

Impact Forces During Heel-Toe Running

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Impact forces have been speculated to be associated with the development of musculoskeletal injuries. However, several findings indicate that the concepts of "impact forces" and the paradigms of their "cushioning" may not be well understood in relation to the etiology of running injuries and that complex mechanisms may be responsible for injury development during running. The purposes of this paper are (a) to review impact mechanics during locomotion, (b) to review injuries and changes of biological tissue due to impact loading, and (c) to synthesize the mechanical and biological findings. In addition, directions for future research are discussed. Future research should address the development of noninvasive techniques to assess changes in the morphology and biochemistry of bone, cartilage, tendon, and ligaments; researchers should also try to simulate impact loading during activities such as running, focusing on the interaction of the various loading parameters that determine the acceptable windows of loading for biological tissues.

Running/jogging has developed in the last 30 years from a physical activity of a few eccentric people to an activity with millions of people involved in most industrialized countries (Cavanagh, 1980; Krissoff & Ferris, 1979). Running has become increasingly popular probably because of its easy access and its improvement of muscular and cardiovascular fitness. In Canada, running more than doubled between 1978 and 1983, from 15% to 31%, but decreased to about 18% in 1988 (Stephens & Craig, 1990; Walter, Hart, Sutton, McIntosh, & Gould, 1988). A possible reason for this decrease may be the high incidence of injury; it has been shown that 37 to 56% of all runners are injured during a year of running activity based on epidemiological studies with more than 500 subjects (Mechelen, 1992) and that running injuries make up the majority of sport-related injuries in the young (31.5%) and the old (40.5%) physically active population (Matheson et al., 1989).

The substantial numbers of running-related injuries have led to speculations and suggestions about their etiology. Major reasons for running injuries proposed in the literature include previous injuries, excessive mileage, excessive impact

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Definitions

Active forces in human locomotion are forces generated by movement that is entirely controlled by muscular activity.

Impact forces in human locomotion are forces that result from a collision of two objects, reaching their maximum earlier than 50 ms after the first contact of the foot with the ground.

Impact peak is the maximal amplitude of the impact force during the impact phase.

Cushioning describes the reduction of the amplitude of impact forces.

Load rate or loading rate is the time derivative of the force-time function. The maximal loading rate is reached between touchdown and the time of the impact peak.

Shock is a transient condition in which the equilibrium of a system is disrupted by a suddenly applied change of force.

Shock absorption describes the reduction of an impact force through the absorption and dissipation of energy (i.e., viscous behavior).

Shock attenuation describes the reduction of the amplitude of impact forces.

Shock isolation is the temporary storage and release of energy (i.e., elastic behavior).

Shock wave is a spatial propagation of mechanical discontinuity of a system.

forces, and excessive pronation (Clement, Taunton, Smart, & McNicol, 1981; Cook, Brinker, & Mahlon, 1990; James, Bates, & Osternig, 1978; Mechelen, 1992; Robbins & Gouw, 1990). Of those potential etiological factors, impact forces and foot pronation can be influenced by the sport shoe. Consequently, the concepts of "cushioning" and "rearfoot control" have been developed for running shoe construction as a result of the cooperation between researchers and the sport shoe industry, and strategies have been studied to reduce potentially excessive impact forces and foot pronation through appropriate running shoe designs.

The results of studies that related impact forces in running to musculoskeletal injuries and that were most commonly used to justify studies of impact forces and cushioning in running were typically either circumstantial in nature (e.g., James et al., 1978) or derived from experiments using animal models (e.g., Radin et al., 1973). Yet, in a more recent prospective study on short-term running injuries, subjects ($n = 131$) who were assessed at the beginning of the study as having high impact peaks in their vertical ground reaction force did not show an increased number of running-related injuries over the 6 months of monitoring in comparison to subjects with average or small impact peaks (Bahlser, 1989). In the same study, subjects assessed as having a high initial rate of loading in the vertical ground reaction force had significantly fewer running-related injuries than subjects with a low loading rate (Figure 1). In addition, the long-term cartilage degeneration that would be expected in runners, based on animal experiments, was not found to have a higher incidence in a running population in comparison to a nonrunning population (Konradsen, Berg-Hansen, & Søndergaard, 1990). However, this study had a possible selection bias: Runners who experienced knee pain may have quit running. Hence, the sample of runners only included those runners who had little or no knee pain.

These results, even considered in their limitation (limited sample size, type of injuries, uncontrolled boundary conditions), indicate that the concept of "impact forces" and the paradigm of their cushioning to influence the frequency

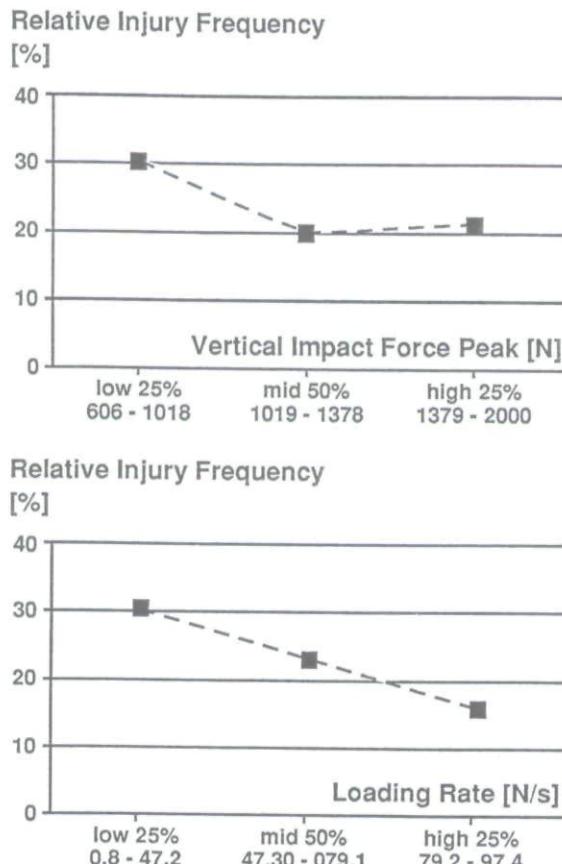


Figure 1 — Relationship between the vertical impact force peak, F_{zi} , the maximal vertical loading rate, G_{zi} , and the frequency of running-related injuries. The graphs are based on a reanalysis of data from 131 subjects (Bahlser, 1989). Their impact forces were assessed for running with a constant speed of 4 m/s at the beginning of the study. Injury occurrences were documented by a physician specializing in sports medicine.

or type of running injuries may not be well understood and that more complex mechanisms may be responsible for injury development during running. Thus, the purposes of this paper are

- to review the impact mechanics during locomotion,
- to review injuries and changes of biological tissue due to impact loading,
- to synthesize the mechanical and biological findings, and
- to discuss possible directions for future research.

Impact Mechanics During Locomotion

The literature on impact forces during human locomotion, specifically during running, uses terms that are also used in classical impact mechanics. Since some

terms are not consistently used or defined, they are discussed in this section and a set of definitions is proposed for use in impact analysis of human locomotion.

The most common terms used in impact analysis include *impact force*, *impulsive force*, and *shock*. Classical impact mechanics uses the term *impact force* for a force due to a collision of two objects that is typically short in duration (Goldsmith, 1960). The term *impulsive force* is used for a force of relatively large magnitude developed in a relatively short time. The term *impulsive force*, consequently, is more comprehensive than and includes the term *impact force*. The landing of the foot on the ground is always a collision of two objects and, therefore, an impact. Consequently, it is proposed that the term *impact force* be used in locomotion for a force due to a collision between the foot and the ground. This definition is applicable for every style of landing (heel, midfoot, or forefoot) and for every movement (running, jumping, etc.). *Impact peak* is the maximal amplitude of the impact force during the impact phase. *Load rate* or *loading rate* is the time derivative of the force-time function. *Maximal loading rate* occurs between touchdown and the time of impact peak.

Classical impact mechanics uses the term *shock* when the kinematics of a system exposed to an impact or an impulsive force is discussed (Crede, 1951). Sometimes the term *shock* is used as equivalent to *acceleration* or *force* and sometimes as a description of a transient condition in which the equilibrium of a system is disrupted by a suddenly applied change of force (Crede, 1951). It is proposed that *shock* be used to describe a transient condition, since the terms *acceleration* and *force* are well defined and don't need an expanded terminology. Consequently, a *shock wave* is a spacial propagation of mechanical discontinuity of a system.

Mustin (1968) described theory and practice of cushion designs and stated that "a cushion is anything interposed between one object and another to mitigate the effects of shock or vibration on the first object" (p. 13). In running research, the term *cushioning* has been used as a general term to describe the reduction of amplitude of an impact force without distinguishing between the different ways this is achieved (increase of the deceleration distance, dissipation of energy, change of kinematics, etc.). *Shock isolation* (Crede, 1951) is the temporary storage and release of energy (i.e., elastic behavior). *Shock absorption* describes the reduction of an impact force through the absorption and dissipation of energy (i.e., viscous behavior).

The literature describes different experimental and theoretical attempts to study impact situations and their mechanical effects, including (a) measurement of ground reaction forces, (b) measurement of segmental accelerations, (c) inverse dynamics estimations of forces acting on internal structures, and (d) simulation models (rigid body, effective mass, and wobbling mass). These approaches and their limitations are discussed in the following paragraphs.

Ground Reaction Forces

Ground reaction forces have been quantified in numerous studies. Force plates are used to quantify the forces between the foot and the ground with an accuracy of a few percentage points. The methodology is well established, and the accuracy of the results is sufficient to study the forces that act on an athlete during running. Ground reaction forces represent the inertial effects of the center of mass of the

body. They do not provide insight into the movement of individual body segments or internal forces without additional information.

Acceleration Measurements

Acceleration measurements use typically lightweight accelerometers ($m \approx 1\text{ g}$) with appropriate range and accuracy. However, acceleration measurements during locomotion may be affected by several problems (Lafortune, 1991). Three of them are discussed in the following paragraphs.

Measurements of accelerations can differ drastically depending on the method used to attach the accelerometer to the subject. Methods of attachment with test subjects can be divided into two groups: strapping the accelerometer to the segment of interest (skin mounted) and mounting the accelerometer onto a pin that is screwed into the bone (bone mounted). The amplitude measured with a skin-mounted accelerometer can be smaller than, equal to, or bigger than the acceleration amplitude measured with a bone-mounted accelerometer. The difference is determined by the mounting of the accelerometer (light or tight strapping) and by the soft tissue mass, among other factors (Nigg, 1994).

Acceleration measurements can be difficult to interpret in a descriptive sense. For example, when runners increase the amount of knee flexion during running stance (e.g., "groucho running"), the peak shank acceleration actually increases while the peak head acceleration decreases (McMahon, Valiant, & Frederick, 1987).

Furthermore, it may be difficult to determine the actual effect of an impact on the acceleration of a body segment due to other factors that influence the measured signal. An acceleration measured on a specific location of a segment provides a signal composed of rotational, translational, and gravitational components (Lafortune & Hennig, 1991; Winter, 1979) of the segment of interest.

$$\mathbf{a}_{\text{tot}} = \mathbf{a}_{\text{tr}} + \mathbf{a}_{\text{rot}} + \mathbf{a}_{\text{gra}}$$

where bold symbols indicate vectors, \mathbf{a}_{tot} = total acceleration measured with an accelerometer mounted at a specific location of a rigid structure, \mathbf{a}_{tr} = contribution to the total acceleration due to the translational acceleration of a point with no rotation, \mathbf{a}_{rot} = contribution to the total acceleration due to the rotation of the rigid structure, and \mathbf{a}_{gra} = contribution to the total acceleration due to gravity.

During running, the contribution of the rotational and gravitational components is about 45% of the total measured acceleration (Lafortune & Hennig, 1991). Acceleration signals measured with accelerometers must, therefore, depending on the question of interest, be decomposed and/or reduced to another point of the rigid structure. This decomposition may, in turn, suffer substantially from errors in the experimental measurements of the other variables required for the decomposition.

Skeletal alignment is important to consider when trying to make inferences about joint loading based on skeletal accelerations. The loading of a segment or part of it is determined by the forces and moments acting on it. The accelerations are a result of the input force at the bottom segment and the geometrical alignment of the chain of segments. A high acceleration measured at segment i may be the result of a high input force and/or a specific geometric alignment. In a rigid body case, the acceleration measured for one segment does not correspond to the

loading experienced by this segment. Knowing the acceleration of a rigid segment does not allow us to estimate, for instance, the joint forces or the material stresses for this segment. For a nonrigid body case, however, the acceleration measured for one segment may provide important information to estimate the relative movement between rigid and nonrigid parts (e.g., brain and skull or bone and muscles in thigh) and may, therefore, allow in specific cases the discussion of actual loading situations.

In summary, acceleration measurements are easy to perform. However, results from acceleration measurements are difficult to interpret properly.

Inverse Dynamics

The experimental quantification of forces in internal structures of the musculoskeletal system is typically not possible due to technical and/or ethical reasons. Force measurements in internal structures in humans have been performed in limited cases (Bergmann, Graichen, & Rohlmann, 1993; Komi, Salonen, Jarvinen, & Kokko, 1987). *Inverse dynamics* is often used to estimate forces in internal structures. The results, however, depend on the assumptions for the model, the mathematical techniques used, the distribution strategy, the quality of the segmental accelerations, and other factors (Herzog & Leonard, 1991). The quality of the segmental accelerations used to determine the inertia forces can be particularly important for the impact phase. The rigid tissues (bone) and soft tissues (muscles) of a segment have different movement patterns. Skin- or bone-mounted markers do not typically provide the appropriate movement to determine the effective acceleration responsible for the inertia effect of the segment of interest in a rigid body model. Consequently, estimations of forces in internal structures during the impact phase of running from inverse dynamics studies using rigid body dynamics may be highly inaccurate. However, results from inverse dynamic calculations may be used to establish trends in comparisons (Nigg & Bobbert, 1990) if most of the shortcomings of this approach are systematic in nature.

Simulation Models

Simulation models can be used to estimate the kinematics and/or kinetics of a system of interest. They allow for the estimation of forces in structures that often could not be estimated or measured using other approaches. Additionally, simulation models provide the opportunity to study the effect of systematically changed input variables on the output variables of interest. However, the results of simulation models depend on the assumptions implemented into the model and its mechanical and conceptual construction. For the study of impact mechanics, simulation models are associated with problems. The human musculoskeletal system is difficult to model, and authors disagree about which mechanical elements are important for impact simulation. During impact, substantial soft tissue movement occurs throughout the body, which may be important to include in the model. Muscle activation is not easy to establish in a two-dimensional model and is extremely difficult to implement in a three-dimensional model. In general, however, simulation models seem to have a unique potential for mechanical impact analysis for selected questions.

The following discussions concentrate on the impact phase during running.

The most common form of running is heel-toe running. About 81% of all runners/joggers impact the ground initially with the heel (Kerr, Beauchamp, Fisher, & Neil, 1983). The center of mass of the body is decelerated vertically for the first half and accelerated vertically for the second half of ground contact. However, for some segments of the runner's body, the deceleration lasts only a few milliseconds, much less than half of ground contact. The foot and the leg, for instance, are typically decelerated in less than 50 ms (Figure 2) (Bobbert, Schamhardt, & Nigg, 1991). The effect of these segmental decelerations can be seen in the vertical ground reaction force curves as a distinct force peak. These vertical impact force peaks were the variable of interest of most research concentrating on impact forces during running over the last 25 years.

In order to understand the propagation of impact forces or accelerations through the human body, the following sections discuss factors that influence these forces and accelerations, namely kinetic energy, material properties, geometrical alignment of body segments, and muscular activation. In each section, the theory behind each factor is first presented with the aid of a simplistic, two-dimensional model. This is followed by selected results of running studies pertaining to each factor, after which the relevance of each factor to the impact forces that occur in running is summarized.

Kinetic Energy

In discussing the mechanics of impact, we begin with a set of n rigid structures connected by hinge joints. In the first case the segments are aligned vertically so that the angle between neighboring segments is 180° (Figure 3, left). When the system contacts the ground, a force is generated at the point of contact. The magnitude of the force will depend on the kinetic energy of the system at the

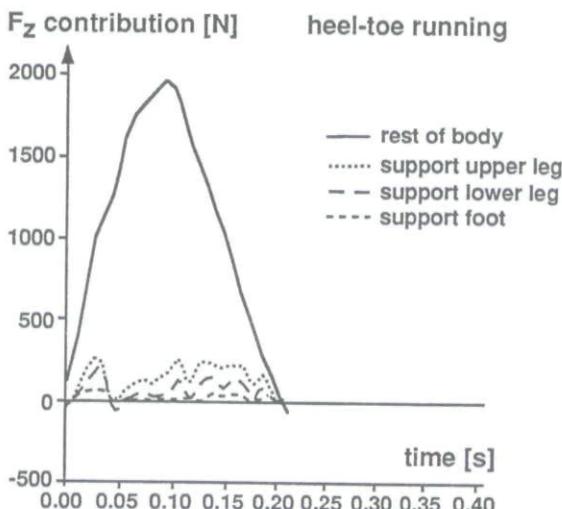


Figure 2 — Illustration of a vertical force-time curve for heel-toe running as measured with a force plate and segmental contributions to this result as determined from kinematic analysis (adapted from Bobbert et al., 1991).

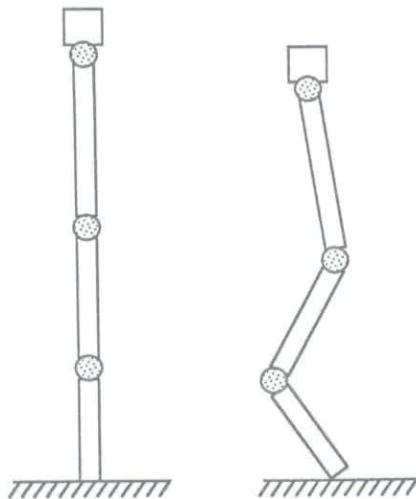


Figure 3 — Two-dimensional representation of a chain of five rigid segments connected with hinge joints in a straight alignment (left) and a not straight alignment (right) to explain the propagation of forces or accelerations.

instant of contact. By assuming linear motion, this energy, obviously, is dependent on the mass and vertical velocity of the system. These effects can be observed in experimental and theoretical studies of impact in running. For example, it has been found that an increase in running speed relates to a substantial increase (more than 100% for running speeds of 3 or 6 m/s) in vertical impact force amplitudes (Frederick & Hagy, 1986; Hamill, Bates, Knutzen, & Sawhill, 1983; Nigg, Bahlsen, Luetli, & Stokes, 1987). Also, an increase in the vertical touch-down velocity of the heel relates to a substantial increase in vertical impact force amplitude and loading rate (Gerritsen, Bogert, & Nigg, in press). And body mass was found to explain 32% of the intersubject variability in impact force peaks in running (Frederick & Hagy, 1986). Of the numerous anatomical and control variables studied, including speed and skeletal alignment (see below), body mass was the most highly correlated with impact forces.

Material Properties

When the distal segment of the system in Figure 3 comes into contact with the ground, a state of disequilibrium exists in the segment due to the instantaneous state of stress/strain at the contact point. Particles in the material move according to the instantaneous stress distribution, and a longitudinal stress wave is propagated. For this simple one-dimensional system, the wave travels through the medium at a speed that is dependent on the elastic modulus and density of the material. When the wave reaches a discontinuity in the medium, a joint, it is transmitted proximally through the joint but also reflected back toward its source. The reflected waves interfere with the waves that are continuing to be propagated from the area of contact. As a result of the finite speed of propagation and the superposition of waves upon one another, a nonuniform distribution of stress will exist within the material. However, if the stress waves travel fast enough to traverse the length of each segment many times over the duration of the impact, then a quasi-static condition is said to exist. Under quasi-static conditions, elastic wave motion in the bodies can be ignored, the total mass of each body is assumed

to move with the velocity of its center of mass at any instant in time, and rigid body dynamics can be used to describe the motion of the system. In the n body system (Figure 2, left) the segments will be uniformly accelerated with the forces in the joints decreasing from the ground up.

When elastic wave effects are not negligible, rigid body dynamics cannot be used to describe the motion of the system. Forces in the joints will still decrease from the ground up; however, the acceleration measured at any point will also decrease from the bottom to the top. The peak ground reaction force that is generated will be less than in the quasi-static situation.

When the material is viscoelastic, the energy of a stress wave can be absorbed and dissipated as heat within the material. The amplitude of the wave will be attenuated as it passes through the material. The amount of attenuation typically depends on the frequency of the input signal, the ground reaction force in this case, which determines the wavelength in the material. Phase shifts in the transmitted signal can also occur.

In the human body, the segments in the model (Figure 2) would consist of bone with cartilage at the joints, except in the upper segment (spine), which would have a much higher content of cartilage. Additionally, the heel pad and midsole of the running shoe would be added as material elements to the distal segment. Theoretical studies of wave propagation effects in nonlinear viscoelastic biomaterials are difficult to conduct. There are a limited number of studies addressing the question of how much the materials in the human body contribute to cushioning and shock absorption during the impact phase in running; a brief discussion of these studies follows.

The speed of longitudinal wave propagation in the long bones of the lower extremity is approximately 3,200 m/s (Chu, Yazdani-Ardakani, Gradisar, & Askew, 1986; Pelker & Saha, 1983). The wavelength of an input signal of 12 Hz, typical of the impact forces in running, would be about 260 m. Within the time frame of the impact phase in running, about 30 ms, these waves would be able to traverse the length of the bones of the lower extremity numerous times, assuming the influence of articular cartilage to be negligible due to its limited thickness. In the spine, the waves will not propagate as quickly due to the additional amounts of cartilage. Radin and Paul (1971a) found, *in vitro*, that although articular cartilage is a more effective shock absorber than an equivalent amount of bone, the long bones absorb greater amounts of impact forces because there is so little articular cartilage in the lower extremity. However, the frequency of the applied force was not presented, making it difficult to compare this study to human running. Using a splinted-knee rabbit model, Paul and coworkers (1978) found that input signals with a frequency content between 3 and 18 Hz were only very slightly attenuated across the ankle joint and along the tibia. Attenuation increased with increasing frequency content of the input force. Chu and coworkers (1986) found a substantial amount of attenuation (40%) of an impact force through the tibia, knee joint, and femur of embalmed human cadavers. However, the wavelength of the input signal was 35 cm, resulting in an input frequency of about 8000 Hz, substantially higher than the frequency content of the impact forces in running. The peak acceleration measured at the head occurs later than the peak acceleration measured at the shank during running (Shorten & Winslow, 1992). This result suggests that there is viscoelasticity somewhere along the skeletal system.

Based on these few results, it is speculated that a quasi-static condition of force propagation exists through the skeletal system of the lower extremity during impact in running. The skeletal system of the lower extremity probably contributes little to attenuation of impact forces in running. Viscoelastic wave propagation effects may occur in the spine due to its additional amount of cartilage, and this may contribute to the attenuation of higher frequency components prior to reaching the head.

Considerably more research exists pertaining to the shock absorption and cushioning properties of the heelpad and the midsole of the running shoe. Selected results follow:

Results from material tests (measurements with an impact tester) showed substantial increases (more than 40%) in the vertical impact force amplitudes with increasing stiffness of the shoe sole (Frederick, Clarke, & Hamill, 1984), whereas results from subject tests (force plate measurements during running) showed only minimal differences in the vertical impact force amplitudes with increasing stiffness of the shoe sole (Clarke, Frederick, & Cooper, 1983; Nigg, Denoth, Lüthi, & Stacoff, 1983). Results from subject tests using kinematic and kinetic information input into a mathematical model to estimate internal forces in the foot and ankle joint complex showed no large differences in impact loading during running for systematic changes in midsole hardness (Cole, Nigg, Fick, & Morlock, 1995). In the same study, impact loading in the ankle joint complex was found to be significantly higher when subjects ran barefoot in comparison to running in shoes.

Results from subject tests (measurements with accelerometers mounted on bone pins screwed into the tibia) showed substantial decreases in the acceleration amplitudes for walking with increasing cushioning in the shoe (Light, MacLellan, & Klenerman, 1979). Results from material and subject tests (impact tester, force plate measurements with runners, and accelerometer measurements with walkers and runners) showed an increase in the time from first contact to maximal amplitude with decreasing shoe sole stiffness (Frederick et al., 1984; Lafontaine & Hennig, 1991; Light et al., 1979; Nigg et al., 1987). Results from subject tests (force plate measurements during running) showed only minimal differences in the vertical impact force amplitudes for changes in the shoe sole thickness (Denoth, Gruber, Keepler, & Ruder, 1985), and results from material tests (measurements with an impact tester) showed no changes in the vertical impact force amplitudes for systematically changed heel flares (Frederick et al., 1984).

Vertical impact force amplitudes were increased substantially in subject tests by removing the lower part of the heel cap and by allowing the heel pad to expand during the impact phase (Jørgensen & Bojsen Möller, 1989). Peak accelerations during pendulum impacts on human cadaver feet were substantially reduced when the heelpad was left intact, compared to when it was surgically removed (Noe, Voto, Hofmann, Askew, & Gradisar, 1993). In vivo pendulum impacts on human feet also suggested that the heelpad acts as a shock absorber (Aerts & DeClercq, 1993). In some contrast to the above results, Instron tests on isolated cadaver calcanei with intact heelpads suggested that the heelpad dissipates between 20 and 25% of the energy in a loading-unloading cycle (Bennett & Ker, 1990).

The results support the concept that the heelpad and the running shoe act

as a shock absorbing and cushioning mechanism. The running shoe adds a substantial amount of cushioning in comparison to the barefoot condition. However, the contradictory results from midsole material and subject tests were certainly unexpected. Specifically, the result that changes in midsole hardness do not seem to affect the external and internal impact force amplitudes is counter to what one would expect and seems to contradict the general feeling of runners that softer shoes are more comfortable (note that not all runners would agree with this statement). The result that the time to the impact peak in the vertical ground reaction force increases with decreasing midsole hardness suggests that runners' perceptions may be related to the frequency content of the loading, or jerk, that is experienced at impact.

Skeletal Alignment

Skeletal alignment refers to an alignment between two neighboring segments that will result in movement at the joint between the segments upon impact (Figure 3, right). In this situation, some of the kinetic energy of the system is transformed into angular motion at impact rather than strictly into linear motion as in the straight alignment case presented above (Figure 3, left). For a given impact velocity, the peak force generated at the point of contact will be less than in the straight alignment case, the forces in the joints will decrease from bottom to top (but differently than in the first case), and the vertical accelerations of the center of mass of each segment will decrease from bottom to top. Assuming that viscous effects in the material are negligible, peak joint forces and segment accelerations will all occur at the same time. This mechanism of impact force reduction is used substantially in running, as illustrated by the following selected results.

Results from subject tests (force plate measurements during running) showed substantial decreases in the vertical impact force amplitudes for increases in lateral heel flare (Denoth et al., 1985; Nigg & Bahlsen, 1988; Nigg & Morlock, 1987). The pronatory movement of the foot acts as an additional cushioning/damping element (Stacoff, Denoth, Kaelin, & Stuessi, 1988). Results of both experimental and theoretical investigations show that the skeletal alignment of the lower extremities, in particular knee flexion, dorsiflexion, and foot inversion, influences external and internal impact force amplitudes and frequency content substantially (Denoth, 1986; Frederick & Hagy, 1986; Gerritsen et al., 1995; Koning, Jacobs, & Nigg, in press; Nigg, 1986). Results from subject tests (force plate measurements during running) showed no differences in the vertical impact force amplitudes for subjects with high and flat foot arches (Nachbauer & Nigg, 1992).

Changing the skeletal alignment at the time of heelstrike in running is an effective means of reducing impact forces. However, there can be a substantial metabolic energy cost in doing so (McMahon et al., 1987).

It is interesting to note that the two postulated running shoe concepts, cushioning and stability (i.e., control of the pronatory movement of the foot), show opposite tendencies. By increasing one of them, one typically decreases the other (Cavanagh, 1980). However, it was proposed recently that this contradiction is caused by the isotropic and homogeneous materials used in the construction

of shoe soles and that other materials (e.g., orthotropic) should not show these negative properties (Stüssi, Stacoff, & Lucchinetti, 1993).

Muscular Activation

The movement that occurs at each joint due to the skeletal alignment described above is dependent, among other things, on the rotational stiffness at each joint. This stiffness is determined by muscular activation. Theoretically, it would be expected that if all other conditions remain the same, increased joint stiffness should result in higher peak impact forces. However, the muscles are also able to absorb some of the rotational energy of the system during impact due to their viscous nature.

The human body is not actually made up of a set of rigid segments as described in Figure 3. The skeletal system is fairly rigid; however, the muscles, internal organs, skin, and adipose tissue are not. These tissues will move relative to the skeletal structures during impact in running. These tissues will not experience the same high accelerations as the skeletal structures, and the effect will be reduced impact forces. Increased muscular activation, both in the lower extremity and in the upper body, should increase the overall rigidity of the system and, hence, the impact forces. The following are some results from the literature.

Using a direct dynamics simulation model of impact in running, Gerritsen et al. (1995) found that inclusion of muscles resulted in a decrease in the peak vertical ground reaction force and an increase in the time to the peak. However, assuming constant resultant joint moments at the time of heelstrike, changing muscular activation had only a small influence on the peak vertical ground reaction force in the same study. More recent simulations of impact in running conducted in our laboratory have suggested that changing muscular activation may have a greater influence on the vertical ground reaction force than previously suggested.

It has been suggested that the activations (EMG) of lower extremity muscles just prior to heelstrike change systematically with changes in shoe sole stiffness (Komi, Hyvärinen, Gollhofer, & Kvist, 1993). However, statistically significant differences in the EMG variables measured were not found. From personal observations of human movements filmed at 400 Hz, there is considerably more soft tissue movement in the lower extremity during the impact phase of running in comparison to landing from a vertical jump, a result that is correlated with the level of muscular activation in each activity. Results from a simulation model using rigid and "wobbling" parts (Gruber et al., 1987) suggested that the wobbling masses contribute to a reduction of the resultant joint forces and moments during landing from a vertical jump. The reduction increases (among other factors) with increasing wobbling mass.

The effects of changing muscular activation prior to heelstrike in running are not well understood. This is viewed as an area for future research because of the potential interaction between boundary conditions in running and muscular activation.

Impact Forces, Injuries, and Tissue Reactions

Studies assessing the effect of impact forces during running on injuries to the musculoskeletal system can be grouped into two major categories: epidemiological studies using actual runners and studies using animal models. Epidemiological

studies typically used a prospective or retrospective strategy assessing factors that were assumed to influence impact loading (shoe sole hardness, running style, surface hardness, etc.). Typically, these studies could not be controlled, and multifactorial influences may have blurred the results. Studies using animal models have the advantage that they can be well controlled. However, the transfer of results from animal models to human runners is rather difficult.

Epidemiological Studies

Results of epidemiological studies assessing the association between impact loading and the development of acute or chronic injuries can be summarized as follows.

A prospective study could not find an increase in the frequency of minor acute injuries for runners with high impact peaks or high impact loading rate compared to those with low impact peaks or impact loading rates, respectively, over a period of 6 months (Bahlsen, 1989).

Peak joint contact force magnitudes in the lower extremity have been estimated using inverse dynamics to be substantially less during the impact phase than the active phase in running due to the increased muscular forces in the active phase (Burdett, 1982; Harrison, Lees, McCullagh, & Rowe, 1986; Scott & Winter, 1990). Based on this result, it was speculated that impact forces are not related to the development of injuries in running (Scott & Winter, 1990). However, joint force estimates during the active phase are more prone to errors resulting from the distribution of resultant joint forces and moments. Therefore, comparisons of joint contact forces between impact and active phases are not necessarily valid, as the errors are not systematic. Additionally, none of these studies considered the temporal aspects of the joint forces, which, based on observations of ground reaction forces, are the main differentiators between impact and active phase loading.

In a review of the related literature, Mechelen (1992) concluded that running on hard surfaces did not increase running injuries compared to running on softer surfaces. McMahon and Greene (1979) suggested, based on circumstantial evidence, that running on "tuned tracks" reduced impact forces, increased running comfort, and was "apparently responsible for a very low rate of running injuries." However, these suggestions were not supported by epidemiological data.

Runners, as a group, do not show a higher incidence of osteoarthritis in comparison to nonrunners (Eichner, 1989; Konradsen et al., 1990; Lane et al., 1986; Panush et al., 1986). This result suggests that the impact forces in running are not related to the development of osteoarthritis. However, this does not preclude the possibility that there is a subgroup of runners with high impact forces and/or high repetitions of these forces who are at risk for development of osteoarthritis.

Shock-absorbing insoles were not effective in reducing the incidence of stress fractures in military recruits (Gardner et al., 1988; Schwellnus, Jordaan, & Noakes, 1990). However, in one study the shock-absorbing insoles were able to reduce the general frequency of injuries (Schwellnus et al., 1990). In studies such as these, it is often not evident whether the tested strategies for impact reduction were ineffective because their cushioning effect was minimal or whether the occurring injuries resulted from factors other than the impact forces.

The ratio of the peak acceleration of the medial femoral condyle to the peak acceleration of the forehead during walking, measured using elastically strapped accelerometers, was found to be lower for subjects with than for subjects without low back pain (Voloshin & Wosk, 1982). This result was interpreted as "a reduced capacity of the musculo-skeletal system . . . to attenuate incoming shock waves" (p. 21). As a result, impact forces were implicated in the etiology of low back pain. However, assuming that the accelerations were measured properly, one may interpret these results differently than did the authors. The accelerations they measured at the forehead were the same for the two groups of subjects, whereas the accelerations measured at the femoral condyles were higher in the control group than in the low back pain group. Thus, the control subjects may actually have been exposed to higher shocks than the subjects in the low back pain group. Additionally, it is possible in cross-sectional studies such as this that pain influences the gait pattern and, as a result, the variables measured. Consequently, conclusions about cause and effect are not appropriate in such studies.

The use of a viscoelastic heel pad was shown to be effective in reducing the symptoms of Achilles tendinitis in an athletic population (MacLellan & Vyvyan, 1981). It was speculated that when the shock resulting from heelstrike is transmitted from bone to soft tissue, "substantial traction, shear and overswing phenomena" take place and that the viscoelastic insert was able to reduce these distortions. However, no evidence was provided to support these speculations.

The results of most of these studies are inconsistent and rather difficult to interpret. In light of this and considering the limitations of most of these studies, one certainly cannot use the results of these studies to support the notion that impact forces are an important factor in the development of chronic or acute running-related injuries.

Hematological Changes Due to Running and Impact Loading

Hematological variations were quantified for runners using hard- or soft-soled running shoes. Running in general was shown to lower haptoglobin levels. Furthermore, soft cushioning running shoes (in this case air-cushioned shoes) were speculated to reduce the hematological effects of acute mechanical damage (Falsetti, Burke, Feld, Frederick, & Ratering, 1983). Additionally, it was shown that after intensive running, reticulocyte counts were significantly higher for runners using hard-soled running shoes compared to runners using soft-soled running shoes (difference in impact peaks of 18%) and that the overall change in the percentage of reticulocytes was significantly correlated to the magnitude of the results of the impact test for the shoe soles (Dressendorfer, Wade, & Frederick, 1992). It was speculated that running shoes with "good cushioning" may help runners maintain their normal red blood cell turnover rate. However, no actual differences in red blood cell count were observed between the two groups of runners in the study.

Tissue Reactions—Changes in Cartilage

Several authors have studied the effect of impact or impulsive loading and/or the effect of running activities on changes in biochemical and physical properties of cartilage. Selected results are summarized in the following paragraphs.

Bovine metacarpal-phalangeal joints were exposed to oscillating low-frequency forces. At the peak of the low-frequency forces, periodic impact forces were superimposed (Radin & Paul, 1971b). The amplitudes of the low-frequency forces were about four times the normal physiological forces and were just below the structural capabilities. Joints exposed only to the low-frequency forces did not show significant wear, while cartilage wear was detectable as early as 12 hours after the onset of the experiment in joints exposed to the combination of low-frequency forces and additional impulsive forces. Several aspects make it difficult to transfer these results to human running. The authors did not apply a force pattern that was typical for impact forces in running. Experiments in which isolated impact forces were applied were not performed. Furthermore, the acting stresses are probably much lower in running than the stresses that were present in this experiment.

A splinted-knee rabbit model was used to study the effect of repetitive impact loading on articular cartilage (Dekel & Weissman, 1978; Radin et al., 1973; Serink, Nachemson, & Hansson, 1977). The rabbits showed changes in their knee cartilage consistent with those of degenerative cartilage disease. However, none of these studies used loading conditions that clearly represented the forces applied in human running. Radin and coworkers (1973) applied impulsive loads of 1 BW (one body weight) with a frequency of 60 Hz, which was somewhat lower than the frequency of typical single limb loading that would occur at average running speeds. Additionally, the temporal characteristics of the loading were not described, and it is not known (based on the published information) if the applied forces corresponded more to impact or active forces in running. Additional detail of the loading conditions was presented by Serink and coworkers (1977). The duration of each load cycle was 420 ms with the peak force occurring at 210 ms; this loading rate was lower than both the impact and active loading rates typically observed in running. Dekel and Weissman (1978) applied a peak force of nearly 10 BW, which is orders of magnitude higher than the peak impact forces in running.

Adult sheep that were exposed to activities on hard (concrete) or soft (wood chip) surfaces for 2-1/2 years showed significant decreases in the hexosamine content of their knee articular cartilage, a result that can also be observed in early osteoarthritis. The group walking on the hard surface had a higher decrease than the group walking on wood chips (Radin, Orr, Kelman, Paul, & Rose, 1982). It was concluded that this change was due to the prolonged walking on the hard surface. Gross pathological changes of osteoarthritis were not observed. Internal forces were not measured, and it cannot be concluded whether the detected changes were the result of increased impact loading or the result of changes in joint geometry (to mention just one other possible reason).

Impact forces were applied to the surface of canine femoral cartilage (Christman, Ladenbauer-Bellis, & Panjabi, 1981). The authors found a fourfold increase in arachidonic acid in the phospholipid pool due to the impact treatment, an indication of early biochemical changes toward osteoarthritis. The magnitude of the forces applied was about 75% of the force that was previously shown to cause fracture and total cartilage necrosis (Repo & Finlay, 1977) and was chosen to simulate a traumatic event such as a severe blow to the knee in an athletic event, rather than to simulate repetitive impact loading that occurs in distance running.

Cartilage and subchondral bone from patellae were subjected to cyclic compression of 1,000 psi (≈ 7 MPa) with a ramp loading of 0.3 s. Primary fissures were detected at 500 cycles, and secondary fissures were observed at 1,000 cycles. However, fissures did not occur for cyclic compressions of 250 to 500 psi (≈ 1.75 to 3.5 MPa) even if the superficial layer of cartilage (100 μm) was removed and the number of loading cycles went over 120,000 (Zimmerman, Smith, Pottenger, & Cooperman, 1988).

A relationship was found between the stiffness of articular cartilage and the typical stress level to which the cartilage was exposed, which was interpreted as an adaptation effect of cartilage to stress (Swann & Seedhom, 1993). Osteoarthritic damages occurred typically in areas of infrequent but excessive stress levels. Furthermore, it has been shown by the same group that the ankle joint exhibited much lower incidence of osteoarthritis than the knee joint even though the two joints were exposed to the same number of loading cycles.

Rabbit knees were exposed to low- and high-frequency loading with the result that "severe changes occurred in joints of high load rate animals significantly more often ($p < 0.001$) than in joints of low load rate animals even though the load magnitudes in the latter group were greater" (Yang et al., 1989, p. 148). Loads were applied at 1 Hz for a duration of 50 ms with peak loads of 0.83 and 0.58 BW in the low and high load rate groups, respectively. Similarly, a highly impulsive load of 1.5 BW developed in 50 ms produced greater cartilage damage in the rabbit knee than a mildly impulsive load of 1.5 BW developed in 500 ms (Anderson, Brown, Yang, & Radin, 1990). The loading conditions in the highly impulsive cases above are relatively similar to the impact observed in the vertical ground reaction force in running, which reaches magnitudes of between 1 and 2 BW within 30 ms. However, certain limitations apply to these studies as well as the previously cited studies using the splinted-knee rabbit model. The relation between the articular contact stresses applied and the stresses that typically occur in the joints of runners was not addressed. Additionally, the loading regime of 40 min/day for 5 days/week is fairly severe even for human runners. If one assumes that cartilage adapts to the stresses acting on it (Swann & Seedhom, 1993), then it would be reasonable to expect different effects of this impulsive loading regime in the rabbit, a fairly sedentary animal, than in humans who have had many years to adapt to it.

A theoretical study using a finite element approach (Anderson et al., 1990) addressed the effect of loading rate (temporal input scaling) on resultant joint force and joint stress. The authors showed that the temporal input scaling affects the resultant joint force substantially in a nonlinear manner.

Fissuring and chondrocyte death of cartilage matrix has been reported for impact stresses greater than 25 MPa (Repo & Finley, 1977) based on in vitro tests. Similar numbers have been proposed by other authors. However, recent in vivo studies with humans provided joint contact stresses for the human patello-femoral joint of up to 40 MPa for an isometric contraction (40% maximal isometric force) and a knee angle of 75° (Ronsky, 1994). Similarly, experimental in situ measurements provided joint contact stresses for the cat patello-femoral joint of up to 40 MPa for an isometric contraction (40% maximal isometric force) and a knee angle of 75° (Ronsky, 1994). Consequently, cartilage seems to behave differently in vivo as compared to in vitro.

The findings on changes in cartilage due to load can be summarized as follows.

Moderate intermittent loading of cartilage stimulates chondrocytes to increase biosynthesis. Specifically, moderate running produces thickening of cartilage and augments glycosaminoglycan, effects which are considered biopositive. Excessive loading of cartilage produces fissures and biochemical changes, effects which are considered bionegative corresponding to changes of early arthritis. It is not well understood how the stresses required to produce degenerative cartilage changes in the repetitive impulsive loading models correspond to the actual articular cartilage stresses experienced during human running. Nor is it well understood how changes in the frequency characteristics of the applied load effect biochemical or mechanical reactions in cartilage for force/stress levels that correspond to force/stress levels in running. Finally, osteoarthritic damages occur frequently in areas of infrequent stress.

Tissue Reactions—Changes in Bone

Bone modeling and remodeling are highly influenced by mechanical stimuli that result from daily activities. Selected results of studies that evaluate changes in biochemical and physical properties of bone in response to mechanical loading are presented in the following paragraphs.

Bedrest studies provide consistent results, with loss of Ca^{2+} and a reduction of bone mineral content typically measured at the calcaneus (Donaldson et al., 1970; Schneider & McDonald, 1984). Studies depriving only well-defined bone locations from mechanical stimuli due to casting indicate that loss of bone mass is specific to bones exposed to these casting regimes (Nillson, 1966; Westlin, 1974). Research with an animal turkey model isolating the left ulna diaphysis from mechanical stimuli for 8 weeks found a 13% decrease in bone mass compared to the intact contralateral control.

Bone is able to increase its mass as a response to increased mechanical stimuli with the adaptation occurring in the area exposed to the high stress (Forwood & Burr, 1993). Ballet dancers showed increased bone mass in the metatarsal bones (Warren et al., 1991). Tennis players at the professional level showed a 33% increase in their forearm cortices for the arm holding the racket compared to the other arm (Jones, Priest, Hayes, Tichenor, & Nagel, 1977). Results from a cross-sectional study suggested that exercise consistently maintained over the entire lifespan is associated with higher skeletal mass (Dalen & Olsson, 1974). A 1-year jogging program (40 km/week) for swine increased the femoral midshaft cross-sectional area by 23%; however, the selected program did not alter the bone mineral content (Woo et al., 1981), a result that has been verified with mice and rats (Kiiskinen & Heikkinen, 1978; Simkin, Aylon, & Leichter, 1989). Effects of mechanical stimuli can be seen rather soon. Ultrasound velocity at the patella, for instance, increased 3.5% between pre- and postmarathon (Rubin & Lanyon, 1987). In summary, intensity and type of exercise influence bone response. Activities such as running, dancing, or weight lifting typically increase skeletal mass, while exercises such as swimming seem to provide this effect less consistently (Gross, 1993).

In a study analyzing the effect of compressive and bending loads on bone remodeling of the radius and ulna of sheep, strain rate was varied (between about

0.25 and 3.3 times the peak strain rate measured during normal locomotion). Strain rate was found to be the best predictor of the variation in the amount of surface bone deposited, explaining 68 to 81% of this variation (O'Connor & Lanyon, 1982). In another study, strain was applied as a sinusoidal signal to bone (McLeod et al., 1990). It was found that a 1-Hz signal was not able to maintain bone mass over an 8-week period, while a 15-Hz signal stimulated substantial new bone formation. Magnitude and strain rate occurring in the bones of the lower extremities during human running are not known. However, if the vertical ground reaction forces and the tibial axial acceleration signals are simplified as sinusoidal waveforms, their frequency content is assumed to be generally between 10 and 20 Hz. Consequently, one might expect a positive osteogenic response due to impact loading in running.

Runners exposed to a rigorous training program of 1,500 km in 5 months showed changes in their bone mineral density that were higher for those running with hard midsoles compared to those running with soft midsoles (Brüggemann, personal communication). The time to the impact peak in the vertical ground reaction force has been shown to decrease with increasing midsole hardness of the running shoe (Nigg et al., 1987), effectively increasing the loading rate and possibly the strain rate for the bones of the lower extremities. The reported change in bone mineral density could, therefore, be explained by the result presented in the preceding paragraph.

Excessive mechanical stimuli, however, seem to affect bone integrity adversely. Intensive "basic training" with military recruits produced stress fractures in 43% of subjects (Leichter et al., 1989). High-intensity running in mice and rats had negative effects on bone integrity (Forwood & Parker, 1991; Kiiskinen & Heikkinen, 1978). Excessive endurance swimming with rats was associated with a loss in trabecular bone in the femur (Bourrin, Ghaemmaghani, Vico, Chappard, & Alexandre, 1992).

The influence of impact loading on the development of tibial stress fractures was investigated using the splinted-knee rabbit model (Burr et al., 1990). Repetitive impulsive loading (1.5 BW, 1 Hz, 40 min/day, 5 days/week, 25 ms rise time) produced stress fractures in the tibial diaphysis generally within 6 weeks of the onset of the experiment. The loading characteristics in this experiment were similar to the characteristics of the impact peaks of the vertical ground reaction forces in heel-toe running. The number of days of loading (5 days/week) is somewhat higher than the typical number for a standard training protocol for runners, particularly since these rabbits would be considered "untrained" (they were not allowed to gradually adapt to the impact condition).

The current discussion about the effects of exercise on bone integrity concentrates on the development of various models that should allow for the prediction of bone adaptation. There seems to be general agreement that bone strains or strain rates between a minimal and a maximal threshold increase the bone modeling and, therefore, increase bone mass, while strains or strain rates outside these thresholds will be associated with bone remodeling and loss of bone mass (Frost, 1986; Grimston & Zernicke, 1993; Martin & Burr, 1989; Whalen & Carter, 1988).

The findings on changes in bone due to load can be summarized as follows. Immobilization of a specific bone is typically associated with a loss of bone mass and integrity, whereas exercise typically increases bone mass and integrity.

There is evidence to suggest that impact loading is better able to increase bone mass than nonimpact loading (e.g., running versus swimming), but there is also evidence that excessive impact loading may result in stress fractures. Specifically what constitutes "excessive loading" in terms of magnitude of load, rate of loading, frequency of load application, duration of loading cycles, and rest time between loading cycles is not well understood.

Methodologically, it is difficult to assess the influence of impact loading during running on bone integrity for a number of reasons. Strain magnitudes and strain rates are not presently known and cannot be measured without using highly invasive procedures. Strains are influenced not only by the external forces (e.g., ground reaction force) but also by internal forces (e.g., tendon and ligament forces), bone geometry, and material properties of the bones. It is our opinion that even qualitative estimates of the strain histories that are important for bone remodeling are difficult to make based on observation of the external loading conditions.

Nevertheless, the body of knowledge presented in the literature gives reason for some speculative comments. Stress fractures are common running injuries (Clement et al., 1981; Matheson et al., 1989). One theory of the etiology of stress fractures suggests that the fracture begins as an osteogenic response to the loading conditions (bone apposition precedes bone deposition in the remodeling process) and that successive loading cycles occur within a time frame that does not allow completion of the remodeling process so that bone apposition is in excess of bone deposition (Grimston & Zernicke, 1993). The results presented above suggest that impact loading stimulates a greater osteogenic response than nonimpact loading. Consequently, it is reasonable to assume that impact loading could result in stress fractures if the loading cycles were repeated without sufficient rest.

Speculations and Future Research

There is agreement that excessive impact forces may damage the human musculoskeletal system and that there is a window of loading in which biological tissues react positively to the applied impact loads. However, it is not evident whether the forces/stresses acting on cartilages, bones, ligaments, and tendons during specific activities are within or outside this window. The analysis of effects of loads acting on the musculoskeletal system is made even more difficult since the ultimate material properties for biological tissue *in vivo* are not well established.

The boundaries for the acceptable loading windows to avoid chronic injuries associated with running or other physical activities are a complex interaction between magnitude of load, rate of load, frequency of load applications, duration of the loading cycles, and rest time between loading cycles. This interaction is presently not well understood.

Assessment of where impact loading falls in relation to this loading window based on our present knowledge of the impact phase in running is difficult because of the complications that arise in estimating the required stress/strain time histories from measurements of external loading, acceleration measurements, or models.

Assessment of where impact loading falls in relation to this loading window based on epidemiological studies suggests the following.

For moderate running, impact loading of cartilage, bone, and soft tissue structures falls within the acceptable window. For intensive running, impact loading of cartilage and soft tissue structures falls within the acceptable window, but impact loading of bone may sometimes fall outside of the acceptable window. The knowledge base upon which these speculations are made is limited. The two basic scenarios that impact loads during running fall within or outside the acceptable window are both possible, and future research is needed to support or reject them conclusively.

Further research studying the effect of impact forces during running on biological tissues needs to concentrate on the following aspects:

- Development of noninvasive techniques that can be used for *in vivo* investigations with runners which allow the assessment of changes in the morphology and biochemistry of bone, cartilage, tendon, and ligaments.
- *In vivo* studies with runners quantifying the occurrence of osteoarthritis and stress fractures using the above-mentioned refined techniques to assess the changes to the biological tissues.
- Longitudinal studies quantifying the effect of different training programs on changes in biological tissues by systematically changing isolated variables. These studies need to be designed with a long-term component.
- Determination of internal impact loads during running using either direct measurement techniques or more refined modeling techniques.
- *In vivo* studies simulating impact loading similar to the actual impact loading during activities such as running. These studies need to focus on understanding the interaction of the various loading parameters that determine the acceptable windows of loading for biological tissues. This set of studies will help to establish which running conditions will result in a beneficial adaptive tissue response and which will result in injury to the tissue.

The fact that running in soft or hard shoes feels different may not directly be associated with the onset of injuries. Running in soft shoes may provide a different level of comfort than running in hard shoes. One may speculate that comfort is associated with a *tuning* reaction of the muscles to avoid excessive vibrations of the soft tissues. Comfort may also be associated with changes in magnitude or rate of change of skeletal accelerations. Or, the difference between soft and hard shoes may be associated with *fatigue* resulting from changes in muscle activities, changes that may be influenced by impact loading. The fatigue may manifest itself in the form of decreased comfort, lower performance, and/or increased risk of injury.

Future research choosing this avenue of thinking should concentrate on quantification of muscular activity during (*tuning*) and after (*fatigue*) activities with different impact loading situations (e.g., different shoes or surfaces). Changes in muscle activity (amplitude, timing, and frequency) may provide an indication of the "comfort aspect" of impact forces.

Research on impact forces during running has made substantial progress in the last 2 decades but has also shown signs of the path of sleepwalkers (Koestler, 1968). It seems, however, that the time is right to make substantial progress toward a more comprehensive understanding of the effects of impact

forces during running, which, in turn, could be transferred to impact loading during other physical activities.

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