ELSEVIER

Journal of Biomechanics 32 (1999) 849–856

JOURNAL OF BIOMECHANICS

www.elsevier.com/locate/jbiomech

The effect of muscle stiffness and damping on simulated impact force peaks during running

Benno M. Nigg*, Wen Liu

The Human Performance Laboratory, Faculty of Kinesiology, The University of Calgary, 2500 University Drive, NW Calgary, Alberta Canada T2NIN4

Received 15 February 1999

Abstract

It has been frequently reported that vertical impact force peaks during running change only minimally when changing the midsole hardness of running shoes. However, the underlying mechanism for these experimental observations is not well understood. An athlete has various possibilities to influence external and internal forces during ground contact (e.g. landing velocity, geometrical alignment, muscle tuning, etc.). The purpose of this study was to discuss one possible strategy to influence external impact forces acting on the athlete's body during running, the strategy to change muscle activity (muscle tuning). The human body was modeled as a simplified mass–spring–damper system. The model included masses of the upper and the lower bodies with each part of the body represented by a rigid and a non-rigid wobbling mass. The influence of mechanical properties of the human body on the vertical impact force peak was examined by varying the spring constants and damping coefficients of the spring–damper units that connected the various masses. Two types of shoe soles were modeled using a non-linear force deformation model with two sets of parameters based on the force–deformation curves of pendulum impact experiments. The simulated results showed that the regulation of the mechanical coupling of rigid and wobbling masses of the human body had an influence on the magnitude of the vertical impact force, but not on its loading rate. It was possible to produce the same impact force peaks altering specific mechanical properties of the system for a soft and a hard shoe sole. This regulation can be achieved through changes of joint angles, changes in joint angular velocities and/or changes in muscle activation levels in the lower extremity. Therefore, it has been concluded that changes in muscle activity (muscle tuning) can be used as a possible strategy to affect vertical impact force peaks during running.

Keywords: Impact; Force peak; Running; Wobbling mass; Muscle tuning

1. Introduction

Impact forces are forces due to the collision of two objects. In running, impact forces occur due to the collision of the foot with the ground. In running, the vertical ground reaction force shows usually two peaks, a first peak, the impact force peak and a second peak, the active force peak. The distinction between impact (they were initially called "passive") and active force peaks was made because the time duration of an impact force is shorter than the time required for a muscle to react to the sensory signal generated during the impact (Nigg, 1978, p. 12). It was proposed that impact forces during running and other sport activities are one major factor responsible for the development of running injuries. Furthermore, it was assumed that the improvement of running shoes and surfaces would reduce the frequency of impact related injuries (Miller, 1978; Nigg, 1978; McMahon and Green, 1979; Cavanagh and Lafortune, 1980). Consequently, one of the most important goals of running shoe construction in the last 20 yr was to reduce impact force peaks.

However, a series of experimental studies showed unexpected and contradictory results. For instance, results from studies where the heel of fixed foot-shoe combinations (Aerts and De Clercq, 1993) or fixed human legs (Lafortune et al., 1996) were exposed to impact forces showed a substantial decrease of impact force peaks with decreasing shoe-sole hardness. However, impact forces acting on actual runners when running in hard or soft soled running shoes showed no (Clarke et al., 1983; Nigg, 1983; Snel et al., 1985) or only small differences (DeVita

* Corresponding author. Tel.: 403-220-3436; fax: 403-284-3553.
E-mail address: uheinz@kin.ucalgary.ca (B.M. Nigg)

0021-9290/99/$ - see front matter
PII: S0021-9290(99)00048-2

and Bates, 1987). One proposed explanation for the small differences in the running studies was that the change in midsole hardness between soft and hard soled running shoes represents only a small difference in the total compliance of the foot–shoe-ground-system and would, therefore, produce only small changes in the measured impact force peaks (DeVita and Bates, 1987; Bates, 1989). However, the differences between the results from the isolated impact tests (Aerts and De Clercq, 1993; Lafortune et al., 1996) and the running experiments (Clarke et al., 1983; Nigg, 1983; Snel et al., 1985; DeVita and Bates, 1987) cannot be explained using this line of arguments.

Impact forces acting on a human foot can be influenced by many different factors. They include inertia properties of the tested system, masses of the various body segments involved in the deceleration process, joint angles between body segments, the coupling between soft and rigid masses and joint stiffness. Several of the listed factors may have been different in the various tests discussed above. Muscle activity, for instance, which is partially responsible for joint stiffness and the coupling between soft (wobbling) and rigid masses, may have been different for the impact test and the running experiment. Changes in muscle activity in the lower extremities in the pre-landing phase alter the mechanical characteristics of the lower extremities. Such changes may be used to control the skeletal movement during ground contact and/or the impact forces during landing. The movement control aspect has been discussed in the literature (Bobbert et al., 1992). However, the effect of changes in muscle activity (muscle tuning) on impact forces during running has not been discussed comprehensively.

Thus, the purpose of this study was to examine the influence of changes in mechanical properties of spring and damper elements of a simplified lumped mass-spring-damper model of the human body on the impact force peak during running using a simulation approach.

The hypotheses tested in this study were:

*H*1: A hard shoe sole increases the impact force peak during running if nothing else changes in the system.

*H*2: Changes in the mechanical coupling properties between soft and rigid masses can be used to compensate for differences in shoe-sole hardness.

2. Methods

The approximation of the human body with rigid segments is only justified for movements with no impacts (Denoth et al., 1984). Simulation models of impact situations must include non-rigid masses, which are mechanically coupled to the rigid masses. Such non-rigid masses have been called "wobbling" masses (Gruber, 1987).

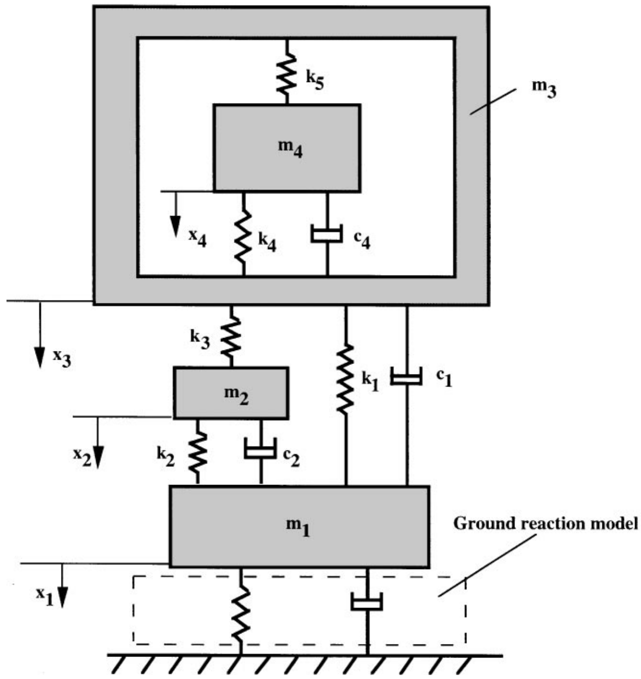

Fig. 1. Illustration of a simplified spring–damper–mass model of the human body used in this study to simulate the dynamics of impact during heel–toe running.

A simplified lumped mass–spring–damper model of the human body (Fig. 1) was used in this study. It consisted of four masses. The upper body and the swing leg were represented by a combination of a rigid mass (M_3) and a non-rigid (wobbling) mass (M_4). The shank, leg and foot of the supporting leg were represented by a combination of a rigid mass (M_1) and a non-rigid (wobbling) mass (M_2). Similar, slightly less complex structures without wobbling masses have been used in previous modeling studies of human movement (Alexander et al., 1986; Ker et al., 1989; Nigg and Anton, 1995). Structures with wobbling masses have been used in a few simulations of human locomotion (Gruber, 1987; Cole, 1995).

The values for the four masses were adapted from Cole (1995), who determined body segment masses and their distribution to skeletal (bone) and soft tissues using data and formulas from Clauser et al. (1969) and Clarys et al. (1986). A total body mass of 75 kg was used. The upper rigid mass was 6 kg, the upper wobbling mass 50.3 kg, the lower rigid mass 6.1 kg, the lower wobbling mass 12.6 kg.

Spring–damper units coupled mechanically the upper and lower rigid masses, and the rigid and wobbling masses (Fig. 1). The wobbling mass of the upper body was attached to the upper rigid mass through a spring ($k_5 = 18$ kN/m) at one end, representing the serial elastic element in a muscle–tendon unit, and a spring–damper combination (k_4, c_4) at the other end, representing the elastic and damping properties of the contractile elements of the muscle–tendon units. The values of these two spring constants, k_4 and k_5 were estimated from a linear approximation of the passive muscle–tendon stiffness estimated from the maximal isometric muscle force, muscle slack length and ultimate deformation ratio of the muscle (Cole, 1995). The damping coefficient, c_4, was estimated using a critical damping assumption (Hörler, 1972) for the upper wobbling mass. Similarly, the wobbling mass of the lower body was connected to the upper and the lower rigid bodies by means of springs and dampers (k_3, k_2, c_2) representing muscle–tendon units, which affected the coupling between the upper and the lower rigid bodies. A compressive spring (k_1) and a damper (c_1) connected the upper and the lower rigid bodies. The combined stiffness of k_1, k_2 and k_3 was equivalent to a "leg stiffness" as defined by McMahon and Cheng (1990).

A non-linear visco-elastic force function (Cole, 1995) was used to simulate the ground reaction model:

$$F_g = A_c[a(x_1)^b + c(x_1)^d(v_1)^e], \quad (x1 > 0),$$

$$F_g = 0 \quad (x_1 \leqslant 0),$$

where the contact force, F_g, was a function of the deformation, x_1, and the deformation velocity, v_1, of the contact element. The constant A_c was 2, assuming that the average contact area during the simulated stance phases was approximately two times of the area reported in the previously mentioned pendulum test (Aerts and De Clercq, 1992). The specific parameters (a, b, c, d, and e) for the hard and the soft shoe-sole models were determined using a least square fit procedure to fit the simulated force–deformation curves of pendulum impacts to the experimental data by Aerts and De Clercq (1992) (Fig. 2a).

The outcome variables discussed in this paper were the variables frequently discussed in research papers dealing with impact loading (Clarke et al., 1985; Nigg, 1986; Lafortune and Henning, 1988). The vertical impact force peak, F_{zi}, was included to address the maximal loading situations. The maximal vertical loading rate, G_{zi}, was included to address the possible eccentric loading of muscle–tendon units. The maximal vertical deceleration of the lower rigid mass, a_z, was included to address possible shock waves travelling through the locomotor system. The maximal vertical displacement of upper body, d_z was included to address possible impairment of vision during running.

Numerical simulation was executed on an Indigo2 R10,000 station (Silicon Graphics Inc.) using a multibody dynamic simulation software (DADS, version 8.5, CADSI, Oakdale, Iowa). The ground reaction model was implemented to the DADS simulation program using the 'USERFORCE' option. The movement was simulated for the first 250 ms after touchdown. However, since no force producing elements were implemented in the model, only results for the first 100 ms were discussed.

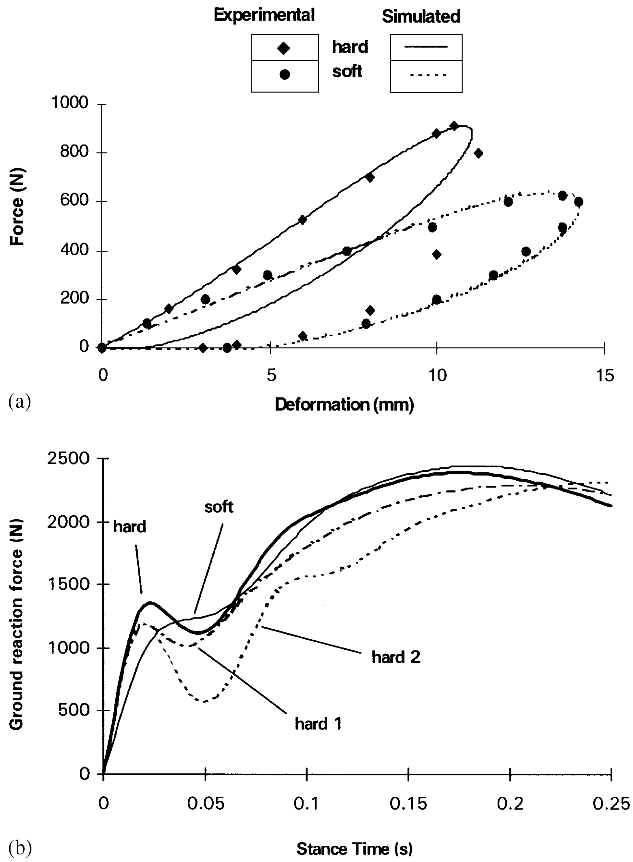

Fig. 2. (a) Force–deformation curves from simulated pendulum impacts with two different ground reaction models that represented the hard (solid line) and soft (dotted line) shoes. The points illustrate the selected digitized points from the force–deformation curves for a hard (◆) and soft (●) shoe sole during an impact of a pendulum (Aerts and De Clercq, 1993); (b) Illustration of the simulated ground reaction forces for different conditions: the hard (hard) and soft shoe (soft) running conditions for a typical system; the hard shoe condition with reduced k_1 (1 kN/m) and c_1 (100 kg/s) (hard1), and the hard shoe condition with reduced k_2 (2 kN/m) and c_2 (250 kg/s) (hard2). The impact force peaks in both hard1 (1194 N) and hard2 (1194 N) are close to the value of the impact force peak in soft shoe running (1187 N).

A vertical touch-down velocity of 0.6 m/s was used for all four masses. First, simulations were performed for the hard and the soft shoe-sole conditions using a system with $k_1 = 6$ kN/m, $k_2 = 6$ kN/m, $k_3 = 10$ kN/m and $k_4 = 10$ kN/m, and $c_1 = 300$ kg/s, $c_2 = 650$ kg/s and $c_4 = 1900$ kg/s. Second, simulations were performed for the hard shoe-sole condition varying the spring constants (k_1, k_2 and k_4) and damping coefficients (c_1, c_2 and c_4) to examine (a) possibilities of producing impact force peaks similar to the soft shoes running and (b) the effect of changes in spring constants and damping coefficients.

Results from previous running experiments showed that the leg spring stiffness varied from 6 to 16 kN/m when subjects changed their stride frequency from − 30 to + 40% (Farley and Gonzalez, 1996). The variations used in the current study for the stiffness of k_1, k_2 and k_4 were all within this range. The variations of damping coefficients were determined by trial and error until predicted impact force peaks in the hard shoe running situation were reduced to a corresponding level in the soft shoe running situation. Simulations were done with a pair of mechanical characteristics (one spring constant and one damping coefficient) varying while the other mechanical characteristics remaining constant.

3. Results

For the typical system, the simulated results showed (Fig. 2b) that the vertical impact force peak, F_{zi}, was higher in the hard (1339 N) than in the soft shoe-sole condition (1187 N). Similarly, the maximal loading rate was much higher in the hard (90.7 N/ms) than in the soft shoe condition (54.4 N/ms). The simulated vertical impact force peak, F_{zi}, in the hard shoe condition could be reduced to 1194 N by decreasing k_1 and c_1 (illustrated as hard1 condition in Fig. 2b), or by decreasing k_2 and c_2 (illustrated as hard2 condition in Fig. 2b). However, the maximal loading rate, G_{zi}, was almost identical (90.7 N/ms) in the hard shoe condition with varied system parameters.

The deceleration of the lower rigid and wobbling masses during the impact phase was higher in the hard than in the soft shoe-sole condition (Fig. 3). Decreasing in k_1 and c_1 (hard1) or k_2 and c_2 (hard2) generated increases in the deceleration of the lower rigid mass, and decreases in the deceleration of the lower wobbling mass.

Within the range of variations of the coupling characteristics k_1, k_2, c_1 and c_2, the vertical impact force peak, F_{zi}, ranged from 1.6 to 2.3 times of body weight (Fig. 4). The vertical impact peak force, F_{zi}, increased with increasing stiffness (k_1 or k_2) or increasing damping coefficients (c_1 or c_2). Changes in damping coefficients had a stronger influence on changes in the vertical impact force peak than changes in stiffness.

Simulated maximal deceleration (a_z) of the lower rigid mass ranged from 2.8 g to 4.1 g for the simulated range of variations of the coupling characteristics k_1, k_2, c_1 and c_2 (Fig. 5). An increase in each of these parameters produced a decrease in the simulated maximal deceleration, a_z. The simulated maximal vertical displacement (d_z) of the upper rigid body ranged from 7.2 to 19.7 mm (Fig. 6). An increase in stiffness (k_1 or k_2) or damping coefficients (c_1 or c_2) produced a decrease of the simulated maximal vertical displacement, d_z.

A variation of k_4 and/or c_3 had almost no effect on F_{zi} or G_{zi}, and only a small effect on a_z and d_z.

4. Discussion

An abstract lumped mass model rather than a multibody segment model has been used in this study. The rationales behind this approach were the following. First, the difficulty in estimating the co-contraction of antagonistic muscles during functional activity has not been resolved due to indeterminate nature of the problem. Second, experimental approaches to study the effect of muscle activity on soft tissue coupling are difficult. Third, the complexity of a multi-body segment model (e.g. Gerritsen et al., 1995; Cole, 1995) makes it difficult to interpret simulated results for the question at hand. Finally, simple models with masses and spring-damper units (Alexander et al., 1986; Ker et al., 1989; Nigg and Anton, 1995) have been successfully used in simulations of human running and hopping to derive general principles.

The current model included soft tissue wobbling masses so that the influence of wobbling masses on the vertical impact forces could be examined. Previous simpler models with 2 masses and 2 spring-damper units but no wobbling masses (Alexander et al., 1986; Ker et al., 1989) could not be used for this purpose. Models of human body including wobbling masses have been developed in the past (Gruber, 1987; Cole, 1995). Comparing results between a model with only rigid masses and a model with rigid and wobbling masses for a drop jump movement, Gruber (1987) found a substantial difference in the predicted impact forces in the knee and hip joint. Cole (1995)

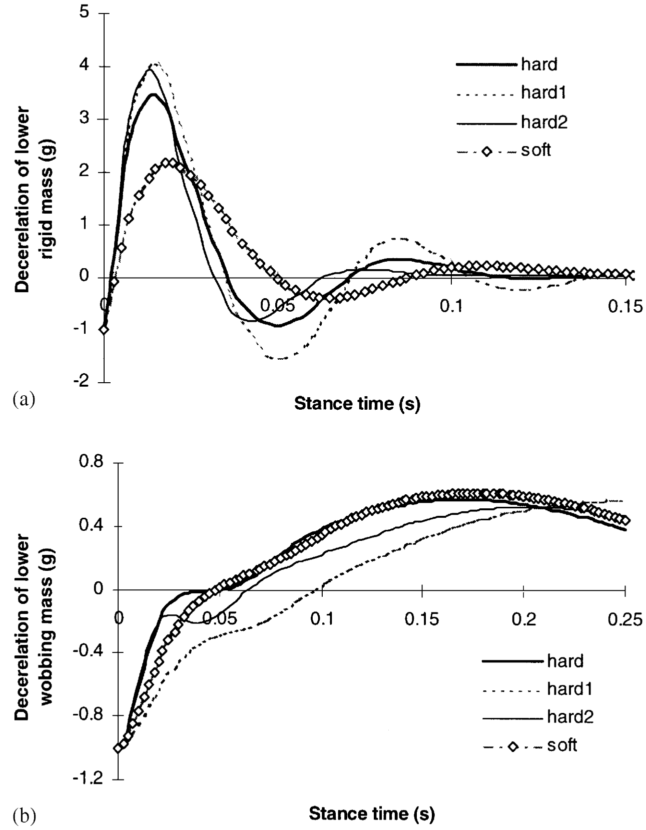

Fig. 3. Simulated deceleration of the lower rigid mass (a), and the lower wobbling mass (b) for four different situations (hard, soft, hard1 and hard2). The unit of the deceleration is g (gravity acceleration).

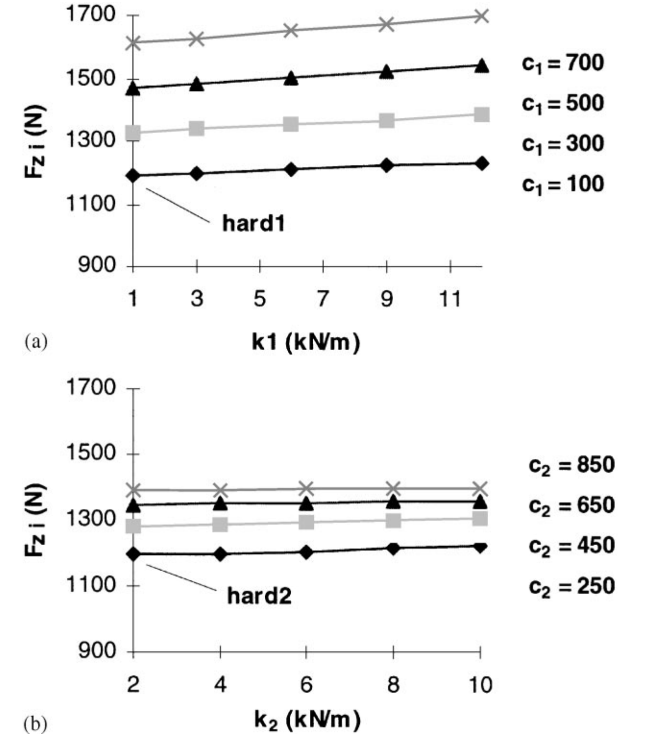

Fig. 4. Illustration of the influence of variations in the characteristics k_1 and c_1 (a) and k_2 and c_2 (b) on the impact force peak, F_{zi}. The impact force peak, F_{zi}, increased with an increase in either the stiffness (k_1 or k_2) or damping coefficients (c_1 or c_2). The damping coefficient had a stronger influence on the impact force peak than the stiffness.

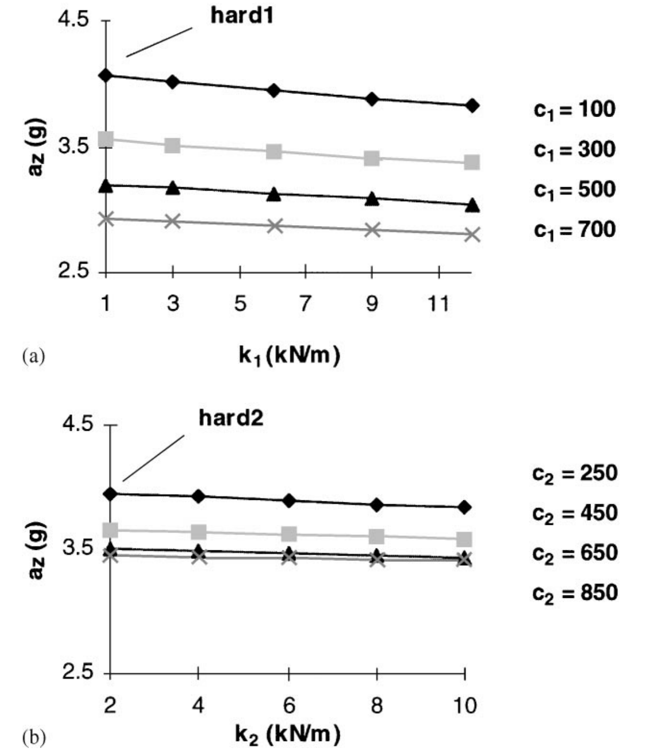

Fig. 5. Illustration of the influence of variations in the characteristics k_1 and c_1 (a) and k_2 and c_2 (b) on the maximal vertical acceleration, a_z, of the lower rigid body. The vertical acceleration, a_z, decreased with an increase in each of these parameters.

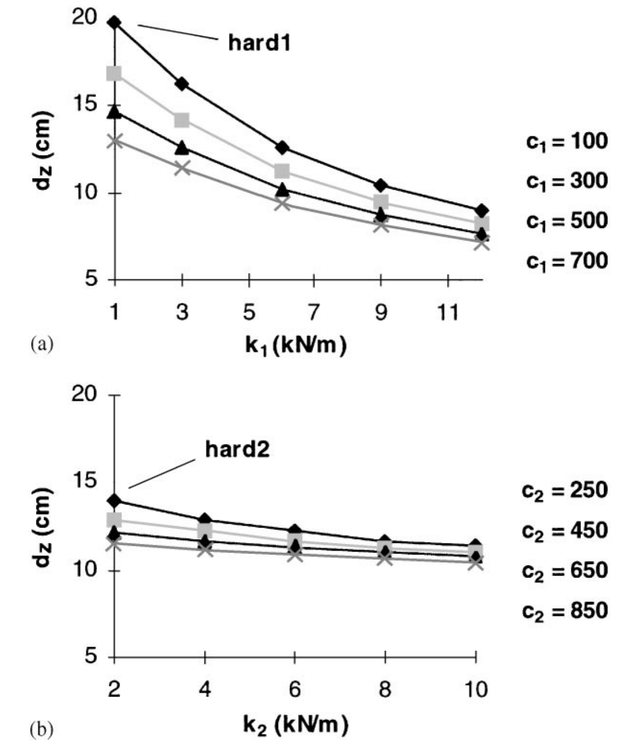

Fig. 6. Illustration of the influence of variations in the characteristics k_1 and c_1 (a) and k_2 and c_2 (b) on the maximal vertical displacement, d_z, of the upper rigid body. The maximal vertical displacement, d_z, decreased with increasing in either stiffness (k_1 or k_2) or damping coefficients (c_1 or c_2).

compared three types of multi-body segment models: one model with only rigid segment masses, a second model with rigid segment masses with muscles, and a third model with rigid segment masses, muscles and soft tissue masses. He found that the third model with rigid segment masses, muscles and soft tissue masses predicted the ground reaction force peak better than the other two models. However, he did not examine the influence of changes in the mechanical coupling between the wobbling and the rigid masses on the vertical impact force peak in relation to different shoes.

The simulated results of this study predicted a vertical impact force peak and a maximal loading rate during the impact phase of running corresponding to the actual range of the experimentally measured results with human subjects (Nigg, 1986). The maximal loading rate during the impact increased with increasing hardness of the shoe-sole, but did not change when changing spring and/or damping characteristics of the system. This result indicated that the mechanical property of shoe-sole and/or the ground was a dominant factor for the magnitude of the loading rate. This result is in agreement with results from experimental studies (Clarke et al., 1983; Frederick et al., 1984; de Wit et al., 1995) and model simulations (Wright et al., 1998). In these studies, the loading rate of the vertical impact force has been shown to increase with increasing shoe-sole hardness. The similarities in the results for the vertical impact peak forces and their loading rates for the simulated and the experimental results provide support for the chosen approach of the lumped mass–spring–damper model and its application to running.

The simulated vertical displacement in this study corresponded to the displacement of the body center of mass along the axis of the leg. The simulated results of vertical displacement (14.2 cm) were close to the measured leg displacement (about 12 cm) during running at the preferred stride frequency by Farley and Gonzalez (1996). Assuming a person with a starting leg length (l_0) of 1 m and an initial leg angle (θ_0) of 24°, a leg displacement of 14.2 cm corresponds to 5.6 cm vertical displacement of the leg. This result is within the range of the corresponding data for human running (McMahon and Cheng, 1990).

The limitations of the current model include a one-dimensional movement, lumped body masses and constant stiffness and damping characteristics during the simulated time period. Therefore, the simulated results must show unrealistic results during the second part of the ground contact because this model did not have any changes in muscle activity to produce the take-off movement. Specifically, the model predicted too high active force peaks, too low decelerations of the lower rigid body, too long stance phases and a time delay between the active force peak and the maximal vertical displacement if compared to the results of running experiments.

The simulations of this study were, among other aspects, done to study the feasibility of the "muscle tuning" concept. It was postulated that subjects pre-activate (tune) their muscles differently when midsole hardness changed (Nigg, 1997) and that these changes may have been one of the possible reasons for the fact that the impact force peaks in the running experiments did not change with changing midsole hardness. There are some evidences to support this concept. The force–length relationship of human muscles results in spring-like behavior with the stiffness depending on muscle activation level (Winters, 1990; Soest and Bobbert, 1993). The force–velocity relationship of human muscles can be modeled as a non-linear dash-pot modulated by the muscle activation level (Hogan et al., 1987; McMahon, 1990). Thus, the change in muscle activity may alter the stiffness and damping of the human body, therefore, leading to a corresponding change in the impact force peaks during running. The results of this study showed that changes in midsole hardness could be offset by changes in the coupling between the wobbling and the rigid masses of the lower extremities (k_2 or c_2). Thus, the results of this simulation support the concept that "muscle tuning" is (at least theoretically) a possible strategy to influence impact forces (hypothesis H2).

A pilot experimental study (Nigg, 1997) has shown that muscle activity changed before landing when running on hard, medium and soft running surfaces. The muscle preactivation increased its median frequency from about 60 Hz to about 90 Hz with decreasing surface hardness from concrete to the soft surface. The integral of the power spectral density was minimal for the medium surface hardness (the most comfortable one) and about 300% higher for the very hard and the very soft surfaces. A muscle-tuning strategy was suggested to be related to the changes of muscle activity.

A previous modeling study (Gerritsen et al., 1995) showed a small change of about 80 N in the impact force peak during running when altering the muscle activation levels for the simulated muscles with the total muscle force remaining the same. In their study, they assumed a low level of co-contraction of simulated muscles during the impact phase. The runner was modeled using only rigid segments and the effect of wobbling masses was not accounted for. Thus, their study addressed primarily the influence of body configuration on impact forces. However, their model cannot be used to study the influence of a changed muscle activity on the body's response (involving wobbling masses) to different shoe-sole hardness during running.

An interesting finding of the current study was that changes in damping had a greater influence on the impact force peak than the changes in stiffness of the system. In running experiments conducted by Farley and Gonzalez (1996), the leg stiffness was shown to change with decreasing in stride frequency and that leg stiffness was not strongly correlated with the impact force peak. The simulated results of the current study showed that the influence of stiffness of the system on the impact force peak could be overridden by the changes in the damping. The predicted deceleration of the lower rigid and wobbling masses may provide an explanation for the influence of changed stiffness and damping of the system on the impact force peak. Comparing systems with reduced stiffness and damping (hard1 and/or hard2) to the reference system (hard) (Fig. 3), a decrease in the impact force peak was accompanied with an increase in the deceleration of the lower rigid mass and a decrease in the deceleration of the lower wobbling mass. A decrease in the deceleration of the lower rigid body during the impact indicated less interaction of the lower rigid mass with the upper rigid body and the lower wobbling mass. A reduced interaction force resulted in a decreased impact force peak. The changes in the stiffness and damping of the system, therefore, caused changes in the interaction force during the impact leading to reductions of the impact force peaks. However, the major component of the impact force peak resulted from the deceleration of the lower rigid mass (Fig. 3). The impact force peak of the lower rigid mass in the hard shoe condition was much higher than the soft shoe condition (hypothesis H1).

There are various strategies possible to influence impact forces during heel-toe running including changing the geometrical position of foot and leg (e.g. knee angle), changing joint stiffness and/or changing the tuning of the wobbling masses. Experimental evidence showed that different runners responded differently to changes in midsole hardness (Clarke et al., 1983; Nigg, 1983; Snel et al., 1985; Nigg et al., 1987). The simulated results of this study showed that a decrease in the impact force peak by reducing (k_1, c_1) or (k_2, c_2) was accompanied by an increase in the maximal acceleration of the lower rigid body and the maximal vertical displacement of the upper rigid body. Thus, it may be possible that a runner runs with a relatively small impact force peak, but that the same runner may feel an increased shock due to an increased acceleration transmitted to their body. Assuming different individual thresholds of body sensation, one would expect different individual strategies in response to changes in midsole hardness.

The results of this study showed that the regulation of the soft tissue coupling to the rigid structures had an important influence on the impact force peak. The simulation allowed producing the same impact force peaks for shoes with different midsole hardness by altering the mechanical properties of the system. This theoretical result corresponded to the experimental results discussed earlier. In the real running situations this result can be achieved through a change in the joint angles (geometry), a change in joint stiffness, and/or a change of the coupling of the wobbling masses to the rigid masses of the lower extremities. The results of this study do not provide any support for one or a combination of these strategies. However, the results of this study provide support for the possibility that "muscle tuning" is one possible strategy that can be applied.

This study had the goal to discuss the feasibility of one possible strategy to influence external impact forces acting on the athlete's body during running, the strategy to change muscle activity (muscle tuning). The improved understanding of possible body responses to changes in boundary conditions (e.g. shoe or surface) during running is, in the view of the authors, important since it provides the basis for understanding the dynamics and the control of human running. If our speculation of "muscle tuning" is correct, a subject specific response to impact forces (magnitude and frequency content) should be expected. Impact force "attenuation" may then become among other possible factors a question of tuning a system mechanically to make it critically damped. This would also have an effect on energy and performance.

If one understands how different individuals respond to changes in shoe characteristics (e.g. soft–hard or elastic–viscous) one may be able to design subject-group specific shoes. Possible consequences of such an approach include an enhancement of running performance, a reduction of the risk of injury and an improved comfort level.

References

Aerts, P., De Clercq, D., 1993. Deformation characteristics of the heel region of the shod foot during a simulated heel strike: the effect of varying midsole hardness. Journal of Sports Science 11, 449–461.

Alexander, R.M., Bennett, M.B., Ker, R.F., 1986. Mechanical properties and function of the paw pads of some mammals. Journal of Zoology 209, 405–419.

Bates, B.T., 1989. Comment on the influence of running velocity and midsole hardness on external impact forces in heel-toe running. Journal of Biomechanics 22, 963–965.

Bobbert, M.F., Yeadon, M.R., Nigg, B.M., 1992. Mechanical analysis of the landing phase in heel-toe running. Journal of Biomechanics 25, 223–234.

Cavanagh, P.R., Lafortune, M.A., 1980. Ground reaction forces in distance running. Journal of Biomechanics 13, 397–406.

Clarke, T.E., Frederick, E.C., Cooper, L.B., 1983. The effects of shoe cushioning upon ground reaction forces in running. International Journal of Sports Medicine 4, 247–251.

Clarke, T.E., Cooper, L.B., Clarke, D.E., Hamill, C.L., 1985. The effect of increased running speed upon peak shank deceleration during ground contact. In: Winter, D., Normal, R., Wells, R., Hayes, K., Patla, A. (Eds.), Biomechanics IX-B. Human Kinetics, Champaign, IL, pp. 101–105.

Clarys, J.P., Marfell-Jones, M.J., 1986. Anthropometric prediction of component tissue masses in the minor limb segments of the human body. Human Biology 58, 761–769.

Clauser, C.E., McConville, J.T., Young, J.W., 1969. Weight, volume, and center of mass of segments of the human body. Wright Patterson Air Force Base. Ohio, AMRL-TR-69–70.

Cole, G.K., 1995. Loading of the joints of the lower extremities during the impact phase in running. Ph.D. dissertation, . Department of Mechanical Engineering, The University of Calgary, Calgary, Canada.

Denoth, J., Gruber, K., Ruder, H., Keppler, M., 1984. Forces and torques during sports activities with high accelerations. In: Perren, S.M., Schneider, E. (Eds.), Biomechanics Current Interdisciplinary Research. Martinuis Nijhoff, Dordrecht, pp. 102–106.

DeVita, P., Bates, B.T., 1987. The effects of time on selected ground reaction force parameters. In: Jonsson, B. (Ed.), Biomechanics X-B. Human Kinetics Publisher, Champaign, IL, pp. 909–912.

de Wit, B., De Clercq, D., Lonoir, M., 1995. The effect of varying midsole hardness on impact forces and foot motion during foot contact in running. Journal of Applied Biomechanics 11, 395–406.

Farley, C.T., Gonzalez, O., 1996. Leg stiffness and stride frequency in human running. Journal of Biomechanics 29, 181–186.

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

Gerritsen, K.G.M., van den Bogert, A.J., Nigg, B.M., 1995. Direct dynamics simulation of the impact phase in heel-toe running. Journal of Biomechanics 28, 661–668.

Gruber, K., 1987. Entwicklung eines Modells zur Berechnung der Kräfte im Knie- und Hüftgelenk bei sportlichen Bewegungsabläufen mit hohen Beschleunigungen. Ph.D. thesis. Universität Tübingen, Germany.

Hogan, N., Bizzi, E., Mussa-Ivaldi, F.A., Flash, T., 1987. Controlling multijoint motor behavior. Exercise and Sport Sciences Reviews 15, 153–190.

Hörler, E., 1972. Mechanisches Modell zur Beschreibung des isometrischen und des dynamischen Muskelkraftverlaufs. Jugend und Sport, Nov, 271–273.

Ker, R.F., Bennett, M.B., Alexander, R.M., Kester, R.C., 1989. Foot strike and the properties of the human heel pad . Proceedings of the Institute of Mechanical Engineering [H] 203, 191–196.

Lafortune, M.A., Henning, E.M., 1988. Effects of velocity and uphill slope on tibial shock during running. In: Cotton, C.E., Lamontagne, M., Robertson, D.G.E., Stothart, J.P. (Eds.), Proceedings of the Fifth Biennial Conference and Human locomotion Symposium of the Canadian Society of Biomechanics. University of Ottawa, Dept. of Kinanthropology, Ottawa, pp. 94–95.

Lafortune, M.A., Henning, E.M., Lake, M.J., 1996. Dominant role of interface over knee angle for cushioning impact loading and regulating initial leg stiffness. Journal of Biomechanics 29, 1523–1529.

McMahon, T.A., Greene, P.R., 1979. The influence of track compliance on running. Journal of Biomechanics 12, 893–904.

McMahon, T.A., 1990. Muscles: reflexion and locomotion. Princeton University Press, Princeton, NJ, pp. 148–161.

McMahon, T.A., Cheng, G.C., 1990. The mechanics of running: how does stiffness couple with speed? Journal of Biomechanics 23 (suppl.), 65–78.

Miller, D.I., 1978. Biomechanics of running — what should the future hold? Canadian Journal of Applied Sport Science 3, 225.

Nigg, B.M., 1978. Die Belastung des menschlichen Bewegungsapparates aus der Sicht des Biomechanikers (Load on the human locomotor system from a biomechanical point of view). In: Nigg, B.M., Juris, (Eds.), Biomechanische Aspekte zu Sportplatzbelägen. Verlag, Zurich.

Nigg, B.M., 1983. External force measurements with sport shoes and playing surfaces. In: Nigg, B.M., Kerr, B.A. (Eds.), Biomechanical aspects of sport shoes and playing surfaces . University Printing, Calgary, pp. 169–176.

Nigg, B.M., 1986. Biomechanical aspects of running. In: Nigg, B.M. (Ed.), Biomechanics of Running Shoes. Human Kinetics Publishers, Champaign, IL, pp. 15–19.

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

Nigg, B.M., 1997. Impact forces in running. Current Opinion in Orthopedics 8, 43–47.

Snel, J.G., Delleman, N.J., Heerkens, Y.F., van Ingen Schenau, G.J., 1985. Shock-absorbing characteristics of running shoes during actual running. In: Winter, D.A., Norman, R.W., Wells, R.P., Hayes, C., Patla, A.E. (Eds.), Biomechanics IX-B. Human Kinetics Publishers, Champaign, IL, pp. 133–138.

Soest, A.J., Bobbert, M.F., 1993. The contribution of muscle properties in the control of explosive movements. Biol. Cybern. 69, 195–204.

Winters, J.M., 1990. Hill-based muscle models: a systems engineering perspective. In: Winters, J.M., Woo, SL-Y. (Eds.), Multiple muscle systems. Biomechanics and movement organization. Springer, Berlin Heidelberg, New York, pp. 69–93.

Wright, I.C., Neptune, R.R., van den Bogert, A.J., Nigg, B.M., 1998. Passive regulation of impact forces in heel-toe running. Clinical Biomechanics 13, 521–531.