Modification of soft tissue vibrations in the leg by muscular activity

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Wakeling, James M., and Benno M. Nigg. Modification of soft tissue vibrations in the leg by muscular activity. J Appl Physiol 90: 412–420, 2001.—Vibration characteristics were recorded for the soft tissues of the triceps surae, tibialis anterior, and quadriceps muscles. The frequency and damping of free vibrations in these tissues were measured while isometric and isotonic contractions of the leg were performed. Soft tissue vibration frequency and damping increased with both the force produced by and the shortening velocity of the underlying muscle. Both frequency and damping were greater in a direction normal to the skin surface than in a direction parallel to the major axis of each leg segment. Vibration characteristics further changed with the muscle length and between the individuals tested. The range of the measured vibration frequencies coincided with typical frequencies of impact forces during running. However, observations suggest that soft tissue vibrations are minimal during running. These results support the strategy that increases in muscular activity may be used by some individuals to move the frequency and damping characteristics of the soft tissues away from those of the impact force and thus minimize vibrations during walking and running.

impact forces during locomotion (e.g., at heel strike) should be expected to produce vibrations in the soft tissues of the body. However, observations suggest that impact-related vibrations are minimal in the muscular soft tissues of the lower extremities during running. Prolonged exposure to vibrations can have detrimental effects on the soft tissues, including reductions in nerve conduction velocity, and reduced peripheral circulation and/or attenuated sensory perception (7, 11). Thus it has been speculated that the body may use strategies to minimize soft tissue vibrations during locomotion (22).

Impact forces in heel-toe running have a maximum between ~5 and 30 ms (1, 4, 21), and such an input force can be described in terms of amplitude and frequency (typically 10–20 Hz). Both the amplitude and frequency of the external impact force signal can be altered by changes in leg geometry and changes in joint stiffness at ground contact (6, 16, 20). A simple wobbling mass model predicts that impact forces can also be modified as a result of changes in the coupling between the soft and rigid tissues of the body (24), although this idea has not been experimentally verified. All three strategies of modifying the input force are the consequence of changing muscle activity patterns and thus are not mutually exclusive. A combination of these strategies may be used to minimize soft tissue vibrations during locomotion, in what has been termed “tuning” (22).

Changes in the coupling between the soft and rigid tissues in the leg may be effective in minimizing soft tissue vibrations if 1) the natural vibration frequencies of the soft tissues coincide with the frequencies of the input signal and 2) muscular activation can shift the resonant frequencies of the soft tissues substantially away from the frequency of the input signal. However, the natural vibration characteristics of the soft tissues of the leg and the effect of muscular activity on vibrations have not been quantified yet. It may be expected that the frequency and damping coefficients of free vibrations in the soft tissues depend on muscle activity, muscle length, shortening velocity, relative contribution of different motor units, and the contraction history (e.g., Refs. 9 and 29). Furthermore, the vibration characteristics of soft tissue are determined not only by the mechanical properties of the muscle it contains, but also by the properties of the fat, connective tissue, and vascular components of the soft tissues and the coupling between these components.

The purpose of this study was to quantify the free vibration frequencies and damping coefficients of selected soft tissue masses as a function of muscle length, shortening velocities, and muscle force production. In addition, the feasibility of the soft tissue tuning concept is discussed in light of the experimental evidence.

METHODS

Seven men [age 28.1 ± 1.5 (SE) yr; height 178.8 ± 2.2 cm; mass 78.2 ± 2.5 kg] and seven women (age 26.1 ± 1.1 yr; height 169.0 ± 1.8 cm; mass 64.2 ± 3.0 kg) participated in

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this study. Subjects gave their informed consent in accordance with the university’s policy on research using human subjects.

**Measurements.** Soft tissue vibrations were measured in the tissues surrounding and including the gastrocnemius, tibialis anterior, and the vastus lateralis muscles of the leg at a position one-third of the muscle length from the distal end of the muscle belly. Vibrations were measured with skin-mounted tri-axial accelerometers made up of three piezoelectric sensors [EGAX accelerometer (nominal frequency response of 0–600 Hz) Entran Devices] aligned in an orthogonal arrangement within an aluminum frame (13 x 13 x 14 mm, total mass = 15.9 g) as described previously (30). The accelerometers were aligned with the x-axis normal to the skin surface, the y-axis parallel to the long axis of the leg segment, and the z-axis perpendicular to the x-y plane. Analog signals from the accelerometers were sampled at 1,000 Hz (14-bit DI210 data acquisition card, Data Instruments).

Muscular force was generated by voluntary contractions. Torques in the knee and ankle joints were measured with a dynamometer (System 3, Biodex Medical Systems) during separate experiments. The dynamometer also measured the joint angle and the joint angular velocity. Dynamometer signals were recorded at 1,000 Hz.

Isometric tests were performed for the quadriceps for the following angles of knee flexion: 0, 20, 40, 60, and 80°, where 0° represents a straight leg position. For these tests, the subject was seated with the thigh held horizontally and the hip flexed at 85°. Isometric tests were performed for the triceps surae and tibialis anterior for the following ankle extension angles: −20, 0, 20, and 40°, where positive angles describe plantar-flexion and an angle of 0° occurs when the foot is at right angles to the lower leg. For these tests, the subject was seated with the lower leg horizontal, the knee flexed at 25°, and the hip flexed at 85°. Three levels of muscle activation were tested for each position: 0, 50, and 100% of maximum voluntary contraction. Subjects monitored their force production on a visual display. The minimum torque measured during the 0% maximal voluntary contraction trials was taken as zero torque for each position, thus negating the effect of gravity on the apparatus.

Isotonic tests were performed at a torque that was close to the maximum torque possible for joint movement. Isotonic tests were also performed at discrete torques in the range of 50–100% of this submaximal torque for the ankle plantar-flexion and the knee extension exercises. Vibrations were recorded as the ankle and knee joints passed through angles of 23 and 41°, respectively. A zero torque value appropriate to the joint position for each isotonic test was determined by quadratic interpolation from the zero torque measurements in the isometric study. Joint power was taken as the product of joint torque and angular velocity during these isotonic tests.

During each muscle contraction, an oscillation was initiated by a strike with a wooden mallet, delivered to a marked position that was 40 mm proximal to the accelerometer. The repeatability of this impulse was monitored by the amplitude of the initial accelerations measured on the tissue. At least six vibrations were measured for each combination of soft tissue region, joint torque, and joint angle for the isometric contractions, and, in isotonic torque, for the isotonic contractions.

The effect of muscular fatigue was determined by repeating the tests on a soft tissue region at the beginning and the end of the experiment. The nature of the vibration motion was investigated by mounting a second accelerometer (80 mm proximal to the first) on the quadriceps of two subjects. The motion and vibration characteristics from the two accelerometers were compared.

**Analysis.** The dominant frequency and the damping coefficient at that frequency were estimated for the vibration signals from each axis of the accelerometer. These coefficients were determined using a least squares minimization (Levenberg-Marquart method) of the following equation

\[ s = Ae^{-ct} \sin (2\pi ft + \varphi) \]

where \( s \) is the measured signal, \( A \) is the amplitude of the vibration, \( t \) is time, \( c \) is the damping coefficient, \( f \) is the damped natural frequency, and \( \varphi \) is a phase coefficient. The fit between the measured and predicted vibration was determined using the Rank-Pearson correlation coefficient \( r \) and fits were excluded where \( r < 0.85 \) (30). Least-squares fitting of the equation (1) to the signal results in the major frequency component and the damping at that frequency being resolved. Noise in the signal which occurs at different frequencies are thus eliminated by this method, and therefore no prior filtering of the signal was used.

The damped oscillation model (equation) fitted to the data is that for a simple linear system. In such a system, the vibration frequency and damping characteristics in the soft tissues would be the result of the interaction between the mallet and the tissue during the initial impulse but, on cessation of the impulse, are entirely due to the biomechanical properties of the tissue. The forced excitation that occurred during impact lasted <50 ms, thus the results were recorded for the free vibration stage for each test, which was defined as 50–250 ms after the initial soft tissue impact (Fig. 1; Ref 30).

![Fig. 1. Accelerations measured in the triceps surae during free vibrations at 0, 50, and 100% maximum voluntary contraction (MVF) for one subject with an ankle plantar-flexion of 40°. Accelerations are normalized to the maximum acceleration at the start of the free vibration to highlight the change in damping. Dashed lines, data for the accelerations in the x direction; solid lines, model decay (from equation in Analysis) and the exponential nature of the decay.](http://jap.physiology.org/)
Statistics. The free vibration frequency $(f)$ and damping coefficient $(c)$ were tested using a general linear model (GLM) ANOVA. The following parameters were used as covariates: identity number, gender, soft tissue tested, trial number, direction of acceleration, joint torque, joint angle, and joint angular velocity. The nature of the significant interactions was observed from planar least-squares regression of $f$ and $c$ against joint torque and angular velocity. Joint torques and powers from all three directions were ranked in ascending magnitude, and means $\pm SE$ for $f$ and $c$ were determined from the first and fourth quartiles of this ranking. The effect of the mounting position for the accelerometers on the quadriceps tissue was tested using $\chi^2$ tests with the null hypothesis that accelerometer signals were identical at the two measurement positions. Student’s $t$-tests were used to measure the significance of differences between isolated conditions. Statistical tests were deemed significant at the 95% confidence level. Values are expressed as mean $\pm SE$, and $n$ refers to the number of samples in each mean or test.

RESULTS

Validation. The repeatability of the applied impact can be judged from the amplitudes of the measured accelerations at the start of each free vibration. The impact was delivered perpendicular to the skin surface (in the $x$ direction), and the initial free accelerations in this direction were $13.85 \pm 0.17$ ($n = 1,086$), $9.27 \pm 0.42$ ($n = 386$) and $10.68 \pm 0.18$ ($n = 1,325$) m/s$^2$ for the soft tissues containing the triceps surae, tibialis anterior, and quadriceps muscles, respectively. The SE of these impacts was thus $<5\%$ of the initial mean value for the tibialis anterior and $<2\%$ for both the triceps surae and the quadriceps tissues. The amplitude of the accelerations recorded in the $y$ and $z$ directions were less than those in the $x$ direction. The maximum accelerations at the start of the free vibration in the $y$ direction were $5.54, 3.34,$ and $3.42$ m/s$^2$ and, in the $z$ direction, were $7.89, 6.40,$ and $3.84$ m/s$^2$ for the soft tissues containing the triceps surae, tibialis anterior, and quadriceps muscles, respectively.

Frequency and damping data were measured in the quadriceps tissues of two subjects from two triaxial accelerometers that were placed equidistant from the impact site, with the additional accelerometer proximal to the original. For individual, knee angle, and knee torque, 43 of 90 combinations showed significant differences between the measured frequency at the two sites, and 30 showed significant differences for the measured $c$. Observation of these differences showed that there was no systematic bias of higher or lower values of $f$ or $c$ for the more proximal accelerometer. Separate $\chi^2$ tests for $f$ and $c$ showed that the proportions of higher and lower values between the distal and proximal recording sites were not significantly different from random.

Isometric tests. The GLM ANOVA showed no significant correlation between the $f$ and both the identity number and the soft-tissue group for the isometric tests ($n = 6,815$). However, significant effects occurred between all other combinations of $f$ and $c$ with identity number, gender, soft tissue group, trial number, direction of acceleration, joint torque, and joint angle. Values for $f$ and $c$ at both low and high joint torques are shown in Table 1. These mean coefficients, taken from the first and fourth quartiles of the joint torque, ranged from $10.5 < f < 43.2$ s$^{-1}$ and $12.1 < c < 55.8$ s$^{-1}$ for the triceps surae tissues, from $16.3 < f < 62.6$ s$^{-1}$ and $13.7 < c < 104.9$ s$^{-1}$ for the tibialis anterior tissues and from $8.2 < f < 41.3$ s$^{-1}$ and $8.4 < c < 61.5$ s$^{-1}$ for the quadriceps tissues. In general, higher $f$ and $c$ values occurred in the $x$-direction, which was normal to the skin surface, than in the tangential $y-z$ plane. In the quadriceps tissues, this dependency on direction was most pronounced at the highest joint torques, when the muscle was most active (Table 1).

The effects of muscle length and force production on soft tissue vibrations are shown in Figs. 2 and 3. There was a general increase in $f$ and $c$ with an increase in torque for any joint angle. The mean $f$ at a knee torque of 60 Nm (Fig. 3; calculated from linear regression) decreased by $34\%$ between the knee flexion angles of 0 and 80°, and the corresponding mean $c$ decreased by $52\%$ across this range. There was a smaller range of $f$ and $c$ at the low joint torques than for the high torques, which can be seen in Table 1.

The isometric ankle plantar-flexion tests were repeated for 2 trials with 10 subjects. There were no significant differences in the maximum voluntary ankle torque produced for any subjects between these two trials. The effect of the trial was tested by using the change in frequency or damping for each joint angle and each joint torque. These differential values showed significant between-trial differences for some individuals for $f$ ($n = 2$) and $c$ ($n = 7$) and resulted in the significant effects detected in the GLM ANOVA. However, the mean values for $f$ and $c$ across all the subjects showed no significant differences between the two trials. Additionally, there was a significant effect of gender on the values of $f$ and $c$ for the isometric tests.

Isotonic tests. The GLM ANOVA showed no significant correlation between $f$ and the trial number for the isotonic tests ($n = 629$). However, significant effects occurred between all the other combinations of $f$ and $c$ with the covariates identity number, gender, soft tissue group, trial number, direction of acceleration, joint torque, and joint velocity. Finally, no significant effect occurred between either $f$ or $c$ and joint angle.

Least-squares planar regression for the data from the quadriceps soft tissue showed that increases in the joint angular velocity produced significant increases in both $f$ and $c$ (Fig. 4). In the triceps surae tissue, similar increases were observed for the damping; however, planar regression was unable to resolve significant increases in $c$. The joint angles tested for the isotonic experiments were nearly constant at $23.2 \pm 0.8°$ ($n = 406$) for the ankle plantar-flexion tests and $41.4 \pm 1.4°$ ($n = 223$) for the knee extension exercises. Therefore, these data were unable to resolve significant effects between the joint position (and thus muscle length) and $f$ and $c$.

Magnitudes of $f$ and $c$, as they relate to the joint powers, are shown in Table 2. Both $f$ and $c$ were greater in the $x$ direction, normal to the skin surface, than in
both soft tissue groups that were tested. The vibration amplitudes were reduced to 50% in 15–70 ms and to 5% within 60–300 ms. In the time taken for the amplitude to reduce to 5%, there were typically no more than 2 oscillations in the vibration. Thus the measured free vibration characteristics were under muscular control and were of under-damped vibrations with a rapid decay.

The dynamometer measured joint angles and joint torques. These can be translated into muscle lengths and forces by knowing the moment arm about which the muscles act on the joint. Estimates of maximal muscle forces using mean moment arm values for each joint angle (18, 28) are shown in Table 1. The maximum voluntary isometric muscle force changed with joint angle and, thus, with muscle length. Individual moment arm lengths are not available for each subject, and variation in moment arms between subjects resulted in individual variation in muscle forces for a given joint torque. Furthermore, the degree of coactivation in antagonistic muscle groups was not measured, and thus a given joint torque is not produced by a unique muscle force. Nonetheless, the conclusion that individuals can use increases in joint torque to cause an increase in the joint and soft tissue vibrations also holds true for increases in muscle force. Similarly, the conclusion that individuals can use increases in joint angular velocity to result in increases in the joint and soft tissue vibrations is also true for increases in muscle shortening velocity. Changes in the vibration characteristics of the soft tissues are due to changes in the mechanical properties of those tissues. When a muscle is activated, it generates tension. The tension depends on both the number of motor units activated and the level of activation within those motor units, and this tension results in the generation of intramuscular pressures. Hill (14) suggested that high intramuscular pressure is caused by curved muscle fibers exerting an inward pressure during contraction. Such a mechanism is sufficient to explain the intramuscular pressure during contractions in the vastus medialis (26). The tense fascia has also been shown to be significant in the generation of intramuscular pressure. Variations in intramuscular pressures between subjects will be affected by the individual differences in muscle geometry, muscle-tendon compliance, the patterns of recruitment between the motor units, amounts of subcutaneous fat, and blood pressure. On an immediate time scale, the body may actively change the muscle recruitment patterns, which can alter the mus-

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Values are means ± SE pooled data taken from the 1st and 4th quartiles of joint torque ($T$). Estimated tendon force ($F$) is calculated using moment arms from Refs. 18 and 28. $n$, No. of contractions; $f$, frequency; $c$, damping coefficient.
cle force, muscle length, and local fiber orientations. However, changes to the structural properties of the muscle-tendon complex, as well as the fat composition and blood pressure, all take durations longer than that required for a single muscle contraction. Mechano-myographic signals are vibrations that occur as a consequence of muscle contractions and have major frequency components <100 Hz (e.g., 25). Such mechano-myographic signals occur in the absence of external impact forces and are due to mechanisms that are different from the vibrations measured in this study. This study measured joint torques rather than muscle-tendon forces or myoelectric activity patterns. Therefore, the conclusions cannot be extended to different patterns of muscle recruitment. However, the general conclusion that soft tissue vibration \( f \) and \( c \) can be increased with increased joint torque and angular velocity holds true across the range of subjects that we tested.

In the present study, the soft tissue vibrations were most likely of smaller amplitude than those typically observed during running. Furthermore, the vibrations were initiated perpendicular to the skin surface, whereas the vibration during impact in running may occur over a range of angles relative to the tissue. Nevertheless, it is suggested that the observations of under-damped vibrations influenced by changes in muscle activation are generally valid and that the potential soft tissue behavior can be extrapolated to locomotor tasks such as walking and running.

The vertical ground reaction impact force that occurs during each heel-strike has a frequency content typically in the range of 10–20 s\(^{-1}\). Vibrations resulting from such an input will probably occur in any soft tissue structure of the human locomotor system. Damping of such vibrations becomes an important aspect for situations in which the frequency of the impact force and the natural frequency of the soft tissue structures are close. In these specific situations, resonance might occur and the muscles must be activated to actively change the vibration response of the tissue to avoid potential resonance effects.

During heel-toe running (between 2.7 and 4.5 m/s) maximum ankle plantar flexion and knee extension...
moments have been measured between 165 and 221 and 188 and 288 Nm, respectively (3, 5, 13, 19, 31). These maximal moments are typically achieved midway through the stance phase. The corresponding maximum moments at the end of the impact phase are probably <50% of these values. Thus the mean moments during impact may be ~45 and 60 Nm for the ankle and knee joints, respectively. Both of these values fall within the range of joint torques tested in our experiments. The mean accelerations recorded at the start of the free vibrations from this study were 13.85 and 10.68 m/s² for the soft tissues containing the tri-

Fig. 3. }f{left) and }c{right) for the soft tissue containing the quadriceps muscle for free vibrations during isometric knee flexion contractions at 0 (A), 40 (B), and 80° (C). Data are shown for 3 joint angles tested, and the average number of data used for each symbol was }n{ = 6.1 ± 0.5, 6.9 ± 0.5, and 7.1 ± 0.4 for 0, 40, and 80° of knee flexion, respectively.
ceps surae and quadriceps muscles, respectively. The free vibration was assumed to start 50 ms after the impact and thus shows a reduced amplitude compared with the maximum achieved at impact. These recorded values are lower than those measured during walking [25 m/s² (Ref. 15)] and running [110 m/s² (Ref. 16)] using bone-mounted accelerometers; however, it can be expected that soft tissue accelerations in the leg are lower than those for the tibia. First, the viscoelastic coupling between the soft tissues and the skeleton result in soft tissue accelerations that are lower than skeletal values. Furthermore, impact shock is attenuated through the body (23) and is therefore reduced between the lower and upper leg. Therefore, $f$ and $c$ were measured at joint torques that occur during running and for amplitudes that are smaller but close to those found for walking and running. The vibration model used to fit the data is that for a simple damped linear oscillation in which $f$ and $c$ are independent of the magnitude of the initial impact. Thus it is suggested that the results from this model can be extrapolated to walking and running situations.

Differences in the vibration characteristics were recorded between the individuals in the study. Both $f$ and $c$ are similar at low torques but showed individual variation at the higher muscle forces; this can be seen in Figs. 2 and 3 and the SE magnitudes in Table 1. Subjects varied in the extent to which they could modify their soft tissue $f$ and $c$ and also in the relative amount of muscle activity required to generate a given vibration response in their system. Significant effects between the subject gender and $f$ and $c$ were also recorded; however, significant correlations between gender and mass and height prevented the effects of these three covariates from being distinguished in this study.

Increases in mass will decrease both the $f$ and $c$ of the system for a simple damped oscillation. The $f$ and $c$ measured during the isometric experiments were typically lowest for the quadriceps tissues and highest for the tibialis anterior tissues in both the relaxed and active states. This variation reflects the differences in mass between the soft tissues, with the quadriceps being the largest and the tibialis anterior being the smallest.

Joint angle can influence the vibration characteristics in a number of ways. The maximum force achieved by a muscle depends on the muscle fiber length (8, 12),...
as shown by the length-dependent change in the maximum joint torques for the triceps surae and quadriceps tissues in Table 1. We found a significant effect of joint angle on the vibration characteristics that was independent of the muscle force production. This can be seen by the change in $f$ and $c$ for both the triceps surae and quadriceps (Table 1) tissues for the zero torque measurements. These changes in the vibration characteristics are due to passive mechanisms such as the change in tension, which was induced in the soft tissues as the skin and connective tissue were stretched. The joint angular and muscle contraction velocity also influence the vibration coefficients, with increases in angular velocity resulting in increases in $f$ and $c$. Therefore, any model used to predict the vibration properties of the soft tissues must consider both the static and dynamic nature of the joint position.

Mean input frequencies of 15.5 s$^{-1}$ have been measured for tibial accelerations during running at 4.5 m/s. During walking, the faster rise times of tibial shock (14) may result in even greater input frequencies in the leg. These input frequencies coincide with the natural free vibration frequencies of the soft tissues of the leg at the joint torques experienced during walking and running for some individuals (Figs. 2 and 3; Table 1). The present study shows that the natural soft tissue vibration frequencies can be increased by increases in muscle force production. During walking and running, the muscles in the leg are activated before ground impact to generate the leg stiffness which is required at landing. Increases in the co-activation of antagonistic muscle groups before impact may, for some individuals, be used to increase the resonance frequencies of the soft tissues of the leg and optimally tune the system to minimize vibrations.

The results of this study show that the soft tissues of the leg vibrate at a range of frequencies that coincide with the frequencies of the impact forces occurring during locomotion (e.g., walking and running). However, observations suggest that soft tissue vibrations are minimal during locomotion. Such vibrations could be minimized by shifting the free vibration frequencies away from those of the impact forces. This study shows that such tuning of the mechanical properties of the soft tissues can result from changes in muscle activity. In particular, we have demonstrated that the frequency and damping of free vibrations in the soft tissues were influenced by changes in muscle activation, muscle length, and contraction velocity. Therefore, the results of this study show that muscle tuning is a possible strategy for minimizing muscle and soft tissue vibrations. However, it remains to be shown whether, and when, such muscle tuning strategies can actually be utilized during locomotion.

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