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Passive regulation of impact forces in heel–toe running

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Abstract

Objective. the purpose of this study was to determine whether passive mechanisms can account for impact force regulation with changing shoe hardness.

Design. A three-dimensional musculoskeletal model of the lower extremity was developed to simulate impact in running with two different shoe hardnesses.

Background. Considerable research has focused on developing shoe cushioning to reduce impact forces. However, only minimal changes in peak external impact force have been observed with changes in shoe hardness. It is hypothesized that passive mechanisms can regulate impact forces with changing shoe hardness, without changing muscle activities.

Methods. Initial kinematic inputs for the simulations were measured from nine male subjects performing heel–toe running. Simulations were performed with initial conditions and muscle stimulation patterns held constant while shoe hardness was varied between a hard and a soft condition.

Results. There was no significant difference between the soft and hard shoe peak impact forces. Peak rates of loading were greater for the hard shoe than the soft shoe. Muscle forces changed with shoe conditions. For some muscles (including the tibialis anterior) the forces were greater for the hard shoe, whereas for other muscles (including the peroneus) forces were greater for the soft shoe condition.

Conclusions. Peak impact forces with changing shoe conditions can be regulated by passive mechanical changes without changing muscle activities or kinematics before touchdown.

Relevance

Potential injury causing loads on internal structures (e.g. muscles, tendons, etc.) during the impact phase of running can depend upon shoe hardness, but are not reflected in changes in external ground reaction force.

Keywords: Running; Impact; Footwear; Simulation; Musculoskeletal modelling

1. Introduction

Impact forces during human locomotion are the forces that result from the foot colliding with the ground, reaching a maximum (impact peak) within 50 ms after the first contact [1]. Impact forces in running are characterized by a peak in the vertical ground reaction force, which is small for toe landing and 1 to 3 times body weight for heel landing. Impact forces during heel–toe running, the form of running which is most common [2], have been associated with the development of injuries [3–6].

Cushioning is defined as the reduction of the amplitude of the vertical ground reaction force during impact [1]. Impact tests of shoes performed by dropping weights on shoes [7] or by dropping pendulums against foot–shoe systems [8] have shown that with softer shoes, there is a reduction in the peak vertical ground reaction force and its time derivative (rate of loading). However, vertical ground reaction forces during actual running did not show the same result as these simple mechanical tests. Surprisingly, peak vertical impact forces have been found to have no [7,9] or a negative [10] correlation with shoe hardness. As expected, vertical loading rates have been shown to increase [10,11] with increasing shoe hardness. These

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results indicate that there must be some mechanism by which the body regulates the magnitude of external impact force during running.

There are at least two possibilities to explain the surprising result that the external forces remained constant for different material properties of the shoe soles: (1) the changes in material properties were too small to have an effect; or (2) there must be an adjustment in the kinematics of the leg. A change in kinematics, however, would have an effect on the internal loading situation and may increase or decrease internal forces and stresses [11–14]. This paper will concentrate on the second possibility.

Several experimental studies have shown changes in kinematics after heel strike with increasing shoe hardness, such as increased rates of knee flexion [9], pronation [10,13,14] and increased amounts of pronation [15]. These changes may be active or passive changes. Active changes may consist of altered muscle stimulations or initial conditions prior to heelstrike and have often been suggested to account for the observed impact force regulation [11,16,17]. However, there is little experimental evidence of changes in either kinematics [11,18] or muscle activity [19] prior to heelstrike.

The lack of evidence of active changes with changing shoe hardness would then lead to the hypothesis that impact forces may be regulated passively. Passive changes are a mechanical reaction of the system to the altered shoe hardness. Passive changes in rates of pronation have been proposed to contribute to impact force regulation [20], but that proposed mechanism could not account for increased loading rates with increased shoe hardness. It has not been shown that passive mechanisms are sufficient to account for experimentally observed impact force regulation with changing shoe conditions, nor has it been shown what effects such passive impact force regulation would have on internal forces and stresses. Therefore, the purposes of this study were

1. To determine whether impact forces could be regulated without changing muscle stimulations or initial segment kinematics; and
2. To determine how such a passive impact force regulation affects the internal loading situation.

2. Methods

A three-dimensional musculoskeletal model was developed to simulate the impact phase in heel–toe running. The model anthropometrics, muscle properties and ground contact model were integrated into a forward dynamic simulation model using DADS (version 8.00, CADSI, Coralville IA, USA). Initial kinematic conditions for the simulations were based on averaged trials from nine subjects. Muscle stimulation patterns were determined by minimizing differences between the measured and simulated movements for one subject. Simulations were then performed with two different shoe hardnesses, while the initial conditions and muscle stimulation patterns were held constant to see if peak external vertical ground reaction forces changed.

2.1. Anthropometric model

The three-dimensional model consisted of a single leg with six rigid bodies representing the toes, foot, talus, shank, patella and thigh (Fig. 1). Segment lengths were taken from an existing model [21] representing a male of 180 cm height. The segment masses and inertial properties were determined using regression equations [22,23] for a male with a mass of 75 kg. The torso, head, arms and other leg were represented by a visceral mass and a rest-of-body segment [24]. The rest-of-body segment was fixed to the pelvis while the visceral mass was attached to the rest-of-body segment by a linear elastic spring with a stiffness of 100 kN/m (ref. [24]).

The joint models were based on existing work [21], but developed further to better simulate movements in three dimensions. The subtalar joint and ankle joint were represented by revolute joints aligned with joint orientations presented in the literature [25]. Passive non-linear stiffnesses were applied as moments about the subtalar joint and talo-crural joint axes to represent the effects of passive soft tissue and bony constraints on movements about these axes [26]. For the knee, the centre of rotation of the shank relative to the thigh moved in the sagittal plane as a given function of flexion angle [21]. Internal–external rotation and adduction–abduction were permitted at the knee and were limited by rotational springs with stiffnesses based on published data [27]. The flexion–extension and anterior–posterior translation of the patella relative to the femur were functions of superior–inferior movement of the patella relative to the femur [21] with the patellar tendon assumed to be inextensible. The hip joint was modelled as a spherical joint with three degrees of freedom.

2.2. Muscle model

The model was actuated by 12 different muscles, each muscle representing one or several functionally similar muscles. These muscle groups were the gluteus maximus, illiopsoas, rectus femoris, hamstrings, vasti, gastrocnemius, soleus, flexor digitorum, tibialis posterior, tibialis anterior, extensor digitorum and peroneals (Table 1). Each muscle had an origin and

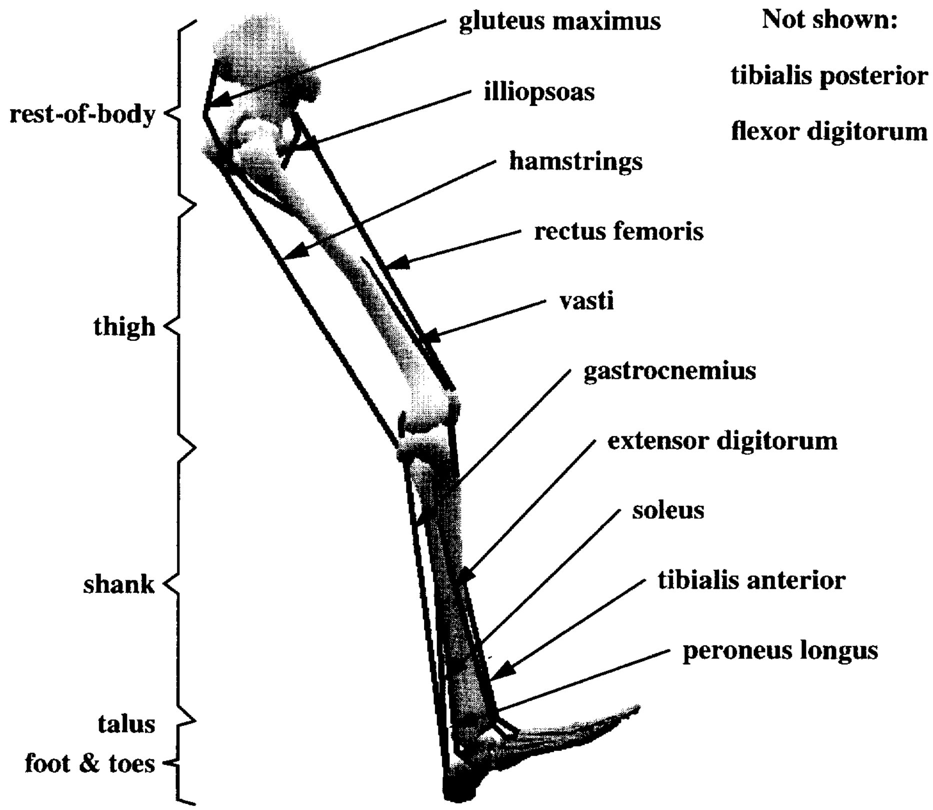

Fig. 1. The body segments and muscles represented in the simulation model.

insertion fixed relative to model segments. Several muscles had additional points fixed to bodies in the model, through which the muscle passed to prescribe anatomically correct muscle paths [21]. For some muscles, additional non-fixed wrapping points were used to represent underlying bones and muscles [28].

The force–length–velocity-activation characteristics of the muscles were modelled using a Hill based muscle model used in previous simulation models [24,29,30]. Muscle specific parameters such as maximum isometric force, contractile element length and series elastic element lengths were based on published muscle model data [21] (Table 1). The peak force of the tibialis anterior and digital extensors were increased to ensure sufficient strength to prevent rapid slap-down of the foot during impact. The model was

Table 1
Parameters describing muscle properties

Muscle model	Maximum force (N)	Principle function(s)	Muscles represented
Gluteus maximus	1752	Hip extension	Gluteus maximus*
Illiopsoas	800	Hip flexion	Iliacus*, psoas
Rectus femoris	780	Hip flexion, knee extension	Rectus femoris*
Hamstrings	1769	Hip extension, knee flexion	Semimembranous*, semitendinosus, biceps femoris (long head)
Vasti	4400	Knee extension	Vastus lateralis, vastus intermedius*, vastus medialus
Gastrocnemius	1605	Knee flexion, plantar flexion	Gastrocnemius* (lateral and medial heads)
Soleus	2830	Plantar flexion	Soleus*
Flexor digitorum	630	Plantar flexion	Flexor digitorum longus*, flexor hallicus longus
Tibialis posterior	1270	Inversion	Tibialis posterior*
Tibialis anterior	2000	Dorsiflexion	Tibialis anterior
Extensor digitorum	2000	Dorsiflexion	Extensor digitorum longus*, extensor hallicus longus
Peroneals	1015	Eversion	Peroneus longus*, peroneus brevis

*Muscle geometry used[28].

controlled by muscle stimulation patterns which were modelled as step functions. The stimulation onset was permitted to be before the simulation started so that the muscle activation could be increasing at the initiation of the simulation.

2.3. Ground contact model

The contact between the foot and the ground was modelled by 66 discrete independent contact elements. Each element represented the mechanical properties of a region of the sole of the shoe and the underlying soft tissue and permitted deformation perpendicular to the floor. When a contact element was deformed by the amount p_z (greater than zero), the vertical force of that element (F_z) was calculated as the greater of either zero or:

$$F_z = f\cdot(a\cdot(p_z)^b + c\cdot(p_z)^d\cdot(v_z)^e) \qquad (1)$$

where v_2 was the rate of deformation, and a, b, c, d and e were shoe-specific parameters. This form of the equation includes a non-linear stiffness term ($a\cdot(p_z)^b$), and a displacement dependent, non-linear damping term ($c\cdot(p_z)^d\cdot(v_z)^e$). The factor f scaled the mechanical properties of each element based on the area and length of the perimeter of the region. Horizontal forces that resisted slipping of the shoe relative to the ground were determined using an approximation to a Coulomb friction model used previously [24,30].

Two sets of shoe specific parameters were obtained from the literature [8] to represent a hard (EVA 65 Asker C) and a soft (EVA 40 Asker C) running shoe (Table 2). The shoe specific parameters were selected so that a simulated pendulum impact test (with the same conditions as the experimental investigation [8]) produced similar force–deformation curves as those presented in the above study (Fig. 2). The factor f (eqn (1)) scaled the mechanical properties of each region relative to the mechanical properties measured using the pendulum impact. It was assumed that each region was square, and that an additional area outside the region equal to a constant w times the

Table 2
Parameters describing shoe properties

	a	b	c	d	e
Soft shoe	6 000 000	2.20	−16 000	0.80	1.50
Hard shoe	6 500 000	2.00	−20 000	0.80	1.50

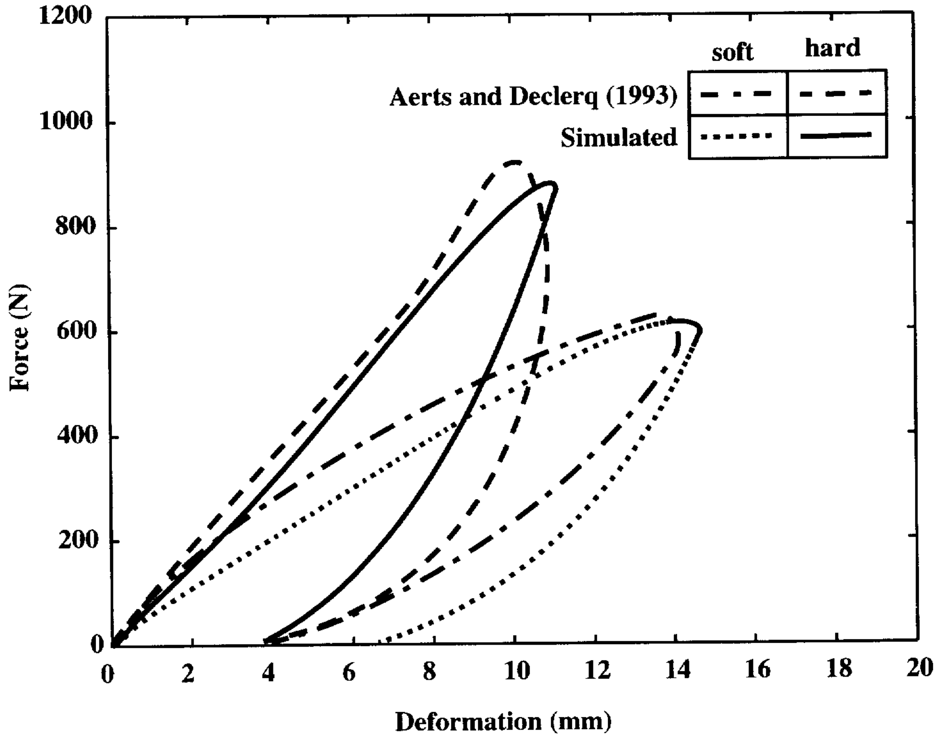

Fig. 2. The simulated and experimental deformation-force response of the hard and soft shoes to the impact of a pendulum of mass 11.615 kg with a velocity at first contact of 0.96 m/s and a contact area of 32.2 cm^2.

length of the perimeter is deformed during contact. Therefore f was calculated as:

$$f=\frac{A+4w\sqrt{A}}{A_p+w\sqrt{\pi A_p}} \quad (2)$$

where A is the area of the region, A_p was the area of the pendulum in the literature [8], and w was set at 0.01 m.

2.4. Data collection

To determine the initial conditions for the simulations, measurements were made of nine healthy male subjects performing heel–toe running at a speed of 4.0 ± 0.4 m/s while wearing the same model of shoe (Adidas Equipment Integral) sized to fit comfortably. Each subject signed an informed consent form prior to the data collection which outlined the data collection protocol. Ten trials were collected for each subject and averaged to produce subject specific initial conditions.

An opto-electronic system (Motion Analysis Corp., Santa Rosa CA, USA) was used to track the three-dimensional movement of the body segments. Three retro-reflective markers were attached to the subjects right shoe (lateral head of the fifth metatarsal, posterior heel, superior lateral aspect of the navicular), shank (head of fibula, anterior mid-shaft of tibia, and distal fibula just proximal to the lateral maleolus), thigh (greater trochanter, anterior mid-thigh, lateral femoral epicondyle) and pelvis (left and right anterior superior illiac spines, and right posterior superior illiac spine). Four infra-red cameras were used in conjunction with motion tracking software (EVA, Motion Analysis Corp. Santa Rosa CA, USA) to collect the marker trajectories at 240 Hz. The marker data were smoothed using quintic splines with a cut-off frequency of 10 Hz [31] and used to reconstruct the positions and orientations of each of the body segments.

To determine the time of touch-down, ground reaction force data were collected simultaneously with the kinematic data at 2400 Hz using a force platform (Kistler Instumente AG, Winterthur, Switzerland). The time of touch-down was determined when the vertical ground reaction force first exceeded 20 N, and toe-off was indicated when the vertical ground reaction force fell below 20 N.

2.5. Muscle stimulation determination

Muscle stimulations were optimized by minimizing the difference between the measured and simulated movement (using the soft shoe) for the subject that most closely matched the model in height and weight. The measure of the closeness of fit of the simulated movement to this averaged movement data was evaluated as the cost function I:

$$I=\sum_i\left\{\frac{[(F_{\text{meas}})_i-(F_{\text{sim}})_i]^2}{[\bar{\sigma}_F]^2}+\sum_j\left\{\frac{[((\theta_{\text{meas}})_j)_i-((\theta_{\text{sim}})_j)_i]^2}{[(\bar{\sigma}_\theta)_j]^2}\right\}\right\} \quad (3)$$

where i are the sampled intervals over the trial (of which there are 25), j are the joints, $(F_{\text{meas}})_i$ is the average measured vertical ground reaction force at interval i, $(F_{\text{sim}})_i$ is the simulated vertical ground reaction forces at interval i, $\bar{\sigma}_F$ is the mean of the vertical ground reaction force standard deviations, $((\theta_{\text{meas}})_j)_i$ is the average measured angle of joint j at interval i, $((\theta_{\text{sim}})_j)_i$ is the simulated angle of joint j at interval i, and $(\bar{\sigma}_\theta)_j$ is the mean standard deviation of angle j. The eight joint angles examined were the subtalar joint angle, the talo-crural joint angle, knee flexion/extension, knee internal/external rotation, knee ab/adduction, hip extension/flexion, hip ab/adduction, and hip internal/external rotation.

The muscle stimulation onsets and magnitudes were the design variables used to minimize I. To reduce the number of design variables, the stimulations for the digital flexors and extensors were made the same as those for the soleus and tibialis anterior muscles respectively, as these muscles had similar functions at the ankle, and the movements of the toes were not expected to affect impact forces. Therefore, there were ten independent stimulation profiles each with onset time and magnitude for a total of 20 design variables.

A simulated annealing [32,33] optimization algorithm was used to determine the muscle stimulation patterns which minimized the cost function I. The same set of muscle stimulations was used for all nine sets of subject-specific initial conditions since the simulations using these stimulations were reasonably close to the measured movements.

2.6. Evaluation procedure

Simulations were performed on nine different subject-specific sets of initial conditions with both the hard and soft shoe conditions to examine the effect of the changing shoe conditions on the ground reaction forces and kinematics. To ensure that the simulations were reasonable reproductions of real movements, the simulations using the initial conditions for each subject were compared to the experimentally measured movements for the same subjects. Peak vertical ground reaction forces in the first 35 ms, and the mean rates of loading over the first 35 ms were determined. Statistical analysis was performed using a paired Student's t-test at $p < 0.05$ to determine whether there were significant differences between soft and hard shoe peak vertical impact forces. Peak muscle forces were determined and compared for the different shoe conditions.

3. Results

3.1. Fit of model to measured movement trials.

The optimal muscle stimulations produced impact simulations which fit the measured data reasonably well. The ground reaction force profile fell within two standard deviations of the measured data for the selected subject (Fig. 3). The subtalar joint, talo-crural joint, knee flexion and hip flexion angles were all within two standard deviations of the mean for the selected subject until the peak impact force (Fig. 4). After the peak impact force, the simulated knee flexed, the talo-crural joint dorsi-flexed and the subtalar joint supinated more than the actual subject.

The simulations performed for the eight other subjects had joint angle and ground reaction force errors of comparable magnitude to that of the subject for which the muscle stimulations were optimized. The mean of the cost function I for the nine subjects was 706 ± 385 while the cost function for the optimized simulation was 400. The greatest joint angle errors occurred near the end of the simulations after the peak in the impact force had occurred. The averaged time histories of the joint angles for the simulations were found to lie within the range of joint angle values observed for the experimental subjects (Fig. 5).

3.2. Effects of shoe hardness

The peak impact force was not found to differ significantly ($p < 0.05$) between the soft and the hard shoe conditions. For five subject-specific simulations, greater impact forces were found for the hard shoe condition, and for four subject-specific simulations, greater impact forces were found for the soft shoe condition (Fig. 6, top left). Initial rates of loading were found to be greater for the hard shoe condition than for the soft shoe condition for all nine subject-specific simulations (Fig. 6, top right). Additionally, rates of knee flexion were greater for the hard shoe than for the soft shoe for all nine subject specific sets of initial conditions (Fig. 6, bottom left).

Muscle forces changed with changing shoe hardness. However, not all muscles changed in the same way with changing shoe hardness. For some muscles, the hamstrings for example, the shoe hardness did not have a consistent effect on peak force across all subjects (Fig. 7, top left). For other muscles, such as the peroneus muscle, peak forces decreased with increasing

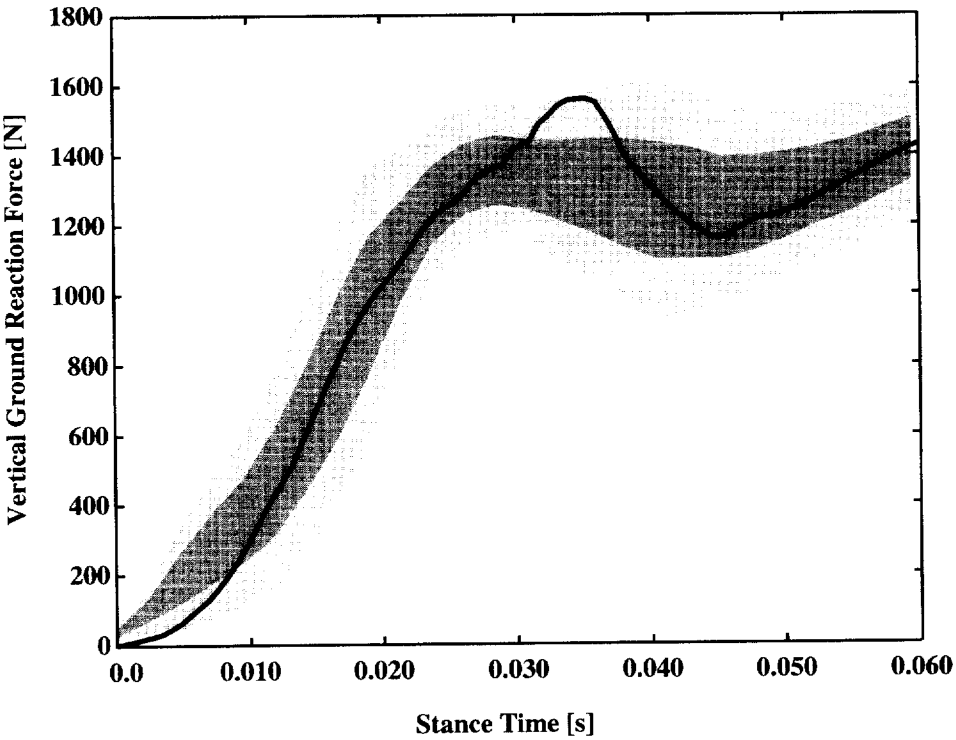

Fig. 3. The vertical ground reaction force profile of the simulation (solid line) compared to 10 measured trials (dark shaded region is bounded by 1 standard deviation, light shaded region is bounded by 2 standard deviations) for one subject.

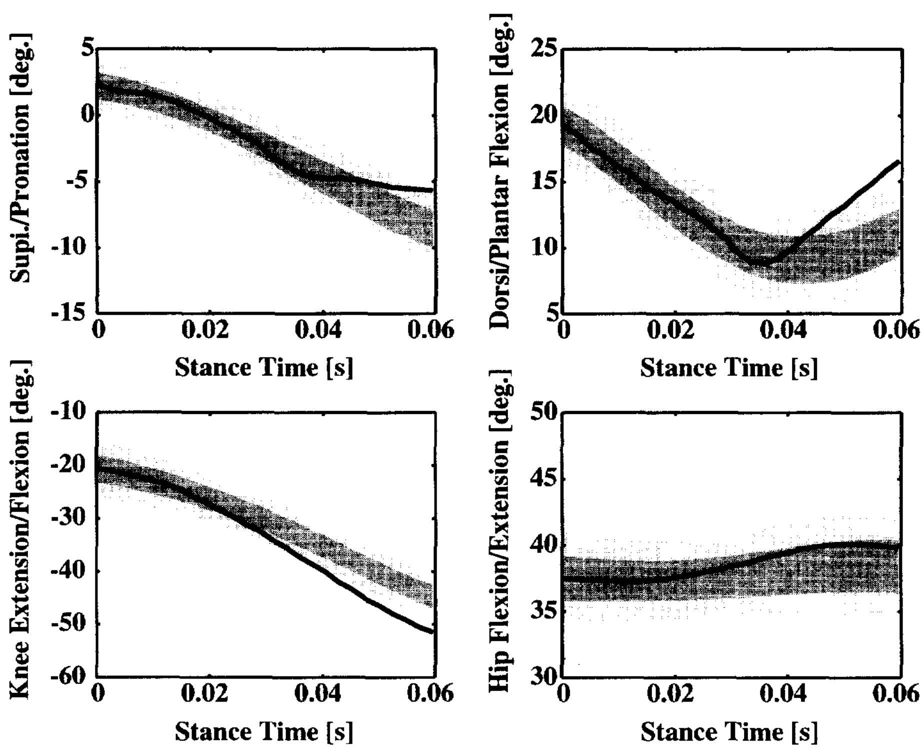

Fig. 4. The foot pronation/supination angle, foot dorsi/plantar flexion angle, knee extension/flexion angle and hip flexion/extension angle of the simulation (solid line) compared to 10 measured trials (dark shaded region is bounded by 1 standard deviation, light shaded region is bounded by 2 standard deviations) for one subject.

shoe hardness for all subject specific sets of initial conditions (Fig. 7, top right), whereas the opposite was true for some other muscles, including the tibialis anterior for example (Fig. 7, bottom left).

4. Discussion

4.1. Model accuracy

The simulations used in the current study had kinetics and kinematics within the range observed in the impact phase of real subjects while heel–toe running. The simulated vertical ground reaction forces (Fig. 3) and the joint kinematics (up until the time of peak impact force) (Fig. 4) were within two standard deviations for the ten trials for the single subject for which the muscle stimulation patterns were optimized. This indicated that the simulated movement was within the range of movements experienced for that subject. Furthermore, the mean simulated joint angle profiles were within the range of joint angle profiles observed for the nine subjects examined (Fig. 5). The difference between the shape of the simulated and experimental vertical ground reaction force profile (Fig. 3) may have been due to the rigid nature of the limb segments used in the simulation model. The joint angle profiles of the simulations after the time of peak impact force did not match the experimental measurements within these same tolerances. These inaccuracies were probably due to the simple step-function muscle stimulation patterns. However, since these discrepancies occurred after the peak impact force, it was not expected that they would influence the response of the model when examining peak vertical ground reaction forces. These results suggested that the simulated movements represent realistic movements in heel–toe running and were sufficient for the examination of passive impact dynamics.

The results showed that the simulation model responded to changes in shoe hardness similarly to human subjects. The peak impact force magnitudes did not change with different shoe conditions which agreed with experimental observations [7,9,11,13,18]. The increase in initial loading rates and initial rate of knee flexion following touch down also matched experimental results [9–11]. These findings further support the utility of the simulation model for investigation of impact dynamics.

The simulation model used in the current study has many parameters that may have influenced the performance of the movement. The values used in this study were based on previously published work, but the accuracy or appropriateness of the selected parameters

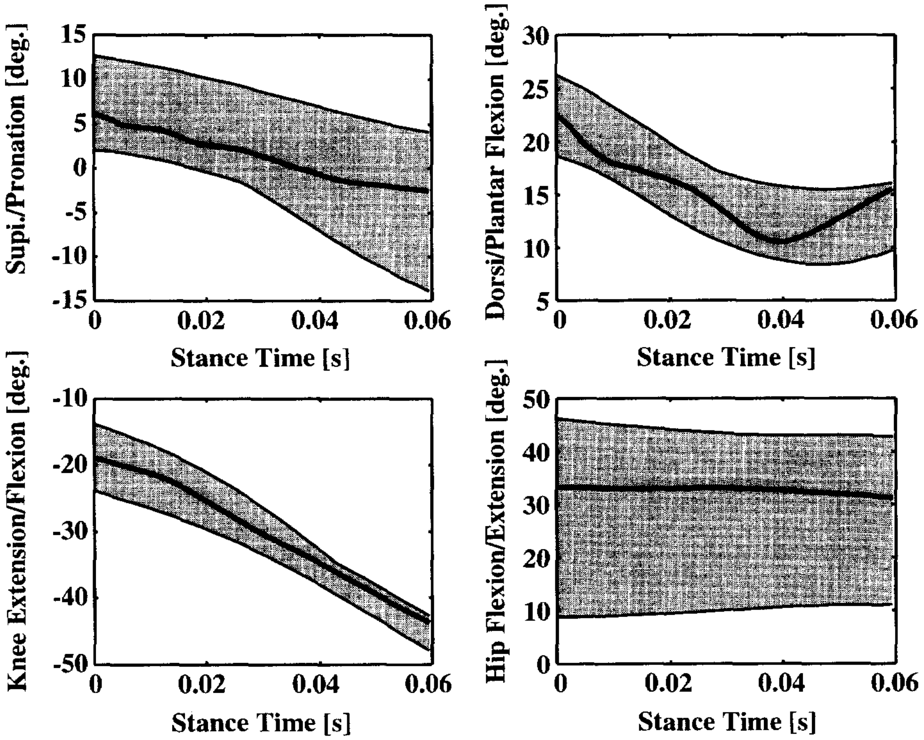

Fig. 5. The mean simulated foot pronation/supination angle, mean foot dorsi/plantar flexion angle, mean knee extension/flexion angle and mean hip flexion/extension angle of all simulations (solid line) compared to the range of the experimentally measured values of the same angles for the nine subjects (dark shaded region).

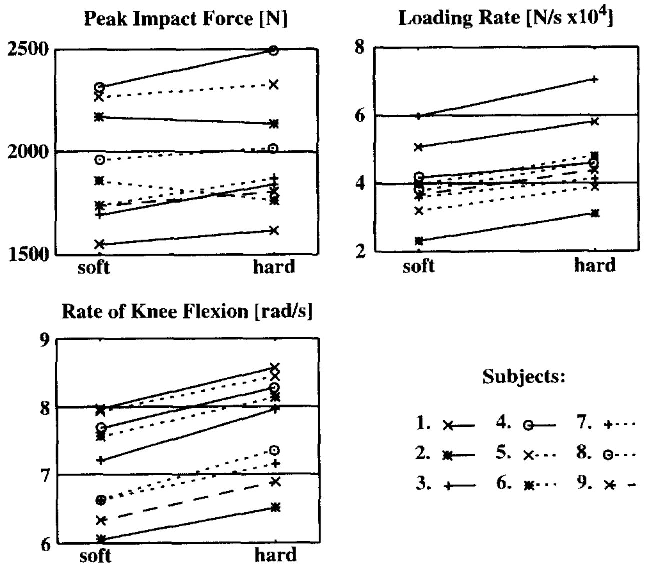

Fig. 6. The effect of shoe hardness on peak impact force (top left), rate of loading (top right), and rate of knee flexion (bottom left) for the nine subject-specific sets of initial conditions.

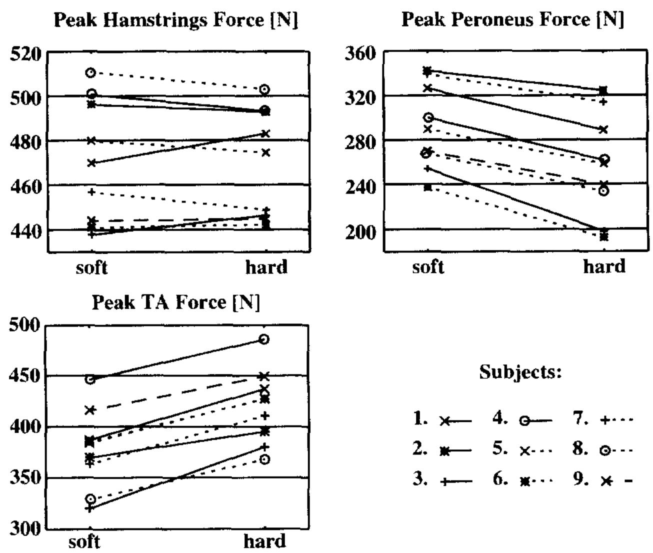

Fig. 7. The effect of shoe hardness on peak hamstrings force (top left), peak peroneus force (top right), and peak tibialis anterior force (bottom left) for the nine subject-specific sets of initial conditions.

is not certain. However a previous sensitivity analysis during the simulation of impact [30] found that variations in initial kinematic variables (such as touchdown velocity and knee flexion angle) have a much greater effect on impact forces than variations in muscle co-contraction. Therefore the initial conditions for several subjects were used for the current investigation. Other parameters, such as limb segment inertial properties and the geometry of the joints and the shoe may affect the magnitude and therefore the absolute accuracy of the simulated movement. However, the parameters selected are close enough to real values that it is reasonable to conclude that the observed mechanism of impact force regulation is possible, though it may not occur in all people.

4.2. *Passive impact force regulation*

The impact simulations showed no significant change in peak vertical impact force with changing shoe sole hardness despite constant initial kinematics and muscle stimulations. Since there was no possibility of active adaptation to the shoe properties, the impact forces were regulated by a passive mechanical response of the system. Therefore it appears that impact forces *can* be regulated by passive mechanisms. The results of this study showed that active adaptation is not required to produce the observed regulation of impact forces, and the lack of experimental evidence [11,18,19] of active changes suggests that it may be passive mechanisms only that account for observed impact force regulation.

Many of the simplifying assumptions used in the model (e.g. rigid limb segments, lack of any reflexes or changes in muscle activity with changing sets of initial conditions) make the simulated impacts more like the passive impact tests and less like real subjects. Therefore, since the model responded to changes in the shoe hardness more like real subjects than like passive impact tests, the conclusion that impact forces can be regulated by passive means is strengthened.

The results of this study do not agree with two previous studies examining the effect of changing shoe hardness on peak impact force magnitudes. The first study used a two dimensional sagittal plane model of the leg during impact, and found increases in impact forces with increases in shoe hardness [30]. However, only a single set of initial conditions were used. Since in the present study different trends in peak vertical impact forces were observed for different sets of subject-specific initial conditions, the results of that study are not conclusive. The lack of agreement between the two studies may also be related to the ground contact model that had only two points of contact, and the different force-deformation–velocity properties of the shoe model used in the previous study. A second experimental study indicated that the effects of knee flexion angle on impact forces were less significant than the effects of surface properties [34]. However, that study did not accurately reproduce the limb segment orientations, velocities and angular velocities observed in running and the impacts were made barefoot upon surfaces of differing hardness, so their findings may not be representative of running with shoes of varying midsole hardness.

For the peak impact forces to remain constant despite changes in shoe hardness, a change in the movement had to occur. This change in the movement could be thought of as a change in the effective mass. The effective mass has previously been defined as the external force divided by the tibial acceleration and is a function of segment masses and joint angles, the latter of which change throughout the movement [35]. The effective mass could alternatively be defined as the external impulse (during the impact phase) divided by the touchdown velocity. This definition encompasses the entire impact phase and is not an instantaneous quantity. Considering this definition, for the initial rates of loading to increase while peak impact forces are held relatively constant with increased shoe hardness (as observed in the current simulations), the effective mass must be reduced. Since the segment masses are held constant in the current simulations, the differences in effective mass between the two shoe hardness conditions must be a consequence of changing joint angles. As knee flexion angle has been shown to influence effective mass [35], and increased rates of knee flexion were observed with the hard shoe condition relative to the soft shoe condition, it may have been that changes in knee flexion rate were the mechanism of impact force regulation for the current simulations. A trend toward increased rates of knee flexion following touchdown with increased shoe hardness has been observed experimentally [9], and changing a rate of knee flexion has been proposed as a mechanism of impact force regulation [9,17]. The current simulation study showed that these kinematic adaptations can be passive mechanical consequences of, rather than active adaptation to, changing shoe sole hardness.

4.3. *Passive changes in internal forces*

Although peak impact forces were not found to vary with shoe hardness, internal forces were affected. Muscle forces were examined as representative of internal forces, and for some muscles, there was no trend in the changes in peak force whereas others increased, and others decreased with increased shoe hardness. These muscle forces changed because of kinematic changes since there were no change in muscle stimulations. As rates of joint flexion changed, so did the velocities of muscles, and as a consequence, the muscle forces changed. These changes in muscle forces resulted in changes in tendon and joint forces, and it is repetitive loading of these internal structures that may eventually lead to overuse injury [11–14]. Furthermore, changes in muscle forces and velocities may affect energy expenditure and comfort or feel of the shoe. Changes in muscle forces and velocities have been observed in the present simulation study without any active adaptation to shoe properties. If there is additional active adaptation to shoe properties, there may be greater changes in internal forces or energy expenditure. Our observations indicate that ground reaction forces do not provide information about the effects of shoe properties on loads within the musculoskeletal system.

4.4. Relevance

The simulations of impact in heel–toe running showed no significant change in peak external impact force magnitude as a function of shoe hardness. It can be concluded that impact forces can be regulated by entirely passive mechanisms since initial limb segment kinematics and muscle stimulation patterns were held constant with different shoe hardness conditions. The rate of knee flexion increased with increased shoe hardness and this may be the passive mechanism of impact force regulation. These same passive mechanisms may account for impact force regulation when people run in shoes of varying hardness.

This passive impact force regulation results in changes in internal forces with changing shoe sole hardness. As protection against overloading is one of the goals in shoe design, an understanding of how these internal forces are affected is of great importance. Since it was found that changes in internal forces do not correspond to changes in external ground reaction forces, external ground reaction forces should not be the only measure of the effectiveness of shoe materials to attenuate forces on the musculoskeletal system.

Acknowledgements

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