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Impact Forces and Muscle Tuning: A New Paradigm

B. M. Nigg and J. M. Wakeling

Human Performance Laboratory, Faculty of Kinesiology, University of Calgary, Calgary, Alberta, Canada

NIGG, B.M. and J.M. WAKELING. Impact forces and muscle tuning: a new paradigm. *Exerc. Sport Sci. Rev.*, Vol. 29, No. 1, pp 37-41, 2001. *We propose that repetitive impact forces during physical activities are not important from an injury perspective but are the reason for changes in myoelectric activity (muscle tuning) to minimize soft tissue vibrations. Changes in myoelectric activity (intensity, frequency, timing), comfort, and performance provide supporting evidence for this new paradigm.* **Keywords:** impact forces, muscle tuning, tissue vibrations, running

INTRODUCTION

Repetitive impact forces in activities such as running typically have a maximum of about one to three times body weight and a major frequency component between 10 and 20 Hz. They have been proposed as a major reason for sports (especially running)-related injuries (7,13). The concept of "cushioning" was proposed in the late 1970s for footwear and sport surfaces to reduce impact loading during athletic activities. However, results of many studies on impact forces and their consequences were surprising and did not support the association between impact forces and sports-related injuries (6). Thus, a new way of thinking about the effect of repetitive impact forces during physical activities is needed.

The purposes of this report were to summarize the current knowledge of the effects of repetitive impact forces and to propose a new paradigm for the understanding of reactions of the locomotor system to repetitive impact forces, with special consideration of running.

IMPACT FORCES: SUMMARY OF RESULTS

Initial research on impact forces during athletic activities (mostly for heel-toe running) provided findings, as summarized in the next few paragraphs, that were mostly surprising to the authors.

External Impact Forces

(a) External impact force peaks were relatively insensitive to changes in the hardness of running shoe midsoles (2), (b) changes in midsole hardness affected the external loading rate $(df_z/dt)_{max}$, and (c) running velocity substantially affected the vertical impact force peaks.

Internal Impact Forces (as Determined through Model Calculations with Experimental Input)

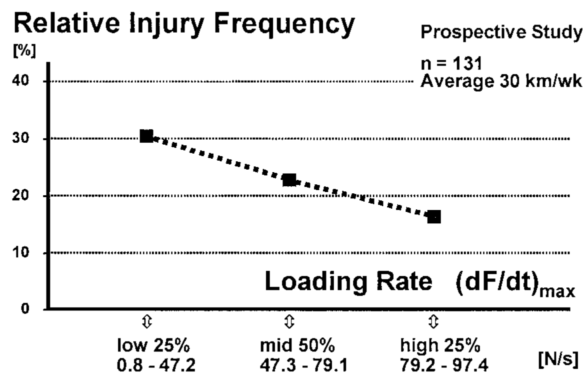
(d) Impact force peaks in the talocrural and talocalcaneal joints were relatively insensitive to changes in hardness of running shoe midsoles, and (e) the magnitude of joint contact forces in the talocrural and talocalcaneal joints was substantially (two to five times) less during the impact than during the active phase of running (1).

Epidemiology

(f) The frequency of osteoarthritis was about equal in runners and nonrunners (8), (g) running on hard surfaces did not result in an increase of running injuries compared with running on softer surfaces (12), (h) shock-absorbing insoles were not effective in reducing the incidence of stress fractures in military recruits for which they were originally designed (4), (i) results of a prospective study (11) showed no significant difference in short-term running injuries between subjects with high-, medium-, and low-impact force peaks (Figure 1), and (j) the same prospective study showed that subjects with a high vertical loading rate $(df_z/dt)_{max}$, had significantly fewer (only about 50%) running-related injuries than did subjects with a low loading rate (Figure 1).

Address for correspondence: Benno M. Nigg, Dr. sc. nat., Professor of Biomechanics, Director: Human Performance Laboratory, Faculty of Kinesiology, The University of Calgary, Calgary, Alberta, Canada T2N 1N4.
Accepted for publication: October 12, 2000

0091-6631/2901/37–41
Exercise and Sport Sciences Reviews



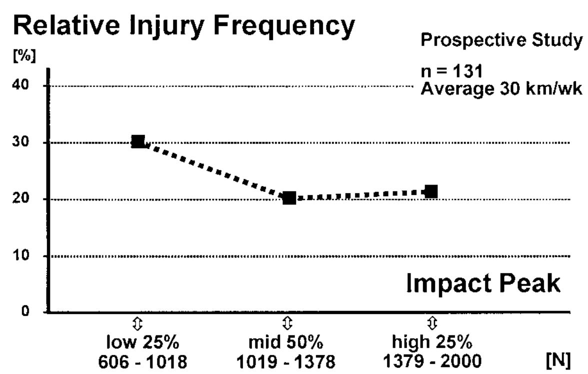


Figure 1 Relationship between the vertical impact force peak (F_{zi}), the maximal vertical loading rate (G_{zi}), and the frequency of running-related injuries. Data are from a prospective study. The impact forces of the participating runners (weekly average of 30 km) were assessed at the beginning of the study. A sports-medical physician documented injury occurrences (from Nigg 1997 [11] with permission).

Energy and Muscles

(k) Model calculations for systematic variations of the mechanical properties of the shoe-surface interface (10) suggested that selected soft and viscous materials required less work than hard and elastic materials for running for certain subject characteristics. Initial experimental results in our laboratory of assessment of oxygen consumption during treadmill running supported this theoretical suggestion for more than 50% of all test subjects. (l) Changes in impact force input into the foot produced substantial and subject-specific changes in myoelectric activity before and/or during ground contact. The changes were subject specific.

Biological Reactions

(m) High-impact activities such as running, gymnastics, and dancing typically increase skeletal mass, whereas low-impact activities such as swimming do not seem to provide the same positive effect (5). (n) It was found that the controlled repeated application of a force with a 1-Hz signal frequency was not able to maintain bone mass over an 8-wk period, whereas the same procedure with a 15-Hz signal frequency stimulated substantial new bone formation (9). The frequency of 15 Hz corresponds approximately to the frequency of the impact forces during heel-toe running. (o) Results for biological reactions of impact loading on cartilage were inconsistent and are still open to discussion (12).

Based on the presented results, one cannot conclude that repetitive impact forces are a major factor in the development of chronic and/or acute running-related injuries. Excessive impact forces may produce damage to the human musculoskeletal system. However, there is a window of loading in which biological tissues react positively to repetitive impact loading. Based on the current knowledge, it is speculated that repetitive impact loading for cartilage and soft tissue structures falls within the acceptable window for moderate and intensive running and that impact loading for bone may sometimes fall outside the acceptable window for intensive running with inappropriate recovery periods. However, the knowledge base on which these speculations are made is limited.

Nevertheless, different impact situations (e.g., changes in shoe midsole stiffness) do affect the human locomotor system. Every runner has experienced this. The question is not whether changes in impact forces do affect the locomotor system. The appropriate question is which parts of the human body are affected by changes in repetitive impact force input. The external and internal forces that have been the focus of many impact-related studies might not be the relevant variables.

THE MUSCULOSKELETAL SYSTEM AND IMPACT FORCES

When landing during heel-toe running, an impact force with a major frequency content of about 10 to 20 Hz acts on the athlete's foot. This impact force produces a shock wave that travels through the athlete's body. This shock wave is sensed by the many sensory receptors of the lower extremities, and this information is transmitted to the central nervous system.

The athlete's leg consists of the skeleton and groups of soft tissues that can be associated with major muscle groups (e.g., triceps surae, hamstrings, or quadriceps): the *soft tissue packages*. The natural frequencies of bones are rather high (200 to 900 Hz) and clearly outside the frequency range of the impact forces and the resulting shock wave. Thus, the acting impact forces are not expected to produce resonance phenomena for the skeleton. The natural frequencies of the soft tissue packages are between about 5 and 65 Hz, depending on the activation, length, and contraction velocity of the major muscles involved (14). The input frequencies of impact forces are therefore close to the natural frequencies of the soft tissue packages. Thus, resonance phenomena in the soft tissue packages might be possible.

However, experimental observations during running indicate that the soft tissue vibrations are very short and heavily damped independent of the surface shoe combination of a given running situation. Additionally, results from laboratory experiments showed that the amplitudes of soft tissue vibrations were typically below 5% of the initial amplitude after two oscillations. Thus, it is proposed that the different soft tissue packages must have had natural frequencies different

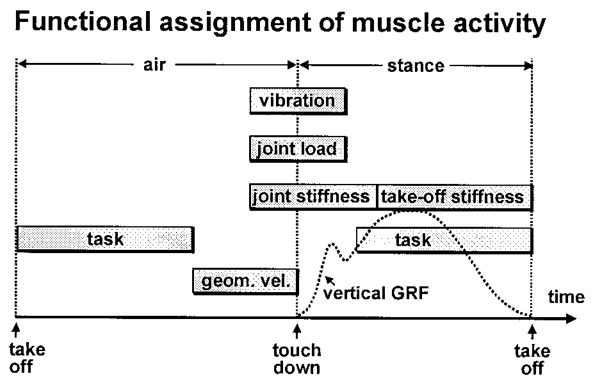


Figure 2 Illustration for the proposed functional reason and the time of occurrence for muscle activities in the lower extremities during running. GRF, ground reaction force.

than the input frequencies or their natural frequency and damping characteristics must have changed through changed muscle activation (Figure 4) to avoid resonance effects. Such changes, however, are tuning effects as they are defined for the purpose of this report.

MUSCLE ACTIVITY AND ITS PURPOSE DURING RUNNING

Muscle activity during a running stride is a complex phenomenon and has been described frequently. Muscle activity affects many aspects of locomotion, including skeletal position and velocity of the lower extremities, joint stiffness of the lower extremities, vibrations of soft tissue packages, joint loading in the lower extremities, stability during ground contact, and propulsion for the movement task at hand.

The muscle activity pattern as measured, for instance, in an electromyographic (EMG) signal is the sum of all muscle activities for the listed functions. The suggested time courses of the specific muscle activities are illustrated in Figure 2.

Muscle activities shortly before ground contact have the major goal of preparing the locomotor system for the landing and the subsequent ground contact. They are predetermined through the repetitive impact signal (amplitude, frequency, and time) experienced during previous landings. The tuning function of muscle activity suggests that shortening of the muscle-tendon unit is of minor importance in the precontact phase.

Muscle activities during ground contact have the major goal to execute the movement task at hand (e.g., heel-toe running or jumping). Muscle activity during ground contact (a) produces joint torques for movement and (b) adjusts joint stiffness during take-off. Shortening of the muscle-tendon unit is import to produce the joint torques for movement.

It is difficult to determine muscle activities responsible for specific tasks from measured muscle activity patterns (e.g., an EMG signal). However, such a procedure would be necessary to understand the actual control mechanism of locomotion and, specifically, the actual effect of repetitive impact loading on myoelectric activity and muscle tuning. Thus, protocols should be developed that allow isolation of such single muscle functions. This has not been done yet except in pilot studies. Currently, one can only speculate about the specific functions and their hierarchy during a running stride.

The authors propose that the dominant precontact muscle adjustment is made by tuning the muscles to minimize vibrations first. Specifically, muscles are tuned for soft tissue packages that are endangered with respect to resonance oscillations. Based on this initial muscle tuning, adjustments of joint stiffness and joint geometry are made with the activation of endangered soft tissue packages as overriding activity. It is proposed that there is no specific adjustment made to reduce joint loading because the soft tissue movement will reduce joint loading compared with a rigid body situation and small changes in soft tissue coupling would not make a substantial difference in joint loading. Furthermore, it is suggested that a hierarchy where joint stiffness is the dominant preactivation factor would not work because subsequent changes in muscle activity to minimize vibrations would change joint stiffness.

A NEW PARADIGM FOR IMPACT LOADING AND MUSCLE ACTIVATION

Based on these considerations, a new paradigm for understanding the reactions of the human locomotor system to repetitive impact forces is proposed.

• Impact forces are an input signal characterized by amplitude, frequency, and time.

• These signals are sensed, and the central nervous responds by tuning, if necessary, the activation of the corresponding major muscle groups.

•The tuning is done to minimize soft tissue vibrations.

• The effects are subject specific and depend on the characteristics of every single soft tissue package.

• Effects of this muscle tuning should be seen in performance, fatigue, and comfort characteristics of specific impact-subject combinations.

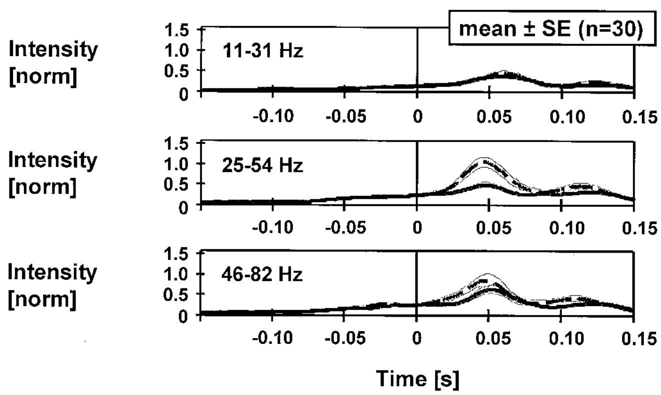


Figure 3 Normalized EMG intensity of the vastus medialis for one subject experiencing heel strike impacts on a pendulum apparatus (30 trials) for three different frequency bands (wavelets). Impact occurred at the time 0s. The experiments were performed with two different shoes: one with a soft viscous heel (*solid line*) and one with a medium-hard elastic heel (*dashed line*). The graphs illustrate how one subject reacts to changes in the input signal differently in different frequency bands (mean and SE).

INITIAL EVIDENCE

Evidence for Possible Muscle Adjustment

Evidence for the adjustment of myoelectric activity to vibrations at the workplace have been documented previously (3). Evidence for changes in myoelectric activity due to changes in footwear has been presented earlier by various authors.

Evidence that Mechanical Characteristics Can Be Adjusted by Myoelectric Activity

There is strong evidence that natural frequency and damping characteristics of "muscle packages" change with changing myoelectric activity (14). The mechanical characteristics were measured while the leg performed isometric and isotonic contractions. The myoelectric activity (EMG) needed to change frequency and damping characteristics was substantial and subject specific.

Evidence that Myoelectric Activity Changed When Input Signal Changed

Myoelectric activity changed in controlled pendulum experiments when the heel of the shoe was changed from a soft and viscous material to a medium-hard and elastic material. Changes occurred in the timing, frequency content, and intensity (and presumably the pattern of motor unit recruitment). These changes were substantial (up to 154%), not systematic, and subject and muscle specific (Figure 3 and Figure 4).

Initial evidence for subject- and muscle-specific changes in EMG activity during running has been found in a pilot study (Figure 5). The subjects performed heel-toe running on a treadmill at an individually set speed corresponding to a running intensity slightly above the aerobic threshold. They used two sets of shoes that were identical except for the material of the heel. One shoe had a soft viscous heel material, and the other shoe had a medium-hard elastic heel material. EMG was quantified for the preactivation phase corresponding to the last 50 ms before landing. The relative comparison of the EMG-RMS showed subject- and muscle-specific changes in muscle activation of up to 60%.

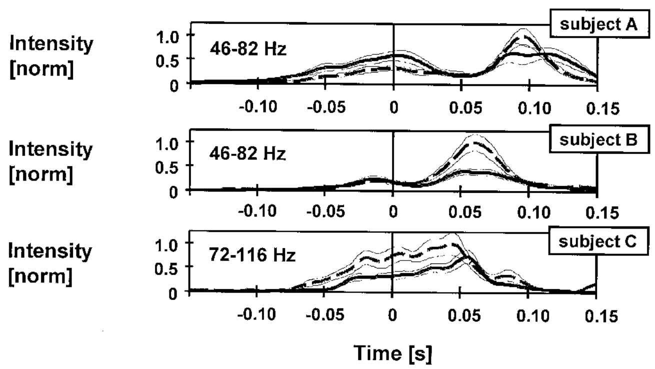


Figure 4 EMG intensity (mean and SE) of the hamstrings and the vastus medialis for different subjects experiencing heel strike impacts on a pendulum apparatus (30 trials). Impact occurred at the time 0s. The experiments were performed with two different shoes: one with a soft viscous heel (*solid line*) and one with a medium-hard elastic heel (*dashed line*). The graph illustrates that changes in myoelectric activity did occur by changing the shoe sole and consequently changing the input signal. The changes occurred with respect to intensity, frequency, and time.

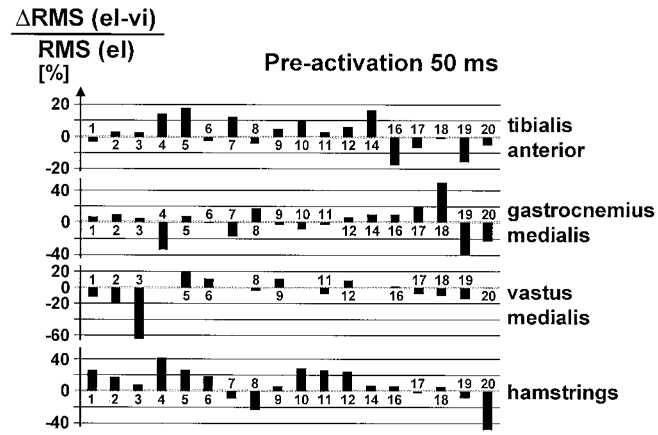


Figure 5 Relative changes in muscle preactivation (last 50 ms before first contact) for the muscles tibialis anterior, medial gastrocnemius, vastus medialis, and hamstrings for the medium elastic and the soft viscous shoe condition. Positive values indicate that the myoelectric activity (RMS) was higher for the elastic than for the viscous shoe condition. Each mean RMS was determined as an average of 24 running strides.

Evidence for Changes in Performance

Evidence for changes in performance when the input signal is changed has been determined in the same pilot study in our laboratory. Oxygen consumption during repeated treadmill running changed when the heel of the running shoes was changed from soft and viscous to medium-hard and elastic. The changes were subject specific. Some test subjects had higher oxygen consumption with the soft and viscous heel; others with the medium-hard and elastic heel; and a third group did not show a difference for the two shoe conditions. The maximal differences were about 2%.

The theoretical and experimental evidence for this proposed paradigm is certainly not conclusive. Further research should be used to solidify or reject the paradigm. Careful experiments should be designed to distinguish among myoelectric responses with respect to their components in intensity, frequency, and time. Furthermore, experiments should be designed to determine a hierarchy of strategies for muscle recruitment (if there is any). Possible evidence could result from (a) experiments that quantify muscle preactivation (intensity, frequency, and timing) shortly before impact for different impact loading, (b) experiments that quantify comfort, fatigue, and/or performance for different impact loading situations, (c) comparison of natural frequencies of soft tissue structures with frequency content of impact signals, and (d) theoretical prediction and experimental measurement of oxygen consumption for combinations of various impact and local natural frequencies.

Acknowledgments

Research related to the topic of impact loading was supported by NSERC (Natural Science and Engineering Research Council of Canada), AHFMR (Alberta Heritage Foundation for Medical Research), Adidas, and Mizuno.

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