeT} ($p = 0.055$).

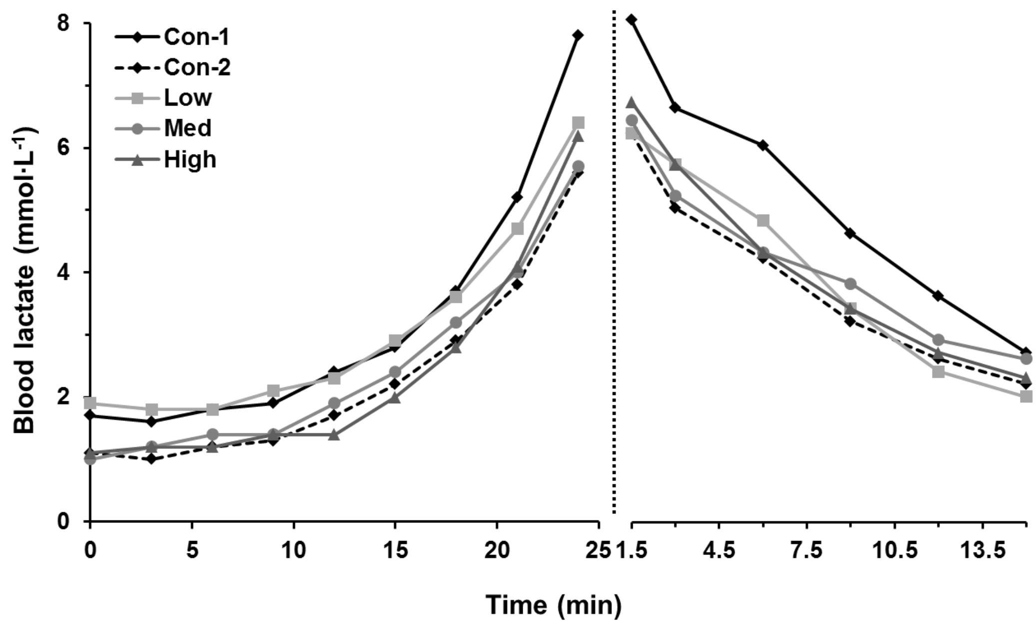

Figure 4.3.1: Typical lactate response curves from one subject and each testing condition. The left part of the figure shows lactate data collected during the incremental running test, during which running speed was increased every three minutes. The right part of the figure shows lactate data collected during the recovery phase at which the subjects were walking at a constant speed of 5.6 km/h. Wearing compression apparel did not significantly affect performance or recovery characteristics that may be inferred by changes in the lactate response curves.

Table 4.3.1: Average ± SD percentage changes relative to Con-1 are shown for S_{AeT} (speed at aerobic threshold), S_{AnT} (speed at anaerobic threshold), La^{-}_{max} (maximal lactate at the final running stage), $La^{-}_{50\%}$ (lactate recovery time) and HR_{max} (maximal heart rate) for each condition. * indicates a significant difference between Con-1 and Con-2. No significant differences existed between conditions Con-2, Low, Med and High.

	Compression Suit			
	Con-2 [%Δ]	Low [%Δ]	Medium [%Δ]	High [%Δ]
S_{AeT}	+1.60 ± 3.04	+1.05 ± 4.84	+1.11 ± 4.25	+3.29 ± 4.68
S_{AnT}	+0.55 ± 2.09	-0.22 ± 2.30	+1.36 ± 4.78	+0.61 ± 2.75
La^{-}_{max}	-14.10 ± 13.19*	-7.70 ± 15.25	-8.44 ± 14.00	-8.12 ± 18.85
$La^{-}_{50\%}$	-4.24 ± 16.85	-4.74 ± 20.61	+0.33 ± 16.74	-7.80 ± 13.46
HR_{max}	-2.57 ± 2.31*	-2.53 ± 2.11	-2.61 ± 2.47	-2.23 ± 2.30

4.4 Discussion

Running speeds at the estimates of the aerobic and anaerobic thresholds trended to be higher in most conditions when using compression apparel relative to Con-1 (Table 4.3.1). However, the differences were not significant for any of the compressive suits and the same trend was also observed for the Con-2 condition, leading to the rejection of the first hypothesis. The trend to improved thresholds in the Con-2 condition indicates a potential learning effect to the study protocol, rather than a potential benefit induced by compression apparel. Maximal blood lactate at the final running stage decreased significantly from Con-1 to Con-2, which may be the result of the test familiarization, and no significant differences between the compression suits and the Con-2 condition were found. This finding is similar to previous research that showed no changes in blood lactate during prolonged high-intensity intermittent running (Sear et al., 2010), running to exhaustion (Varela-Sanz et al., 2011) or intensity/time controlled running (Ali et al., 2010). To the best of our knowledge, only one study showed significantly improved aerobic and anaerobic threshold running speeds when using compression apparel by analyzing effect sizes, however, the effect sizes were small (d = 0.28 for S_{AeT}; d = 0.22 for S_{AnT}) (Kemmler et al., 2009).

Lactate recovery time was not significantly shorter when wearing compression apparel compared to a control condition. Similar to the results of the lactate thresholds, a trend for improvement in recovery time was found, however, the same trend was present for the Con-2 condition. Therefore, the hypothesis that compression improves recovery time in terms of lactate clearance needs to be rejected. The current findings agree with previous studies, which typically did not find improved lactate clearance after exercise using compression apparel (Berry et al., 1990; Duffield et al., 2008; Burden and Glaister, 2012; Rimaud et al., 2010). Only two studies found improved lactate recovery characteristics. One of them is probably the earliest study about compression apparel and its effect on lactate clearance showed a significantly decreased lactate level 15 min after completing a high intensity running test (Berry and McMurray, 1987). The other study showed decreased lactate during the recovery phase after intensive cycling in elderly subjects (Chatard et al., 2004). However, it was speculated that the reduced lactate measured during recovery was based on a reduced diffusion of lactate from the muscular bed rather than enhanced lactate removal (Berry and McMurray, 1987).

Maximal heart rate was significantly lower in the Con-2 condition compared to the Con-1 condition but no significant differences were found between Con-2 and any of the compression conditions. Although it is known that heart rate can be affected by factors such as psychological stress and nutrition, the relatively consistent decrease in maximal heart rate in all conditions (Table 4.3.1) does not suggest a potential benefit by compression apparel which agrees with previous studies (Ali et al., 2011; Duffield et al., 2008; Higgins et al., 2009; Sear et al., 2010).

All variables tested in this study showed a trend in the speculated direction in most of the tested conditions, however, the same trend was also consistently observed for the second control condition. This study used compression suits to increase blood flow volume, which is needed to potentially elevate muscle oxygenation and facilitate metabolic by-product removal. Two concepts were employed to achieve this goal. Firstly, a full-body design was chosen to maximize the area of compressed muscle tissue of the body, since the muscles of the whole body are involved in energy metabolism. Secondly, different levels of compression were used as it has been shown that the level of compression determines the increase in blood flow volume (Lawrence and Kakkar, 1980; Bochmann et al., 2005). Despite the approach to improve blood flow in two different ways, no effects on running speed at the estimates of aerobic and anaerobic thresholds were found in this study.

It may be that as a result of compression apparel blood flow volume is not actually improved during exercise. Most studies that investigated the effect of compression on blood flow velocity and volume were conducted at rest (Bochmann et al., 2005; Agu et al., 1999; Lawrence and Kakkar, 1980). Very few studies investigated the effects of compression on blood flow during dynamic exercises, such as walking (Ibegbuna et al., 2003) and no study has been done during moderate or high intensity exercise. Future studies need to verify whether or not compression improves blood flow during high intensity exercise as blood flow is a key element to potentially elicit performance benefits. The results also showed that the familiarization to the exercise protocol was stronger than any potential effect of compression apparel.

4.5 Conclusions

In conclusion, this study showed no benefit of compression suits on the running velocities at the aerobic and anaerobic lactate thresholds, maximal lactate at the final running stage and lactate recovery, independent of the level of compression. There are multiple potential reasons for the ineffectiveness of compression suits to improve running performance. Validating the capability of compression apparel to increase blood flow volume during moderate and high intensity exercise may be the most critical factor in contributing to improvements in endurance exercise performance.

4.6 Practical Implications

- This study clearly demonstrated the need for multiple familiarization sessions using the exact same testing protocol when a repeated measures design is used.
- Learning adaptations even occurred for well-trained athletes such as the ones used in the current study.

- No improvements in running performance related physiological variables were found when using full-body compression apparel and effects on blood flow during exercise need to be evaluated.

Chapter five: Vibration transmission to lower extremity soft tissues during whole-body vibration

5.1 Introduction

Whole-body vibration (WBV) training by means of standing on a vibrating platform is an increasingly popular way to strengthen muscles and bones, and has been reported to be associated with benefits in postural control (Bogaerts et al., 2007), paediatric rehabilitation (Semler et al., 2008), pain treatment (Rittweger et al., 2002), and muscle strength (Marin and Rhea, 2010a). With respect to muscular performance, the effectiveness of WBV training largely depends on the capability to elicit or intensify muscle activation. There are numerous proposed suggestions about the mechanisms responsible for the vibration induced muscle activation during WBV exercise, but are usually related to augmented excitatory input from muscle spindles exposed to vibration (Nordlund and Thorstensson, 2007). It has been shown that increasing vibration amplitude and frequency elevates muscle activity during WBV exercise (Hazell et al., 2007; Pollock et al., 2010; Ritzmann et al., 2013), with one known exception (Cardinale and Lim, 2003). This suggests that the strongest increase in muscle activation, and presumably the strongest training stimulus during WBV exercise, may be achieved at high amplitude – frequency combinations.

However, the exposure to high amplitude – frequency combinations may lead to discomfort and potential safety issues. Indeed, this concern emerged with the introduction of WBV platforms because it is known that vibrations transmitted to the spine and the head are risk factors for employees exposed to vibrations within their work environment (Bovenzi et al., 2002; Paddan and Griffin, 1998). Therefore, numerous studies have investigated the transmissibility of vibrations to the skeleton, i.e. the ratio of vibrations measured at the bony structures of the body relative to the vibrations at the WBV platform (Kiiski et al., 2008; Pel et al., 2009; Muir et al., 2013). In contrast, the transmissibility of vibrations to soft tissues packages of the lower limbs has largely been neglected, despite the reported short-term adverse effects during WBV exercise, such as erythema, edema and itching (Rittweger et al., 2000). A number of researchers have already investigated the potential effects of long-term exposure to vibrations in settings different from WBV training. For example, pathological changes of soft tissues due to long-term exposure to vibrations were shown in employees that use hand-held vibrating tools, including arteriosclerosis and lipid deposition, together with a loss of nerve fibers and myelin sheath (Takeuchi et al., 1986). Furthermore, invasive animal studies confirmed that vibrations were related to adverse effects in muscles (Necking et al., 1992; Necking et al., 1996), nerves (Govindaraju et al., 2006), and the vascular system (Govindaraju et al., 2006) at certain amplitude – frequency combinations and/or exposure durations. Although many studies showed the long-term benefits of WBV training (Marin and Rhea, 2010a; Nordlund and Thorstensson, 2007; Mester et al., 2006) the reported long-term (Takeuchi et al., 1986; Govindaraju et al., 2006) and short-term (Rittweger et al., 2000; Necking et al., 1996) adverse effects of vibrations should not be disregarded. The quantification of vibrations transmitted to soft tissue packages during WBV exercise is an initial, yet important step in order to eventually assess the potential risk of soft tissue injury when using WBV platforms as a long-term training intervention.

Therefore, this study was designed to quantify vibration frequency, amplitude and acceleration of soft tissue compartments of the lower limbs while standing on a WBV platform. In addition to transmissibility, absolute values of acceleration and movement amplitude of the soft tissue compartments will be reported to allow for comparisons to previous literature on vibration induced soft tissue damage.

5.2 Methods

5.2.1 Subjects

Sixteen healthy, physically active male participants (age: 26.7 ± 3.4 yrs; body mass: 78.6 ± 10.1 kg; height: 180.3 ± 8.3 cm; mean ± SD) were recruited for this study. All participants gave their written informed consent according to the institution's policy on research using human participants. The University of Calgary's Conjoint Health Research Ethics Board approved the study.

5.2.2 Protocol

The participants were instructed to stand freely on a side-altering (i.e., oscillation about a horizontal antero-posterior central axis) vibration platform (Galileo Advanced, Novotec, Germany), in an upright posture without shoes and socks. The participants' knees were flexed at 5-10° (0° corresponding to full knee extension) and the feet were placed parallel and 29.5 cm apart (heel-midlines) on the platform for each trial, corresponding to a vertical vibration amplitude of 2.5 mm (i.e., 5 mm peak-to-peak displacement). Adhesive tape with a rough surface was placed on the vibration platform to avoid skidding. The tested vibration frequencies were 10 Hz, 17 Hz, and 28 Hz. This combination of vibration frequencies and amplitude was chosen to encompass a range of commonly used vibration frequencies during WBV training (Marin and Rhea, 2010b; Nordlund and Thorstensson, 2007).

Each participant was familiarized to each vibration frequency for a period of 30 seconds. Following familiarization, the three frequency conditions were tested in a randomized order. Participants were exposed to the vibrations for 60 seconds per condition and 5 minutes of rest were given between conditions.

5.2.3 Data analysis

All participants were instrumented with two miniature tri-axial accelerometers (model: ADXL 78, range: ± 35 g, nominal frequency response: 0-400 Hz, weight: 2 grams, dimensions: 10/6/3 mm, Analog Devices USA). 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). The first accelerometer was secured to the most prominent bulk of the triceps surae muscle and the second one was placed on the quadriceps femoris muscle compartment, half way on a line between the trochanter major and the medial epicondyle of the femur. The accelerometers were fixed to the skin using stretch medical adhesive tape (Cover-Roll stretch, BSN Medical, Hamburg, Germany). The axes of the accelerometers were oriented to be (1) axial (i.e. parallel to the long axis of the tibia/femur), (2) medio-lateral (i.e. perpendicular to the tibia/femur in the frontal plane) and (3) antero-posterior (i.e. normal to the skin surface). A third tri-axial accelerometer was fixed on the platform, in line with the heel-midlines of the participants' feet to identify the exact vibration input parameters of the platform. All accelerometer data were recorded at 2400 Hz using a 12-bit data acquisition system (National Instruments, USA).

The raw acceleration data were filtered using a wavelet band-pass filter with cut-off frequencies between 6 Hz and 90 Hz. In order to account for balance corrections by the participants upon vibration start, only seconds 15 to 45 within the 60 second trial were used for analysis.

During WBV exercise, the muscles are simultaneously exposed to accelerations in three dimensions. Therefore, the resultant acceleration vector was calculated as the sum of the three individual acceleration vectors acquired by the tri-axial accelerometer according to the three-dimensional form of the Pythagorean Theorem. Afterwards, the 30 second data segment containing the resultant acceleration vector was split into 60 windows of 500 ms each. Finally, peak acceleration was quantified by detecting the maximum acceleration within each window, so that 60 peak acceleration values per condition were quantified, and the mean value was used for further analysis. Splitting the 30 second data segment into 60 individual segments eliminated the chance of reporting outliers. The results of all participants were then pooled according to the tested condition.

The acceleration data of the individual vibration axes were transformed into amplitude [mm] by double integration of the acceleration data in frequency space (Mallat, 1999). Accuracy of this method was tested using a custom model simulating three overlaid sinusoids with randomized amplitudes, frequencies and phase shifts. The error of the calculated amplitude from 1000 model iterations was 0.1±1.65 % (mean ± SD) of the known maximal input amplitude. The peak amplitude was then calculated analogue to the peak acceleration. The transmissibility of acceleration and vibration amplitude was calculated by dividing the values measured at the muscles by the ones measured at the platform for each participant and condition.

The frequency content of the entire 30 second data segment was calculated from the vibration amplitude signal by means of a Fast Fourier Transform (FFT; 0.033 Hz resolution) and determined individually for each vibration axis. The assessment of whether the input frequency by the platform was also the dominant frequency of the muscles was done in three steps. First, the exact input frequency of the platform was determined by using the FFT. Second, an analysis window was defined by the identified platform frequency ± 1 Hz. Third, the defined analysis window was applied to each muscle power spectrum and the maximal power within this window was compared to the maximal power outside of it. If the maximal power within the defined window was higher than any power outside the window, the muscle was classified as being dominantly vibrating at the same frequency as the vibration platform.

5.2.4 Statistical analysis

For statistical analysis, the primary focus of this study was to compare the peak acceleration and peak amplitude between the vibration platform and the two tested muscles individually for each frequency. Independent Student's t-tests with a level of significance of $\alpha = 0.05$ were sufficient, since multiple comparisons were not conducted. Vibration frequency was analyzed as follows: In order to test if the maximal power within the defined analysis window (as described above) was higher than any power outside of it, a Wilcoxon signed-rank test was used to account for the non-normal data distribution for this variable. All statistical analyses were conducted using SPSS (version 19.0.0, IBM SPSS, USA) and the level of significance was $\alpha = 0.05$.

5.3 Results

Raw acceleration data of one participant is shown in Figure 5.3.1. For each vibration frequency, the resultant peak acceleration and peak amplitude measured at the triceps surae muscle were significantly higher than the input peak acceleration and peak amplitude of the platform. Peak acceleration and peak amplitude at the quadriceps were significantly higher at 10 Hz, not significantly different at 17 Hz, and significantly lower at the 28 Hz condition when compared to the platform (Figure 5.3.2; Figure 5.3.3).

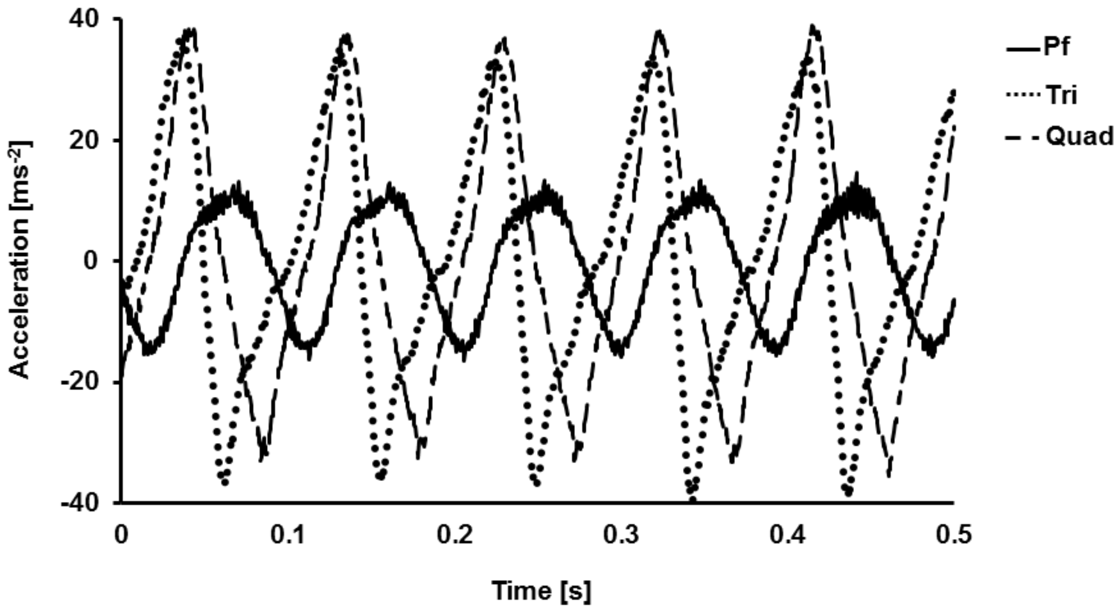

Figure 5.3.1: Raw acceleration data in axial direction measured at the platform (solid line), the triceps surae (dotted), and the quadriceps muscle compartments (dashed) at a frequency of 10 Hz produced by the vibration platform. At this frequency, the acceleration at the soft tissues is significantly higher than the acceleration produced by the vibration platform.

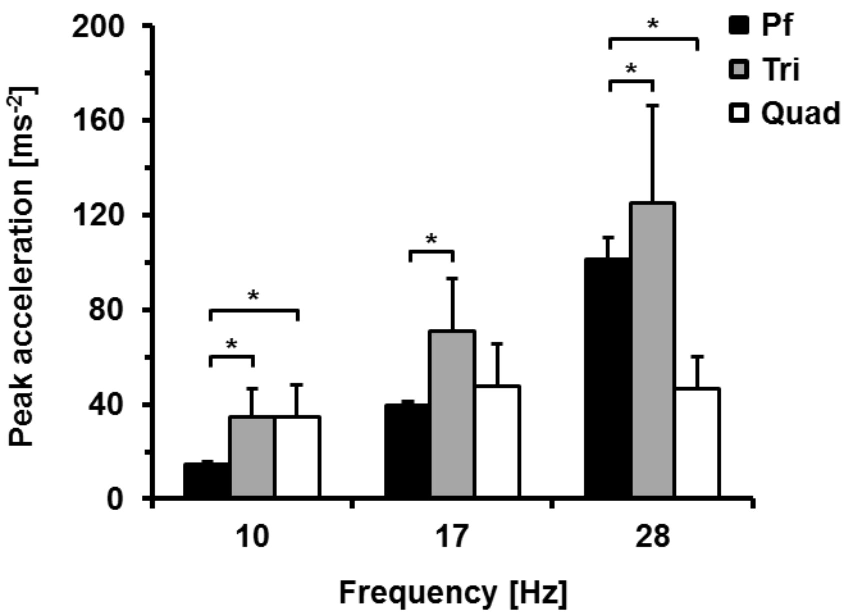

Figure 5.3.2: Resultant peak acceleration (mean ± SD) for three input frequencies, measured at the WBV platform (Pf, black), the triceps surae (Tri, gray), and the quadriceps (Quad, white) from all participants. * denotes significant differences with $p < 0.05$.

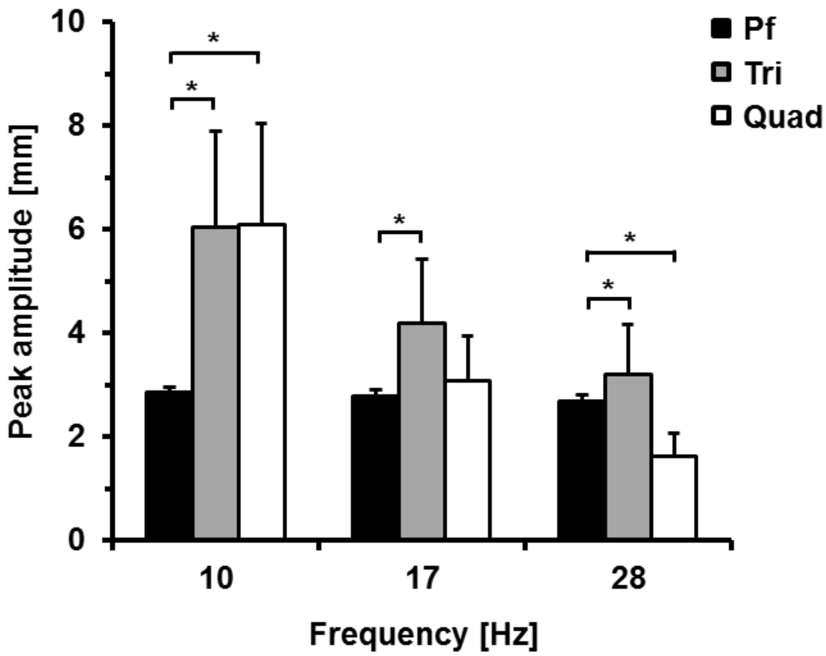

Figure 5.3.3: Resultant peak vibration amplitude (mean ± SD) for three input frequencies, measured at the WBV platform (Pf, black), the triceps surae (Tri, gray), and the quadriceps (Quad, white) from all participants. * denotes significant differences with $p < 0.05$.

Transmissibility of the peak acceleration and peak amplitude from the platform to the muscles decreased as input frequency increased (Table 5.3.1). The dominant frequency of the two muscle groups for all measured vibration directions and conditions was consistent with the vibration platform frequency (Table 5.3.1).

Table 5.3.1 Transmissibility (i.e., output at the muscle divided by input by the platform) of resultant peak acceleration and peak amplitude for the triceps surae and quadriceps muscles for three WBV platform frequencies. Values in brackets indicate standard deviation. Values for the "Dominant frequency" section indicate how many out of the 16 participants had the muscles predominantly vibrate at the frequency given by the WBV platform. * denotes that the dominant frequency at the muscle is equal to the frequency of the platform with $p < 0.05$.

Muscle	**Platform frequency**	**Transmissibility**		**Dominant frequency**		
		Peak acceleration	Peak amplitude	axial	medio-lateral	antero-posterior
Triceps surae	10 Hz	2.33 (0.91)	2.10 (0.68)	16/16*	16/16*	16/16*
	17 Hz	1.88 (0.56)	1.54 (0.42)	16/16*	16/16*	16/16*
	28 Hz	1.23 (0.41)	1.17 (0.33)	16/16*	16/16*	15/16*
Quadriceps femoris	10 Hz	2.33 (1.02)	2.09 (0.71)	16/16*	16/16*	16/16*
	17 Hz	1.17 (0.45)	1.10 (0.32)	16/16*	16/16*	16/16*
	28 Hz	0.46 (0.14)	0.60 (0.17)	16/16*	15/16*	15/16*

5.4 Discussion

Three main findings can be deducted from the results of this study. First, the triceps surae and quadriceps femoris vibrate at substantially larger peak accelerations and peak amplitudes compared to the input provided through the WBV platform at the 10 Hz condition (Figures 5.3.2 and 5.3.3). Second, the vibration transmissibility of the peak acceleration and peak amplitude decreases as the input frequency through the WBV platform increases and third, the muscles predominantly vibrate at the same frequency as the vibration platform (Table 5.3.1).

One possible explanation for the high transmissibility of acceleration and vibration amplitude at low frequencies may be related to soft tissue resonance. If the excitation frequency is close to the natural frequency of the soft tissue package, then resonance occurs, leading to an amplification of the input amplitude (transmissibility > 1). The natural frequency of the relaxed triceps surae is between 10 Hz and 20 Hz and between 8 Hz and 11 Hz for the quadriceps (Wakeling and Nigg, 2001). This may be the reason for the high transmissibility of acceleration and vibration amplitude at the two lower input frequencies, which were at 10 Hz and 17 Hz. If the input frequency diverges from the tissue's natural frequency (corresponding to the 28 Hz condition), resonance diminishes and thus may explain the reduced transmissibility at higher platform frequencies.

Next to the natural frequency, the transmissibility of vibrations is further affected by the damping properties of muscle tissue. Therefore, another plausible explanation for the results is the finding that the damping coefficient of muscles increases together with muscle activation (Wakeling and Nigg, 2001). Muscle activation levels were shown to increase with ascending vibration frequency (Hazell et al., 2007; Pollock et al., 2010; Ritzmann et al., 2013). Consequently, the decreased transmissibility of vibrations at higher input frequencies may also be explained by the increased damping of a more activated muscle (Wakeling and Nigg, 2001).

It is likely that the two described explanations are related to each other, since an expected change in muscle activation affects both, the damping properties (Wakeling and Nigg, 2001) and the intramuscular pressure of a muscle (Aratow et al., 1993). A change in the intramuscular pressure modulates the stiffness of a muscle (Aratow et al., 1993), which directly influences any mechanical system's natural frequency.

Further, less obvious influences that affect vibration transmissibility are possibly associated with the interface between the feet and the vibration platform. Since the feet are not rigidly attached to the platform, they may not perfectly follow the sinusoidal waves of the platform (Rittweger, 2010), leading to air-borne phases of the feet that may affect the transmission of vibrations to the feet. Nonetheless, in this study the vibration frequency of the WBV platform was the dominant frequency measured at both soft tissue packages in any condition and vibration axis. This means that the tissue was mainly vibrating at the frequency provided by the platform, whilst acceleration and amplitude at the tissue diverged significantly from that of the platform. It may be speculated that air-borne phases of the feet only start to occur at higher platform amplitude – frequency combinations than the ones used in the current study.

A limitation of this study is that the knee angle of the participants was only visually controlled during each testing condition. Future studies may use live feedback goniometers placed on ankle, knee and hip joints to ensure an equal posture for each testing conditions. Also, a goniometer at the back and trunk region may be employed, as upper body movements could potentially shift the participants' centre of mass. Thus, a person could potentially lean forward during less comfortable (i.e., high frequency) conditions and put more weight on the forefoot. This would engage the triceps surae muscles more in the damping process compared to a lesser contribution when in mid-foot stance (Rittweger, 2010). If this was the case in the current study, it may have contributed to the low transmissibility of vibrations to the quadriceps at high vibration frequencies.

To the best of our knowledge, there is only one study that previously quantified the transmissibility of vibrations to soft tissues on a WBV platform (Cook et al., 2011). The results of the current study agree with the previous findings in terms of decreasing transmissibility of acceleration with increasing vibration frequency. Other comparisons are limited due to differences in the type of platform used (side-altering vibration in the current study vs. synchronous), body posture (less knee bending in current study) and analysis (vector sum of all three axes versus individual axes). Those methodological differences may also explain the substantially larger maximal accelerations in the current study measured at both of the muscles, which demonstrates dependency of platform type/setting and posture on vibration transmissibility.

In order to assess the potential risk of WBV, the International Organization for Standardization defined limits of vibration exposure duration and intensity that people can be expected to safely tolerate within the ISO-2631 (ISO, 1997). While this standard may be applicable to health concerns for persons that are experiencing WBV in a seated position, the effects of WBV on health for vibrations transmitted through a standing posture are not known (ISO, 1997). Although this standard was used to speculate about the risks of WBV training (Muir et al., 2013), the applicability of this standard to WBV training and the effects on lower limb soft tissues is strongly limited, as was discussed previously (Rittweger, 2010). Another approach to assess the potential risk of WBV training for soft tissues may be the comparison to animal studies that have shown soft tissue injury induced by vibrations. For example, muscle fibre swelling (Necking et al., 1996) and necrosis (Necking et al., 1992) were shown in rat hind limbs after vibrating the limb at 45 ms^{-2} for five hours per day for two and five consecutive days, respectively. Also, endothelial cells of rat tail arteries showed signs of injury after being exposed to a single session of four hours of continuous vibrations at 49 ms^{-2} (Curry et al., 2002). The measured acceleration acting on the triceps surae at 28 Hz in the current study was about 160 % higher than the acceleration acting on the rat tail (125 ms^{-2} vs. 49 ms^{-2}), however, WBV exercise does typically not exceed 30 minutes (Marin and Rhea, 2010b) and thus the vibration dose is limited. Clearly, the anatomical differences between a human calf and a rat tail, together with the differences in the study protocols are substantial, and inferences on injury risk for humans would be speculative. Therefore, neither the ISO standard nor the comparison to animal studies seem appropriate to speculate about potential long-term injury risks of WBV training. Future studies need to evaluate the effects of vibrations on soft tissues using vibration properties and exposure durations that are relevant for WBV training. The current data, in combination with previous work (Cook et al., 2011), could be used as a reference to set proper input parameters for more realistic animal or computational muscle models with the goal to assess injury risk of muscles, nerves and the vascular system in humans. This may be an important step, since many vibration-related soft tissue diseases or injuries only occur after exposure to vibrations over years (Takeuchi et al., 1986). The assessment of transmissibility was a critical step needed to eventually address the potential injury risk for soft tissues by WBV training, but clearly more work needs to be done to provide sufficient WBV training exposure regulations.

5.5 Conclusions

The transfer of vibrations to lower extremity soft tissue compartments was strongly dependent on the platform frequency and the particular muscle. The study showed that the soft tissues were exposed to substantially higher accelerations and vibration amplitudes than the vibration platform settings may have suggested, particularly for the triceps surae. Inferences on potential injury risk for humans during WBV training seem inadequate at this time, as neither the ISO regulations nor the comparison to animal studies are directly applicable. The acceleration acting on the triceps surae was higher than levels related to soft tissue injury in animal studies, which emphasizes the need to study long-term effects of WBV training. Animal or computational muscle models may be used to evaluate potentially unwanted side effects of WBV training and the current data may serve as a reference for proper model input parameters.

5.6 Practical Implications

- The acceleration experienced by the lower limb muscles can substantially exceed the acceleration provided by the vibration platform.
- Compared to the acceleration used in animal studies, the accelerations experienced by the lower limb muscles during WBV exercise are higher than accelerations related to injury in animals, but the long-term effects for humans are unknown.
- The current results provide the basis for more realistic animal and computational muscle models.

Chapter six: Local metabolic rate during whole-body vibration

6.1 Introduction

The number of applications incorporating whole-body vibration (WBV) training has increased dramatically over the last two decades. Researchers have investigated the effects of WBV interventions on bone quality (Flieger et al., 1998; Gusi et al., 2006; Rubin et al., 2001; Verschueren et al., 2004; von Stengel et al., 2011), muscle performance and training adaptation (Cochrane et al., 2004; Delecluse et al., 2003; Roelants et al., 2004; Torvinen et al., 2002), responses of the neuromuscular system (Armstrong et al., 2008; Cardinale and Bosco, 2003; Cardinale and Lim, 2003; Kipp et al., 2011; Ritzmann et al., 2011), hormonal responses (Bosco et al., 2000; Erskine et al., 2007; Kvorning et al., 2006), as well as diseases such as Parkinson's (Ebersbach et al., 2008; Haas et al., 2006), and multiple sclerosis (Jackson et al., 2008; Schuhfried et al., 2005). Considering this broad scope, it is not surprising that training protocols incorporating WBV differ substantially. Nevertheless, numerous long-term WBV studies with similar research questions, albeit using different training protocols and different vibration platform settings, have been compared in the literature (Rehn et al., 2007; Rittweger, 2010). As a consequence, the validity and conclusions drawn within those reviews may be uncertain, given the importance of platform protocols and settings.

A notable reason for the large variety of platform settings that are currently used in long-term studies may be the apparent absence of WBV training standards. One approach to address this problem is to gain a thorough understanding of acute physiological effects of WBV. By understanding the acute effects, one would have more information about the training intensity that is imposed by WBV on the body, so that the appropriate vibration settings needed to trigger training effects can be determined. However, with respect to muscle tissue, it is not known which training intensities, in the form of the muscles' metabolic demand, correspond to the diverse WBV platform settings that are currently used.

The first attempts to understand the acute metabolic demand of the body when using WBV as an exercise intervention were made by systematically changing vibration platform frequencies (Rittweger et al., 2000), as well as amplitudes and carried loads (Rittweger et al., 2002). Whole-body metabolic rate was estimated from whole-body oxygen uptake, which was then used to assess the training intensity imposed on the body during WBV (Rittweger et al., 2000; Rittweger et al., 2002). Although whole-body training intensity is a valuable assessment, it yields quite general and undetailed information about the training intensity for individual muscles, since it is not possible to determine an individual muscle's contribution and change in metabolic rate. The metabolic rate of individual leg muscles is likely to be vastly different during WBV if it was compared to the metabolic rate of the entire body. In order to understand the effects of WBV on muscle tissue more specifically, it is imperative to collect metabolic information about individual muscles and muscles groups locally to determine which muscles are being exercised and at what intensity.

Recent technological advances have allowed for the use of near infrared spectroscopy (NIRS) in quantifying the oxygen demand and thus metabolic rate of individual muscles. Some studies investigated the oxygen saturation (OS) during vibrations versus a non-vibration control condition. For instance, NIRS was used to study OS of the vastus lateralis muscle while squatting on a WBV device (Yamada et al., 2005) and it was shown that the OS was lower when vibrations were superimposed to the exercise compared to the control condition. Furthermore, changes in OS were tested in two leg muscles while standing on a WBV platform at different vibration frequencies (Cardinale et al., 2007). Arguably, quantifying local OS alone does not reflect the actual metabolic demand of these muscles. Oxygen saturation merely represents the relation between oxygen demand (oxygen used by the muscles) and supply (oxygen supply by the blood stream) (Boushel et al., 2001). In order to exclusively quantify the oxygen used by individual muscles in the leg, oxygen supply via the blood stream must be stopped, for example by inducing limb ischemia (Cheatle et al., 1991). The quantification of oxygen demand during ischemia then allows for the determination of the muscle's metabolic rate (Boushel et al., 2001; Hamaoka et al., 1996; Homma et al., 1996; Praagman et al., 2003), which was the goal of this study. This method was already used for investigating oxygen demand of the gastrocnemius medialis during vibration exercise (Coza et al., 2011; Zange et al., 2009). Therefore, the goal of this study was to quantify the metabolic rate of two different leg muscles during WBV at different vibration frequencies during limb ischemia. The muscles of interest were the gastrocnemius medialis (GM) and the vastus lateralis (VL), as they are important for daily activities like walking, lifting and standing.

It was hypothesized that (H1) the metabolic rate of each muscle would be higher during vibrations compared to a non-vibration control condition and that (H2) the metabolic rate would increase with higher vibration frequencies.

6.2 Methods

6.2.1 Subjects

Fourteen healthy, physically active male volunteers (age: 26.8 ± 3.4 yr; body mass: 74.9 ± 9.5 kg; height: 178.8 ± 8.1 cm; mean ± SD) were recruited for this study. All subjects gave their written informed consent to participate and the University of Calgary's health research ethics board approved the study.

6.2.2 Vibration platform setup

The subjects were instructed to stand freely and without shoes or socks on a side-altering vibration platform (Galileo Advanced, Novotec, Germany) in an upright posture. The subjects' knees were flexed at 5-10° (0° corresponding to full knee extension) and the flexion angles were controlled by means of templates given before each measurement (Ritzmann et al., 2011). The feet were kept in a constant position 29.5 cm apart (heel-midlines) on the platform for each trial, corresponding to a 5 mm peak-to-peak vertical vibration displacement. The vibration frequencies tested were 0 Hz (control), 10 Hz, 17 Hz, and 28 Hz. This combination of vibration frequencies and amplitude was chosen to encompass a range of commonly used vibration frequencies during WBV training and to closely correspond with parameters from previous systematic studies (Cardinale et al., 2007; Rittweger et al., 2002). Adhesive tape with a rough surface was placed on the vibration platform to avoid skidding.

6.2.3 Near-infrared measurements

Muscle metabolic rates were estimated using data from near-infrared spectroscopy, obtained with a NIRO 200-NX (Hamamatsu Photonics, Japan). After shaving and cleaning the skin with alcohol swabs, the spectrometer's optodes were placed over the belly of the GM and VL muscles. The distance between the emitter and detector was 4 cm. Baseline data for changes in oxyhemoglobin (O_2Hb) and estimated total hemoglobin index (nTHI) were collected with a sampling frequency of 20 Hz while the subjects were standing on the platform. Ischemia was induced by arterial occlusion using a pneumatic blood pressure cuff (width: 15.5 cm; Caliber large, Mabis/DMI, IL) that was placed at the level of the gluteal tuberosity of the femur.

6.2.4 Protocol

Once the setup was completed, the participants were exposed to each vibration frequency for a period of 30 seconds to familiarize them with the procedure. Following familiarization, the four conditions were tested in a randomized order. Before vibration was commenced, the oxygen saturation of the muscles was monitored carefully to confirm that a steady state was reached. Once steady state oxygen saturation was observed, the pressure cuff was rapidly (< 10 sec) inflated to a pressure of 300 mmHg. After inflation, the vibration platform was turned on for 60 sec. After this time, the vibrations stopped and the pressure cuff was released. At least 5 minutes of rest were given to the subjects between conditions to allow blood flow to reach homeostasis.

6.2.5 Data analysis

Muscle metabolic rate in this study was defined by the O_2Hb utilization rate during ischemia (Boushel et al., 1998; Hamaoka et al., 1996; Praagman et al., 2003), and was determined by performing a regression analysis of the linear portion of the O_2Hb slope over a 30 second period for each tested vibration frequency (Figure 6.2.1). Although the vibrations lasted for 60 sec, only the center 30 sec of that window were analysed due to occasionally observed nonlinear changes of O_2Hb in the beginning and end of the trials. The O_2Hb readings are affected by changes in blood volume (BV) and therefore BV needs to be as constant as possible. The nTHI serves as an indicator of BV and was used to quantify the maximal change in BV as a percentage of the initial BV during the (30 sec) data analysis window.

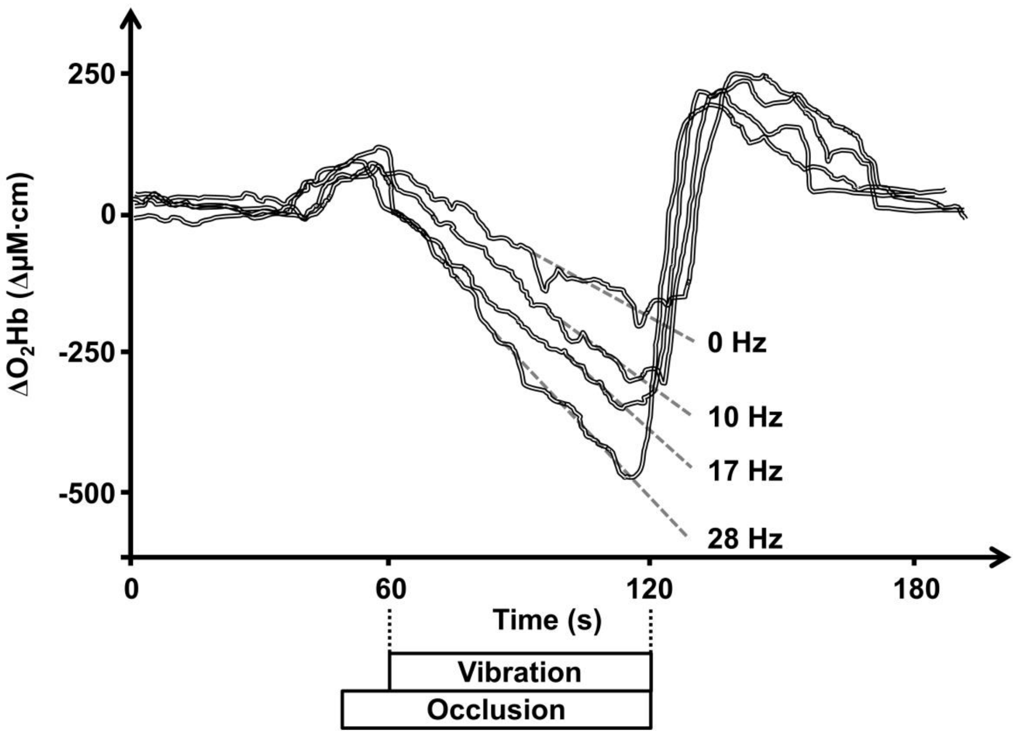

Figure 6.2.1: Changes in vastus lateralis O_2Hb concentration before (0-60 sec), during (60-120 sec) and after (120-180 sec) the intervention at four different trials with different vibration frequencies for one subject. The dashed lines are a visual aid to represent the muscle's metabolic rate.

6.2.6 Statistical analysis

The metabolic rates of the GM and VL muscles were analyzed independently since interactions between the two muscles were not the focus of this study. Thus, a one-way repeated measures analysis of variance (ANOVA) was used to analyse the two datasets. Mauchly's tests were used to satisfy the assumptions of sphericity, and where violated, the degrees of freedom for the F-test were adjusted using a Greenhouse-Geisser correction. If significant main effects were observed by the ANOVA (level of significance $\alpha = 0.05$), paired student's t-tests were used to identify which of the four conditions within a dataset were significantly different. The level of significance for the post-hoc student's t-tests was adjusted using a Bonferroni correction for a total of six comparisons per dataset, and was $\alpha = 0.0083$. In the case of significant post-hoc differences, effect sizes were calculated using Cohen's d procedure. All statistical analyses were conducted using SPSS (version 19.0.0, IBM SPSS, USA).

6.3 Results

The average changes in blood volume during the measurement period as indicated by nTHI were 6.51 ± 5.17 % for the GM and 4.15 ± 2.56 % for the VL. The metabolic rate of the GM and the VL at different vibration frequencies is shown in Figure 6.3.1. Statistical analysis revealed a significant main effect for the metabolic rate of the GM ($F_{(1.54, 52)} = 55.42$; $P < 0.01$) and the VL ($F_{(1.97, 52)} = 39.32$; $P < 0.01$). The following post-hoc analysis showed significant increases in mean GM and VL metabolic rate with each increase in vibration frequency (Table 6.3.1; Figure 6.3.1). Effect sizes were large (> 0.8) for the GM and inconsistent for the VL (0.42-2.28) for the post-hoc comparisons (Table 6.3.1).

Table 6.3.1: P-values with respect to H1 (comparison to control condition) and H2 (comparison to next lowest vibration frequency) and their respective effect sizes for each tested vibration frequency and muscle.

Muscle	Frequency [Hz]	P-value H1	Cohen's *d* for H1	P-value H2	Cohen's *d* for H2
Gastrocnemius medialis	0	-	-	-	-
	10	0.046	0.83	0.046	0.83
	17	< 0.01	1.85	0.048	1.03
	28	< 0.01	3.23	< 0.01	2.06
Vastus lateralis	0	-	-	-	-
	10	0.046	0.76	0.046	0.76
	17	< 0.01	1.41	0.032	0.42
	28	< 0.01	2.28	< 0.01	1.16

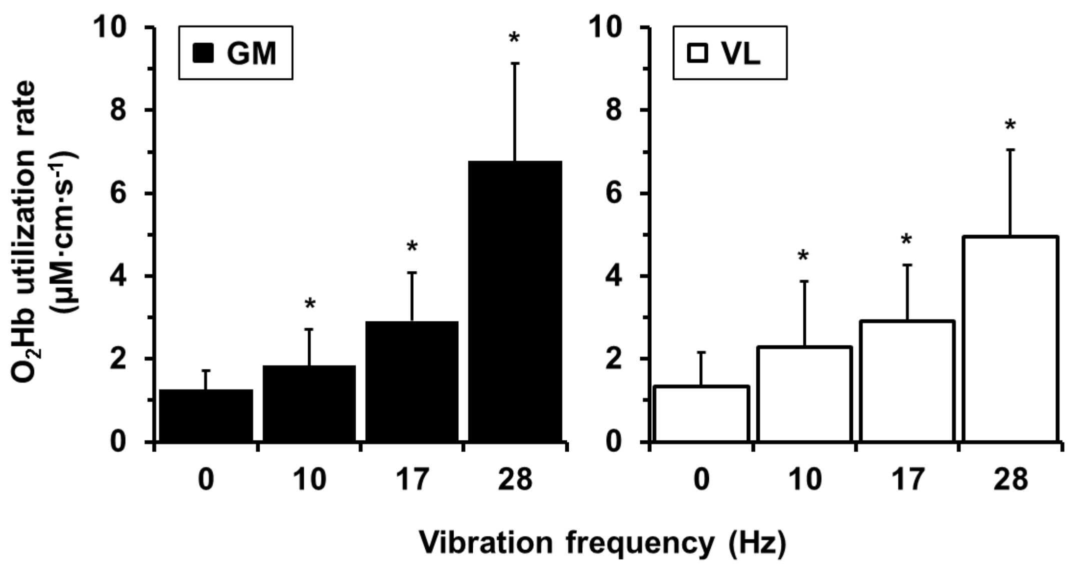

Figure 6.3.1: Average O_2Hb utilization rate of the gastrocnemius medialis (GM, black bars, mean ± SD) and vastus lateralis (VL, white bars) at four different vibration frequencies for all

subjects. Each increase in vibration frequency resulted in a significantly higher O_2Hb utilization rate for the GM and VL, respectively. * indicates significant differences ($P < 0.05$) to the next lowest frequency.

6.4 Discussion

To the best of our knowledge, this is the first study to investigate the local metabolic demand during WBV at different vibration frequencies. The key findings of this study were that both muscles had significantly higher metabolic rates during vibration exposure at all vibration frequencies, compared to the control condition (0 Hz), in support of H1. Furthermore, metabolic rates in both muscles increased significantly with each higher vibration frequency, supporting H2 (see Table 6.3.1 and Figure 6.3.1).

WBV platforms are currently being used for training, rehabilitation and even weight loss at gyms, sports clubs and clinics. In a study that investigated whole-body metabolic rate during stance on a vibration platform (Rittweger et al., 2002), it was demonstrated that WBV increased the metabolic rate to about 2.0 times resting metabolic rate or metabolic equivalents (METs, i.e., work metabolic rate divided by resting metabolic rate) at 18 Hz. They furthermore showed that the metabolic rate increased to 2.4 METs at 26 Hz and 2.8 METs at 34 Hz (Rittweger et al., 2002). However, according to the recommendations by the Centers for Disease Control and Prevention (CDC) and the American College of Sports Medicine (ACSM), every (US) adult should daily accumulate 30 minutes or more of moderate-intensity physical activity. Moderate-intensity is therein defined as exercising at 3 to 6 METs, corresponding to walking briskly at 4.8 – 6.4 km/h (Pate et al., 1995). This means that standing on a vibration platform may not exclusively fulfill the demands of moderate-intensity training for the whole body as, firstly, the minimum MET is not achieved and, secondly, most WBV training sessions take less than 30 minutes of net vibration time (Marin et al., 2010b).

In the current study, however, the local metabolic rate at the highest frequency (28 Hz) was on average 537 % (GM) and 369 % (VL) of the control metabolic rate. This means that on a local (muscular) level, WBV induced roughly 5.4 local metabolic equivalents (LMETs) for the GM and about 3.7 LMETs for the VL at 28 Hz. One may speculate that even higher LMETs could be observed in leg muscles that are involved in postural control (e.g., ankle stabilizing muscles), as WBV creates an unstable standing environment. As a consequence, WBV training may not seem to be an effective training when considering the muscles of the whole body, but may be sufficient to locally induce moderate-intensity training for some leg muscles. Specifically, improvements in muscle power (Marin and Rhea, 2010b) as well as in muscle strength (Marin and Rhea, 2010a) were shown in the literature and can be expected when using WBV platforms as a training intervention. This may be particularly relevant for people with balance or mobility issues due to age, disease or obesity since WBV training is still possible under such conditions.

However, the estimated metabolic equivalents for whole-body oxygen uptake (METs) may not be directly comparable to the previously discussed local metabolic equivalents (LMETs) since the work metabolic rate was not divided by the resting (sitting) metabolic rate, but rather the one during quiet standing. In addition, no comparative data for LMETs during typical movements such as walking or running exists, but should be quantified in future studies to allow for valid comparisons to the MET scale and to common training regimes. Also, the reported LMETs may vary, depending on the tested population's muscle fiber type compositions. It may be speculated that people with predominantly fast-twitch fibers consume different amounts of oxygen during WBV training compared to people with predominantly slow-twitch fibers for the same vibration settings due to different metabolic characteristics of certain fiber types. Furthermore, the values of LMETs are based on changes in O_2Hb, and O_2Hb is directly affected by changes in blood volume. Therefore, the need to report these readings has to be strongly emphasized and in the current study they were relatively small (4-7 % on average), as indicated by nTHI readings, but no comparative values are available from previous studies.

The question that remains is how the current findings can help to overcome a major difficulty when designing a WBV study, which is the large number of vibration platform parameters that can be adjusted. This difficulty becomes obvious when considering the parameters that usually need to be set, involving: (A) vibration amplitude; (B) vibration frequency; (C) vibration duration; (D) training session frequency; (E) posture or exercise specificity during vibration, and

(F) population. In addition, the metabolic demand and perceived comfort will also depend on the type of vibration platform used. It is known that side-altering platforms (as used in the current study) induce acceleration magnitudes twice as high when compared to synchronous vibration platforms, in conjunction with a smaller vibration dampening effect at the ankle joint and higher muscle activity (Pel et al., 2009; Ritzmann et al., 2013). Although the vibration type and platform parameters are a crucial factor when examining acute and long-term effects of WBV training, there are still no common recommendations for the most effective combinations. A long-term goal of WBV research should be to identify platform parameters that maximize the muscles' metabolic demand and are tolerable in order to establish standards and/or recommendations for the most effective vibration platform parameters. The current study was one step in this direction, but more muscles and conditions need to be tested systematically in the future. Once the acute effects are better understood, then more consistent, comparable and effective long-term WBV intervention studies may be possible. This includes all other observed physiological benefits of WBV training, including improvements in bone quality, postural control, skin and muscle perfusion, flexibility and pediatric rehabilitation (Rittweger, 2010).

6.5 Conclusions

In conclusion, this study showed an increase in local metabolic demand of two leg muscles with each increment in vibration frequency. On a local muscular level, WBV training may represent a significant training stimulus for particular muscles. Further systematic studies need to assess the local metabolic demand of muscles involved in WBV training to achieve and optimize long-term training benefits.

Chapter seven: Reduced elbow extension torque during vibrations

7.1 Introduction

Soft tissue vibrations occur when participating in activities involving impacts, such as racquet sports, running, skiing, and vibration platform workouts. Tissue vibrations trigger several physiological responses that have been studied in recent decades. Effects on bones (Gemne and Saraste, 1987; Verschueren et al., 2004; Gusi et al., 2006; Rubin et al., 2001), nerves (Goodwin et al., 1972; Bovenzi 1998; Burke et al., 1976; Govindaraju et al., 2006), muscles (Bosco et al., 1999; Mischi and Cardinale, 2009; Humphries et al., 2004; Issurin and Tenenbaum, 1999; Necking et al., 1996), and other biological and physiological variables (Rittweger, 2010; Kvorning et al., 2006) have been observed. It is generally agreed that the effects of vibrations on the locomotor system depend largely on the interplay between the vibration frequency, amplitude, exposure time, transmission method, input location, and the task performed by the subject (Cardinale and Wakeling, 2005; Issurin, 2005; Rittweger, 2010). A clear understanding of the effects during vibration exposure on the body is needed to potentially minimize fatigue, reduce injury risk or to determine the most effective vibration settings to fine-tune vibration training devices. Understanding muscle behaviour during vibration exposure is critical, as muscles control movement, posture, and loading of joints and bones. Any modification of contraction properties due to vibrations could have a significant impact on muscle-controlled tasks. Many studies investigating the effects of vibrations on muscles used direct vibratory muscle stimulation of the muscle belly or tendon (e.g., Eklund and Hagbarth, 1996; Burke et al., 1976; Humphries et al., 2004). Activities like racquet sports, cycling or operating vibrating tools (lawn mower, jackhammer) transmit vibrations directly to the hand and forearm, and previous research may apply to the muscles in this area. However, vibrations are also transmitted indirectly from the forearm via the elbow joint to the upper arm muscles, which may be the muscles controlling limb movement or position while the forearm is being vibrated. Therefore, the goal of this study was to identify the acute effect of vibrations on elbow extension torque by exposing the forearm to vibrations similar to those occurring during daily activities. It is known that vibrations can increase muscle activity and induce involuntary contractions by means of muscle spindle stimulation, which is elicited by a form of the spinal stretch reflex, called tonic vibration reflex (TVR) (Eklund and Hagbarth, 1966). It has been shown that the intensity of the TVR increases when the vibration frequency (Eklund and Hagbarth, 1966; Gottlieb and Agarwal, 1979; Dietz et al., 1981) and/or the initial length of the receptor-bearing muscle increases (Burke et al., 1976).

Since vibration frequency and initial muscle length affect the TVR, we hypothesized that (H1) higher vibration frequencies and (H2) lower elbow flexion angles (thereby increasing the initial muscle length of the force-generating muscles), will increase the maximal torque and the electromyographic activity of the elbow extensor muscles.

The outcome of this study will contribute to understand the potential advantages/disadvantages of vibrations during athletic competitions, training and occupational settings.

7.2 Methods

7.2.1 Subjects

Fifteen right-handed, healthy, and recreationally active female subjects with no specific upper extremity weight or endurance training (age 25.9 ± 2.8 yrs; weight 62.6 ± 6.3 kg; height 168.6 ± 5.7 cm; mean ± SD) were recruited for this study. All subjects gave their informed, written, consent to participate, and the University of Calgary's health research ethics board approved the study.

7.2.2 Protocol

The subjects were seated in a dynamometer (Biodex System 3 Pro, Biodex Medical Systems, NY, USA) with the right elbow placed on an arm support and the wrist strapped tightly to a wrist support so that the forearm was constantly in a supinated position. The effects of vibrations were tested at elbow angles of 60 deg (flexed), 90 deg, and 120 deg (extended), which were measured between forearm and upper arm using an analog goniometer to assess the 90 deg angle and using the dynamometer's control panel to induce the changes of ± 30 deg. At each angle, the subjects were instructed to perform four maximum voluntary isometric elbow extensions (MVC). The four randomized trials per angle consisted of one control trial (no vibrations) and three trials with vibrations at different vibration frequencies. The vibrator was started three seconds before the contraction and remained on throughout the five-second contraction. Two minutes of rest were given between trials, and five minutes between conditions (i.e., elbow angles). Subjects performed test trials at each vibration frequency to become familiar with the procedure.

7.2.3 Vibratory stimulation

The vibrations were generated by a pneumatic vibrator (Figure 7.2.1, b and c; model VOA-100, Chicago Vibrator Products, Westmont, IL, USA) and were applied perpendicularly to the long axis of the forearm in the sagittal plane. The frequency was adjusted by modifying the vibrator's input air-pressure, which resulted in a frequency range between 20 Hz and 45 Hz and a total displacement of 6.0 ± 2.2 mm, measured by a tri-axial accelerometer (ADXL 78, range ± 35 g, nominal frequency response 0 Hz to 400 Hz, Analog Devices, USA). The accelerometer was firmly attached to the skin above the distal end of the radius bone using stretch medical adhesive tape in order to quantify solely vibration frequency and amplitude acting on the forearm. Torque was measured by the dynamometer (Figure 7.2.1).

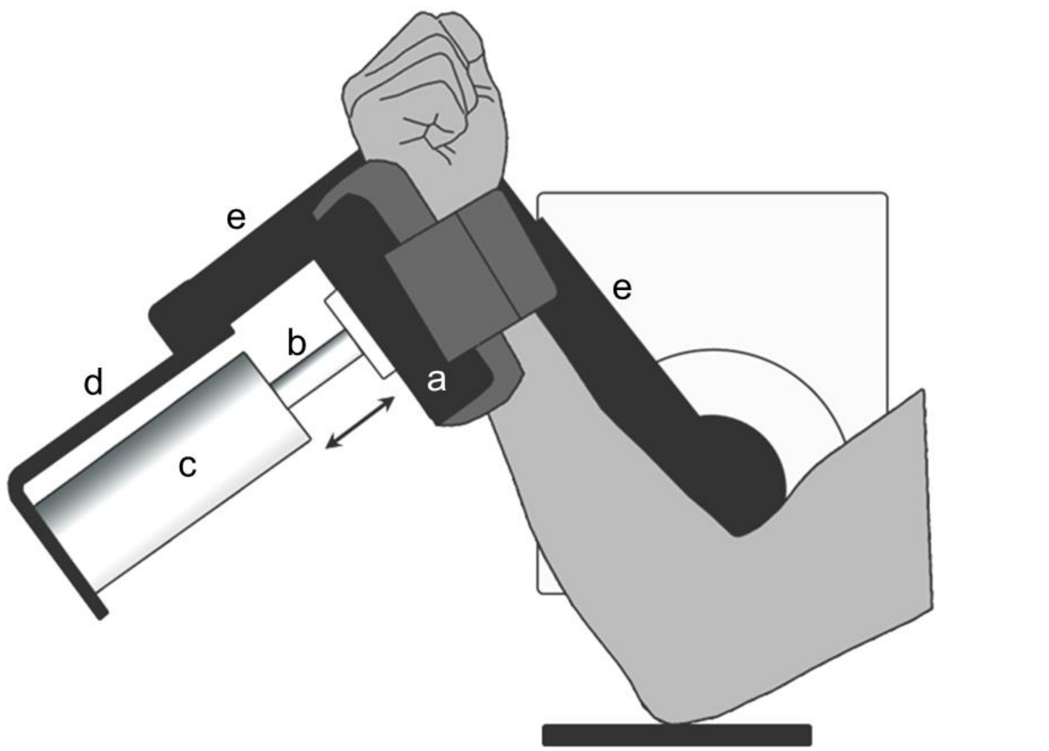

Figure 7.2.1: The forearm was strapped to the wrist support and the subject performed maximal isometric elbow extensions against the vibrating wrist support. The force was transferred from the wrist support (a) to the piston (b), then to the vibrator (c) and its mounting brace (d), and from there onto the dynamometer's lever (e); the lever passed behind the wrist support (a) and was only connected to the mounting brace (d) and the dynamometer.

7.2.4 Data collection and analysis

Vibration frequency and amplitude, elbow extension torque, and electromyographic (EMG) activity from biceps and triceps brachii muscles were recorded simultaneously at 2,400 Hz using a DAQCard-6062E 12-bit A/D data acquisition system (National Instruments, USA) and a custom made software interface using a data acquisition toolbox from Matlab R2010a (The MathWorks Inc., Natick, MA, USA).

The raw torque signal was processed using a fifth-order, low-pass Butterworth filter with a cutoff frequency of 5 Hz. The filter parameters were determined by frequency analysis of pilot data and were set to securely remove vibration artifacts in the torque signal, while maintaining the full torque signal strength generated by the subjects. Bipolar surface electrodes (Ag/AgCl) with an inter-electrode spacing of 22 mm (Norotrode 20, Myotronics Inc., USA) were positioned on the muscle belly of the biceps brachii (long head) and triceps brachii (long head) according to SENIAM recommendations (Hermens et al., 2000). The skin was prepared using alcohol swabs and the electrodes (including cables) were secured using medical adhesive tape. The EMG signal was pre-amplified at source (gain 2500; band-pass filter 10-500 Hz; Biovision, Wehrheim, Germany). A band-stop wavelet filter between 20 Hz and 45 Hz was applied to the EMG signals of each trial to remove possible, vibration-induced motion artefacts. After rectification, the activation intensity was expressed using the root mean square (RMS) of the signal using 100 ms windows. A 500 ms interval at the plateau region of the torque-time trace was used to quantify the applied torque and all other variables. Torque and EMG results were then normalized to the control (non-vibration) trials. Pearson's correlation coefficient was calculated to quantify the strength of the relationship of torque over frequency and EMG over frequency. The dependent measures torque, triceps EMG and biceps EMG were analyzed using one-way repeated measures analyses of variance (ANOVA). Mauchly's tests were used to prove that assumptions of sphericity have been met and if violated, the degrees of freedom for the F-test were adjusted using a Greenhouse-Geisser correction. Paired Student's t-tests were used to test for significance of the correlations as well as post hoc the ANOVA to test for differences between vibration and control conditions and for comparisons of different angles, while applying a Bonferroni correction for multiple comparisons. The level of statistical significance was set at a $p < 0.05$ level. All statistical analyses were conducted using SPSS (version 19.0.0, IBM SPSS, USA).

7.3 Results

The effects of vibrations on raw torque development and muscle activity are shown in Figure 7.3.1.

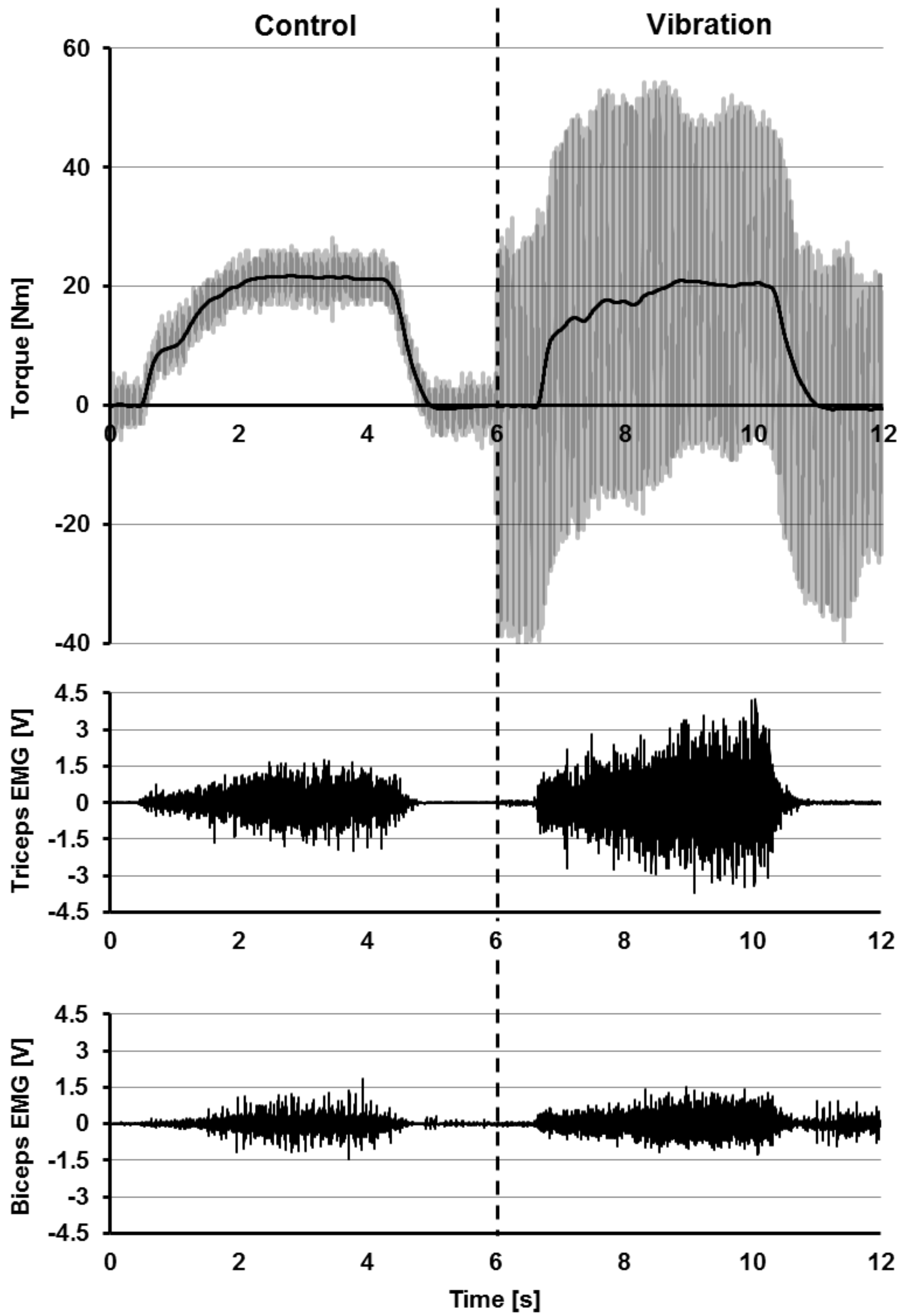

Figure 7.3.1: The top panel shows the raw torque data (grey) for the non-vibration control condition in the first six seconds and the raw data during vibration exposure in the final six seconds. The black line shows filtered torque data using a firths-order, low-pass Butterworth filter with a cutoff frequency of 5 Hz. The center and bottom panel show raw EMG data from the

triceps and biceps brachii muscles, respectively. The EMG data was pre-filtered at source using a bandpass filter (10-500 Hz) and amplified with a gain of 2500. Note that there was an effective break of at least two minutes in between the shown control and vibration conditions.

The correlation between torque and frequency at the individual angles was R = 0.256 (p = 0.09), R = 0.117 (p = 0.449), and R = 0.241 (p = 0.114) at 60 deg, 90 deg, and 120 deg, respectively and not significant (Figure 7.3.2 A). The result of the one way repeated measures ANOVA indicated a significant difference between the torque produced without vibrations and with vibrations at the three different elbow angles ($F(2.37,40) = 19.09$; $p < 0.01$). Relative to the corresponding non-vibration conditions, the average torque decrease was smallest and not significant at the flexed position of 60 degrees, with a decrease of 1.8% ± 5.7 % (p = 0.20). The average torque decrease was greatest at 90 degrees (7.4 % ± 7.9 %, p < 0.01) and in between at 120 degrees (5.0 % ± 8.2 %, p < 0.01) (Figure 7.3.2 B).

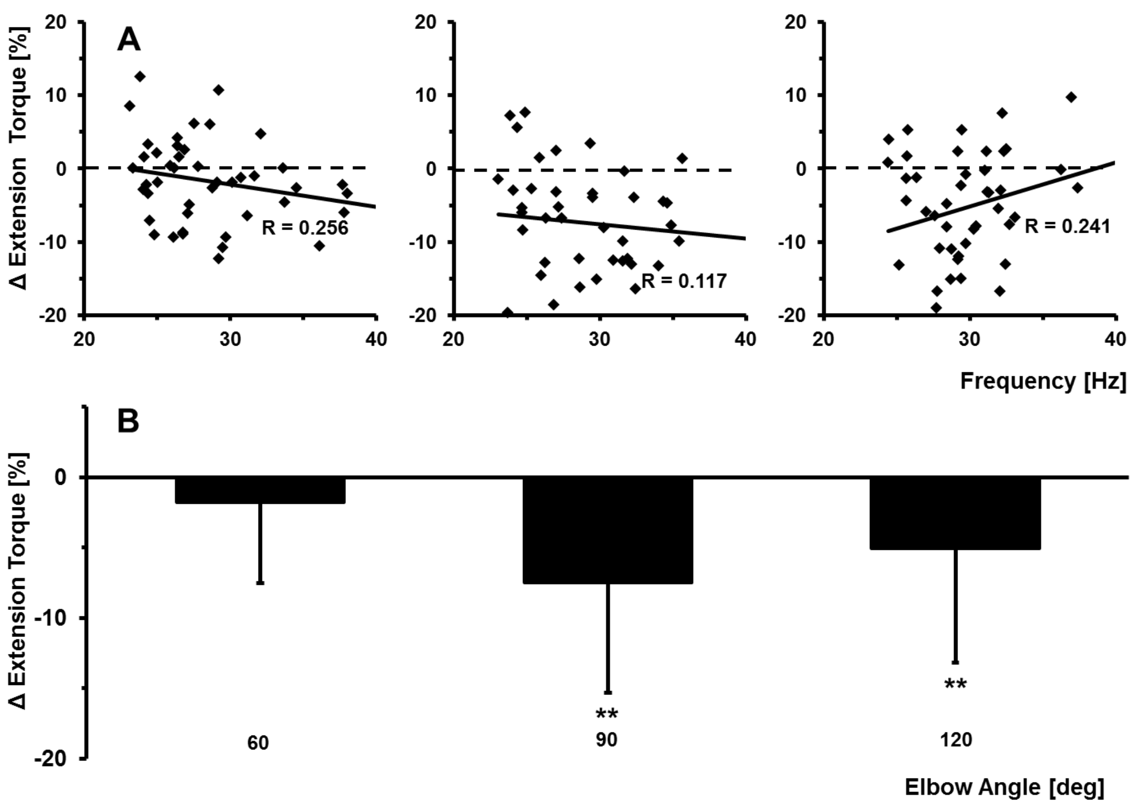

Figure 7.3.2: (A) Torque generated from each individual trial during vibrations (diamonds) relative to non-vibration control (dashed baseline) as a function of vibration frequency. Data shown from each elbow angle (left: 60 deg [flexed]; centre: 90 deg; right: 120 deg [extended]). (B) Average extension torque during vibrations relative to non-vibration control using data seen in (A). ** indicates $p < 0.01$.

The correlation between triceps muscle activity and frequency at the individual angles was $R = 0.075$ ($p = 0.625$), $R = 0.261$ ($p = 0.09$), and $R = 0.000$ ($p = 1.00$) at 60 deg, 90 deg, and 120 deg, respectively and not significant (Figure 7.3.3 A). Average muscle activity of the triceps showed significant differences during vibrations at the different elbow angles (F (2.37,40) = 19.09; $p < 0.01$). The increases were 42.6 % (±43.8 %, $p < 0.01$), 35.8 % (±34.1 %, $p < 0.01$) and 37.7 % (±34.7 %, $p < 0.01$), at elbow angles of 60 deg, 90 deg, and 120 deg during vibration exposure relative to the non-vibration control trials, respectively. Biceps muscle activity also changed significantly (F (2.52,40) = 12.97; $p < 0.01$). The muscle activity increased by 38.8 % (±43.8 %, $p < 0.01$), 34.2 % (±46.4 %, $p < 0.01$) and 29.6 % (±33.7 %, $p < 0.01$) during vibration exposure at 60 deg, 90 deg, and 120 deg, respectively. The increase between triceps (agonistic) and biceps (antagonistic) muscle activity did not differ significantly (F (3.09,39) = 0.81; $p = 0.49$) (Figure 7.3.3 B).

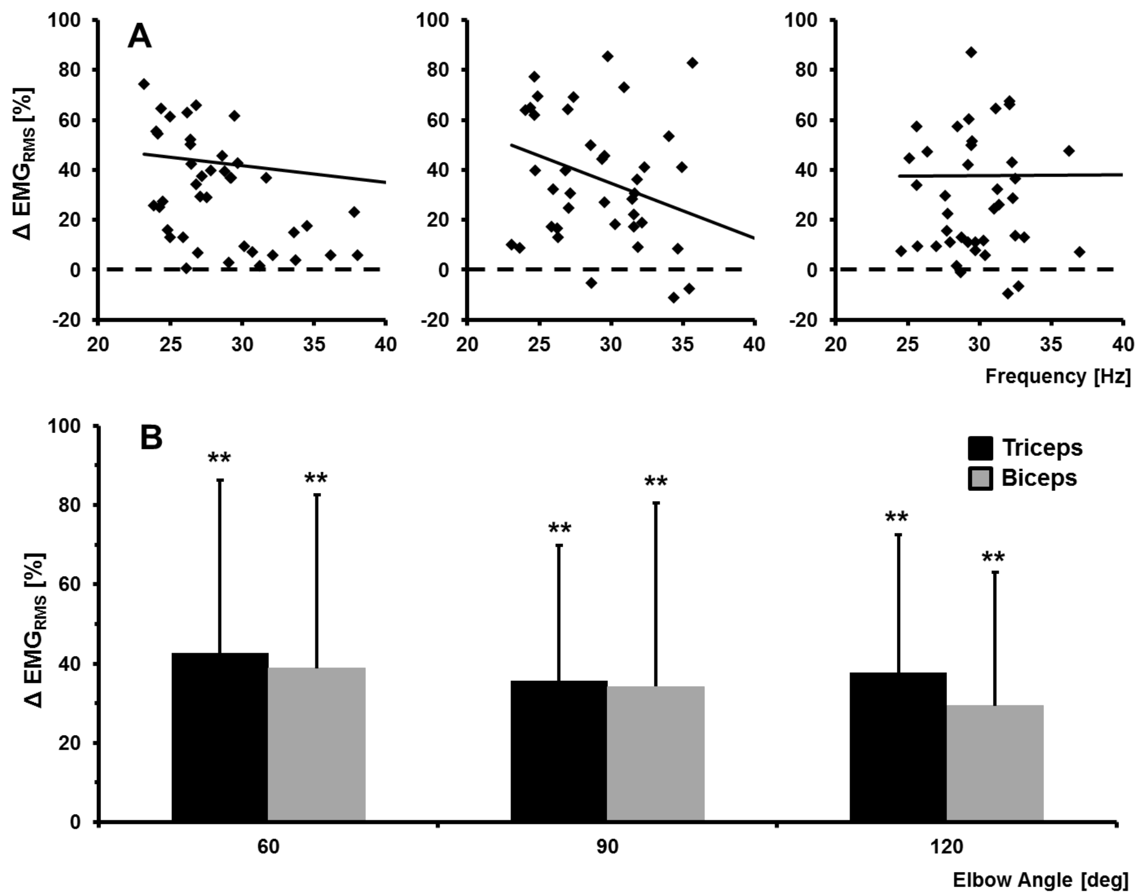

Figure 7.3.3: (A) Root mean squared triceps activity (EMG_{RMS}) from each individual trial during vibrations (diamonds) relative to non-vibration control activity (dashed baseline) as a function of vibration frequency. Data shown from each elbow angle (left: 60 deg [flexed]; centre: 90 deg; right: 120 deg [extended]). (B) Average EMG_{RMS} during vibrations relative to non-vibration control. ** indicates $p < 0.01$.

7.4 Discussion

Firstly, the results of this study showed that changes in vibration frequency did not correlate with maximal torque or muscle activity in the tested frequency band. Secondly, the average torque decreased during vibration exposure, depending on the elbow angle, while muscle activity increased. The influence of changing vibration frequency (H1) and changing elbow angle (H2) are discussed separately below.

7.4.1 Influence of changing vibration frequency

Maximal extension torque or muscle activity did not increase as vibration frequency increased, so our hypothesis that increasing vibration frequency would increase maximal torque and muscle activity of the target muscles (H1) needs to be rejected. A possible explanation for the low correlation between torque, muscle activity and vibration frequency may be the rather narrow frequency range used. In fact, the majority of vibration trials were frequencies between 23 Hz and 38 Hz, corresponding to an angular velocity of approximately 50 deg/s to 150 deg/s about the elbow joint. This frequency range may be too narrow to affect stretch reflexes enough to significantly modify the resultant torque, even though a linear relation between stretch velocity and spinal stretch reflex in the human triceps surae has been demonstrated (Gottlieb and Agarwal, 1979). Besides the narrow range, the maximum angular velocity is comparatively low when compared to angular velocities during extreme situations that may cause injury, such as falling forward where angular velocities of up to 1,000 deg/s may occur about the elbow joint (Dietz et al., 1981).

Another explanation for the lack of effect of vibration frequency on torque and muscle activity may be that series elastic elements compensated for the imposed muscle stretches. Although a recent study showed that whole body vibration affects the contractile length of the medial gastrocnemius (Cochrane et al., 2009), the fascicle and muscle spindle length in this particular study may have been only minimally affected, independent of vibration frequency, thus suppressing frequency-dependent reflex strength.

7.4.2 Influence of changing elbow angle

At two of the three angles, the torque generated during vibration exposure was significantly lower compared to the non-vibration control, while at the same time muscle activity significantly increased. One reason for reduced torque could be a greater increase in biceps muscle activity (antagonist) relative to the triceps (agonist) muscles. Unexpectedly, the results indicate that antagonist and agonist muscle activity increased by about the same percentage (30 % to 40 %) during vibrations compared to the control. However, it needs to be considered that the resulting moment about the elbow joint is not only a function of the muscle activity and the related forces of agonist and antagonist, but also of the moment arms of those muscles about the joint. While muscle activity (and potentially force) in both muscles increased by about the same amount, the moment arms of both muscles in fact changed over the range of motion (Murray et al., 1995) (Figure 7.4.1).

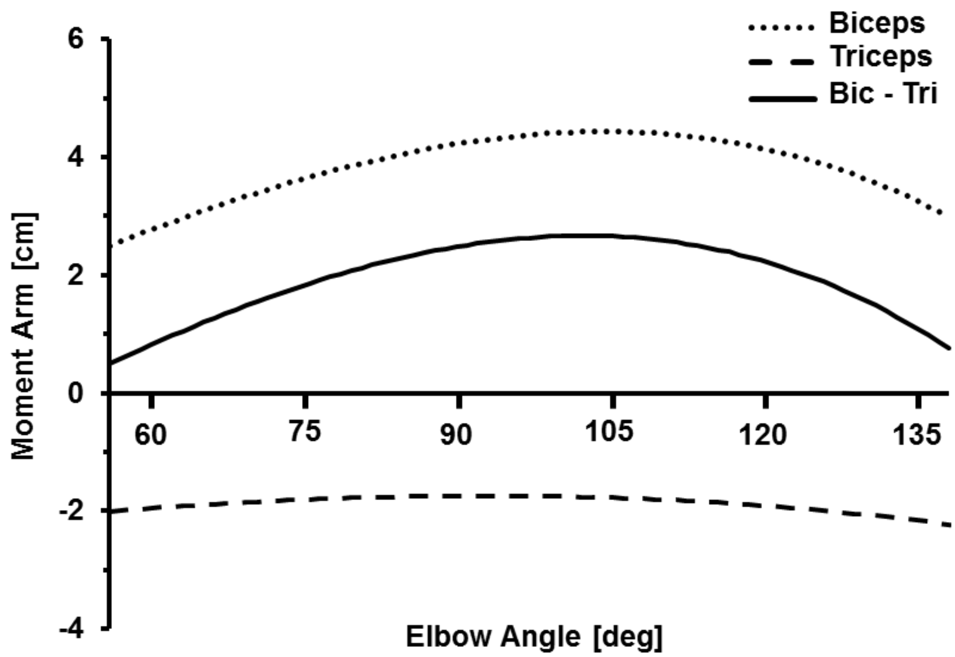

Figure 7.4.1: Average female moment arms of biceps (dotted) and triceps (dashed) muscles about the elbow joint. The difference between the two moment arms (Bic: Biceps; Tri: Triceps; solid line) changes as a function of the elbow angle. Data reproduced with permission (Murray et al., 1995).

The difference in the moment arm between the biceps and the triceps at the three angles studied is small for the 60 deg angle (0.82 cm), and relatively large for the 90 deg and 120 deg angles (2.51 cm 2.15 cm, respectively). In fact, the moment arm of the biceps is up to 2.4 times larger than the moment arm of the triceps at 90 deg and may be one reason for the elbow-angle-dependent decrease in elbow extension torque. Indeed, the larger the difference in moment arms about the elbow joint, the larger the loss in extension torque during vibrations. The proposed explanation involves a number of assumptions in order to hold true. Probably the most important assumption is that the increased muscle activity increased the force of both muscles by roughly the same absolute amount during vibration exposure. In other words, if the biceps activity during elbow extension had been, for example, only one percent of its maximum in the control condition, a 40 % increase of that one percent would have only minimal effects on the muscle's absolute force outcome. However, the absolute biceps muscle activity was about 37 % relative to the triceps's activity at the time of maximal effort, which indicates that the biceps was fairly active during the task. The high level of biceps activity is likely related to both stabilizing the joint and to the supinated forearm position during the experiment, which is known to cause biceps activation (Buchanan et al., 1989). It needs to be noted that the biceps is only one of the major elbow flexors, and the EMG activity of other flexors, such as the m. brachialis and m. brachioradialis, would need to be quantified together with the individual heads of the triceps if the proposed mechanism was to be verified.

An alternate explanation for the findings of this study is that vibrations may have an effect on cross-bridge dynamics, so that the actin-myosin interaction is negatively influenced by the quick, repetitive, and longitudinal muscle stretches. The stretch amount of half sarcomeres was therefore approximated using the following algorithm: Firstly, the change in the elbow angle due to vibrations imposed on the forearm was calculated using the known vibration amplitude and an average estimated ulna length (Murray et al., 2000). Given a change in elbow angle, the stretch of the triceps muscle can be estimated by considering its average moment arm about the elbow joint (Gerbeaux et al., 1996). Then the stretch of a half sarcomere can be calculated, given an average fascicle length, pennation angle and average sarcomere length (Langenderfer et al., 2004). The approximations showed that the vibrations in our study stretched half sarcomeres between 10 nm and 12 nm. Stretches exceeding 10 nm per half sarcomere were shown to potentially disrupt cross-bridges (Huxley and Simmons, 1971; Burton, 1995) and thus limit the maximal force production. This argument may explain losses in torque, but not the dependency on elbow angle. As the triceps moment arm changes with the elbow angle (Gerbeaux et al., 1996; Murray et al., 1995), the vibrations impose different stretch amounts on the sarcomeres. The moment arm, and thus the stretch amount, is smallest at the 60 deg angle (calculated stretch: 9.99 nm per half sarcomere), and the imposed stretch may fall below the critical stretch mark of 10 nm per half sarcomere in some subjects, causing little or no cross-bridge disruption. However, these estimations involve a number of assumptions and the potential effects on single sarcomeres remain highly speculative. If cross-bridge disruption is in fact the major factor responsible for the observed decrease in torque, then vibration amplitude should negatively correlate with maximal torque production. With respect to the stated hypothesis, consistent increases in muscle activity and maximal torque as initial muscle length increased were not observed, so H2 needs to be rejected. Further in vivo studies involving changes in vibration amplitude, as well as in situ protocols using isolated intact muscle and single myofibril preparations are needed to understand the acute (and consequently long-term) behaviour of muscle contraction during vibration exposure and to test the cross-bridge disruption theory.

Both proposed mechanisms, antagonist muscle activation and cross-bridge disruption, are affected by the changes in the muscle's moment arms with elbow angle. A male population might react slightly different in this experiment compared to females because the moment arms over the range of motion do not change equally for both genders (Murray et al., 1995).

7.5 Conclusions

This study showed that (a) maximal torque and muscle activity is not frequency dependent at the tested range, but that (b) torque is reduced when exposed to vibrations depending on the elbow angle, although (c) muscle activity was significantly elevated. The findings suggest that longitudinal vibrations may decrease mechanical output while simultaneously increasing neural input. This means that while vibrations should preferably be avoided if maximal extension torque is needed, elevated activation levels may be used to intensify strength training and to accelerate desired neurophysiological adaptations.

Summary

Vibrations of soft tissue compartments occur naturally during locomotion and can also be artificially induced by machines at the workplace or by whole-body vibration platforms. The spectrum of possible effects of vibrations ranges from degenerative processes in soft tissues and related pain to possible benefits in muscle performance and bone strength. This variety of effects triggered the interest of many, including our research group. It was proposed that soft tissue vibrations are damped by muscle activation in certain situations, which may have implications on muscle performance, comfort and fatigue ("muscle tuning" concept). The concept was challenged in many studies and yet many questions remain unanswered. This thesis attempted to elaborate on open questions related to tissue vibrations in running and during whole-body vibration training. Specifically, the questions we tried to answer were:

(a) Does muscular fatigue affect the vibration characteristics of soft tissues during prolonged running?
(b) Does compression apparel positively affect running performance and recovery?
(c) To which extent are vibrations transmitted to lower extremity soft tissues during whole-body vibration training?
(d) What is the oxidative metabolic demand of individual lower extremity muscles during whole-body vibration training at different vibration frequencies?
(e) Can longitudinally applied vibrations increase maximal elbow extension torque?

8.1 Changes in tissue vibrations in a run to fatigue

In this study, the participants were asked to run at their 10 km race pace on an outdoor track until the required lap time could not be maintained. Soft tissue accelerations of the triceps surae muscle compartment were quantified by means of a 3-dimensional accelerometer mounted to the skin. It was hypothesized that the vibrations of the soft tissue compartment due to impact will change with fatigue, under the assumption that a fatigued muscle is less capable to damp vibrations. The results showed that the vibration intensity, which was calculated by means of a Fast-Fourier Transformation, increased significantly in the fatigued state relative to the non-fatigued state in axial (parallel to the tibia) and medio-lateral directions. Furthermore, the peak intensity occurred later after heel-strike, as quantified by means of a wavelet transformation of the vibration signal. The vibration frequency of the soft tissue compartment was unchanged and the results lead to the conclusion that fatigue increases vibration intensity and potentially vibration duration of soft tissues during prolonged running.

8.1.1 Critical considerations

The interpretation of the results, which were that the soft tissues might be at an increased risk of injury due to increased vibration intensities (caused by decreased muscular damping), should be taken with caution. The change in tissue vibration characteristics may have multiple factors such as changes in tissue temperature throughout the run, relaxation of soft tissues due to the repeated stretches, and changes in running technique. Those factors were not quantified in this study, and the amount of muscular fatigue was not directly assessed.

8.1.2 Future directions

This study was part of a larger study with the general aim to assess the effects of fatigue on psychological, physiological, and biomechanical variables during running. Some of the above shortcomings, such as the assessment of muscle activity and kinematics (running technique) were in fact quantified for some of the participants. Therefore, future work in this area could use the available data and try to correlate the changes in vibration properties and muscle activity and running technique, with the goal to identify the actual cause of the changed tissue vibration properties in a prolonged run. Equally important, the actual risk by soft tissue vibrations that occur during running needs to be assessed. As of now, this study only showed changes in tissue vibrations in a run to fatigue but the implications of this finding remain speculative.

Computational, animal, and even human models may be used to study the physiological effects of vibrations, using the exact same vibration frequencies and intensities as quantified in this study.

8.2 Compression apparel and its effect on running performance and recovery

This project was designed to identify potential performance benefits of compression apparel during running at different speeds. Two factors may provide a benefit when wearing compression apparel during running. Firstly, compression apparel was shown to reduce soft tissue vibrations and may therefore aid the muscles with the vibration damping process according to the muscle tuning concept. Secondly, compression was shown to increase blood flow and may thus facilitate oxygen supply to muscles and metabolic by-product removal. The level of compression needed to optimally facilitate blood flow is unknown for dynamic exercises, thus three levels of compression were tested. The results showed that the speed at which the aerobic and anaerobic lactate thresholds occur were not affected by any of the compression apparel prototypes. No effect on maximal heart rate during the final running stage and speed of lactate clearance after the run was found when using compression apparel. The findings in this study generally agree with previous studies that tested the effects of compression apparel on running performance.

8.2.1 Critical considerations

One reason for the lack of improvement in running performance and recovery may be related to the initial assumptions made. Compression apparel was shown to reduce soft tissue vibrations after landing from a jump, but this effect was not yet shown during running. Assuming that compression apparel would have this effect, it is still unknown whether at all, or to which extent this would affect the needed muscle activity per stride, and thus potentially the energy requirements for running at a certain speed. Another reason may be related to the blood flow enhancing effect of compression apparel. Beneficial effects on blood flow were predominantly shown in quasi-static situations, and a few studies showed benefits during low intensity exercise.

The effects of compression apparel on blood flow during high intensity exercises, such as running above the aerobic and anaerobic thresholds were not yet tested.

8.2.2 Future directions

Therefore, one needs to (a) study the effects of compression on vibrations and muscle activity during running at different speeds and different levels of compression and (b) evaluate the effects on blood flow during high-intensity exercise. The first aspect, tissue vibrations and muscle activity, were addressed by an intern at our laboratory. Initial results for tissue vibrations showed reduced vibration intensity for the triceps surae area at speeds faster than 3.58 ms^{-1} but not for the vastus lateralis and no systematic damping of vibrations with increasing levels of compression. Muscle activity was even elevated in the gastrocnemius medialis muscle at speeds faster than 4.47 ms^{-1} when using the high level compression suit.

Different tests to assess running performance, such as a time to exhaustion test or time needed to complete a certain distance may be more direct indicators of performance. Finally, one may use near-infrared spectroscopy to potentially quantify hemodynamic changes when using compression apparel during exercise, which may provide some insight into the effects of compression apparel on blood flow during high-intensity exercise.

8.3 Transmissibility of vibrations on a whole-body vibration platform

Soft tissue vibrations occur naturally during walking and running but are purposely generated during whole-body vibration training. Vibration platforms gained significantly in popularity over the last decade and the efficacy of this training intervention in terms of improving muscle performance and mechanical bone properties was studied. Vibrations have the potential to induce damage to soft tissue compartments, depending on vibration amplitude, frequency, exposure duration, and exposure site. Although it is known that vibrations may induce damage to soft tissues over a matter of years, the transmissibility of the vibrations form the platform to the soft tissue compartments has hardly been studied. One project within this thesis therefore tested the vibration transmissibility at three different vibration frequencies (10 Hz, 17 Hz, and 28 Hz) and a frequently used vibration amplitude (2.5 mm). The results showed high transmissibility (2.1-2.3) of peak acceleration and peak amplitude (tissue displacement) at 10 Hz, moderate transmissibility at 17 Hz (1.1-1.9), and low transmissibility at 28 Hz (0.5-1.2). The peak acceleration was reached at 28 Hz and was about 125.4 ms^{-2} and 46.5 ms^{-2} for the triceps surae and quadriceps femoris, respectively. The results also showed that the largest displacement occurs at the low frequency and the highest acceleration at the high frequency. This study showed for the first time the vibration characteristics of soft tissue compartments while standing on a side-altering vibration platform. The absolute acceleration at 28 Hz was substantially higher than the acceleration used in rat studies that were related to soft tissue injury. Based on those findings, further research about long term effects of vibration training on soft tissues seems necessary.

8.3.1 Critical considerations

This finding does not necessarily mean that vibrations during WBV training have harmful effects. The biggest difference between the study within this thesis and the rat studies which showed negative effects of vibrations is the exposure duration, which was always substantially higher in the animal studies compared to typical vibration exposure durations in WBV training. Nonetheless, it is unknown whether WBV training can induce injury in a matter of years, for example when this type of training is used on a regular basis.

8.3.2 Future directions

Future studies can use the acceleration and tissue displacement data acquired in this study in computational or animal models to determine whether soft tissue accelerations that are common during WBV training are harmful.

8.4 Metabolic demand of individual muscles during whole-body vibration training

The vibrations induced by a whole-body vibration platform increased muscle activity as was shown by means of surface electromyography in previous studies. The ongoing dispute about proper analysis of the EMG signal during WBV training emphasises an interest in alternative approaches to quantify the activity of the muscle. Using near-infrared spectroscopy (NIRS), the oxidative metabolic demand of the gastrocnemius medialis and vastus lateralis muscle was quantified at vibration frequencies of 10 Hz, 17 Hz, and 28 Hz and an amplitude of 2.5 mm. The results showed that with each increase in vibration frequency, a significant increase in the oxidative metabolic demand of the muscle occurred. On a local muscular level, WBV training may represent a significant training stimulus for particular muscles. The results also represent a first step in finding the optimal vibration platform settings to improve training efficacy by means of near-infrared spectroscopy.

8.4.1 Critical considerations

The stance position in terms of hip, knee, and ankle angles was not controlled in this study. Especially the condition with a vibration frequency of 28 Hz was uncomfortable in an upright position and some subjects adapted a more flexed leg position. This visually observed change in posture likely affected the transmissibility of vibrations (as previous research showed) and thus local oxygen consumption. Furthermore, the determination of local oxygen consumption required the occlusion of the entire limb, which, according to subject statements and personal experience, induces moderate to large discomfort in the subjects and also induced a feeling of numbness. The numbness may lead the shifts in the body weight to one or the other leg and could affect the quantification of the metabolic demand by NIRS.

8.4.2 Future directions

In the future, the quantified metabolic cost of individual muscles needs to be assessed for multiple muscles and for other exercises in order to compare it to the results acquired during vibration training. One vision would be to generate a database consisting of the average metabolic cost of various lower extremity muscles during various exercises from a large sample population. This database could be used to customize vibration training individually depending on the goal (maintain or improve muscle mass and performance) of the trainee or patient.

8.5 Do vibrations affect maximal strength?

Previous and part of the research within this study showed that vibrations can affect muscles in many different ways. One, if not the most, fundamental property of a muscle is to generate force. Some previous work showed that strength was not affected, however one study found improved strength during biceps flexion when vibrations were applied to the muscle. It seems obvious that the means by which vibrations are transmitted is crucial for the effects they have on any soft tissue. A custom-made vibratory device allowed the quantification of the effects of vibrations on muscle strength in a scenario very close to a real-life situation. The effects of longitudinally applied vibrations on maximal isometric elbow extension torque was independent of the tested frequency range (approx. 20 Hz to 40 Hz) but changed with alterations in the elbow angle. Maximal extension torque decreased significantly at 90° and 120° elbow angles by -7.4 ± 7.9 % and -5.0 ± 8.2 %, respectively. The reasons for this decrease in elbow extension torque were speculated to be related to (a) an increase in muscle activity of agonistic and antagonistic muscles and/or (b) disturbance of cross-bridge formation on the sarcomere level due to the vibrations. An increase in muscle activity would favor the biceps (antagonist) at certain elbow angles, since the lever of this muscle about the elbow joint is larger than the one of the triceps brachii and changes as a function of the elbow angle.

8.5.1 Critical considerations

It remains unclear whether the vibrations actually affect the strength of a muscle, since torque was measured and the experimental setup involved agonistic as well as antagonistic muscles about the elbow joint. The vibrations were unfamiliar to the subjects, especially during a maximal effort, which may have prevented some subjects from producing maximal torque during the vibration conditions. If this psychological inhibition was not the case, then, from a practical point of view, this experiment showed that vibrations transmitted to the elbow extension muscles may limit the effective torque output which may be relevant in certain sports (tennis, motocross, windsurfing) and occupations (drilling).

8.5.2 Future directions

Future studies may isolate single muscles or muscle fibers and expose them to longitudinal vibrations in order to gain a deeper understanding of the effect of vibrations in muscle force production. If practical, in-vivo experiments were of future interest, then a familiarization phase of the subjects would be needed, together with an electrical stimulation of the muscle during the maximal contraction phase to quantify the extent of muscle inhibition.

In summary, this thesis tried to answer a variety of questions related to soft tissue vibrations. The main findings were that (a) soft tissue vibrations increased in a run to fatigue, (b) compression apparel did not aid running performance and recovery, (c) whole-body vibration platforms transmit high intensity vibrations to soft tissues, (d) increasing vibration frequency increases localized, oxidative metabolic demand, and (e) vibrations decreased maximal isometric elbow extension torque.

Vibrations appear to interfere with the regular functionality of muscles, since the energetic demand increased and maximal torque output about the elbow joint decreased. Future studies may focus on quantifying the amount of energy solely needed by the muscles to damp vibrations during walking and running, whether compression apparel indeed can help to reduce tissue vibrations and whether single muscle fibers exposed to vibrations show a decreased force production.

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