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Muscle activity reduces soft-tissue resonance at heel-strike during walking

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Abstract

Muscle activity has previously been suggested to minimize soft-tissue resonance which occurs at heel-strike during walking and running. If this concept were true then the greatest vibration damping would occur when the input force was closest to the resonant frequency of the soft-tissues at heel-strike. However, this idea has not been tested. The purpose of this study was to test whether muscle activity in the lower extremity is used to damp soft-tissue resonance which occurs at heel-strike during walking. Hard and soft shoe conditions were tested in a randomized block design. Ground reaction forces, soft-tissue accelerations and myoelectric activity were measured during walking for 40 subjects. Soft-tissue mass was estimated from anthropologic measurements, allowing inertial forces in the soft-tissues to be calculated. The force transfer from the ground to the tissues was compared with changes in the muscle activity. The soft condition resulted in relative frequencies (input/tissue) to be closer to resonance for the main soft-tissue groups. However, no increase in force transmission was observed. Therefore, the vibration damping in the tissues must have increased. This increase concurred with increases in the muscle activity for the biceps femoris and lateral gastrocnemius. The evidence supports the proposal that muscle activity damps soft-tissue resonance at heel-strike. Muscles generate forces which act across the joints and, therefore, shoe design may be used to modify muscle activity and thus joint loading during walking and running.

Keywords: Vibration; EMG; Joint loading

1. Introduction

Ground reaction forces act between the ground and the foot of a person during walking and running. Ground reaction forces typically have an impact peak after heel-strike due to the collision of the foot with the ground. The impact force occurs within 50 ms after first contact and causes shock waves to travel through both the soft-tissues and skeletal components of the body. The impact force should be expected to cause oscillations in the wobbling structures of the body, and the tissues may resonante if their natural frequencies are close to the frequency of the input force.

It has been proposed that lower extremity muscle activity adapts to the impact force which occurs during heel-strike in order to minimize soft-tissue resonance (Nigg et al., 1995). If such a situation occurs then it may be possible to use shoe materials to modify the impact force and thus vibration load on the body in order to cause specific alterations in the muscle activity and thus joint loading. Previous studies have shown that the hardness of a shoe midsole causes changes in the time to peak impact force at heel-strike (Light et al., 1980; Frederick et al., 1984; Nigg et al., 1987; Lafortune et al., 1996). This time and the associated loading rate are a correlate of the major frequency content of the impact force (typically 10–20 Hz for running) and thus indicate that different shoes will result in different vibration loads on the tissues. The natural frequencies of the soft-tissues in the lower extremity range between 10 and 50 Hz (Wakeling and Nigg, 2001a), and so may potentially resonate due to the heel-strike impacts. However, both the natural frequency and damping coefficients of the soft-tissues of the lower extremity

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change with altered muscle activity (Wakeling and Nigg, 2001a), indicating that muscle has the capability to alter the vibration response of the soft-tissues and the potential to minimize the soft-tissue resonance. During controlled vibration experiments, it was shown that muscle activity was used to increase vibration damping when an input frequency approached the resonance frequency of the tissue (Wakeling et al., 2002a). It has been shown that lower extremity muscle activity does adapt to the altered impact forces during running with different shoe midsole materials (Wakeling et al., 2001b, 2002b). However, it has not been shown that the muscle adaptation which occurs in response to the impact force during walking and running is related to the soft-tissue vibration frequencies.

The purpose of this experiment was to measure changes in the muscle activity and soft-tissue vibrations which occurred during walking with different shoe conditions. The hypothesis to be tested was that when the input frequency of the ground reaction force approached the vibration frequency of the soft-tissues then the muscle activity within those tissues would increase in order to minimize the vibrations.

2. Methods

2.1. Subjects

Twenty male (age 26.9 ± 1.0 yr; mass 78.24 ± 3.20 kg; mean ± SEM) and 20 female (age 25.8 ± 1.1 yr; mass 67.61 ± 1.60 kg) subjects were tested. The subjects were students at the University of Calgary. Subjects gave their informed, written, consent to participate in accordance with the University of Calgary's Conjoint Health Research Ethics Board policy on research using human subjects.

Muscle and soft-tissue masses within the lower extremity were estimated from a series of length, breadth and skin fold measurements using methods previously described (Wakeling et al., 2002a).

2.2. Protocol

Subjects were instructed to walk at a brisk pace along a 20 m indoor track. A force plate (Kistler AG, Winterthur, Switzerland) was placed in the centre of the track, level with the ground. Initial trials determined the starting position, which would result in the fifth step of the right foot striking the force plate. During the experiment, subjects were requested not to focus on the position of the force plate. Timing lights, spanning the force plate, were used to measure the walking velocity. The timing of heel-strike was determined by an accelerometer attached to the heel-cup of the right shoe.

The control condition was a hard-soled leather shoe (Asker Shore C 77, 320 g). A soft (Asker Shore C 28) heel-cup insert was used as the insert condition. This insert had mass 26.2 g, and a maximum thickness of 4.5 mm which compressed to approximately 1.5 mm with body weight. The starting condition was randomized. Each condition was tested in blocks of 10 consecutive trials. The blocks were tested in the order ABBAAB, where A and B represent the two different conditions. This block design was used to minimize bias in the results due to muscle fatigue. For each trial, data were collected from a standing posture for 2 s, a subsequent walk of 10 double paces, and a final 2 s stand. Data were analysed for the middle eight steps from each trial.

2.3. Outcome measures

The ground reaction force, the myoelectric signals from the lower extremity muscles and the soft-tissue accelerations were recorded at 2400 Hz using a 12-bit data acquisition system. Soft-tissue vibrations were measured from the muscle bellies of the vastus lateralis, biceps femoris (long head), tibialis anterior and lateral gastrocnemius using skin-mounted tri-axial accelerometers (EGAX accelerometer, nominal frequency response 0–600 Hz; Entran devices). The axes were orientated to be parallel to the long axis of the segment, normal to the skin, and medio-lateral. Accelerometers (<5 g) were attached to the skin surface, 1 cm distal to the EMG electrode, using Hollister medical adhesive glue, and a stretch adhesive bandage preloaded the accelerometer to improve the congruence of motion with the soft-tissues (Wakeling and Nigg, 2001a, b). Myoelectric activity was recorded from the rectus femoris, biceps femoris (long head), tibialis anterior and lateral gastrocnemius muscles. Myoelectric activity was measured from the muscle bellies using round bipolar surface electrodes (Ag/AgCl) after removal of the hair and cleaning of the skin with isopropyl wipes. Each electrode was 10 mm in diameter and had an interelectrode spacing of 22 mm and was placed midway between the motor end plate and distal myotendinous junction. A ground electrode was placed on the lateral condyle of the knee. EMGs were preamplified at source (Biovision, Wehrheim, Germany).

2.4. Analysis

Initial analysis showed high frequency oscillations (>300 Hz) in the vertical ground reaction forces. These oscillations were removed by a 100 Hz low-pass filter, and the mean ground reaction force calculated from the 30 trials for each subject-condition combination and the vertical impact force was quantified for each trial by its maximum loading rate, $\dot{F}_{\mathrm{GRF,max}}$, and

Table 1
Coefficients for the wavelets used to analyze high- and low-frequency components from the myoelectric signal

Frequency	f_c (Hz)	s	τ (ms)	Frequency band (Hz)
High	30.91	0.0997	2.1	12–63
Low	150.95	0.0302	7.1	71–274

Center-frequency f_c, scale s, time resolution τ.

its peak value, $F_{GRF,max}$. An effective input frequency, f_{GRF}, was estimated from four times the period between the maximum loading rate and the peak impact force.

The mean tissue acceleration during the 2 s static standing period prior to each walking trial was subtracted from the acceleration records, to reference the accelerations to a vertical standing posture. The mean acceleration traces were calculated from the 240 steps from each subject-muscle-axis-condition combination. Inertial tissue forces were calculated for each direction from the product of the mean referenced acceleration in that direction and the soft-tissue mass. Inertial forces were quantified by their maximum absolute loading rate after heel-strike, the maximum and minimum peak forces surrounding the maximum inertial loading rate. The vibration frequency was estimated from twice the period between the maximum and minimum forces.

Myoelectric signals were resolved into their intensities in time-frequency space using EMG specific wavelet techniques (von Tscharner, 2000). The intensity is the power of the EMG signal contained within a particular frequency band. The total intensity over the frequency band 11–432 Hz was calculated using a filter-bank of 11 wavelets with time resolutions from 45 to 12 ms. The intensity was also calculated for specific high and low-frequency bands using two wavelets $W(f)$ which were defined as the following function of frequency, f:

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

where f_c is the centre frequency of the wavelet, and s is a scaling factor. Parameters defining these two wavelets are given in Table 1. The intensities at the high and low-frequency bands were calculated in the same manner as the intensity for each wavelet from the previously reported filter-bank (von Tscharner, 2000).

The mean intensity trace was calculated for the 240 steps from each subject-muscle-condition combination, and was calculated for 50 ms time windows before and after heel-strike for the total intensity, and for 10 ms time windows spanning 50 ms before to 50 ms after heel-strike for the high- and low-frequency bands.

2.5. Statistics

The effects of the shoe condition on the measures of the GRF, vibration and EMG were determined using multifactorial analyses of variance. In each test the subject identity and the shoe condition were used as factors. Relative changes in parameters between the shoe conditions were calculated as the difference between the insert and control relative to the control value. All tests were considered significant at the $\alpha = 0.05$ level. Mean values are presented with the standard error of sample mean (SEM).

3. Results

The subjects walked at a velocity of $2.10 \pm 0.01\,\mathrm{m\,s^{-1}}$ (mean $\pm$ SEM) with a stance duration of 549 ± 2 ms. Analysis of variance showed that there were significant differences in both the walking velocity and stance duration between subjects, however, there were no significant effects of the shoe condition on these parameters.

3.1. Ground reaction forces

The vertical ground reaction force showed an impact peak at 22 ms, and this impact peak was followed by a second peak at 50 ms for 35 of the 40 subjects (Fig. 1A). The ground reaction forces then showed large and broad peaks which are characteristic of walking. Superimposed on this characteristic pattern there were oscillations of small magnitude and high frequency (> 300 Hz). Parameters quantifying the impact peak were calculated after these high-frequency oscillations were filtered and are shown in Table 2.

Analysis of variance of the GRF parameters showed that there were significant effects of the shoe condition on the loading rate (15.9% decrease for the insert condition), and the effective input frequency (9.4% decrease), however there were no significant effects of the shoe condition on the magnitude of the impact peak.

3.2. Soft-tissue vibrations

Transient peaks of inertial force occurred in all soft-tissue groups after heel-strike (e.g. Figs. 2 and 3). The magnitude of the peak forces (Table 3) varied with their direction relative to the leg segment. When walking with the control condition, the mean vibration frequencies weighted by the magnitude of the peak inertial forces were 26.13 ± 1.02, 24.26 ± 1.28, 38.54 ± 1.16 and 28.62 ± 1.03 Hz ($N = 40$) for the quadriceps, hamstrings, tibialis anterior and triceps surae soft-tissue groups, respectively.

There were significant effects of the insert condition on the maximum inertial loading rate for 10 of the possible 12 tissue-direction combinations (Table 3) with a reduction in the maximum inertial loading rate with the insert condition (e.g. 23.3% for the axial direction in the tibialis anterior). There were significant effects of the insert condition on the maximum inertial force for five of the possible 12 tissue-direction combinations (reduction for the insert condition; greatest reduction of 19.5% for the axial direction in the tibialis anterior). There were significant effects of the insert condition on the vibration frequency for four of the possible 12 tissue-direction combinations (reduction for the insert condition; greatest reduction of 16.2% for the axial direction in the hamstrings).

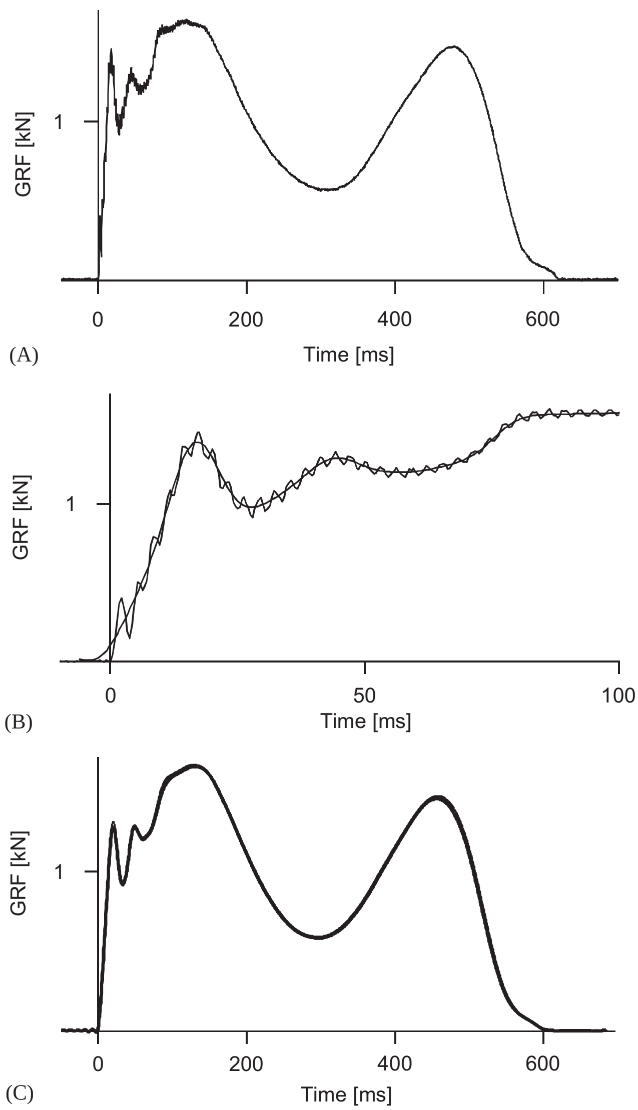

Fig. 1. Vertical ground reaction force (GRF) during one stance phase of walking (A). The time around heel-strike is shown at a larger scale in B. Mean and standard error of sample mean of the vertical GRFs for 30 trials (C). Raw traces are shown in A and B, and filtered traces are used for B and C.

3.3. Myoelectric activity

The insert condition resulted in changes to the myoelectric activity for all four muscles tested (Fig. 4). Analysis of variance showed that there was no significant effect of the shoe condition for the total intensity for the rectus femoris in both the 50 ms windows before and after heel-strike. Significant effects of the shoe condition were observed for the total intensity for the biceps femoris, tibialis anterior and lateral gastrocnemius (increases in the total intensity with the insert condition; Fig. 4).

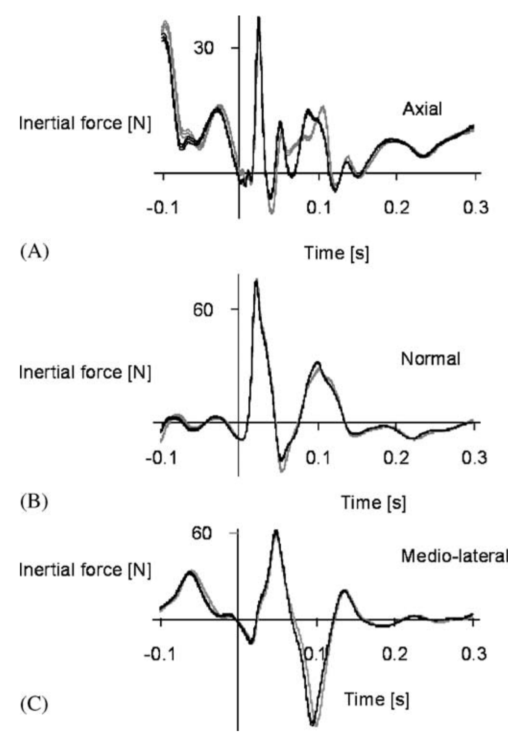

Fig. 2. Inertial forces in the hamstrings for a 57 kg subject. Traces are shown for the control condition (black lines) and insert condition (grey lines), as mean ± SEM ($N = 240$). Heel-strike occurred at a time of 0.

Table 2
Parameters describing the impact peak of the vertical ground reaction force

Condition	Loading rate, $\dot{F}_{GRF,max}$ (kN s^{-1})	Impact force, $F_{GRF,max}$ (kN s^{-1})	Time to peak force t_{GRF} (ms)	Input frequency, f_{GRF} (Hz)
Control	63.85 ± 0.67	1.036 ± 0.005	22.23 ± 0.16	34.83 ± 0.21
Insert	53.70 ± 0.57	1.036 ± 0.005	22.70 ± 0.19	31.57 ± 0.21

Values are given as mean ± SEM ($N = 1192$).

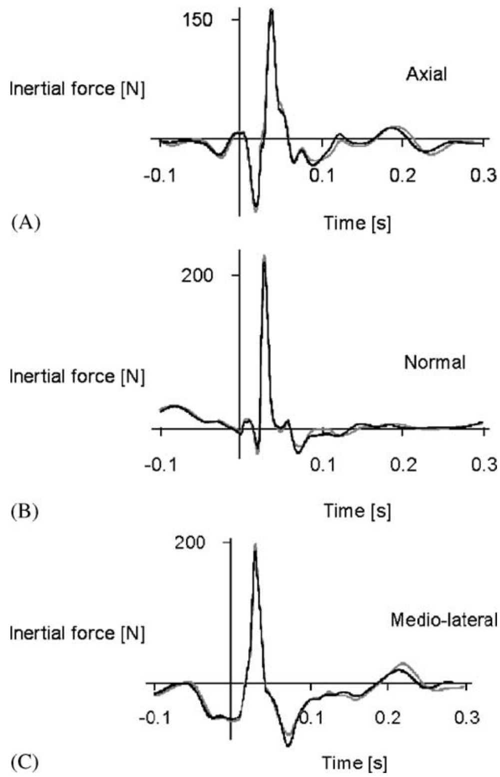

Fig. 3. Inertial forces in the triceps surae for a 57 kg subject. Traces are shown for the control condition (black lines) and insert condition (grey lines), as mean $\pm$ SEM ($N = 240$). Heel-strike occurred at a time of 0.

ANOVA results showed the only significant decreases in the high-frequency myoelectric activity to occur at -40 to -30 ms and -10 to 0 ms for the rectus femoris, and for -50 to -30 ms for the tibialis anterior (Fig. 4). All other significant effects of shoe condition on the myoelectric frequency at the high- and low-frequency bands were for increases with the insert condition. The greatest changes in myoelectric activity occurred for the lateral gastrocnemius muscle (24% increase extending from -50 to $+50$ ms). The biceps femoris showed significant shoe increases in myoelectric activity which reached 15% for the interval -20 to $+20$ ms. The tibialis anterior showed a gradation from decreased activity before heel-strike to increased activity after heel-strike, with the change from decreasing to increasing intensity occurring approximately 20 ms later for the high- than for the low-frequency band. The shoe condition showed little significant effect on the myoelectric activity from the rectus femoris.

4. Discussion

4.1. Limitations to the experimental design

Muscle force production during walking changes throughout a stride. The frequency and damping coefficients of the soft-tissues change with muscle force (Wakeling and Nigg, 2001a) and, therefore, will also be expected to change throughout the stride. If dynamic muscle force cannot be accurately predicted from myoelectric activity, then so too the frequency and damping coefficients cannot be accurately modelled throughout a stride. The purpose of this study was to investigate the soft-tissue vibrations immediately following heel-strike and so the soft-tissue vibrations were characterized by their inertial loading rate, peak inertial force and effective frequency. These measures are taken directly from the acceleration traces, and therefore can be made without invoking assumptions about the muscle-force-dependent changes in the vibrations which would have to be made for more complex modelling.

The high- and low-frequency bands used for myoelectric analysis in this study (Table 1) were developed to have short time resolutions (<10 ms), and to resolve the frequency bands where distinct myoelectric signals have previously been observed in man (Wakeling et al., 2001a). The short time resolutions were necessary to minimize the effects of movement artefacts related to the heel-strike impact from the measured myoelectric activity. We can be confident that the resolved signal before heel-strike using the two-frequency band approach is independent of impact related artefacts. The presented results were similar to myoelectric intensities calculated using pooled high- and low-frequency bands (wavelets 2–3 and 6–8 from the filter-bank of 11 wavelets; time resolutions <40 ms), but confirm that changes in muscle pre-activation are a real effect.

4.2. Soft-tissue resonance

The lower extremities experienced oscillating input forces (the ground reaction force) during walking (Fig. 1). The high-frequency oscillations measured in the ground reaction force (>300 Hz) are likely due to the resonance of the force plate and had frequencies an order of magnitude higher than the vibration frequencies of the soft-tissues. Their contribution to soft-tissue resonance can, therefore, be assumed to be negligible. The impact force showed a couple of force peaks within the first 50 ms after heel-strike. The time from heel-strike to the forefoot striking the ground was approximately 108 ms, and therefore the two force peaks occurring within the first 50 ms can be considered impact related events. The effective frequency of this impact force was between 31 and 35 Hz and was close to the vibration frequencies of the tissues.

When the frequency of an input force is close to the natural frequency of a system, then that system will have a tendency to resonate. During forced vibrations in a spring-damper system, the transmissibility β_t can be calculated as the ratio of the amplitude of the transmitted force to the amplitude of the applied

Table 3
Parameters describing the inertial loads on the soft-tissue after heel-strike

Soft-tissue	Direction	Condition	Loading rate $\dot{F}_{T,max}$ (kN s^{-1})	Inertial force $F_{T,max}$ (N)	Vibration frequency f_T (Hz)
Quadriceps	Axial	Control	35.35±3.43*	196.10±13.09*	28.71±1.10*
		Insert	30.05±2.84	173.96±12.73	27.16±1.08
	Normal	Control	27.46±2.69*	67.06±13.47	19.89±1.85
		Insert	23.58±2.33	64.75±10.98	19.08±1.53
	Medio-lateral	Control	25.14±2.53*	115.93±14.87*	26.98±1.78
		Insert	22.18±2.04	100.80±14.47	28.01±1.96
Hamstrings	Axial	Control	7.24±0.80	46.69±3.95	32.66±2.40*
		Insert	6.30±0.60	46.18±2.95	27.38±1.89
	Normal	Control	11.86±1.00*	23.19±5.62	19.80±0.82
		Insert	10.82±0.85	23.16±5.31	19.94±0.84
	Medio-lateral	Control	8.75±0.72*	44.26±6.85	20.91±1.26
		Insert	7.92±0.65	41.15±6.65	20.18±0.99
Tibialis anterior	Axial	Control	7.12±0.62*	29.98±2.13*	49.08±1.29*
		Insert	5.46±0.46	24.14±1.76	46.62±1.47
	Normal	Control	5.31±0.60*	35.77±2.70*	28.62±1.22*
		Insert	4.46±0.49	32.05±2.41	26.83±1.09
	Medio-lateral	Control	3.24±0.44	4.44±0.70	51.35±5.41
		Insert	2.82±0.35	4.36±0.58	52.91±14.48
Triceps surae	Axial	Control	29.36±2.70*	115.95±7.94*	27.16±0.99
		Insert	25.15±2.17	108.56±7.05	26.40±1.00
	Normal	Control	25.28±2.35*	26.93±5.27	44.89±2.10
		Insert	21.57±2.06	24.75±4.77	42.38±1.90
	Medio-lateral	Control	27.77±3.07*	41.52±3.46	23.40±1.14
		Insert	23.68±2.48	40.40±3.24	22.53±0.95

Values are given as mean ±SEM ($N = 40$). Asterisks denote where there was a significant effect of the shoe condition (ANOVA, degrees of freedom = 79).

force (Shabana, 1996):

$$\beta_t = \frac{\sqrt{1 + (2r\xi)^2}}{\sqrt{(1 - r^2) + (2r\xi)^2}},$$

where ξ is the damping factor for the system, and r is the ratio of the input frequency to the natural frequency of the system. Damping factors for the lower extremity muscles take the values $0.14 < \xi < 0.73$ (Wakeling and Nigg, 2001b). The transmissibilities for this range and also for the range experienced during walking with the control shoe are illustrated in Fig. 5.

The frequency ratio between the ground reaction force and soft-tissues vibrations can be estimated from the effective input frequency and the weighted tissue vibration frequency:

$$r \approx \frac{f_{GRF}}{f_T}.$$

The mean frequency ratios during walking with the control shoe were 1.33, 1.44, 1.22 for the quadriceps, hamstrings, and triceps surae soft-tissue groups, respectively. The 9.4% reduction in effective input frequency with the insert condition would result in frequency ratios closer to unity for these tissues. If the mechanical properties of these tissues remained the same then it would be expected that there would be greater force transmission and thus higher inertial forces in the soft-tissues with the insert condition. However, the opposite result was observed. The significant changes due to the insert condition occurred as decreases in the inertial loading rate and peak inertial forces for these tissues (Table 3). Therefore, we must conclude that the mechanical properties of the tissues changed in response to the shoe conditions, and changed in a manner which reduced the potential resonance of those tissues.

4.3. *Damping of soft-tissue resonance*

The body can minimize its soft-tissue vibrations by different mechanisms, namely by reducing the input force transmitted to the soft-tissues, by changing their resonance frequencies or by increasing the damping of the vibrations within those tissues. The impact shock is attenuated as it travels towards the head, and this attenuation can be changed by adjusting joint stiffnesses and kinematics. Increasing the knee flexion angle reduces the effective axial stiffness of the body (McMahon et al., 1987; Lafortune et al., 1996). However, altering shoe hardness during running can

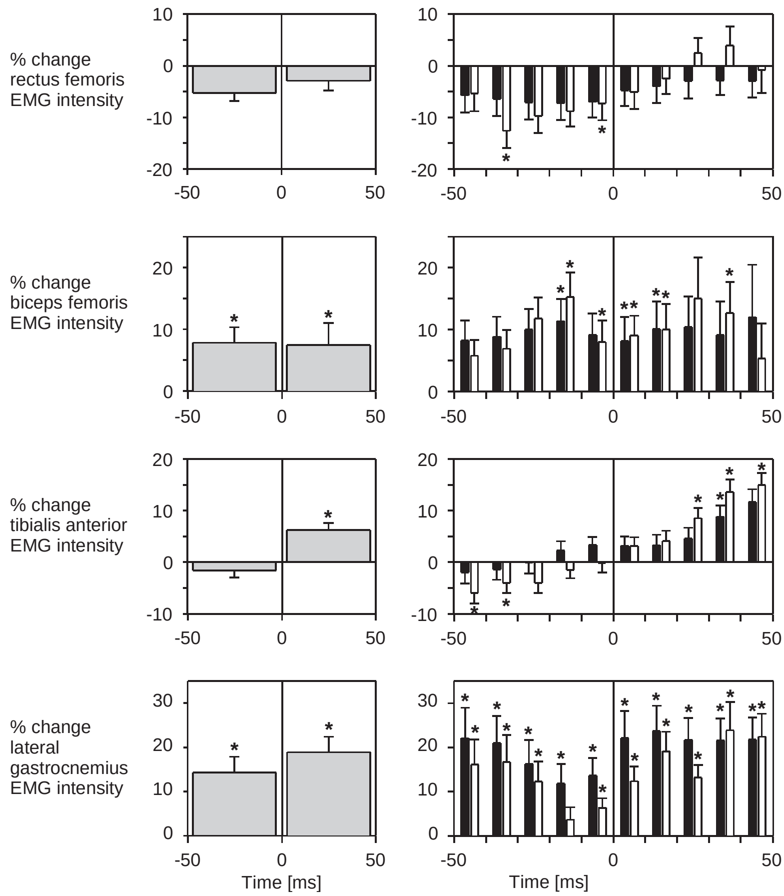

Fig. 4. Changes in myoelectric intensity with the insert condition relative to the control condition. Heel-strike occurred at time 0. Grey bars on the left panel show the changes for the 50 ms windows before and after heel-strike for the total intensity. The right panel shows the changes in intensity for 10 ms time windows for low myoelectric frequencies (black bars) and high myoelectric frequencies (white bars). Bars shown the mean change, error bars show the SEM ($N = 40$). Asterisks denote significant effects of the shoe condition from post-hoc univariate tests.

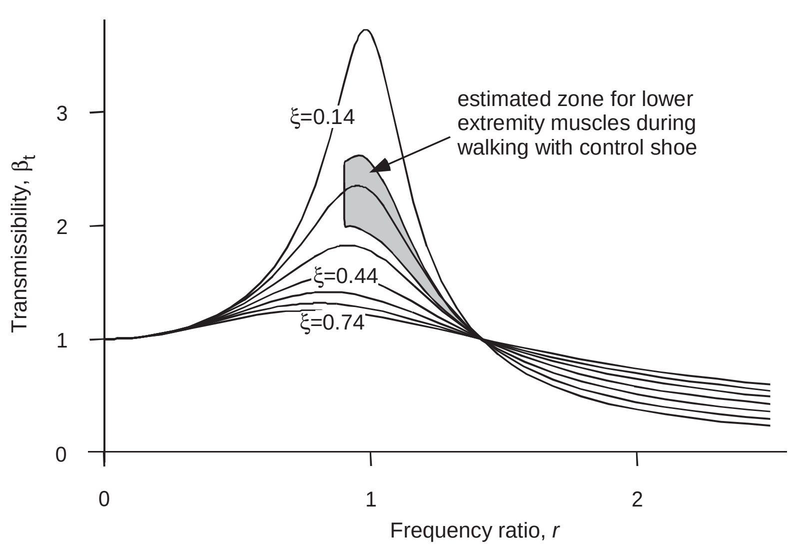

Fig. 5. Transmissibility between the amplitude of the transmitted force and the amplitude of an applied force for a forced vibration. Transmissibilities are shown for damping factors which span those found in the lower extremity muscles (data from Wakeling and Nigg, 2001a, b). The shaded region shows the region of transmissibilities which cover the frequency ratios, r, and estimated damping factors, ξ, for the lower extremity muscles during walking with the control shoe.

result in no significant difference to femur and knee angles at touchdown (Clarke et al., 1983). The shock transmission changes, while running at different speeds, to result in a constant level of head acceleration regardless of the shock input (Nigg et al., 1974; Hamill et al., 1995). In this study there was no difference in the magnitude of the impact force, it is likely that it was attenuated to the same level at the head, and thus that the soft-tissues experienced similar input loads for the two conditions. The segmental kinematics were not measured in this study, however, the experiment was designed to keep the gross kinematics constant. A constant stride length was forced by walking a constant distance to the force platform and there were no significant effects of the insert condition on the stance duration and the walking velocity. Thus, it is unlikely that the decreases in inertial forces in the soft-tissues with the soft insert condition were the result of a reduced vibration load on the tissues.

Increases in muscle activity increase both the resonance frequency and damping coefficients in the soft-tissues (Wakeling and Nigg, 2001a, b). Previous studies suggested that the role of muscle activity in damping soft-tissue resonance may be more by changing its damping coefficient than by changing its resonant frequency (Wilson et al., 2001; Wakeling et al., 2002a). Muscle can damp soft-tissue vibrations either due to contractions which occur during a steady vibration known as the tonic vibration reflex (Hagbarth and Eklund, 1966) or through being activated in an anticipatory manner immediately prior to the impact. Short latency reflex responses to stretch have been shown to take 41 ms in the triceps surae (Nardone and Schieppati, 1998), and may be as low as 34 and 39 ms in the biceps femoris and rectus femoris, respectively (Schillings et al., 1999). It may take a further 20–60 ms for the muscle to generate force (Burke et al., 1971) after being activated. In this walking study the greatest soft-tissue vibrations occurred within 50 ms of heel-strike, which is a shorter time period than a muscle could produce force after a reflex activation. It is, therefore, unlikely that the tonic vibration reflex plays a significant role in the damping of the soft-tissue vibrations at heel-strike. Anticipatory muscle activation, on the other hand, can occur before the impact and have sufficient time for the muscle to generate force to damp the vibration. Muscle activation in anticipation to transient impact forces has been previously been observed for jumping (Santello and McDonagh, 1998), heel impacts on a pendulum apparatus (Wakeling et al., 2001b) and pulsed displacements on a vibration platform (Wakeling et al., 2002a). The results from this current study showed increases in muscle activity in the biceps femoris and lateral gastrocnemius (Fig. 5) even before heel-strike, which indicate that anticipatory muscle activation played a role in the damping of the soft-tissue vibrations.

Greater damping of the soft-tissues occurs in response to pulsed inputs when the input frequency is close to the resonant frequency (Wakeling et al., 2002a). Such responses are anticipatory and occur with specific increases in muscle activity in a high myoelectric frequency band (indicating the activity of faster motor units; Wakeling et al., 2002a). Muscle preactivation events have been similarly been observed before landing from different heights (Santello and McDonagh, 1998) and in response to cyclic heel-strike impacts on a human pendulum experiment (Wakeling et al., 2001b). Results from the current study also showed changes in the myoelectric activity before heel-strike in an anticipated response to the impact force. However, the results from the current study (Fig. 4) do not provide further evidence that faster motor units (indicated by higher myoelectric frequencies) are preferentially recruited for the vibration damping task.

4.4. Muscle tuning during walking and running

During walking the body experiences an impact shock at heel-strike which travels through the body and places a vibration load on the soft-tissues. The body adapts to the input force, and changes its mechanical properties in order to reduce the potential soft-tissue vibrations. The increases in muscle activity for the lateral gastrocnemius and biceps femoris in this study indicate that, certainly for these muscles, that vibration damping by increased muscle activity is a plausible mechanism. This study is the first time that such muscle adaptation to impact forces at heel-strike has been linked to changes in tissue resonance and soft-tissue vibrations.

Causing specific changes in the lower extremity muscle activity may alter the joint loading and thus provide a mechanism for the treatment and prevention of musculoskeletal disorders. It has been previously shown that articular cartilage can adapt to changes in its mechanical environment (Adams, 1989; Brandt et al., 1991; Jurvelin et al., 1986; Setton et al., 1994) and that osteoarthritic changes can result from disruption of normal joint mechanics through surgical removal of the anterior cruciate ligament, meniscetomy or resection of the tibial plateau (Pond and Nuki, 1973; McDevitt et al., 1977; Moskowitz et al., 1979). The lower extremity muscles generate forces which act across the knee and ankle joints and these forces contribute to joint loading. Therefore, it is possible that mechanisms which cause an alteration in the lower extremity muscle forces may affect the onset and progression of osteoarthritis at the knee and ankle joints. The results from this study support the possibility that using shoe materials to cause alterations in lower extremity muscle activity during walking and running may be a way for alleviating these musculoskeletal disorders.

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References

Adams, M.E., 1989. Cartilage hypertrophy following canine anterior cruciate ligament transection differs among different areas of the joint. Journal of Rheumatology 16, 818–824.

Brandt, K.D., Braunstein, E.M., Visco, D.M., O'Connor, B., Heck, D., Albrecht, M., 1991. Anterior (cranial) cruciate ligament transection in the dog: a bonafide modal of osteoarthritis, not merely of cartilage injury and repair. Journal of Rheumatology 18, 436–446.

Burke, R.E., Levine, D.N., Zajac, F.E., 1971. Mammalian motor units: physiological-histochemical correlation in three types in cat gastrocnemius. Science 174, 709–712.

Clarke, T.E., Frederick, E.C., Cooper, LB., 1983. Biomechanical measurement of running shoe cushioning properties. In: Nigg, B.M., Kerr, B.A. (Eds.), Biomechanical Aspects of Sports Shoes and Playing Surfaces, University Printing, Calgary, pp. 25–33.

Frederick, E.C., Clarke, T.E., Hamill, C.L., 1984. The effect of running shoe design on shock attenuation. In: Frederick, E.C. (Ed.), Sport Shoes and Playing Surfaces. Human Kinetic Publishers, Champaign, IL, pp. 190–198.

Hagbarth, K-E., Eklund, G., 1966. Motor effects of vibratory stimuli in man. In: Granit, R. (Ed.), Muscular Afferents and Motor Control. Almquist & Wiksell, Stockholm, pp. 177–186.

Hamill, J., Derrick, T.R., Holt, K.G., 1995. Shock attenuation and stride frequency during running. Human Movement Science 14, 45–60.

Jurvelin, J., Kiviranta, F., Tammi, M., Helminen, H.J., 1986. Softening of canine articular cartilage after immobilization of the knee joint. Clinical Orthopaedics and Related Research 207, 246–252.

Lafortune, M.A., Lake, M.J., Hennig, E.M., 1996. Differential shock transmission response of the human body to impact severity and lower limb posture. Journal of Biomechanics 29, 1531–1537.

Light, L.H., MacLellan, G.E., Klenerman, L., 1980. Skeletal transients on heel strike in normal walking with different footwear. Journal of Biomechanics 13, 477–488.

McDevitt, C., Gilbertson, E., Muir, H., 1977. An experimental model of osteoarthritis; early morphological and biochemical changes. Journal of Bone and Joint Surgery 59B, 24–35.

McMahon, T.A., Valiant, G., Frederick, E.C., 1987. Groucho running. Journal of Applied Physiology 62, 2326–2337.

Moskowitz, R.W., Howell, D.S., Goldberg, V.M., 1979. Cartilage proteoglycan alterations in an experimentally induced model of rabbit osteoarthritis. Arthritis and Rheumatism 22, 155–163.

Nardone, A., Schieppati, M., 1998. Medium-latency response to muscle stretch in human lower limb: estimation of conduction velocity of group II fibres and central delay. Neuroscience Letters 249, 29–32.

Nigg, B.M., Neukomm, P.A., Unold, E., 1974. Biomeckanik und sport. Über beschleunigungen die am menschlichen körper bei verschiedenen bewegungen auf verschiedenen unterlagen auftreten. Orthopäde 3, 140–147.

Nigg, B.M., Bahlsen, H.A., Luethi, S.M., Stokes, S., 1987. The influence of running velocity and midsole hardness on external impact forces in heel-toe running. Journal of Biomechanics 20, 951–959.

Nigg, B.M., Cole, G.K., Brüggemann, G-P., 1995. Impact forces during heel-toe running. Journal of Applied Biomechanics 11, 407–432.

Pond, M.J., Nuki, G., 1973. Experimentally induced osteoarthritis in the dog. Annals of the Rheumatic Diseases 32, 387–388.

Santello, M., McDonagh, M.J.M., 1998. The control of timing and amplitude of EMG activity in landing movements in humans. Experimental Physiology 83, 857–874.

Schillings, A.M., van Wezel, B.M.H., Mulder, T., Duysens, J., 1999. Widespread short-latency stretch reflexes and their modulation during stumbling over obstacles. Brain Research 816, 480–486.

Setton, L.A., Mow, V.C., Muller, F.J., Pita, J.C., Howell, D.S., 1994. Mechanical properties of canine articular cartilage are significantly altered following transection of the anterior cruciate ligament. Journal of Orthopaedic Research 12, 451–463.

Shabana, A.A., 1996. Vibration of Discrete and Continuous Systems. Springer, New York, 393pp.

Von Tscharner, V., 2000. Intensity analysis in time-frequency space of surface myoelectric signals by wavelets of specified resolution. Journal of Electromyography and Kinesiology 10, 433–445.

Wakeling, J.M., Nigg, B.M., 2001a. Modification of soft tissue vibrations in the leg by muscular activity. Journal of Applied Physiology 90, 412–420.

Wakeling, J.M., Nigg, B.M., 2001b. Soft tissue vibrations in the quadriceps measured with skin mounted transducers. Journal of Biomechanics 34, 539–543.

Wakeling, J.M., Pascual, S.A., Nigg, B.M., von Tscharner, V., 2001a. Surface EMG shows distinct populations of muscle activity when measured during sustained exercise. European Journal of Applied Physiology 86, 40–47.

Wakeling, J.M., von Tscharner, V., Nigg, B.M., Stergiou, P., 2001b. Muscle activity in the leg is tuned in response to ground reaction forces. Journal of Applied Physiology 91, 1307–1317.

Wakeling, J.M., Nigg, B.M., Rozitis, A.I., 2002a. Muscle activity in the lower extremity damps the soft-tissue vibrations which occur in response to pulsed and continuous vibrations. Journal of Applied Physiology 93, 1093–1103.

Wakeling, J.M., Pascual, S.A., Nigg, B.M., 2002b. Altering muscle activity in the lower extremities by running with different shoes. Medicine and Science in Sports and Exercise 34 (9), 1529–1532.

Wilson, A.M., McGuigan, M.P., Su, A., van den Bogert, A.J., 2001. Horses damp the spring in their step. Nature 414, 895–899.