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Gravitational force modulates muscle activity during mechanical oscillation of the tibia in humans

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Abstract

Mechanical oscillation (vibration) is an osteogenic stimulus for bone in animal models and may hold promise as an anti-osteoporosis measure in humans with spinal cord injury (SCI). However, the level of reflex induced muscle contractions associated with various loads (g force) during limb segment oscillation is uncertain. The purpose of this study was to determine whether certain gravitational loads (g forces) at a fixed oscillation frequency (30 Hz) increases muscle reflex activity in individuals with and without SCI. Nine healthy subjects and two individuals with SCI sat with their hip and knee joints at 90° and the foot secured on an oscillation platform. Vertical mechanical oscillations were introduced at 0.3, 0.6, 1.2, 3 and 5 g force for 20 seconds at 30 Hz. Non-SCI subjects received the oscillation with and without a 5% MVC background contraction. Peak soleus and tibialis anterior (TA) EMG were normalized to M-max. Soleus and TA EMG were < 2.5% of M-max in both SCI and non-SCI subjects. The greatest EMG occurred at the highest acceleration (5 g). Low magnitude mechanical oscillation, shown to enhance bone anabolism in animal models, did not elicit high levels of reflex muscle activity in individuals with and without SCI. These findings support the g force modulated background muscle activity during fixed frequency vibration. The magnitude of muscle activity was low and likely does not influence the load during fixed frequency oscillation of the tibia.

Keywords

Mechanical oscillation; Reflex; Spinal cord injury

1. Introduction

While the optimal parameters for osteogenesis are currently unknown, studies using animal models can form a basis for research in human subjects.

Large-amplitude loads, in excess of one times body weight, produced by muscle contractions, increase bone density in people with spinal cord injury (SCI) (Dudley-Javoroski and Shields, 2008a,b; Granhed et al., 1987; Shields and Dudley-Javoroski, 2006); but low load, imperceptible vibratory stimuli predominate bone's daily strain history (Fritton et al., 2000). Before the mechanisms of low oscillatory gravitational loads (g force) on bone density can be assessed in humans, we must understand if these stimuli also induce reflexive muscle forces on the skeletal system.

Basic research findings support that low magnitude loads at certain oscillatory frequencies ($\square$ 30 Hz) exert a robust osteo-regulatory influence. For example, numerous animal studies support that these oscillatory interventions trigger anabolism of osteoblasts in isolated bone segments (Flieger et al., 1998; Garman et al., 2007a,b; Rubin et al., 2001, 2002), even in the absence of load (Garman et al., 2007a,b). To our knowledge, the mechanical oscillatory stimulation parameters (30 Hz; 0.3–1*g* force) that increase osteogenesis in animal models (Christiansen et al., 2008) have not yet been assessed for reflexive muscle activity in humans.

Previous human studies have employed whole body vibration (WBV) rather than localized limb segments, in which a subject stands atop a vibrating surface (Gilsanz et al., 2006; Iwamoto et al., 2005; Torvinen et al., 2003; Verschueren et al., 2004; Ward et al., 2004). These previous studies did not determine whether the muscles in the vibrated limbs were active in a reflex-mediated manner (Abercromby et al., 2007). Indeed, if these oscillation patterns trigger reflexive muscle contractions (Hazell et al., 2007), then a mechanical oscillatory stimulus along with an applied force from the muscle contraction would contribute to the mechanical dose of stress on the bone.

Individuals with spinal cord injury (SCI) experience severe osteoporosis, and are known to have heightened muscle reflex responses to perturbations (upper motor neuron lesions); while those with lower motor neuron lesions, are unable to trigger any reflex muscle contractions. If various *g* forces during fixed frequency mechanical oscillation training triggers reflexive muscle contractions, then the dose of strain delivered to human bone may vary from person to person.

Our long term goal is to determine if longitudinal mechanical oscillatory interventions alone, or in combination with muscle contractions, influence bone density in humans with and without SCI. Accordingly, the purpose of this study is to determine whether a fixed frequency mechanical oscillation (30 Hz), with various gravitational forces (*g* force), increases muscle activity in individuals with and without SCI. We hypothesized that muscle activity will be elicited with increasing gravitational force (*g*), but at low levels in those with and without SCI.

2. Methods

2.1. Subjects

Nine healthy subjects (5 females and 4 males: age = 33.5 ± 5.0 - years; height = 170.2 ± 4.6 cm; weight = 68.1 ± 9.0 kg) and 2 individuals with chronic complete spinal cord injury (ASIA-A [Association, 2002]) participated in the study. Subjects were screened for the following exclusion criteria: pressure ulcer; systemic illness; recent history of fracture to the lower extremity; acute inflammation in the lower extremity; implants; gallstones; kidney or bladder stones; any disease of the spine; peripheral vascular disease; or pregnancy. The absence of lower motor neuron injury was confirmed by eliciting soleus evoked contractions. All subjects were right leg dominant. Subjects provided written informed consent before data collection. The study protocol was approved by the Human Subjects Office Institutional Review Board at the University of Iowa. Subject characteristics are shown in Table 1.

2.2. Instrumentation

2.2.1. Mechanical Oscillation Device—Mechanical oscillation was produced by Ling model 721 servo-controlled vibration generator (Ling Dynamic Systems, Royston, Herts, England). Subjects approached the system from a height-adjustable chair and placed one

foot on a platform affixed to the top of the vibration drum. The hip, knee, and ankle were positioned at 90° of flexion. The heel was centered on the vibration platform. Straps were placed at the chest, waist and dorsum of the foot for additional stabilization (Fig. 1). A mono-axial accelerometer at the center of the platform measured acceleration (g) during testing. An LDS digital sine controller (Dactron COM 200) delivered the oscillatory stimulus to the limb segment. The system could be terminated with a manual abort switch.

2.2.2. Justification for Oscillation Parameters—Future therapeutic oscillation protocols will likely target frequency and amplitude parameters that mimic normal physiologic values, as these may hold the greatest osteogenic potential. Animal studies support that oscillation frequencies that mimic the intrinsic frequency spectrum of muscle contraction (0.1 to ~50 Hz) (Fritton et al., 2000; Huang et al., 1999) can serve as an anabolic stimulus to bone (Flieger et al., 1998; Garman et al., 2007a,b; Ozcivici et al., 2007; Rubin et al., 2002). For the present study, we selected a 30 Hz vibratory stimulus because it showed a 32% increased in bone density in sheep (Rubin et al., 2002), and, the authors speculated that the 30 Hz oscillation frequency simulates the important “fusion frequency of skeletal muscle”. However, in a different preparation, this intervention elicited muscle reflex activity (Cardinale and Lim, 2003), supporting the need for this study.

Previous animal and human studies have employed a range of acceleration magnitudes (0.3 to >2 g) (Flieger et al., 1998; Garman et al., 2007a,b; Gilsanz et al., 2006; Judex et al., 2003, 2006; Rubin et al., 2002; Ward et al., 2004). We administered a 30 Hz oscillation over a comparable range of g values: 0.3, 0.6, 1.2, 3.0 and 5.0 g . These accelerations elicited peak to peak displacements of 0.16, 0.33, 0.66, 1.66, and 2.76 mm, respectively. Fig. 2A shows examples of oscillation sine wave signals in 4 continuous events at 0.3, 1.2 and 5 g .

2.2.3. Electromyographic (EMG) Recordings—EMG was recorded using active bipolar surface electrodes (silver-silver chloride discs of 8 mm in diameter spaced 20 mm between centers) for the soleus (Sol) and tibialis anterior (TA) of the right leg. The location of the electrode on Sol was 2 cm lateral to midline of the posterior calf and 2 cm distal to the distal most palpable border of the gastrocnemius. On TA, the electrode was placed at the upper 1/3 and 2.0 cm anterior to a line between the tip of the fibular head and the tip of the lateral malleolus.

The skin at these locations was abraded with sandpaper and cleaned with an alcohol swab. A common ground electrode was placed on the anterior bony ridge of the tibia of the right leg. The EMG signals from all electrodes were preamplified on site with a gain of 35 and then differentially amplified (input impedance of 15 M Ω at 100 Hz; frequency response 15–4000 Hz; common mode rejection ratio 87 dB at 60 Hz) with an overall gain of 1000. A representative example of raw EMG of Sol and TA appears in Fig. 2B.

2.2.4. Peripheral Nerve Electrical Stimulation—The posterior tibial nerve was electrically stimulated using a constant current stimulator (DS7A, Digitimer, UK) with a range of 1–400 V and a constant current up to 200 mA. The stimulator delivered a square wave with a pulse width that ranged from 250 to 1000 ms. A double-pronged surface stimulating electrode was placed at the popliteal fossa with the cathode proximal to the anode. To obtain maximal M wave (M-max) of Sol, the stimulus intensity was increased gradually until the peak amplitude of Sol M-max no longer increased. To ensure supramaximal stimulation for M-max, the stimulation intensity was then doubled. TA m-waves were also measured during SOL M-max acquisition. A representative trial of Sol M-max is shown in Fig. 2B.

2.2.5. Force (Maximal and Submaximal) recordings—A force transducer (1500ASK-200, Interface, Scottsdale, AZ) was placed on the distal, anterior surface of the thigh (above the knee joint) while the limb was secured on the oscillating platform (Fig. 1). Non-SCI subjects performed 3 isometric plantar flexion maximal voluntary contractions (MVCs) for 5 s, with each contraction separated by 1 min. A target force of 5% MVC was calculated from the highest force level among the 3 MVCs and displayed on a computer screen. Subjects received visual feedback of active plantar flexor force in comparison to this target contraction level.

2.3. Procedure

Baseline EMG was obtained in resting and (for non-SCI subjects) during a 5% MVC contraction. Five vertical oscillation stimuli (0.3, 0.6, 1.2, 3.0 and 5.0 g) were then randomly introduced via the platform. Each epoch of oscillation lasted 20 s and was followed by a 1 minute rest. After an additional one minute rest, Non-SCI subjects received a second random oscillation sequence as they maintained a 5% MVC contraction. The cumulative vibration exposure for each subject was less than 4 minutes.

2.4. Data Analysis

M-max, EMG, and oscillation signals were sampled at 2000 Hz in Datapac 2K2 (RUN Technologies, Mission Viego, CA). Raw EMG signals were processed in the following steps: passive demeaning, Butterworth band-stop filtering then full wave rectified at 0 v. Power spectrum analysis was applied to identify any frequency of the motion artifact and its superior harmonics (Fratini et al., 2008). A band-stop filter was then applied between 30–31, 60–61 and 90–91 Hz to eliminate the motion artifacts induced by mechanical oscillation. One oscillation event was defined and identified based on one sinusoidal wave (wave of acceleration *g* force). One second (30 events) of EMG activity in the middle of the trial was averaged and the peak value was identified for further analysis. Peak EMG of tested Sol and TA were normalized to M-max of each muscle and expressed as% M-max.

2.4.1. Statistical Analysis—Sol and TA peak EMG were compared among 6 oscillation magnitudes in each condition (resting and 5% MVC (when applicable)) for Non-SCI by repeated measures ANOVA (SAS 9.1.3, Cary, NC). Alpha was set at .05. The SCI subjects are presented descriptively.

3. Results

For the non-SCI subjects under resting conditions, Sol peak EMG was greater at 5 *g* than at 0, 0.3, 0.6 and 1.2 *g* (all $p < .05$) (Fig. 3A). No differences in TA peak EMG appeared for any *g* force (all $p > .05$). When maintaining a 5% MVC, Sol peak EMG was greater at 1.2, 3 and 5 *g* than at 0 *g* force (all $p < .05$) (Fig. 3B). Likewise, Sol peak EMG during 5% MVC was greater for 5 *g* than for 0.3 and 0.6 *g* (all $p < .05$). Unlike during resting conditions, Greater TA peak EMG occurred for the 5 *g* than for 0, 0.3, 0.6 and 1.2 *g* force (all $p < .05$). Fig. 4 depicts the Sol EMG average for 30 oscillation events in the resting (Fig. 4A) and 5% MVC conditions (Fig. 4B). In both conditions, the magnitude of EMG increased with increasing acceleration, especially during the latter part of the oscillation cycle. Under all conditions, the increased background EMG levels were only a small fraction of M-max ($< 2.5\%$ in all cases).

Fig. 5 depicts the Sol EMG average for 30 oscillation events in one subject with SCI. Unlike for non-SCI subjects, neither Sol nor TA peak EMG increased with increasing acceleration, nor did the EMG pattern differ across the oscillation cycle. This trend was similar for both subjects with SCI.

4. Discussion

The purpose of this study was to determine whether oscillation parameters, that are known to have bone-anabolic potential (based on animal studies), increase electromyographic activity in individuals with and without SCI. We hypothesized that reflexive EMG activity would be elicited with increasing *g* force in both cohorts. To our knowledge, this is the first study to quantify reflexive muscle activity during low magnitude, fixed frequency mechanical oscillation of an isolated constrained human limb segment.

Statistically significant increases in background Sol and TA EMG emerged in Non-SCI subjects when the limb underwent oscillation at high amplitudes. However, a more important finding is that these increased background EMG levels were only a small fraction of M-max (< 2.5% in all cases). Thus while increases in SOL and TA EMG were detectable, it is unlikely that this level of muscle contraction could contribute to sizable compressive loads to the tibia.

In human whole body vibration (WBV) studies, EMG is found to be significantly increased when subjects stand on an oscillating surface (Cardinale and Lim, 2003; Roelants et al., 2006). Subjects in previous WBV/EMG reports actively stood and/or performed a variety of squat maneuvers, and therefore maintained higher levels of volitional muscle activation than subjects in the present report. In these studies, the application of an oscillatory load triggered peak EMG increases of up to 50% (Abercromby et al., 2007; Cardinale and Lim, 2003; Roelants et al., 2006). We are aware of no previous WBV studies that compare peak EMG during passive standing (no volitional activation) with and without oscillation, a more valid comparison between WBV and the present study. However, as the magnitude of EMG change was considerable in WBV + squat protocols, it seems likely that background EMG and direct oscillation of the head (vestibular system) plays a role in determining whether oscillation will trigger additional reflex activity. This view is congruent with Fig. 3, in which peak EMG increases were more prevalent during a 5% MVC background contraction. We surmise that excitatory central drive, with or without vestibular input to the alpha motor neuron pool, underlies this facilitation.

4.1. Mechanisms of Reflex Responses

In WBV studies, vestibular activation may play a role in reflex responses to oscillation. Amplification of the platform acceleration can be considerable at the ankle, knee, hip, and spine, and standing subjects exposed to platform displacements >0.5mm reported discomfort (Kiiski et al., 2008). Bite bar studies indicate that during WBV, transmission of accelerations to the head can be considerable (Abercromby et al., 2007). These factors support that WBV perturbs the head, triggering motor cortical drive to the lower extremity muscles. In the present study, subjects were fully supported in a sitting position with the lower leg constrained on the platform. As transmissibility of accelerations to the head is damped by lower extremity joint flexion (Abercromby et al., 2007), we anticipate that vestibular inputs were minimal during this study. This lack of excitatory descending drive to the lower extremity motor neuronal pool likely contributed to the minimal increase in background EMG observed during limb oscillation, even at high platform amplitudes.

Methodological factors may have also led to the small responses of background EMG to limb oscillation. Previous WBV studies of surface EMG with power spectral analysis demonstrated sharp peaks that corresponded exactly with the oscillation frequency and its harmonics (Fratini et al., 2008). These authors suggested that motion artifact is contributory to EMG in WBV studies, particularly when platform displacement is high (4 mm) and frequencies range from 30 to 45 Hz (Fratini et al., 2008; Hazell et al., 2007). In the current study, a band-stop filter was applied to eliminate any possible motion artifact associated

with the vibration stimulus and its harmonics. This procedure thus decreased the risk of over-estimation of reflex muscle activation (Fratini et al., 2008).

While filtering negated the effect of motion artifact in the EMG signal, the level of background EMG appeared to be influenced by the phase of the oscillation (Fig. 4A and B). Particularly during oscillation with a 5% MVC, background EMG increased after the halfway point of the oscillation event, at approximately 16 ms from the onset of the oscillation. The latter half of the oscillation corresponds to the descending movement of the vibration platform. Indeed, this suggests that the accumulation of phase 1 inputs of the oscillation triggers the EMG response observed in the 2nd phase. This most likely does not occur on the first response as only ~ 16 ms separated the phase 1 from phase 2 response. Therefore, the EMG response in phase 2 was triggered by either/both phase 1 or phases 2 events from the *previous* oscillation cycle. This would give the EMG response either a latency of ~ 35 ms, timing consistent with short latency responses. The quiescence of EMG during the initial phase is a consistent response, and may represent a strong consistent inhibition from previous cycles of oscillation, but this is speculative and cannot be specifically addressed in this study.

4.2. SCI vs Non-SCI EMG Responses

In contrast to non-SCI subjects, subjects with SCI demonstrated no increase in EMG activity at any magnitude of oscillation (Fig. 5). Several possible explanations for this difference exist. Subjects with complete SCI lack descending supra spinal input to the lower extremities, unlike the non-SCI group. However, as previously described, we believe that head perturbation was minimal in this experimental design. None of the subjects with SCI were on anti-spasmodic medication, so this study was not constrained by pharmacological management. This feature of diminished EMG during constrained limb segment oscillation differs from previous WBV approaches and should offer unique insights into the effects of oscillation on spinal reflex pathways and bone physiology.

Secondly, SCI is characterized by hyper-reflexia. If rapid muscle length changes were to occur (as has been observed during oscillation on see-saw type systems (Cochrane et al., 2009)), stretch reflex-mediated contractions in SCI subjects would be expected. The absence of reflex-mediated EMG activity in subjects with SCI supports that the mode of oscillation delivered in the present study did not yield noteworthy changes in soleus or TA muscle length. By extension, changes in muscle length were not likely to be the source of peak EMG increases observed in non-SCI subjects.

The tonic vibration reflex (TVR) is mediated by both monosynaptic and poly-synaptic pathways that result in increased motor unit activation (Martin and Park, 1997). After incomplete SCI, direct vibration of tendon or of a muscle belly facilitates muscle contraction (Cotey et al., 2009) and WBV training has recently been shown to augment gait (Ness and Field-Fote, 2009). In complete SCI however, prolonged application of vibration to tendon appears to inhibit electromyographic activity of plantar flexor muscles (Butler et al., 2006). This finding is congruent with the blunted EMG response of SCI subjects to whole limb oscillation in the present report.

4.3. Considerations for Future SCI Studies

Compared to Non-SCI subjects, low magnitude fixed frequency mechanical oscillation of the lower leg induced a similar minimal reflexive muscle activity in the two subjects with SCI. Future work should explore the mechanisms of these differences, such as alterations in post-activation depression of spinal pathways. The results of the present study suggest that constrained limb oscillation after SCI, as an experimental anti-osteoporosis intervention,

would not induce significant compressive loads due to reflex-mediated muscle contractions, although our conclusions must be tempered as we studied only two subjects with SCI. Recent animal work suggests that limb oscillation in the absence of compressive loads may offer an adequate stimulus to trigger bone anabolism (Garman et al., 2007a,b). Therefore, mechanical oscillation could be a powerful osteogenic stimulus for individuals without muscle voluntary contraction, such as those with SCI. Moreover, it may offer a new therapeutic option for people with lower motor neuron injury, who cannot trigger muscle contractions either reflexively or via electrical stimulation. Future studies should proceed to determine if mechanical oscillation can influence bone integrity in individuals with SCI. This study supports that an effect on bone density in future studies will likely be attributable to the mechanical transduction on osteoblasts rather than forces generated through high level muscle contractions.

5. Conclusions

In subjects without SCI, oscillation of the lower limb elicited significant increases in Sol and TA background EMG at high acceleration magnitudes. However, the actual magnitude of the EMG responses was small ($< 2.5\%$ MVC), suggesting that muscular loads upon the tibia were negligible. Subjects with SCI demonstrated no alteration of background EMG, even at high acceleration magnitudes. Together, these findings support that testing of mechanical oscillation for post-SCI bone preservation may not be influenced by extraneous reflex-mediated muscular loads.

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Biographies

Shuo-Hsiu Chang is currently a Research Associates at the University of Texas Health Science Center at Houston. He graduated from the Physical Therapy program at Kaohsiung Medical University in Kaohsiung, Taiwan in 1997. He worked as a physical therapist in Taiwan and received his MS and PhD degrees in Human Movement Science from The University of North Carolina at Chapel Hill in 2003 and 2007. His research interests focus on the neuromuscular control in dynamic balance and training-induced neuromuscular plasticity in elderly and individuals with neurological disorders.

Shauna Dudley-Javoroski received an advanced degree in Physical Therapy and a PhD degree in Rehabilitation Science from the University of Iowa. Her research explores the neuromuscular and skeletal adaptations that occur in humans after spinal cord injury. She has expertise in motor control and skeletal adaptations to mechanical stress. Dr. Dudley-Javoroski received awards and scholarships from the Neurology Section of the APTA, the Foundation for Physical Therapy, and the Clinical Research Award from the Iowa Physical Therapy Association. She has been supported on grants from the NIH, Christopher Reeve Foundation, and the Neilsen Foundation.

Richard K. Shields received advanced degrees in Physical Therapy from the Mayo Clinic and the University of Iowa, and a PhD degree in Exercise Science from the University of Iowa. His research explores the neuromuscular and skeletal adaptations that occur in humans during natural perturbations (fatigue, disuse, trauma, immobility, pathology, paralysis) and unnatural perturbations (vibration, electrical stimulation). The Dept. of Veterans Affairs, Neilsen Foundation, and the National Institutes of Health currently fund Dr. Shields' research. Dr. Shields received the Neurology Section Research Excellence Award, and named a Catherine Worthingham Fellow for his advancement of science, education, and clinical practice in rehabilitation.

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Fig. 1. Illustration of the constrained mechanical oscillation system. The subject was fully supported in sitting and the right foot was placed and secured on the platform. A force transducer was placed above the knee joint. The mechanical oscillation device was controlled by a computer that received feedback of acceleration. A manual abort switch was used to terminate the oscillation if necessary.

Fig. 2.

Representative trials of mechanical oscillation, Sol EMG and Sol and M-max in four continuous oscillation events in one non-SCI. In (A), three magnitudes of g forces are shown. In (B), band-stop filtered Sol EMG was shown compared to M-max (4.5 V peak to peak). Sol EMG in two events was enlarged and shown below the arrow. The magnitude of Sol EMG was about 0.04 V peak to peak.

Fig. 3.

Comparisons in SOL and TA peak EMG among various magnitudes of oscillation in non-SCI subjects. Values of estimated mean and standard errors are presented in resting (A) and 5% MVC (B). In (A), * denotes Sol peak EMG significantly greater in 5g compared to 0, 0.3, 0.6 and 1.2 g. In (B), † denotes Sol peak EMG was significantly greater in 1.2, 3 and 5g compared 0g. ‡ denotes Sol EMG was significantly greater in 5g compared to 0.3 and 0.6g. # denotes TA EMG was significantly greater in 5g compared to 0, 0.3, 0.6 and 1.2g.

Fig. 4.

Average of 30 events in processed Sol EMG in non-SCI subjects in resting (A) and 5% MVC (B). The duration of one oscillation event was 33.33 ms. In both conditions, the magnitude of EMG increased with the 5g force especially in the later part of the oscillation.

Fig. 5.

Average of 30 events in processed EMG of Sol (A) and TA (B) in one SCI subject. The duration of one oscillation event was 33.33 ms. Neither Sol nor TA EMG-peak increased from 0, 0.3, 0.6, 1.2 and 5g, nor did EMG changes occur over the course of the oscillation cycle, as was seen in non-SCI subjects (Fig. 4).

Table 1

Subject characteristics.

Subjects	Age (years, mean $\pm$ standard deviation)	SCI level	ASIA score	SCI years
Non-SCI (female = 5; male = 4)	33.5 $\pm$ 5	—	—	—
SCI-1	27	T10	A	1.2
SCI-2	45	T4	A	1.5