The mechanical loading and muscle activation of four common exercises used in osteoporosis prevention for early postmenopausal women

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Abstract

High impact exercise can reduce postmenopausal bone loss, however stimulus frequency (loading cycles per second) can affect osteogenesis. We aimed to examine the effect of stimulus frequency on the mechanical loading of four common osteoporosis prevention exercises, measuring body acceleration

and muscle activation with accelerometry and electromyography (EMG), respectively. Fourteen early postmenopausal women completed randomised countermovement jumps (CMJ), box-drops (BD), heel-drops (HD) and stamp (STP) exercises for continuous and intermittent stimulus frequencies. Sacrum accelerometry and surface electromyography (EMG) of four muscles were recorded. CMJ (mean ± SD: 10.7 ± 4.8 g & 10.0 ± 5.0 g), BD (9.6 ± 4.1 g & 9.5 ± 4.0 g) and HD (7.3 ± 3.8 g & 8.6 ± 4.4 g) conditions generated greater peak acceleration than STP (3.5 ± 1.4 g & 3.6 ± 1.7 g) across continuous and intermittent trials. CMJ and BD generated greater acceleration gradients than STP across continuous and intermittent trials. CMJ generated greater rectus femoris EMG than all other exercises, CMJ and BD generated greater semitendinosus and tibialis anterior EMG than HD across continuous and intermittent trials. CMJ and BD provide greater peak acceleration than STP and remain similar during different stimulus frequencies. CMJ, BD and HD may exceed STP in maintaining postmenopausal bone health.

Introduction

Postmenopausal women experience rapid declines in bone mineral density (BMD) leading to a greater risk of developing osteoporosis (Kroger et al., 1992; Shipman et al., 1999). High impact exercise can reduce postmenopausal BMD loss and in some cases improve BMD during the early years postmenopause, therefore reducing the likelihood of developing osteoporosis and experiencing a subsequent bone fracture (Berard et al., 1997; Wolff et al., 1999; Engelke et al., 2006; Wallace and Cumming, 2006; Kelley et al., 2012). However, there is no consensus on the most osteogenic mode of exercise (Xu et al., 2016), so it is important to establish the most effective exercises to optimise future exercise programmes targeting bone health.

Countermovement jumps (CMJ), box-drops (BD), heel-drops (HD) and stamping (STP) have all been used to stimulate increases in BMD (Hans et al., 2002; Vainionpaa et al., 2006; Young et al., 2007; Zhao et al., 2014). However, a direct comparison of these exercises regarding mechanical loading and stimulus frequencies (number of loading cycles per second) has yet to be undertaken.

Numerous loading parameters quantify the osteogenic potential of exercise, including the magnitude and gradient of impacts (Jamsa et al., 2006; Heikkinen et al., 2007), which are commonly measured with accelerometry and have shown good agreement with ground reaction forces [$R^2 = 0.812$; $p < 0.01$] (Elvin et al., 2007; Pouliot-Laforte et al, 2014). Accelerometer recordings are recommended as a surrogate measure of bone strain during physical activity due to their strong relationship [$R^2 = 0.80$; $p < 0.001$] (Burkhart et al, 2012). Accelerometer recordings at the waist have also shown good agreement with ground reaction forces [$r > 0.96$; $p < 0.001$] and therefore can also reflect site-specific and lower limb mechanical loading (Pouliot-Laforte et al, 2014). Additionally, stimulus frequency (Robling et al., 2001), and number of loading cycles are both associated with bone remodelling (Turner and Robling, 2003). Electromyography (EMG) amplitude has also been linked with femoral compressive forces (Bassey et al., 1997), and muscular forces provide a main stressor for bone remodelling capable of inducing a strong osteogenic response (Robling, 2009). As it is well established that both gravity and muscle forces are sources of bone strain (Beck, 2009), a combination of accelerometry and EMG would consequently provide a good representation of the mechanical loading and would possibly indicate the likelihood of site-specific bone remodelling.

It has been highlighted in a recent review paper (Hart et al., 2017) how mechanical loading can be administered following numerous modalities and programming variables, such as volume, intensity, frequency, distributions, rest and recovery. The same authors highlight how the control of these variables is mostly insufficient in human studies (Hart et al., 2017).

For instance, the correct load-to-rest ratio appears to be crucial for bone remodelling in animal studies. Intermittent stimulus frequencies have enhanced BMD adaptations in rodents when compared to continuous stimulus frequencies. Ten to fourteen second recovery intervals reduce bone desensitisation with repetitive continuous loading cycles and create a more potent osteogenic stimulus for bone adaptation (Robling et al., 2001; LaMothe and Zernicke, 2004; Srinivasan et al., 2007). This has never been examined in a human population however, and therefore it remains unknown if intermittent

stimulus frequencies could potentially supplant continuous mechanical loading for exercise interventions aimed at reducing postmenopausal bone loss.

Further, participation in high impact exercise relates to increases inlongitudinal bone adaptations, bone turnover markers above baseline following exercise (Erickson and Vukovich, 2010; Nilsson et al., 2012), and also greater current BMD status than control populations (Weeks and Beck, 2008; Rantalainen et al., 2010). There are however different methods of reporting loading conditions that target bone health, many of which use instrumentation with low sampling frequencies and low operating ranges, which underestimate the true magnitude of the recorded loading signal (Ziebart et al., 2017). The latter issue, combined with the previously mentioned factors governing bone adaptation, which are regularly disregarded, calls for quantification of a range of loading parameters for high impact exercises using high specification instrumentation (Daly and Bass, 2006; Lester et al., 2009; Ahola et al., 2010; Nilsson et al., 2012).

To examine whether recovery periods increase bone adaptation in early postmenopausal women, firstly it is important to identify the peak acceleration of a range of common osteoporosis prevention exercises that are used in a clinical setting. Secondly, peak accelerations generated during these exercise modes must be evaluated, using higher specification instrumentation, for consistency across continuous and intermittent stimulus frequencies to determine if there are similar body accelerations. These exercises have never been directly compared with regard to acceleration and EMG in the same population of humans, whilst controlling for loading modalities.

Therefore, the aim of this study was the measurement of mechanical loading, via both accelerations and muscle activations of four common continuous and intermittent osteoporosis prevention exercises used in clinical settings, while at the same time controlling for volume, intensity and frequency of the loading stimulus. This has potential to inform future exercise interventions targeted at improving BMD in a population of early postmenopausal women that are at a higher risk of developing osteoporosis. This study will help to identify exercises that will allow future testing of the effects of stimulus frequency

upon bone remodelling in humans. We hypothesise that the magnitude of acceleration and muscular activation will be greater during countermovement jumps (CMJ) and boxdrops (BD) than heel-drops (HD) and stamping (STP).

Methods

Overview

The fourteen healthy early postmenopausal women (mean ± SD: 55.7 ± 3.8 years, 163.0 ± 4.3 cm, 65.8 ± 11.9 kg, 1 - 5 years post-menopause, defined cessation of menstrual cycle for more than 12 months), that undertook the testing procedure were recreationally active. The study required 14 participants in order to detect a difference in peak acceleration between exercises. This was based on peak acceleration values during pilot testing of five participants, an alpha level of 0.05, and a power $(1 - \beta)$ of 0.8 (G*Power 3.0, Faul et al., 2007). Participants completed 10 repetitions each for four exercises in a continuous (stimulus frequency of 0.25 Hz, one repetition every four seconds / 15bpm) and intermittent condition (stimulus frequency of 0.067 Hz, one repetition every 15 seconds / 4bpm) in accordance with previous stimulus frequencies (Robling et al., 2001), in a randomised order. Exercises were performed barefoot in the laboratory (concrete screed floor with a 6mm vinyl composite) with an accelerometer (Noraxon, DTS 3D accelerometer 16 g, Arizona, USA) attached to the sacrum (S1, aligned with the vertical axis) and four surface electromyography electrodes (Noraxon, DTS Desk Receiver System, Arizona, USA) placed over the rectus femoris (RF), semitendinosus (ST), tibialis anterior (TA) and the lateral head of the gastrocnemius (GL) muscles. The protocol was approved by the institutional ethics committee, adhered to the World Medical Association Declaration of Helsinki and informed consent was obtained from all participants prior to testing.

Main Testing Session

Following a warm up of jogging and dynamic stretching, a 30 second static standing EMG trial was recorded to calculate baseline EMG amplitude.

Exercises consisted of 10 repeated CMJ, BD, HD and unilateral STP at both continuous and intermittent stimulus frequencies in a randomised order with 2 - 3 minutes rest between trials to reduce the effect of fatigue and keep a representative number of loading cycles (Carroll et al, 2017). Stimulus frequency, measured by an audible metronome app (Metronome Version 1.3 for iPhone, Marketwall.com) coincided with previous stimulus frequencies (Robling et al., 2001). For CMJ, participants were instructed to "jump for maximum height, using their arms and land with bent knees on the balls of their feet". BD were initiated from a 0.2 m box (Reebok Step, Reebok International Limited, Canton, MA, USA), in order to replicate the height of an average house step (Ministry of Housing, Communities & Local Government, 2013). Participants stepped backwards up onto the box before stepping off the box ensuring a 0.2 m freefall onto the floor, then immediately stepped back onto the box for the rest interval (leading with the right leg). For HD, participants were instructed to "stand as high as they can on their toes before instantly relaxing the leg muscles and dropping onto both of their heels to create an impact with knees slightly flexed". For STP, participants were instructed to stamp with their right leg "as hard as physically possible without generating discomfort". Participants were familiarised with exercises before data was recorded, trials were repeated if necessary.

Sensor Placement

Surface EMG electrodes (Ag/AgCl; Ambu Blue Sensor N, Ambu, Cambridgeshire, UK) were placed over the RF, ST, TA, and GL of the participant's right leg in accordance with SENIAM location recommendations (Hermens et al., 1999). These muscles were selected due to the forces that they exert on the femur across the hip and knee joints and the forces that they exert on the tibia. Two rectangular

electrodes (22 x 30 mm) per muscle placed with an inter-electrode distance of 22 mm. Skin was shaved, abraded and cleansed with a 70% alcohol swab before electrode attachment. The accelerometer was vertically aligned and attached to the participant's sacrum (S1) due to the flatter attachment site, to estimate whole body acceleration in close proximity to the hip and lumbar spine which are at risk of osteoporotic fracture (Kelley et al., 2014; Heilmeier et al., 2016). Accelerometer and EMG wearable hardware were secured with surgical tape and elasticated bandages to reduce signal artefacts.

Accelerometry

Vertical accelerometer data (output from the vertical axis) were filtered optimally using 95% of the signal energy of the data from a base value of 1 g before the impact to the impact peak for each of the 10 highest impact peaks per exercise. Accelerometer data were presented in g while acceleration gradients from 1 g to the peak, were converted to $m \cdot s^{-3}$. The 10 highest peaks in acceleration data were calculated using the peak from each loading cycle across 10 cycles. The 10 corresponding acceleration gradients from 1 g (9.81 $m \cdot s^{-2}$) leading to the peak for each loading cycle were calculated using the following formula (Heikkinen et al., 2007) (Fig.1):

Acceleration gradient = (peak acceleration – 9.81 $m \cdot s^{-2}$) / (t @ peak acceleration – t @ 9.81 $m \cdot s^{-1}$)

(Eq. 1)

Where "t @" denotes the time point at either peak acceleration or at 9.81 $m \cdot s^{-2}$

Peak ACC and Grad ACC were calculated as the average peak acceleration and acceleration gradient across the 10 loading cycles respectively.

Electromyography

For all trials, vertical acceleration and EMG were recorded synchronously at 1500 Hz (EMG: input impedance > 100 MΩ, CMRR > 100 dB, baseline noise < 1μV RMS, base gain = 200, final gain = 500). acceleration and EMG were automatically synchronised with Noraxon hardware (Noraxon, DTS Desk Receiver System, Arizona, USA). EMG data were band-pass filtered (bi-directional Butterworth, 10-500 Hz), full wave rectified and low-pass filtered at 8 Hz to obtain linear envelopes (Shaharudin et al., 2014). Baseline EMG amplitude for each muscle was calculated as 3 standard deviations above the mean value from the standing static trial and was removed from the exercise trials for each muscle leaving EMG activity due to exercise only. Baseline EMG was therefore eliminated both before and after the EMG bursts from each loading cycle, in order to remove the EMG that was recorded during the rest intervals (Hodges and Bui, 1996; Prosser et al., 2011) (Fig.2). EMG Mean amplitude was calculated using the mean of the EMG activity from each loading cycle that was above the baseline EMG amplitude from 2.5 seconds before the first impact peak to 2.5 seconds after the last impact peak as determined by the acceleration data. EMG Mean amplitude was averaged across the 10 loading cycles for each exercise and was normalised to the corresponding CMJ continuous trial, which showed the highest mean values for the RF and ST muscles (Prosser et al., 2011).

Statistical Analysis

Data were processed using a customised MATLAB programme (MATLAB R2011a, Mathworks, Cambridge, UK). Parametric data (ST EMG, GL EMG) were statistically analysed using two-way (4 exercises x 2 stimulus frequencies) repeated measures ANOVAs (Sidak adjustments) with post-hoc pairwise comparisons using SPSS (IBM SPSS Statistics Version 20.0. IBM Corp, NY, USA). Peak ACC, Grad ACC and TA EMG variables were non-normally distributed data but were logtransformed in order to show a normal distribution and analysed using parametric methods. A published spreadsheet was used to calculate the magnitude of the effect using Cohen's *d* effect size thresholds (0 - 0.19 trivial, 0.2 - 0.59 small, 0.6 - 1.19 moderate, 1.2 - 1.99 large, 2.0 - 3.99 very large) (Hopkins, 2006). Magnitude

based inferences were made based on the smallest worthwhile change (calculated as 0.2 multiplied by the between-subject SD). Uncertainty in the population estimates were expressed as 95% confidence intervals. For RF EMG log-transformation did not allow for a normal distribution, therefore for the original non-parametric data, Friedman's tests were used to compare main effects of exercise and stimulus frequency conditions. Wilcoxon signed-rank tests with

Bonferroni corrections were used for pairwise comparisons and post-hoc tests in SPSS. Cliff's Delta (δ) effect size was calculated in R (R Foundation for Statistical Computing 3.2.1, Vienna, Austria) effsize package (Torchiano, 2016), and evaluated using the following scale: 0 - 0.146 trivial, 0.147 - 0.32 small, 0.33 - 0.473 moderate, >0.474 large. Uncertainty in the population estimates were expressed as 95% confidence intervals (Cliff, 1996).

Results

Acceleration

For peak acceleration there was a significant main effect for exercise ($P < 0.001$), but not for stimulus frequency ($P = 0.433$). During continuous conditions, CMJ were statistically significantly higher than HD and STP ($d = 0.83$ [95% CI: 0.27 to 1.4], $P = 0.043$; $d = 2.38$ [95% CI: 1.36 to 3.39], $P = 0.001$). BD and HD were statistically significantly higher than STP ($d = 2.17$ [95% CI: 1.13 to 3.2], $P = 0.003$; $d = 1.54$ [95% CI: 0.71 to 2.37], $P = 0.009$). For peak acceleration during intermittent conditions, CMJ, BD and HD were statistically significantly higher than STP ($d = 2.21$ [95% CI: 1.15 to 3.26], $P = 0.004$; $d = 2.15$ [95% CI: 1.2 to 3.1], $P = 0.002$; $d = 1.88$ [95% CI: 0.98 to 2.79], $P = 0.004$). Intermittent HD were statistically significantly higher than continuous HD ($d = 0.33$ [95% CI:

0.00 to 0.65], $P = 0.047$) (Fig.3).

For acceleration gradients there was a main effect for exercise (P < 0.001) but not for stimulus frequency (P = 0.097). During continuous conditions, CMJ and BD were statistically significantly higher than STP (*d* = 1.89 [95% CI: 1.03 to 2.75], *P* = 0.002; *d* = 1.74 [95% CI: 0.88 to 2.61], *P* = 0.005). For acceleration gradients during intermittent conditions, CMJ, BD and HD were statistically significantly higher than STP (*d* = 1.77 [95% CI: 0.89 to 2.66], *P* = 0.005; *d* = 1.78 [95% CI: 1.02 to 2.54], *P* = 0.001; *d* = 1.57 [95% CI: 0.7 to 2.44], *P* = 0.011) (Fig.3).

Electromyography

ST and GL EMG amplitude were significantly higher during continuous conditions than intermittent conditions for all exercises. RF EMG amplitude was significantly higher during continuous conditions than intermittent conditions for CMJ, BD and STP exercises. TA EMG amplitude was significantly higher during continuous conditions than intermittent conditions for BD and STP exercises. (Table 1).

For rectus femoris EMG amplitude there was a main effect for exercise (P < 0.001) and stimulus frequency (P < 0.001). During continuous and intermittent conditions, CMJ were statistically significantly higher than all other exercises. BD were statistically significantly higher than HD and STP. During only the continuous condition STP were statistically significantly higher than HD (Table 2).

For semitendinosus EMG amplitude there was a main effect for exercise (P < 0.001) and stimulus frequency (P < 0.001). During continuous conditions, all exercises were statistically significantly higher than HD. For intermittent conditions, CMJ and BD were statistically significantly higher than HD (Table 2).

For tibialis anterior EMG amplitude there was a significant effect of exercise (P = 0.006) and stimulus frequency (P = 0.029). During continuous and intermittent conditions, CMJ and BD were statistically significantly higher than HD (Table 2).

For lateral head of the gastrocnemius EMG amplitude there was a main effect for exercise ($P < 0.001$) and stimulus frequency ($P < 0.001$). During continuous conditions, CMJ were statistically significantly higher than BD and STP. For continuous and intermittent conditions, HD were statistically significantly higher than BD and STP. For intermittent conditions only, CMJ were statistically significantly higher than BD (Table 2).

Discussion

Main findings showed that CMJ could provide a greater osteogenic stimulus than HD and STP exercises due to the greater Peak ACC, Grad ACC and RF EMG parameters. However, the CMJ, BD and HD exercises all provided a greater peak acceleration than STP, where CMJ and BD were consistent in continuous and intermittent peak accelerations.

Acceleration

CMJ, BD and HD produced favourable mechanical loading in terms of greater Peak ACC when compared to STP across continuous and intermittent trials, and a potentially greater stimulus for bone adaptation (Turner and Robling, 2003) (Fig.3). Differences highlighted in Peak ACC values were meaningful in surpassing 0.17 g, the minimal detectable change (Turcot et al., 2008). Peak ACC values were higher than reported in previous research but was likely due to the higher sampling frequency in the current investigation (Vainionpaa et al., 2006; Heikkinen et al., 2007). All exercises, apart from STP, consistently surpassed the previously reported osteogenic threshold of 4.9 g (accounting for 1 g being standing in the current study) (Vainionpaa et al., 2006), this would suggest that STP would be insufficient for stimulating an osteogenic response. However, comparisons between studies using different accelerometer instrumentation should be cautiously interpreted, particularly when considering the current study used a sampling frequency of 1500 Hz and the osteogenic threshold of 4.9 g was

established using instrumentation sampled at 400 Hz (Vainionpaa et al., 2006; Ziebart et al., 2017). Whilst CMJ and BD produce higher Peak ACC, HD were similar in mechanical loading parameters and could provide a useful alternative for early postmenopausal women that wish to maintain BMD but cannot tolerate CMJ or BD (Hans et al., 2002). Current participants were able to tolerate CMJ and BD exercises which was confirmed verbally during the testing session. With inconsistent continuous and intermittent Peak ACC during HD, the most preferable exercises for the evaluation of the effects of continuous and intermittent stimulus frequencies on mechanical bone remodelling with early postmenopausal women would be CMJ and BD as they produce consistent continuous and intermittent peak accelerations. Both HD and STP may provide some advantages in terms of injury prevention as they allow for a constant connection with the floor and therefore may reduce the potential for lower limb landing based injuries.

Grad ACC data showed that all exercises surpassed the osteogenic threshold of 1000 $m \cdot s^{-3}$ (Heikkinen et al., 2007), although yet again, comparisons with previous studies may be complicated by different instrumentation (Ziebart et al, 2017). CMJ and BD could provide greater stimulus for bone adaptation than STP, due to greater Grad ACC (Heikkinen et al., 2007) (Fig.3). HD were the only exercise to display greater Grad ACC during the intermittent trial than the continuous trial. Stimulus frequency had an effect on the loading experienced during the HD, which had been uncontrolled previously (Hans et al., 2002). Lower continuous Peak ACC and Grad ACC during HD might be explained by gastrocnemius relaxation difficulties during the brief rest interval with continuous trials, although this would require further investigation.

Electromyography

CMJ could indicate greater femoral compressive forces due to greater RF EMG than BD, HD and STP (Bassey et al., 1997). CMJ and BD generated greater ST EMG and TA EMG when compared to

HD. CMJ generated greater GL EMG when compared to BD, across continuous and intermittent trials (Table.1). The strong linear relationship between EMG amplitude and force output would suggest that higher EMG amplitudes subject the skeleton to greater internal loading forces (Andrade and Andrade, 2012). HD generated the lowest EMG amplitude across RF, ST and TA but the greatest GL EMG. This was expected due to small involvement from joints other than the ankle. While HD may not generate as higher activation in the muscles compressing the femur and spanning the hip as CMJ and BD, HD could therefore provide greater peripheral muscular activation in the lower limb, which could lead to greater peripheral bone adaptations than BD and STP (Hans et al., 2002). However, causal evidence linking higher EMG amplitude with greater BMD adaptations is lacking and as direct internal force measurements were not possible, the EMG amplitude is only suggestive of internal loading. Bone strain cannot be inferred from EMG measures currently, this study adds to the current body of literature in this area but a potential relationship between EMG amplitudes and bone strain remains undetermined. The greater EMG amplitude generated by continuous conditions when compared to intermittent conditions, might have arisen from the shorter rest interval during continuous trials inducing greater fatigue and higher motor unit recruitment (Carneiro et al., 2010), although this requires further research. Despite the acceleration showing no statistical difference between continuous and intermittent conditions for CMJ, BD and STP exercises, the EMG amplitude data indicates that internal muscular activation could be substantially different from altering the stimulus frequency.

Limitations

The peak accelerations are purely an indicator of potential BMD adaptations and do not always indicate changes in BMD or bone geometry. Peak ACC is used as an indicator of bone strain and occasionally may not accurately reflect the internal loading despite being highly related (Edwards et al., 2009; Burkhart et al, 2012). While the acceleration magnitude reflects BMD adaptations (Rantalainen et al., 2009; Ahola et al., 2010; Erickson and Vukovich, 2010; Reiger and Yingling,

2015), other factors that may also govern bone adaptation are usually overlooked e.g. stimulus frequency, acceleration gradient, muscular activation (Robling et al., 2001; Heikkinen et al., 2007; Robling, 2009). Future research should incorporate these measures when considering mechanical loading parameters. Comparisons with previous studies and suggested thresholds for bone adaptation are difficult due to differences in accelerometer instrumentation and processing (Jamsa et al, 2011), with low cut-off filter frequencies, lower sampling frequencies and lower operating ranges eradicating higher frequency components of the signal which are integral to mechanical loading stimuli and osteogenesis (Kelley et al., 2014; Ziebart et al., 2017). It is therefore recommended that future research report instrumentation, attachment and data processing methods precisely to allow better comparison across studies.

Conclusions

CMJ consistently produced the highest Peak ACC, Grad ACC and RF EMG, which would suggest that CMJ could provide a greater osteogenic stimulus than HD and STP exercises. However, CMJ, BD and HD exercises all provided a greater peak acceleration than STP. CMJ and BD provide consistent continuous and intermittent peak acceleration for the assessment of stimulus frequency on bone adaptation. Continuous conditions showed a tendency to generate greater EMG amplitude than intermittent conditions. This study has highlighted appropriate exercises that can inform future high impact training programmes targeted at early postmenopausal BMD maintenance and can provide valuable information for exercise and osteoporosis prevention guidelines throughout the ageing process.

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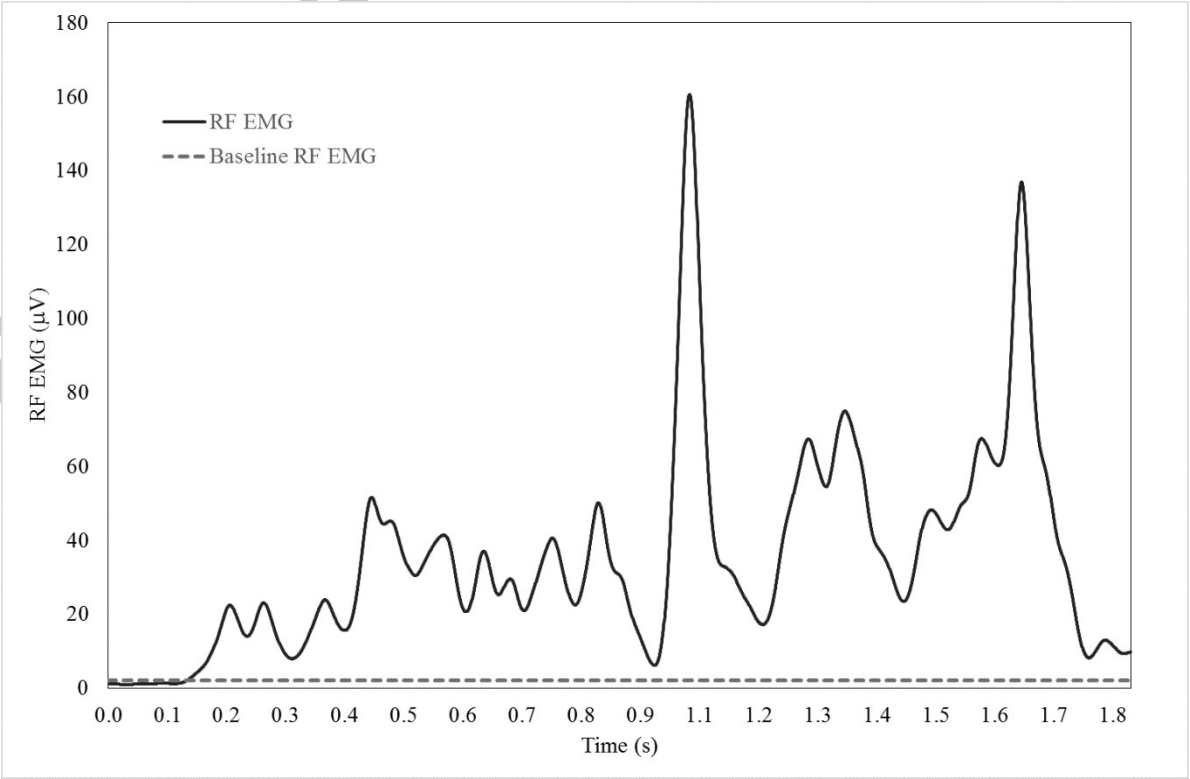
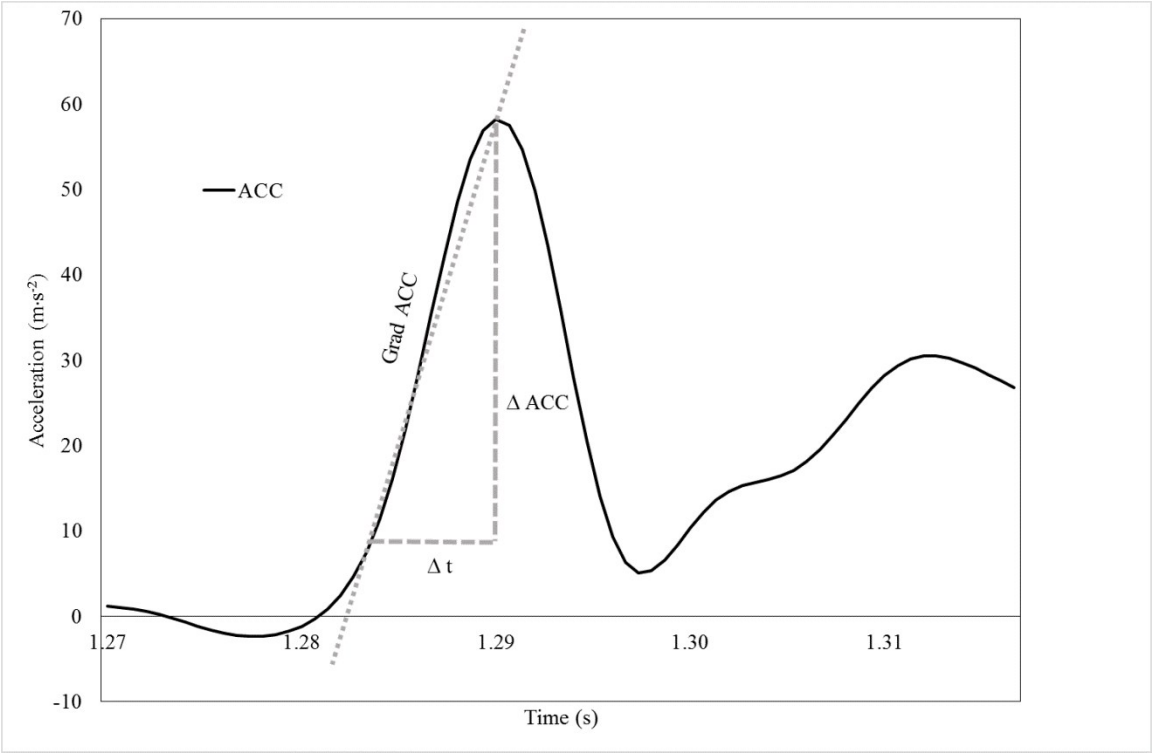
Figures

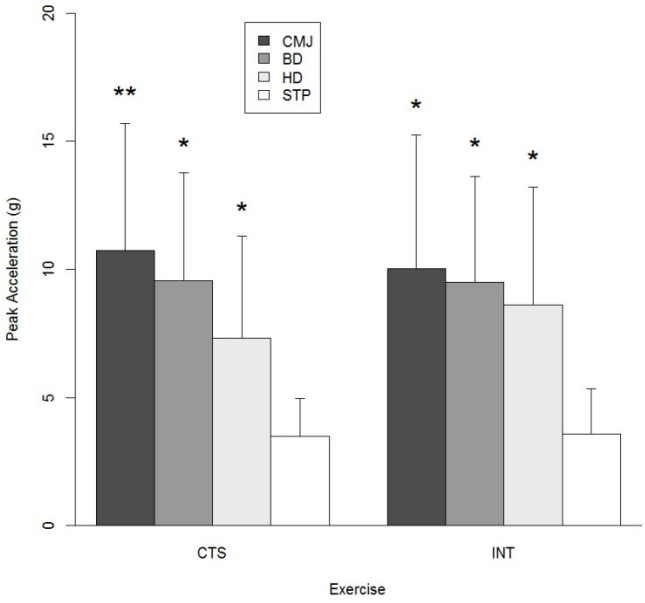
Fig. 1. Acceleration trace for one loading cycle of the box-drop exercise. Solid black line indicates the filtered acceleration trace. □ACC indicates the change in the acceleration, □t indicates the change in time. Grad ACC indicates the acceleration gradient as highlighted by the dotted grey line.

Fig. 2. Rectus femoris electromyography trace for one loading cycle of the box-drop exercise. Solid black line indicates the filtered rectus femoris electromyography trace before normalisation has occurred. Dashed grey line indicates the baseline rectus femoris electromyography amplitude as calculated from a standing static trial.

Fig. 3. Mean (±SD) peak acceleration and acceleration gradient of countermovement jumps (CMJ), box-drops (BD), heel-drops (HD) and stamps (STP) when performed both continuously and intermittently. Bars are means, error bars are SD. ** indicates condition is statistically significantly higher than the corresponding HD and STP conditions, * indicates condition is statistically significantly higher than the corresponding STP condition. n = 14






Peak Acceleration (g)
0
5
10
15
20
CMJ
BD
HD
STP
**
*
*
*
*
*
CTS
INT
Exercise

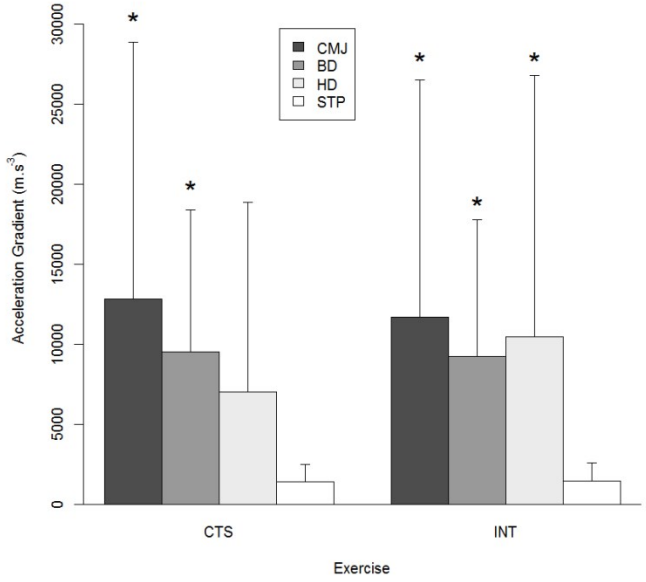

Acceleration Gradient (m.s^{-3})
0
5000
10000
15000
20000
25000
30000
CMJ
BD
HD
STP
*
*
*
*
*
CTS
INT
Exercise

Table 1

EMG amplitude for each of the four muscles across countermovement jump, box drop, heel drop and stamp exercises showing continuous and intermittent stimulus frequency comparisons. EMG amplitude is normalised to the continuous countermovement jump trial and presented as a percentage.

RF	CTS	INT	CTS-INT
CMJ	100 ± 0	87 ± 23	δ = 0.57 [95% CI: 0.27 to 0.77], *P* = 0.009***
BD	85 ± 11	69 ± 10	δ = 0.85 [95% CI: 0.48 to 0.96], *P* = 0.001***
HD	40 ± 26	38 ± 19	δ = 0.04 [95% CI: -0.39 to 0.46], *P* = 0.975
STP	68 ± 24	51 ± 32	δ = 0.41 [95% CI: -0.04 to 0.72], *P* = 0.008**
ST			
CMJ	100 ± 0	78 ± 15	*d* = 1.32 [95% CI: 0.78 to 1.86], *P* < 0.001***
BD	97 ± 23	73 ± 16	*d* = 1.4 [95% CI: 0.57 to 2.22], *P* = 0.003***
HD	62 ± 19	51 ± 18	*d* = 0.58 [95% CI: 0.19 to 0.97], *P* = 0.006*
STP	88 ± 17	63 ± 21	*d* = 1.12 [95% CI: 0.76 to 1.49], *P* < 0.001***
TA			
CMJ			
BD	100 ± 0	93 ± 19	*d* = 0.32 [95% CI: -0.23 to 0.86], *P* = 0.133
HD	106 ± 24	88 ± 22	*d* = 0.68 [95% CI: 0.29 to 1.07], *P* = 0.002**
STP	72 ± 22	71 ± 17	*d* = 0.05 [95% CI: -0.51 to 0.62], *P* = 0.929
GL	94 ± 39	86 ± 46	*d* = 0.23 [95% CI: 0.00 to 0.46], *P* = 0.048*
CMJ			
BD	100 ± 0	78 ± 19	*d* = 1.08 [95% CI: 0.54 to 1.63], *P* < 0.001**
HD	78 ± 14	64 ± 12	*d* = 1.08 [95% CI: 0.6 to 1.57], *P* < 0.001**
STP	125 ± 32	106 ± 37	*d* = 0.47 [95% CI: 0.07 to 0.87], *P* < 0.001*
	69 ± 20	58 ± 23	*d* = 0.42 [95% CI: 0.07 to 0.77], *P* < 0.001*

* Rectus femoris data are presented as medians ± interquartile range with Cliff's Delta (δ) [95% confidence intervals], *p* value

RF, rectus femoris; ST, semitendinosus; TA, tibialis anterior; GL, lateral head of the gastrocnemius; CTS, continuous condition; INT, intermittent condition

CMJ, countermovement jump; BD, box drop; HD, heel drop; STP, stamp; n = 14, magnitude of difference indicated by; * small, ** moderate, *** large, **** very large

Table 2

EMG amplitude for each of the four muscles across continuous and intermittent conditions whilst performing countermovement jump, box drop, heel drop and stamp exercises. EMG amplitude is normalised to the continuous countermovement jump trial and presented as a percentage.

	EMG Amplitude (Mean±SD)			Post-hoc pairwise comparisons (Cohen's *d* [95% confidence intervals], *p* value)					
	CMJ	**BD**	**HD**	**STP**	**CMJ-BD**	**CMJ-HD**	**CMJ-STP**	**BDHD**	**BD-STP**	**HD-STP**
CTS	100 ± 0	85 ± 11	40 ± 26	68 ± 24	δ = 0.86 [95% CI: 0.64 to 0.95], *P* = 0.011***	δ = 0.86 [95% CI: 0.64 to 0.95], *P* = 0.001***	δ = 0.93 [95% CI 0.75 to 1.00], *P* = 0.001***	δ = 0.86 [95% CI: 0.36 to 0.98], *P* = 0.001***	δ = 0.82 [95% CI: 0.48 to 0.94], *P* = 0.002***	δ = 0.56 [95% CI: -0.82 to 0.11], *P* = 0.019***
RF	100 ± 0	97 ± 23	62 ± 19	88 ± 17	*d* = 0.12 [95% CI: -0.66 to 0.90], *P* = 0.998	*d* = 1.83 [95% CI: 1.05 to 2.62], *P* < 0.001***	*d* = 0.62 [95% CI: -0.16 to 1.40], *P* = 0.159**	*d* = 1.69 [95% CI: 0.68 to 2.69], *P* = 0.001***	*d* = 0.46 [95% CI: -0.71 to 1.62], *P* = 0.815	*d* = 1.40 [95% CI: -2.52 to 0.28], *P* = 0.011***
	100 ± 0	106 ± 24	72 ± 22	94 ± 39	*d* = -0.24 [95% CI: -1.02 to 0.54], *P* = 0.995	*d* = 1.16 [95% CI: 0.38 to 1.94], *P* = 0.006**	*d* = 0.14 [95% CI: -0.64 to 0.92], *P* = 0.772	*d* = 1.43 [95% CI: 0.71 to 2.15], *P* < 0.001***	*d* = 0.29 [95% CI: -0.51 to 1.09], *P* = 0.631	*d* = 0.49 [95% CI: -1.39 to 0.41], *P* = 0.700

STA GL INT	100 ± 0	78 ± 14	125 ± 32	69 ± 20	*d* = 1.40 [95% CI: 0.62 to 2.18], *P* = 0.001***	*d* = -0.73 [95% CI: -1.51 to 0.06], *P* = 0.075	*d* = 1.43 [95% CI: 0.65 to 2.21], *P* < 0.001***	*d* = 1.35 [95% CI: -2.17 to 0.53], *P* = 0.001***	*d* = 0.42 [95% CI: -0.50 to 1.33], *P* = 0.697	*d* = 2.60 [95% CI: 1.52 to 3.68], *P* < 0.001***
RF	87 ± 23	69 ± 10	38 ± 19	51 ± 32	δ = 0.69 [95% CI: 0.31 to 0.88], *P* = 0.001***	δ = 0.92 [95% CI 0.48 to 1.00], *P* = 0.001***	δ = 0.89 [95% CI: 0.64 to 0.97], *P* = 0.001***	δ = 0.83 [95% CI: 0.50 to 0.95], *P* = 0.001***	δ = 0.54 [95% CI: 0.08 to 0.81], *P* = 0.002***	δ = 0.22 [95% CI: -0.57 to 0.18], *P* = 0.096

ST	78 ± 15	73 ± 16	51 ± 18	63 ± 21				*d* = 1.12 [95 % CI: 0.56 to 1.68], *P* = 0.001**	*d* = 0.44 [95 % CI: -0.27 to 1.16], *P* = 0.384	*d* = 0.51 [95 % CI: -1.14 to 0.12], *P* = 0.143
TA	93 ± 19	88 ± 22	71 ± 17	86 ± 46	*d* = 0.33 [95% CI: -0.45 to 1.10], *P* = 0.766	*d* = 1.41 [95% CI: 0.47 to 2.36], *P* = 0.003***	*d* = 0.69 [95% CI: -0.13 to 1.50], *P* = 0.120	*d* = 0.88 [95 % CI: 0.10 to 1.66], *P* = 0.016**	*d* = 0.04 [95 % CI: -0.58 to 0.66], *P* = 0.912	*d* = 0.29 [95 % CI: -0.99 to 0.42], *P* = 0.994
GL	78 ± 19	64 ± 12	106 ± 37	58 ± 23	*d* = 0.22 [95% CI: -0.49 to 0.93], *P* = 0.851	*d* = 1.16 [95% CI: 0.36 to 1.97], *P* = 0.004**	*d* = 0.15 [95% CI: -0.50 to 0.81], *P* = 0.524	*d* = 1.06 [95 % CI: -1.85 to 0.28], *P* = 0.006**	*d* = 0.20 [95 % CI: -0.57 to 0.98], *P* = 0.965	*d* = 1.93 [95 % CI: 0.82 to 3.04], *P* = 0.001***
					d = 1.05 [95% CI: 0.01 to 2.08], *P* = 0.046**	*d* = -0.72 [95% CI: -1.46 to 0.03], *P* = 0.061	*d* = 0.77 [95% CI: -0.14 to 1.68], *P* = 0.120			

* Rectus femoris data are presented as medians ± interquartile range with Cliff's Delta (δ) [95% confidence intervals], *p* value

RF, rectus femoris; ST, semitendinosus; TA, tibialis anterior; GL, lateral head of the gastrocnemius; CTS, continuous condition; INT, intermittent condition

CMJ, countermovement jump; BD, box drop; HD, heel drop; STP, stamp; n = 14, magnitude of difference indicated by; * small, ** moderate, *** large, **** very large **Biography**

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- Mechanical loading and bone remodelling.
- Explosive strength and neuromuscular performance.

Published Papers

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