MECHANICAL ANALYSIS OF THE LANDING PHASE IN HEEL–TOE RUNNING

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Abstract—Results of mechanical analyses of running may be helpful in the search for the etiology of running injuries. In this study a mechanical analysis was made of the landing phase of three trained heel–toe runners, running at their preferred speed and style. The body was modeled as a system of seven linked rigid segments, and the positions of markers defining these segments were monitored using 200 Hz video analysis. Information about the ground reaction force vector was collected using a force plate. Segment kinematics were combined with ground reaction force data for calculation of the net intersegmental forces and moments.

The vertical component of the ground reaction force vector $F_y$ was found to reach a first peak approximately 25 ms after touch-down. This peak occurs because, in the support leg, the vertical acceleration of the knee joint is not reduced relative to that of the ankle joint by rotation of the lower leg, so that the support leg segments collide with the floor. Rotation of the support upper leg, however, reduces the vertical acceleration of the hip joint relative to that of the knee joint, and thereby plays an important role in limiting the vertical forces during the first 40 ms. Between 40 and 100 ms after touch-down, the vertical forces are mainly limited by rotation of the support lower leg.

At the instant that $F_y$ reaches its first peak, net moments about ankle, knee and hip joints of the support leg are virtually zero. The net moment about the knee joint changed from $-100$ Nm (flexion) at touch-down to $+200$ Nm (extension) 50 ms after touch-down. These changes are too rapid to be explained by variations in the muscle activation levels and were ascribed to spring-like behavior of pre-activated knee flexor and knee extensor muscles. These results imply that the runners investigated had no opportunity to control the rotations of body segments during the first part of the contact phase, other than by selecting a certain geometry of the body and muscular (co-)activation levels prior to touch-down.

INTRODUCTION

The increase in the number of people who run in order to improve their physical fitness has been accompanied by an increase in the prevalence of musculo-skeletal injuries due to running. Most of these injuries are classified by clinicians as 'overuse injuries'. In biomechanical terms the development of an overuse injury to an anatomical structure may be formulated as follows. During each running step, the stress developed within the structure is so high that biological reactions take place which, in the long term, reduce the maximal stress that can be sustained without failure. When this maximal stress drops below the stress actually encountered during running, microtraumata or macrotraumata occur, which constitute an injury.

In the search for the etiology of overuse running injuries and methods of prevention, it would be ideal if time histories of stresses within structures could be determined during running and compared among injured and noninjured individuals, preferably in prospective epidemiological studies. Unfortunately, this is a utopian situation; it is currently impossible to measure stresses in vivo, and in order to calculate stresses we have to make so many assumptions that even the order of magnitude of the outcome is questionable. A crucial unknown in the calculation of stresses is, for instance, how the total force acting on an anatomical structure is distributed over the cross-sectional area of the structure. The only information that can be obtained reliably and routinely seems to consist of measured kinematics and ground reaction forces, and calculated net intersegmental forces and moments. Given the fact that these variables are indirectly related to stresses in anatomical structures, they may bear some relationship to running injuries. The only way to find out whether such a relationship exists is to conduct an epidemiological study in which time histories of the variables are compared among injured and noninjured runners. If differences are found, it may be useful in the search for methods of preventing injuries to compare the variables under various experimental conditions. In this case, varying shoe construction could be an important means of creating different experimental conditions since the choice of footwear is probably all that can be influenced in recreational runners.

It seems reasonable to assume that some running injuries are related to phenomena that occur during the landing phase of a running stride. In this phase, the muscles involved in reducing the momentum of body segments are forcibly lengthened, which could cause them to develop large stresses. Also in this phase, runners who strike the ground first with their rearfoot...
produce an 'impact force peak': a high-frequency peak in the time history of the vertical ground reaction force \( F_z(t) \) during the first 50 ms of ground contact (Cavanagh and LaFortune, 1980; Frederick et al., 1981; Nigg et al., 1987). These force peaks may be accompanied by high stresses on bones and joint surfaces. Information about ground reaction forces during the landing phase in running is already available in the literature (e.g. Cavanagh and LaFortune, 1980), and the extent to which the magnitude of ground reaction force peaks depends on the properties of running shoes has also been studied (Lees and McCullagh, 1984; Komi et al., 1987; Nigg and Morlock, 1987; Nigg et al., 1987, 1988). However, a mechanical analysis of the landing phase in running, which is needed to improve our understanding of muscle functioning during this phase, has not been made. A crucial question is whether positional data of sufficient accuracy can be obtained for a mechanical analysis: the high-frequency peaks in \( F_z(t) \) reflect high-frequency acceleration peaks of body segments, and these acceleration peaks have to be reconstructed by double differentiation of marker position time histories. In a previous study (Bobbert et al., 1991) the first force peak in \( F_z(t) \) during heel-toe running could be reconstructed with errors less than 10% from positional data, suggesting that the method of collecting and processing these data is sufficiently accurate to be used for a mechanical analysis. The purpose of the present study is to make such a mechanical analysis of the landing phase in heel-toe running, with a focus on the role of segmental rotations in limiting the forces which occur during this phase, and the role of muscle moments in controlling segmental rotations.

METHODS

Subjects and experimental protocol

Three trained male 10 km runners (times ranging from 33 to 38 min) participated in this study. Their characteristics (mean ± standard deviation) were: age 28 ± 4 yr, height 1.82 ± 0.03 m, body mass 71 ± 4 kg. Retroreflective spheres were applied to their bodies in order to define seven body segments: the two feet, the two lower legs, the two upper legs, and one segment comprising head, arms and trunk (HAT). The arms were not incorporated separately; in pilot work it had been established that they do not affect \( F_z \) to a significant extent in the early support phase (see also Bobbert et al., 1991; Hinrichs et al., 1987). The subjects were instructed to run along a 30 m runway, in the middle of which a force plate was mounted. Each subject performed five running trials in which he was allowed to use his preferred speed and style. Subsequently, running speed and running style were varied across a number of additional trials so as to obtain a wide range in orientations and velocities of body segments at touch-down. One of the subjects performed only five additional trials. For each trial the subject adjusted the length of the approach until he landed consistently with his right foot on the force plate. During running, the force plate was used to determine the three orthogonal components and the point of application of the ground reaction force vector, and a high-speed video analysis system was used to monitor the 3-D positions of the retroreflective spheres. For the kinematic analysis, linear and angular displacements, velocities and accelerations of the body segments were calculated from the positional data. Net intersegmental forces and moments were calculated using an inverse dynamics approach, combining kinematic information, the measured ground reaction force vector and its point of application, and values reported in the literature for locations of segmental mass centers, segmental masses and segmental moments of inertia. Methods and procedures are detailed below. For reasons of conciseness, only results of the analysis in the sagittal plane projection will be presented.

Collection and processing of ground reaction force data

The three orthogonal components and the point of application of the ground reaction force vector were determined using a force plate (KISTLER type 9287, Kistler Instrumente AG, Winterthur, Switzerland), which was installed according to the manufacturer's specifications. The force plate was connected to an electronic amplifier unit (KISTLER type 9861A, Kistler Instrumente AG, Winterthur, Switzerland), and the eight output signals of this unit were sampled at 1000 Hz using a data acquisition board (DT2821-F-16SE, Data Translation Inc., Marlborough, MA) and a personal computer (COMPAQ Portable III, Compaq Computer Corp., Houston, TX). The \( x, y \) and \( z \) components of the ground reaction force vector, and the \( x \) and \( y \) coordinates of its point of application, were calculated using standard equations supplied by the manufacturer (Kistler, 1984). The accuracy of the calculated point of force application was subsequently improved using the algorithm developed by Bobbert and Schamhardt (1990), which corrects for symmetrically distributed errors found in a systematic comparison of calculated and known points of force application.

For the purpose of synchronization of force plate and video data, the analog \( F_z \) output of the amplifier unit was fed to an electronic circuit including a light-emitting diode (LED) in view of one of the video cameras. The threshold in the circuit was such that the LED came on when \( F_z \) exceeded 20 N.

Collection and processing of positional data

The method of collecting video data has been detailed elsewhere (Bobbert et al., 1991), and will be described only briefly below. In order to minimize the amount of skin movement relative to the mass center of leg segments, a device was used on each leg. The
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device consisted of two light wooden rods connected by a hinge joint. The hinge joint was aligned with the estimated flexion-extension axis of the knee joint (at the height of the lateral collateral ligament, 2 cm above the tibial plateau). Subsequently, one rod was fastened to the lower leg and one to the upper leg using athletic tape and elastic bandages. Retroreflective spheres with a diameter of 2 cm were fixed to the wooden rods at the height of the estimated plantarflexion/dorsiflexion axis of the ankle joint (on the lateral malleolus, 0.5 cm anterior to its tip), the axis of the hinge joint, and the estimated axis of flexion/extension of the hip joint (2 cm proximal to the greater trochanter). Two retroreflective markers were also applied to the lateral side of the shoe of the support leg, one at the height of the 5th metatarsophalangeal joint, the other at the height of the tuber calcanei. On the upper body, one marker was taped to the skin at the height of the sternum (with long strips of athletic tape running from below the sternum diagonally across the thorax, over the neck, to the shoulder blades) and one marker was fixed on top of the head (using an elastic band passing under the chin). The latter two markers were retroreflective spheres with diameters of 4 cm. The marker positions are shown schematically in Fig. 1.

Video data were collected using a VP310 video recorder (Motion Analysis Corporation, Santa Rosa, CA) and four electronically shuttered cameras (NAC MOS V-14 Camera 60/220 F/S) equipped with 12-120 mm zoom lenses (Angénieux, Paris, France). A 1000 W lamp was placed directly behind each camera. The cameras were zoomed in to a volume of 2 m in the x direction (axis along which subjects were running), 1 m in the y (medio-lateral) direction and 2 m in the z (vertical) direction, with the center of the base area corresponding to the center of the force plate. This volume was subsequently calibrated using 16 control points and Expert Vision 3-D software.

During running, data were collected at 200 Hz, the maximum sampling frequency of the system. For synchronization of video data with force data, the LED described above was used. The first video frame on which it was detected was taken to correspond in time to the second force sample following the one where \( F_y \) had exceeded the 20 N threshold. This way, synchronization errors ranged from -3 to +2 ms.

Time histories of marker positions were determined using Expert Vision 3-D software and smoothed using a Butterworth 4th order (zero lag) filter. A cutoff frequency of 15 Hz was used for smoothing position time histories of head and sternum markers, 50 Hz was used for all other markers. In a previous study (Bobbert et al., 1991) it has been shown that lowering the cutoff frequency for head and sternum markers from 50 to 15 Hz resulted in a slight improvement during the landing phase of the correspondence between \( F_y(t) \) calculated from positional data and \( F_y(t) \) measured using a force plate. The smoothed position time histories were differentiated numerically using a direct 3-point derivative routine in order to obtain time histories of linear velocities and accelerations. Angles of body segments with the horizontal were calculated from the smoothed marker position time histories, and differentiated to obtain time histories of angular velocities and accelerations.

**Kinematic analysis**

The kinematic data were used to produce stick figures in which the linear velocity and acceleration vectors of markers were displayed as a function of position. As mentioned before, only results of the analysis in the sagittal plane (projection on x-z plane) will be presented. The difference in the vertical accelerations of the knee and the ankle joint markers (\( z_k - z_\alpha \)) was analyzed as follows. The vertical distance between the knee and the ankle joint markers (\( z_k - z_\alpha \)) is given by

\[
(z_k - z_\alpha) = l_1 \sin \theta_1, \tag{1}
\]

where \( l_1 \) is the length of the line segment defined by the two markers (length of the lower leg), and \( \theta_1 \) is the angle between this line segment and the horizontal. Double differentiation of both sides of equation (1) with respect to time yields

\[
\begin{align*}
(\dot{z}_k - \dot{z}_\alpha) &= l_1 \dot{\theta}_1 \cos \theta_1 - l_1 \dot{\theta}_1^2 \sin \theta_1 + \dot{l_1} \sin \theta_1, \\
&\quad + 2 \dot{l}_1 \dot{\theta}_1 \cos \theta_1, \tag{2}
\end{align*}
\]

If \( l_1 \) is taken to represent the length of the bones of the lower leg, it should remain virtually constant and all the terms including the derivatives of \( l_1 \) should be negligible, giving

\[
(\ddot{z}_k - \ddot{z}_\alpha) = l_1 \dot{\theta}_1 \cos \theta_1 - l_1 \dot{\theta}_1^2 \sin \theta_1, \tag{3}
\]

with the two terms on the right-hand side representing the vertical components of tangential and centripetal acceleration, respectively. Thus, all variations in \((\ddot{z}_k - \ddot{z}_\alpha)\) due to the terms including \( \dot{l}_1 \) and \( \ddot{l}_1 \) may be considered as experimental errors. Figure 2 gives an example of \((\ddot{z}_k - \ddot{z}_\alpha)\) and \((\ddot{z}_k - \ddot{z}_\alpha)^*\) time histories re-
Fig. 2. Time histories of the difference in vertical acceleration between knee and ankle joint of the support leg in heel-toe running. $(\ddot{x}_k - \ddot{x}_a)$ was calculated directly by double differentiation of the positional data, $(\ddot{x}_k - \ddot{x}_a)$ was calculated assuming a constant distance between knee and ankle markers [cf. equation (3)]. Curves have been plotted from the last frame before touch-down ($t=0$) to take off.

Fig. 3. Stick figures for the first part of the contact phase in heel-toe running. The leftmost stick figure shows the segment orientations on the last frame before touch-down. Time between two subsequent stick figures is 5 ms.

**RESULTS AND DISCUSSION**

**Kinematics and forces**

Except when explicitly mentioned otherwise, the results presented and discussed were extracted from the trials in which the subjects used their preferred running speed and style. The running speeds selected in these trials ranged from 3.8 to 4.5 ms$^{-1}$. All three subjects were so-called rearfoot strikers (Cavanagh and LaFortune, 1980) or heel-toe runners (Nigg et al., 1988), producing 'impact force peaks' in $F_y$ that ranged from 1400 to 1850 N. Although the results varied somewhat from trial to trial and from subject to subject, the overall pattern of kinematic and kinetic changes was the same across trials and subjects. This
overall pattern will be described and discussed using the results of a single trial. It should be stressed here that if any of the other trials had been selected, the outcome would have been qualitatively identical, although quantitatively slightly different. Figure 3 presents stick figures for the phase of interest in this trial, starting on the last frame before touch-down \((t=0)\). For clarity, the stick figures have been separated and the swing leg has been deleted to show velocity vectors of markers at support leg ankle, support leg knee, support leg hip, sternum and head (Fig. 4), acceleration vectors of these markers (Fig. 5), and a free-body...
The relationship between kinematics and variations in the vertical component of the ground reaction force vector is more complex, because the vertical velocity of the large mass of HAT does change considerably. The vertical velocity of the markers on head and sternum was reduced from $-1.0 \text{ m s}^{-1}$ at touch-down to $-0.4 \text{ m s}^{-1}$ at $t=0.080\text{s}$. Figure 6 shows that during this phase the vertical component of the ground reaction force vector increases to reach a first peak at $t=0.025\text{s}$ (the 'impact force peak'), decreases to a local minimum reached at $t=0.040\text{s}$ and, thereafter, increases again. The calculated and the measured $F_z(t)$ curves, and contributions of body segments to the calculated $F_z(t)$ curve, are shown in Fig. 7. As discussed in a previous paper (Bobbert et al., 1991), the first peak in $F_z(t)$ has its origin in upward acceleration of the support leg segments. Its magnitude, however, is determined to a large extent by the contribution of the rest of the body. This is due to its large mass rather than to high vertical accelerations; in Fig. 5, where all marker acceleration vectors are plotted on the same scale, the ones pertaining to head and sternum markers do not even appear until $t=0.055\text{s}$. Note that acceleration vectors do appear for the hip joint marker. Differences between these vectors and vectors for sternum and head markers were found to occur in some parts of the landing phase because of rotational motion in the frontal plane of the pelvis about the support leg hip joint, causing the mass center of HAT to be displaced (accelerated) relative to this joint.

In Fig. 5 it can also be seen that when the sole of the

![Fig. 6. Free-body diagrams showing the vector of the ground reaction force and the vector of the force of gravity. For clarity, diagrams have been separated and the swing leg has been deleted. The leftmost figure ($t=0$) shows the situation on the last frame before touch-down. Time between two subsequent diagrams is 5 ms.](image)
shoe becomes horizontal, the direction of the acceleration vector of the knee joint marker changes from upward to downward. Thus, the downward velocity of the shank no longer decreases but begins to increase. This is a consequence of the fact that the ankle joint can no longer move forward. Since the forward velocity of the knee joint is maintained, motion of the lower leg is limited to rotation about the now stationary ankle joint. This rotation is associated with a centripetal acceleration pointing from the knee to the ankle, with the negative acceleration being relatively large because the lower leg is almost vertical. Similar results have been reported by LaFortune and Hennig, 1989). The change in the direction of the acceleration vector of the knee joint marker from upward to downward is accompanied by a reduction of the upward acceleration of the hip joint marker (vectors in Fig. 5 disappear between \( t = 0.04 \) s and \( t = 0.05 \) s), which corresponds in time to the phase where \( F_y(t) \) goes through a local minimum (Fig. 6).

The observations presented above were subjected to a quantitative analysis. Figure 8 shows the time histories of the vertical accelerations of the knee and the ankle joint markers (\( \ddot{z}_K \) and \( \ddot{z}_A \), respectively), the vertical acceleration difference \( (\ddot{z}_K - \ddot{z}_A)_v \) calculated according to equation (3), and the contributions to \( (\ddot{z}_K - \ddot{z}_A)_v \) of centripetal acceleration \( (I_{\theta})_v \sin \theta_0 \) and tangential acceleration \( (I_{\theta})_v \cos \theta_0 \) of the lower leg [cf. equation (3)]. Curves have been plotted from the last frame before touch-down \( (t = 0) \).

**Fig. 7.** (a) Calculated and measured time histories of the vertical ground reaction force \( (F_y) \). (b) Segmental contributions to \( F_y(t) \). Curves have been plotted from the last frame before touch-down \( (t = 0) \). The mass of the subject was 71 kg.

**Fig. 8.** (a) Time histories of the vertical accelerations of knee and ankle joint markers in the support leg \( (\ddot{z}_K \) and \( \ddot{z}_A \), respectively). (b) Time histories of the vertical acceleration difference \( (\ddot{z}_K - \ddot{z}_A)_v \) calculated according to equation (3), and the contributions to \( (\ddot{z}_K - \ddot{z}_A)_v \) of centripetal acceleration \( (I_{\theta})_v \sin \theta_0 \) and tangential acceleration \( (I_{\theta})_v \cos \theta_0 \) of the lower leg [cf. equation (3)]. Curves have been plotted from the last frame before touch-down \( (t = 0) \).
sion of these contributions than \((\ddot{z}_K - \ddot{z}_A)\) calculated directly by double differentiation of the positional data. It turns out that \((\ddot{z}_K - \ddot{z}_A)^*\) is almost zero during the phase where the first peak occurs in \(F_r(t)\). After \(t = 0.035\) s, the vertical component of centripetal acceleration helps to reduce \(\ddot{z}_K\) relative to \(\ddot{z}_A\). The contribution of centripetal acceleration to \((\ddot{z}_K - \ddot{z}_A)^*\) peaks at \(t = 0.055\) s, just after the ankle joint has experienced its maximum backward acceleration (Fig. 5) and its horizontal velocity has become very small. The results of a similar analysis of the contribution of rotation of the upper leg to the vertical acceleration difference between hip and knee joints, \((\ddot{z}_H - \ddot{z}_K)^*\), are presented in Fig. 9. In this case, the most important contribution comes from the vertical component of the tangential acceleration of the upper leg. Because of this component, \(\ddot{z}_H\) is lower than \(\ddot{z}_K\) during the first 40 ms of the support phase, and \(\ddot{z}_H\) remains positive in spite of the fact that \(\ddot{z}_K\) is negative between \(t = 0.045\) s and \(t = 0.075\) s.

One of the purposes of this study was to determine the role of segmental rotations in limiting the forces that occur during the landing phase. In order to fully appreciate this role in limiting the vertical forces, it should be realized that rotation of a support leg segment affects not only the vertical acceleration of the mass center of that particular segment but also the vertical acceleration of segments proximal to it. Since most of the mass of the body is located proximal to the hip joint, the influence of rotation of a support leg segment on \(F_r(t)\) is determined primarily by its effect on \(\ddot{z}_H\). Looking at the results from this point of view, it becomes clear that if rotation of the upper leg had not occurred during the first 40 ms of the support phase, \(\ddot{z}_H\) would have been higher (up to a maximum of \(\pm 50\) m s\(^{-2}\)), \(\ddot{z}_H\) would have been higher, and the first force peak in \(F_r(t)\) would have been higher (theoretically up to a maximum of \(\pm 3000\) N). Thus, it may be said that rotation of the upper leg, because of the associated tangential acceleration, plays a major role in limiting \(F_r\) during the first 40 ms of the support phase. In a similar way it may be argued that rotation of the lower leg, because of the associated centripetal acceleration, also plays a role in limiting \(F_r\) between \(t = 0.040\) s and \(t = 0.100\) s. In this phase it acts opposite to the clockwise angular acceleration of the thigh, which, as it slows down the anticlockwise angular velocity, tends increase the value of \(F_r\).

**Kinematics and moments**

Figure 10 shows the free-body diagrams of the upper leg and the lower leg for the first 80 ms of ground contact. For the sake of clarity in illustration, intersegmental moments have been shown as circular arrows, with the center of the circle lying at the proximal or distal end of a segment and the angle of the arc corresponding to the magnitude of the moment; a circular arrow pointing anticlockwise represents a moment vector coming out of the paper. In the landing phase, the limits of the ranges of motion in hip, knee and ankle joints are not approached. Thus, the moments exerted by passive structures about these joints will be small, and it seems safe to interpret changes in the net intersegmental moments about these joints as changes in the net muscle moments.

Before touch-down hip extension and knee flexion moments are present, indicating a backward acceleration of the leg segments relative to the rest of the body. This explains why the acceleration of the ankle joint is backward (Fig. 5) although there is no ground reaction force yet (Fig. 6). The moment about the hip joint does not change direction, although it decreases
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Fig. 10. Free-body diagrams of support upper leg and lower leg. Proximal and distal intersegmental forces have been shown as vectors with their origin at the segmental endpoints, proximal and distal intersegmental moments have been shown as circular arrows, with the center of the circle lying at a segmental end point and the angle of the arc corresponding to the magnitude of the moment; a circular arrow pointing anticlockwise actually represents a moment vector coming out of the paper. For clarity, the diagrams have been separated and the swing leg has been deleted. The leftmost figure (t = 0) shows the situation on the last frame before touch-down. Time between two subsequent diagrams is 5 ms.

temporarily during the phase where the knee joint is accelerated downward, i.e. between \( t = 0.040 \) s and \( t = 0.070 \) s (Fig. 10). The moment about the knee joint, however, does not change direction; it switches from flexion to extension between touch-down and \( t = 0.035 \) s. At the instant where the first peak in \( F_{x}(t) \) occurs, the proximal and distal joint reaction forces acting on the tibia are directed along the long axis of the lower leg, and the knee joint moment is almost zero. The same phenomenon was found in trials where the subjects used other-than-preferred running styles, including Groucho running (running while keeping the body's mass center low, as described by McMahon et al. (1987). Between \( t = 0.030 \) s and \( t = 0.055 \) s, the knee extension moment increases from 10 to 200 Nm. At the ankle joint, a dorsiflexion moment occurs during the first 45 ms of ground contact; this may be concluded from the fact that the ground reaction force vector passes posterior of the joint (Fig. 6). The moment values are, however, too small to be plotted on the same scale as the moments about knee and hip joints; the maximum dorsiflexing moment is only 24 Nm. This value is typical of all running trials of the three subjects. Note that during the first 15 ms, \( \ddot{z}_A \) remains downward (Fig. 8), indicating that the moments exerted about the ankle have little effect on the movement.

Neuromuscular control during the landing phase

One of the most interesting questions is whether neuromuscular control during the landing phase in running has any influence on the forces occurring in this phase. Figure 10 shows that changes in the net moments about the joints occur, but since these changes are extremely rapid it is unlikely that they are due to changes in the muscle activation levels. Let us consider, for instance, the change from \(-100 \) Nm (flexion) to \(+200 \) Nm (extension) which occurs in the net moment about the knee joint during the first 55 ms of ground contact (these values are typical of the trials in which the subjects used their preferred running speed and style). If a subject is instructed to contract his knee extensor muscles as rapidly and forcefully as possible while the knee angle is kept constant in a dynamometer, it typically takes more than 100 ms before the knee extension moment has increased from 0 to 100 Nm (see e.g. Viitasalo et al., 1980; Viitasalo, 1982). Moreover, significant changes in the moments due to stretch reflexes do not seem to occur within the first 100 ms after stretch (Melvill Jones and Watt, 1971; Gottlieb and Agarwal, 1979a, b). In order to explain the rapid changes in the moment about the knee joint during running that were observed in this study, the following hypothesis may be put forward. At touch-down, co-contraction of knee extensor and knee flexor muscles is present, with the knee flexors exerting a larger moment than the knee extensors (since a net knee flexing moment is present at touch-down). Because of the collision with the ground, a rapid knee flexion occurs so that the knee flexors experience a quick release, causing the force exerted by them to drop. Simultaneously, the knee extensors experience a sudden stretch, which causes the force produced by them to increase. Both the decrease in the force exerted by the knee flexors and the increase in the force produced by the knee extensors contribute to a rapid change in the net moment about the knee joint from flexion to extension.

The mechanism proposed above assumes spring-like behavior of muscle–tendon complexes during rapid changes in the distance between origin and insertion. The stiffness of muscle–tendon complexes under such conditions varies with the stiffness of the cross-bridge array, which, in turn, depends on the number of cross bridges (Morgan, 1977). Since the changes in the net knee joint moment are relatively large, with a flexion moment of 100 Nm being close to the maximum isometric knee flexion moment (Wickie-
wicz et al., 1984) and an extension moment of 200 Nm possibly even exceeding the maximum isometric knee extension moment (Wickiewicz et al., 1984), we need a strong co-contraction of knee extensor and knee flexor muscles and a concomitant high stiffness for the proposed mechanism to work. Results of electromyographic studies suggest that a strong co-activation of quadriceps femoris and hamstring muscles does occur before touch-down and during the first part of the support phase in running (Elliott and Blanksby, 1979; Mann and Hagy, 1980; Mann et al., 1986). It has been speculated that this co-activation serves the purpose of lower limb stabilization (Elliott and Blanksby, 1979). According to the argument given above, however, co-activation is an essential step in the preparation for the landing phase. Without it, the rate of change of muscle moments about the knee joint would depend on the rate of force development (or decay) following changes in muscle activation. This rate is relatively low because of the interaction between contractile elements and series elastic elements, and possibly also because the active state of muscles may rise or decay slower than the level of muscle activation (Bobbert and van Ingen Schenau, 1990). Thus, the absence of co-activation could lead to a loss of (passive) control of body motion during the landing phase.

**Touch-down conditions and segmental force contributions during landing**

In the previous paragraph it was shown that during the first 25 ms of the landing phase, where \( F_t(t) \) increases to reach its first peak, lower leg rotation does not reduce \( \dot{z}_K \) relative to \( \dot{z}_A \). Thus, the acceleration of the mass centre of the lower leg and, therewith, the contribution of the lower leg to \( F(t) \), depends purely on what happens distal to the ankle joint. It was also shown that \( \dot{z}_A \) does not become positive until \( t = 0.015 \text{s} \), while the moment arm of the ground reaction force vector with respect to the ankle joint remains close to zero during the impact phase, at least in the sagittal plane projection (Fig. 6). It seems, therefore, that the ankle dorsiflexor muscles have little influence on \( \dot{z}_A \), so that the collision of the lower leg (and foot) with the force plate should be comparable to a mass striking a spring, the spring being formed by the elastic structures underneath the calcaneus (calcaneal fat pad and sole of the shoe). In the case of a mass striking a spring, the force developed during the collision increases with the initial velocity, the absolute force level depending on the spring stiffness (McMahon et al., 1987). Figure 11(a) shows for one subject (the one who performed 20 running trials) the contribution of the lower leg to the first peak in \( F_t(t) \) (Fig. 7) as a function of \( \dot{z}_{A,\text{max}} \), the maximal downward velocity of the ankle joint (in the case of the trial used for illustration, \( \dot{z}_{A,\text{max}} \) was reached at \( t = 0.015 \text{s} \)). The coefficient of linear correlation between the two was 0.82 (\( p < 0.01 \)). Figure 11(b) shows for the same trials the contribution of the upper leg to the first peak in \( F_t(t) \) (Fig. 7) as a function of \( \dot{z}_{A,\text{max}} \). Again, a high correlation of 0.77 (\( p < 0.01 \)) was found. The latter finding is not really surprising: \( \dot{z}_k \) equals \( \dot{z}_A \) when the first peak in \( F_t(t) \) occurs, and \( \dot{z}_K \) determines approximately half the vertical acceleration of the mass centre of the upper leg. It seems that the collision of the support leg with the ground can indeed be described as a collision between a mass and a spring. A similar description was proposed earlier by Denoth.
the selection of initial conditions may be improved by
studying what would happen to the mechanics of the
landing phase if the position of the lower leg or pre-
activation levels of muscles were changed. Possibly,
such changes, which could occur when runners be-
come fatigued, are involved in the development of
injuries. For a study of the effects of the initial con-
ditions on the mechanics of the landing phase, forward
dynamics simulation of the landing phase in heel-toe
running with a mathematical model of the musculo-
skeletal system seems indicated.

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tion into the shock absorbency of running shoes and shoe

Final remarks and future research
Based on a mechanical analysis of running trials of
three subjects, it was argued in this paper that during
the first 25 ms of the landing phase in heel-toe run-
ing, where \( F \) increases to reach its first peak, changes
in muscle moments about the joints are so rapid that
they have to be caused by spring-like behavior of pre-
activated muscles. This implies that the runners in-
vestigated had no opportunity to control the rotations
of body segments during this period, other than by
selecting the initial conditions, i.e. by selecting a
certain geometry of the body and muscular (co-
activation levels prior to touch-down. It would be
interesting to know the criteria used in selecting
the initial conditions. One of them seems to be that the
backward acceleration of HAT is minimized during
landing, so that a minimal amount of energy is utilized
for re-increasing the horizontal velocity of HAT. In
the vertical direction, the situation is less clear. For
energy requirements it may be useful to minimize the
vertical excursion of HAT, so that no energy is wasted
in moving up and down. However, this requires high
vertical accelerations of HAT, leading to large forces
and possibly hazardous loads on the locomotor sys-
tem. It is noteworthy that selecting an almost vertical
position of the lower leg at touch-down, rather than a
position past vertical, at the same time allows for
limitation of the vertical acceleration of HAT (because
the contribution of centripetal acceleration helps to
reduce \( \dot{z}_x \) relative to \( \ddot{z}_x \) and keeps the vertical ex-
cursion of HAT small (\( \ddot{z}_x - \dot{z}_x \) is actually slightly
increased during the first 50 ms of ground contact, as
can be seen in Fig. 4). Although selecting this position
of the lower leg at touch-down seems advantageous
for controlling the motion HAT, it also seems to be
partly responsible for the occurrence of the high-
frequency force peak in \( F_y(t) \). Our understanding of
the selection of initial conditions may be improved by

(1986) and Nigg and Morlock (1987). Since the spring
stiffness depends on the stiffness of the shoe sole and
the stiffness of the surface, it is theoretically possible to
reduce the force contributions of the upper and the
lower legs to the first peak in \( F_y(t) \), and therewith
attenuate the high-frequency force peaks 'seen' by the
knee and ankle joints. Whether this is a practical
solution or not depends, of course, on how thick the
shoe sole or surface would have to be for that. As
pointed out in a previous study (Bobbert et al., 1991),
the effects of changes in shoe or surface stiffness on
forces occurring during the impact phase should be
evaluated by determining the segmental contributions
to the first peak in \( F_y(t) \) and not by determining only
the absolute value of this peak. The reason is that the
segmental force contributions of support leg segments
and the rest of the body (Fig. 7) can be varied inde-
pendently. In this context, it is relevant to mention
that the coefficient of linear correlation between the
first peak in \( F_y(t) \) and \( |\ddot{z}_A,\text{max} | \) was only 0.5.


