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Effect of landing stiffness on joint kinetics and energetics in the lower extremity

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ABSTRACT

DEVITA, P. and W. A. SKELLY. Effect of landing stiffness on joint kinetics and energetics in the lower extremity. *Med. Sci. Sports Exerc.*, Vol. 24, No. 1, pp. 108–115, 1992. Ground reaction forces (GRF), joint positions, joint moments, and muscle powers in the lower extremity were compared between soft and stiff landings from a vertical fall of 59 cm. Soft and stiff landings had less than and greater than 90 degrees of knee flexion after floor contact. Ten trials of sagittal plane film and GRF data, sampled at 100 and 1000 Hz, were obtained from each of eight female athletes and two landing conditions. Inverse dynamics were performed on these data to obtain the moments and powers during descent (free fall) and floor contact phases. Angular impulse and work values were calculated from these curves, and the conditions were compared with a correlated *t*-test. Soft and stiff landings averaged 117 and 77 degrees of knee flexion. Larger hip extensor (0.010 vs 0.019 $N \cdot m \cdot s \cdot kg^{-1}$; $P < 0.01$) and knee flexor (−0.010 vs −0.013 $N \cdot m \cdot s \cdot kg^{-1}$; $P < 0.01$) moments were observed during descent in the stiff landing, which produced a more erect body posture and a flexed knee position at impact. The shapes of the GRF, moment, and power curves were identical between landings. The stiff landing had larger GRFs, but only the ankle plantarflexors produced a larger moment (0.185 vs 0.232 $N \cdot m \cdot s \cdot kg^{-1}$; $P < 0.01$) in this condition. The hip and knee muscles absorbed more energy in the soft landing (hip, −0.60 vs −0.39 $W \cdot kg^{-1}$; $P < 0.01$; knee, −0.89 vs −0.61 $W \cdot kg^{-1}$; $P < 0.01$), while the ankle muscles absorbed more in the stiff landing (−0.88 vs −1.00 $W \cdot kg^{-1}$; $P < 0.05$). Overall, the muscular system absorbed 19% more of the body's kinetic energy in the soft landing compared with the stiff landing, reducing the impact stress on other body tissues. The ankle plantarflexors provided the major energy absorption function in both conditions, averaging 44% of the total muscular work done followed by the knee (34%) and hip (22%) extensors.

IMPACT FORCES; VERTICAL FALLS; MOMENT OF FORCE; MUSCLE POWER; BIOMECHANICS; HIP, KNEE, ANKLE; INVERSE DYNAMICS; FORCE PLATFORM

Jumping and landing movements are integral features of many sporting activities and have been investigated by numerous researchers. Generally, the research on jumping seeks to understand the performance aspects of this movement, namely, how the body generates and utilizes the forces and energy necessary to propel itself into the air (2,5,6,20). In contrast, the research on landing has concentrated on the biomechanical implications of impact and the resulting loads placed on lower extremity tissues (9,11,12,19,22,25). This body of work seeks to evaluate the injury potential of various landing situations as opposed to identifying the performance variables necessary to stop the fall of the body center of mass. Additionally, these studies reported only externally applied ground reaction forces and then hypothesized lower extremity tissue loads.

Few landing studies have been reported in which inverse dynamic analytical methods have been employed to identify internal limb kinetics and energetics, and methodological factors limit the reliability and generalizability of these studies. Two of these investigations (23,29) used only one subject. Smith (23) reported knee, gastrocnemius, and patellar tendon forces in landing from a relatively high (1.07 m) platform. Zatsiorsky and Prilutsky (29) measured the ratio of the work done by the muscle moments to the reduction in total mechanical energy during soft and stiff landings. Fukuda (10) compared joint moments and powers in landing from a relatively low height (0.20 m) onto two surfaces with different stiffness characteristics. The analysis was performed with kinematic data only, progressing from hip to ankle joints. Panzer, Wood, Bates, and Mason (21) reported hip and knee compressive and shear forces in landing from single and double back somersaults. Since the horizontal ground reaction forces were largely biased in the posterior direction, these results may not be indicative of those in landing from vertical falls. Additionally, Bobbert, Huijing, and van Ingen Schenau (3,4) reported lower extremity joint moments and powers in several landing movements. The focus of this work, however, was the subsequent jump after landing and the results and discussion emphasized the jumping phase.

While falling from a distance above the floor, a person's total energy will change from potential to kinetic energy. The subsequent landing will involve movements designed to dissipate the kinetic energy and will be characterized by work being performed on the muscles of the lower extremity. Joint moments of force and muscle powers in the lower extremity can be used to describe the mechanisms by which the acquired kinetic energy is dissipated. Additionally, several re-

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searchers (7,17,18) have identified impact forces that occur in less than 30–50 ms as passive forces. The term passive implied that these forces were applied too rapidly to be modified by a reaction response from the human neuromuscular system. Any attempt to reduce these external impact forces must therefore include some activity prior to contacting the landing surface, as stated by Lees (14). Other researchers (8,16,24) have hypothesized that this activity may be a presetting of the lower extremity muscle tensions during the descent phase, which they further state is directly related to the magnitude of the impact ground reaction force. The purpose of this study was to identify and compare ground reaction forces, and joint positions, moments, and muscle powers in the lower extremity during the descent (free fall) and floor contact phases of soft and stiff landings. Soft and stiff landings had relatively large and small amounts of knee flexion, respectively, during the floor contact phase.

METHODS

Subjects. Eight healthy, female, intercollegiate basketball and volleyball players (mean mass: 68.4 ± 8.0 kg, mean age: 20.0 ± 1.3 yr) volunteered as subjects for this study. All subjects signed informed consent forms prior to their participation in accordance with university and American College of Sports Medicine policy.

Instrumentation. An AMTI force platform interfaced to a computer was used to measure the vertical and anteroposterior ground reaction forces (GRF) under the right foot at a sampling frequency of 1000 Hz. Sagittal plane film records of the descent and floor contact phases were obtained with a 16 mm LoCam camera operating at 100 Hz and located 11.3 m from the center of the force platform. The field-of-view of the camera at the force platform was approximately 2.0 m wide and 2.4 m high.

Experimental protocol. Each subject was tested in a single session and wore her own court shoes during the test. All shoes were similarly constructed with leather high-top uppers and were in good condition. Prior to data collection, six landmarks were identified and marked on the subject's right side to aid in the digitizing process: 5th metatarsal head, mid-point of the back edge of the shoe, lateral maleolus, lateral femoral condyle, greater trochanter, and shoulder. Circumference measures at the upper thigh, knee, ankle, and metatarsal heads were taken for later use in the mathematical model of the human body (13).

To standardize the vertical velocity during the descent phase and to limit horizontal velocity for all subjects and trials, the subjects stepped off a 59 cm high platform that was placed 11 cm from the edge of the force platform. The subjects landed with the right foot on the force platform and the left foot on the floor.

For the purposes of describing the tasks to the subjects and monitoring the performances, the soft and stiff landing techniques were defined by maximum knee flexion angles of greater than and less than 90 degrees from full extension, respectively. The order of presentation of the two techniques was counterbalanced across subjects. Prior to testing, the subject practiced the technique, first at a lower height, then at the testing height, until the movement could be performed appropriately and consistently. Each subject was instructed to carefully step off the platform without jumping up or lowering her body prior to leaving the platform. The subject stood in a balanced position near the front edge of the platform with her right foot suspended over the force platform and her right heel resting against the front of the raised surface. This position placed the subject's center of mass as far forward as possible in an attempt to limit horizontal motion. To initiate the movement, the subject simply shifted her weight forward and dropped as vertically as possible while attempting to land evenly balanced on both feet. After landing, the subject returned to her normal standing position. The arms were not constrained during the landings and were generally held up and in front of the subjects for balance. Each subject performed 10 successful trials. All trials were visually monitored to assure that the technique was performed correctly and the right foot was completely on the force platform. All subjects used a forefoot landing style at initial contact with the platform.

Data reduction. For the purpose of this study, the descent phase was defined as the last 100 ms of the fall and the floor contact phase was the period from initial floor contact until maximum knee flexion. The descent phase covered the last half of the fall (approximately 30 cm). The starting point for film digitization was 18 frames before first contact of the force platform with the foot. This phase included the last 100 ms of the descent plus eight extra frames. Alternate frames were digitized in this phase since the acceleration was constant and the angular velocity of the body segments was low. Digitization continued for every frame of the impact phase until four frames after the frame of maximum knee flexion. The extra digitized frames before and after the actual trial were included to improve accuracy of the data at the performance boundaries (28).

The six body points and the front corner of the force platform were digitized in each frame. The platform point was used both as a reference point and to locate the center of pressure in the kinematic reference frame. The data from the film records were smoothed using an interactive cubic spline routine and were interpolated to produce 20 and 280 frames of data for the descent and impact phases, respectively.

The GRF data were scaled and smoothed by using a second order low pass digital filter with cutoff frequencies between 60 and 80 Hz. The smoothed GRF data were then interpolated to 280 points so that each body position had a corresponding applied GRF during ground contact. The accuracy of the interpolation routine was tested to assure that the error introduced by the interpolation was minimal. Maximum and average force values from the interpolated curves were within 1% of the original data. Positive values indicate a vertical GRF acting upward on the body or a horizontal GRF acting to accelerate the body forward. The point of application of the GRFs on the body was calculated as the center of pressure.

The location and magnitude of the lower extremity segmental masses and their moments of inertia were estimated using a mathematical model (13), the average segmental masses reported by Winter (27), and the individual subject's anthropometric data. Joint reaction forces and net joint moments of force were calculated for the lower extremity by using an inverse dynamic analysis that combined the anthropometric, film, and GRF data. Joint moments of force that were extensor at the hip and knee, and plantarflexor at the ankle, were assigned the positive direction.

Joint angular position and velocity were calculated from the kinematic data. Zero degrees at the three joints corresponded to an erect, standing position with the trunk, thigh and leg in a straight line, and the foot at a right angle to the leg. Positive values were assigned for extension at the hip and knee and plantarflexion at the ankle.

The joint muscle power was calculated as the product of the joint moment and joint angular velocity. Positive power phases indicate concentric muscle contractions and work being performed by the muscles on the skeletal system, resulting in positive accelerations of body segments. Negative phases indicate eccentric contractions and work being performed by the skeletal system on the muscles and producing negative accelerations.

The moment and power curves were evaluated with sets of five and three parameters, respectively. The angular impulse and work parameters were assessed by calculating the areas under the respective curves. All kinetic and energetic parameters were normalized for body mass.

Statistical analysis. Mean parameter values across trials were calculated for each subject/condition. The resulting subject means for the two conditions were subjected to a one-tailed correlated *t*-test for directional hypotheses (1) since it was expected that the stiff landing would produce greater parameter values.

RESULTS

Kinematics. Kinematic descriptors of the descent and floor contact phases are presented in Figures 1 and 2, which show representative trials. The stick figures are shown to enable the reader to better visualize and compare the movements. The joint position results indicated that selective adjustments in the descent phase kinematics were made depending on the stiffness of the landing. The subjects flexed more at each joint in preparation for the soft landing. The floor contact phase began in a more flexed body configuration in the soft landing with the hip and knee joints flexed about 9 degrees more and the ankle plantarflexed about 5 degrees less compared with the stiff landing.

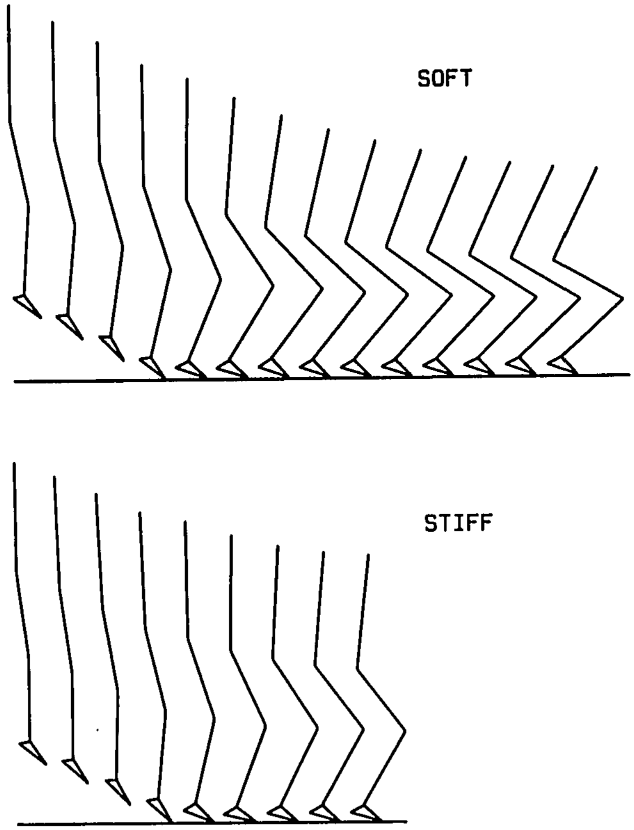

Figure 1—Stick figure representations of typical soft and stiff landings. The descent phase was the last 100 ms of free fall. The floor contact phase (first contact to maximum knee flexion) lasted 342 ms and reached a final knee position of 117 degrees of flexion in the soft landing. Corresponding stiff landing values were 152 ms and 77 degrees. The stiff landing had a more erect body posture throughout the landing.

The stiff landing was characterized by a more erect final position and, therefore, had a smaller range of motion at each joint compared with the soft landing. The mean final knee positions were −117 and −77 degrees for soft and stiff landings, respectively. The mean duration of each floor contact phase, which lasted from initial contact until maximum knee flexion, was 342 and 152 ms for the soft and stiff landings, respectively.

Descent phase kinetics and energetics. The mechanism by which the kinematic adjustments were made was investigated by examining the joint moments and

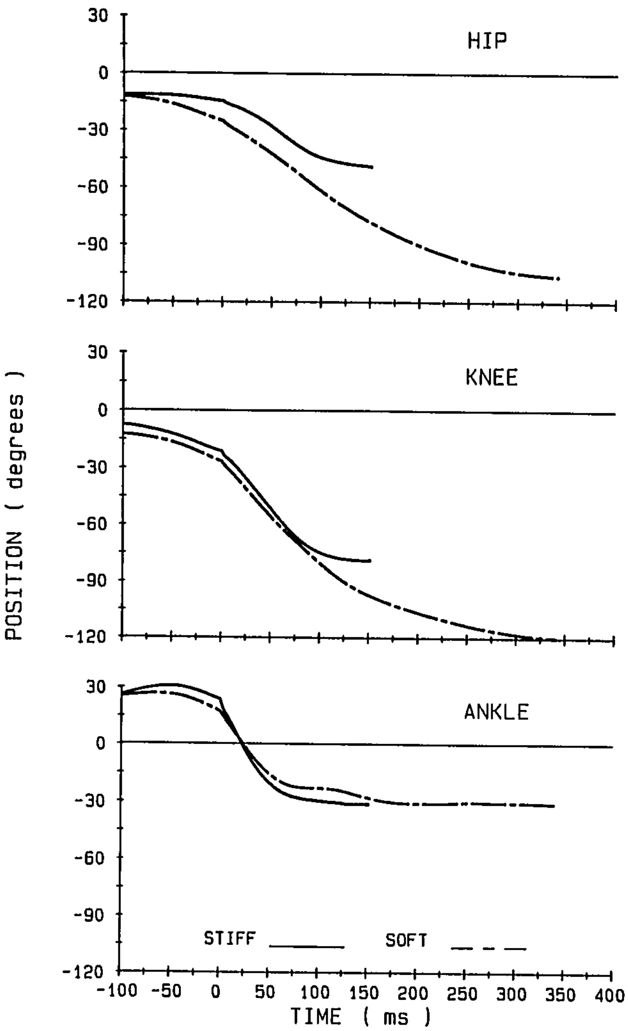

Figure 2—Joint position curves from representative soft and stiff landings. Positive values are extended or plantarflexed positions. Negative, positive times indicate descent and floor contact phases. Subjects were flexed more at all joints during descent and floor contact phases and used a larger range of motion in soft landings.

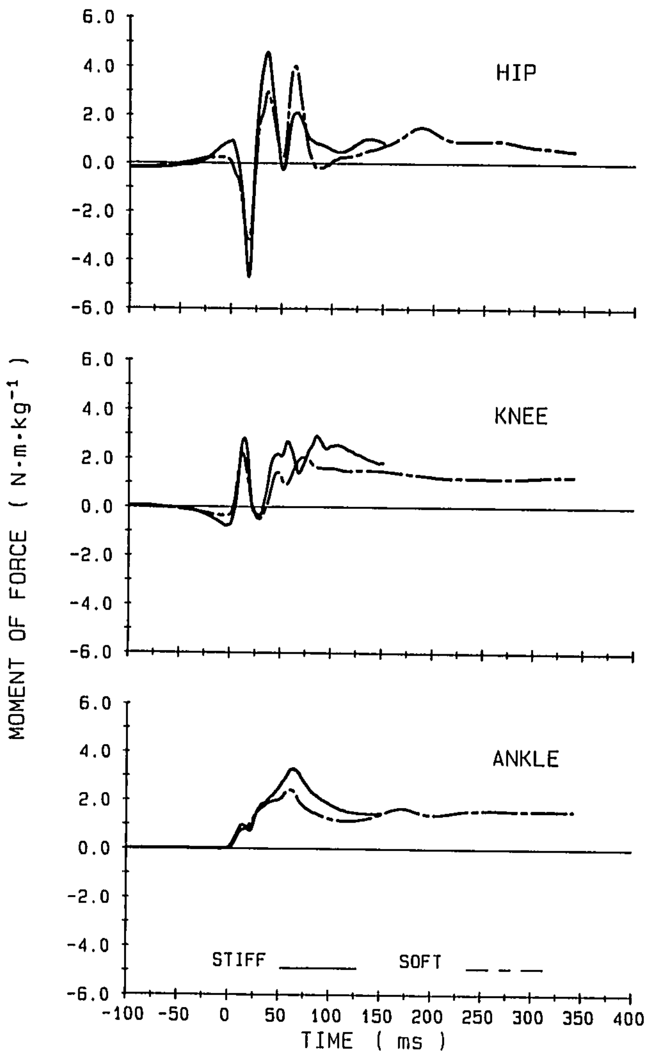

Figure 3—Joint moment of force curves from representative soft and stiff landings. Positive values are extensor or plantarflexor moments. Negative, positive times indicate descent and floor contact phases. Stiff landings had larger hip extensor and knee flexor moments during descent, which produced a more erect body posture at impact. Angular impulse values were similar between conditions at the hip and knee and were larger at the ankle in the stiff landing. The hip flexor moment after floor contact (time 0) may have functioned to rotate the trunk forward, reducing the external flexing moment at the knee and the load of the knee extensors.

muscle powers in the lower extremity during the descent phase. Figures 3 and 4 show representative moment and power curves during the total performance for soft and stiff landings. The magnitude of the moment curves was much less during the descent compared with the floor contract phase. However, they were of sufficient magnitude to rotate the segments since the external gravitational moments and the segmental moments of inertia were also very low. Table 1 lists mean parameter values that were selected to describe and compare these curves between conditions.

The response in the joints moments during the descent phase to the experimental conditions was very consistent across subjects. The moments about the hip were similar in soft and stiff landings and had flexor dominant phases followed by extensor dominance until floor contact. The extensor angular impulse was 93% larger ($P < 0.01$) in the stiff landing than in the soft landing. The muscle power curves at the hip were negative prior to floor contact and indicated the extensor moment worked eccentrically to reduce hip flexion velocity, especially in the stiff landing. The knee moments were initially negligible and then showed flexor dominance for both landings. The stiff landing had a 30% larger ($P < 0.01$) amount of flexor angular impulse during the descent phase. The flexor moments worked concentrically (see muscle power curves) to increase knee flexion prior to floor contact.

Floor contact phase kinetics and energetics. Soft and stiff landing vertical and horizontal GRF curves from representative trials are shown in Figure 5. Both landings had bimodal vertical GRF curves with maxi-

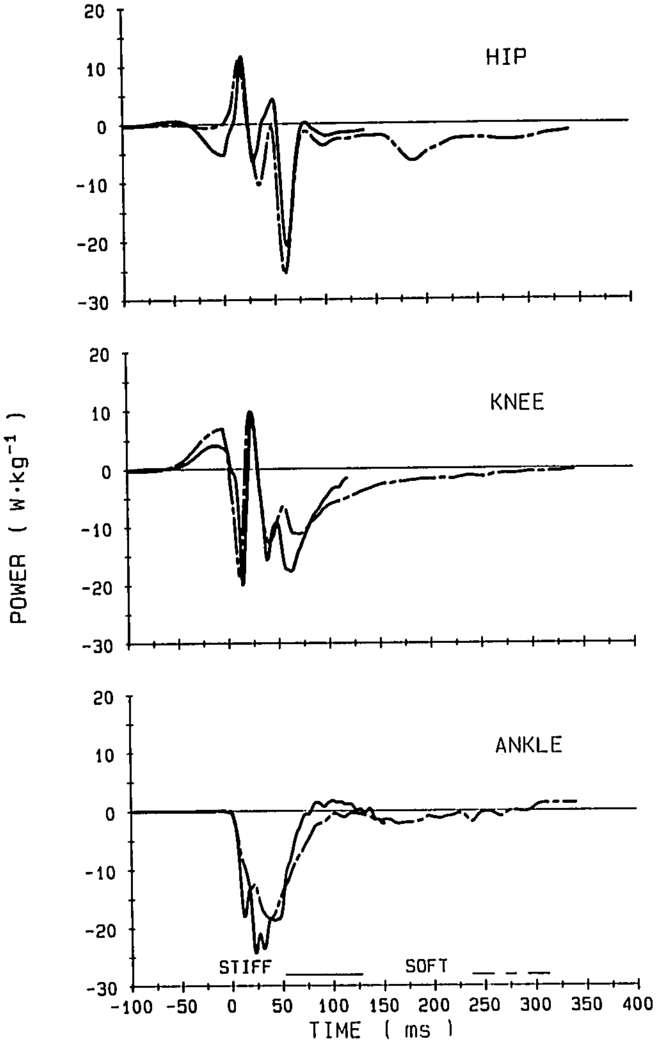

Figure 4—Joint muscle power curves from representative soft and stiff landings. Positive and negative values indicate energy generation (concentric muscle action) and energy absorption (eccentric muscle action). Negative, positive times indicate descent and floor contact phases. The total work was negative at each joint with the hip and knee muscles performing more work in soft landings and the ankle doing more in the stiff landing. The summated work across joints showed the muscles absorbed more kinetic energy in soft (−2.37 J·kg⁻¹) compared with stiff (−2.00 J·kg⁻¹) landings, placing greater loads on other tissues in the stiff landing.

TABLE 1. Mean parameter values across all subjects.

	Soft	Stiff
Descent phase moments		
Hip extensor impulse	0.010 (0.005)	0.019** (0.005)
Knee flexor impulse	−0.010 (0.002)	−0.013** (0.003)
Impact phase moments		
Hip extensor impulse	0.096 (0.021)	0.110 (0.034)
Knee extensor impulse	0.146 (0.027)	0.141 (0.044)
Ankle plantarflexor impulse	0.185 (0.029)	0.232* (0.050)
Impact phase powers		
Work at hip	−0.600 (0.180)	−0.390** (0.140)
Work at knee	−0.890 (0.170)	−0.610** (0.180)
Work at ankle	−0.880 (0.080)	−1.000* (0.140)

Mean, SD values are calculated from mean values over trials for each subject; standard deviations are in ( ); impulse values are in $N \cdot m \cdot s \cdot kg^{-1}$; work values are in $J \cdot kg^{-1}$. ** $P < 0.01$; * $P < 0.05$.

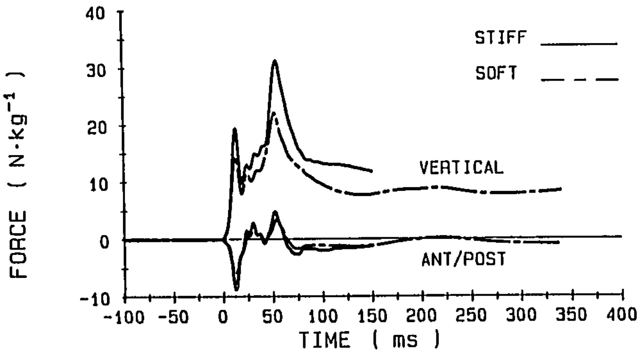

Figure 5—Ground reaction force curves from representative soft and stiff landings. Stiff landing had a 23% larger linear impulse compared with soft landing. Temporal features were similar between landings in both vertical and horizontal GRFs. Impact phase durations (period of large force application) were 125 and 85 ms for these soft and stiff trials and averaged 126 and 113 ms for soft and stiff landing trials.

mum values occurring about 12 and 50 ms after contact. The horizontal GRF placed a significant load on the subjects and also reached maximum posterior and anterior values at about 12 and 50 ms after floor contact.

Although the floor contact phases lasted 342 and 152 ms, as noted above, the period of relatively large force application occurred in a shorter time, especially in the soft landing. This period will be referred to as the impact phase and was defined as the time between initial floor contact and the moment when the vertical GRF reached either a minimum value or a relatively stable value after the second peak force. As can be seen in the vertical GRF curves, the impact phases for these trials ended about 130 ms (minimum value) and 88 ms (relatively stable value) after contact for soft and stiff landings, respectively. The mean impact phase durations for all subjects were 126 and 113 ms for soft and stiff landings. The vertical GRFs and extensor moments that were observed after this time functioned primarily to support the body center of mass as opposed to reducing its downward velocity. All parameter variables for the moment and power curves were only applied to the impact phase.

Impact phases for both conditions were initiated by simultaneous knee extensor and hip flexor moments, which reached maximum values about 15 ms after contact and ended about 25 ms after contact. Work was performed on the knee extensors, as indicated by the negative (absorption) knee muscle power. This action reduced the downward velocity of the body center of mass. In contrast, the hip flexors performed work on the skeletal system during the initial part of the impact phase, accelerating the trunk forward and downward. The remainder of the impact phase had extensor moments at each joint, all of which performed eccentrically and reduced the subjects' downward velocity.

Mean parameter values describing the moment and power curves during the impact phase are listed in Table 1. The angular impulse about the ankle was 25% greater ($P < 0.05$) in the stiff landing condition, whereas the hip and knee moments were very similar between conditions. All muscle power parameters were significantly different between conditions. The amounts of work performed at the hip and knee were 54% larger ($P < 0.01$) and 46% larger ($P < 0.01$), respectively, in the soft landing, indicating that the muscles crossing these joints absorbed more energy in this condition compared with the stiff landing. In contrast to these results, the work performed at the ankle was 14% larger ($P < 0.05$) in the stiff landing, indicating that the ankle plantarflexors absorbed more energy in the stiff landing compared with the soft landing.

DISCUSSION

The purpose of this study was to identify and compare ground reaction forces, and joint positions, moments, and muscle powers in the lower extremity during the descent (free fall) and floor contact phases of soft and stiff landings. Soft and stiff landings had relatively large and small amounts of knee flexion, respectively, during the floor contact phase. The action of landing from a vertical fall applied forces and moments to the lower extremities that accelerated hip and knee flexion and ankle dorsiflexion, thus causing the extremities to collapse. The goal of a successful landing was to resist this collapse by applying counter extensor moments at these joints in such a way that the body's negative velocity was reduced to zero without injury. These extensor moments primarily worked eccentrically to absorb kinetic energy from the skeletal system and stop the person's fall.

Descent phase. The subjects prepared themselves for either soft of stiff landing impacts by producing particular moment patterns during the descent phase. The flexor moment at the hip functioned to rotate the trunk and thigh forward (hip flexion) from vertical in preparation for landing, reducing the impact stress on the spine. The extensor moment at the hip in late descent worked eccentrically to reduce hip flexion velocity. The larger extensor moment in the stiff landing caused a greater reduction in hip flexion velocity and produced a more erect body configuration at impact. The stiff landing was not accomplished by simply increasing the extensor moments at each joint during the impact phase since only the angular impulse about the ankle increased from soft to stiff landings. A primary factor in successfully performing the stiff landing was to initiate the impact phase with a more erect body posture. This posture reduced the moment arms of the external forces, which accelerated joint flexion, especially at the hip and knee. The reduced moment arms partially counteracted the increase in external forces such that the stiff landing was performed with only a 13% increase in angular impulse over all joints even though the GRF impulse increased 23%. This result identifies the critical role of the extensor moment at the hip in late descent in modifying landing stiffness and the magnitude of the impact forces by adjusting the body posture prior to impact.

The observed flexor moment at the knee prior to contact may have been a secondary outcome of the extensor moment at the hip, owing to force development in the hamstrings muscle group. Miyatsu et al. (15) identified EMG activity in the medial hamstrings during the descent phase in landing, which supports this contention. The knee flexor moment was necessary, however, to ensure that the knee would be flexed upon floor contact since the hip extensor moment produced a posterior force on the thigh, which in turn produced a posterior knee joint reaction force (mean impulse: -0.020 and -0.036 $N \cdot m \cdot s \cdot kg^{-1}$ in soft and stiff landings). This reaction forced created an extensor torque around the leg center of mass (mean angular impulse: 0.005 and 0.008 $N \cdot s \cdot kg^{-1}$ in soft and stiff landings) that, if unchecked, would cause knee extension and put the subject into a potentially injurious landing position. Force production in the hamstrings to create the extensor moment at the hip would also produce a force couple acting on the leg. The force couple would then provide a knee flexor moment (mean angular impulse: -0.010 and -0.013 $N \cdot m \cdot s \cdot kg^{-1}$ for soft and stiff landings) to counteract the thigh induced extensor moment and cause knee flexion during the descent phase.

The extensor moment at the hip and the flexor moment at the knee were the principle preset joint moments used to prepare the subjects for soft or stiff landing impacts. This result confirms Lees' (14) statement that impact GRFs are modified by muscular activity during the descent phase. Additionally, these results show that stiffer landings are not performed by simply increasing the muscle tensions on both sides of the joints in the lower extremities but by changing the relative tensions across the joints to produce changes in the net joint moments.

Floor contact phase. The observed landing vertical GRF curves were similar to those presented previously (11,12,26), even though in those studies the subjects jumped up from the floor and did not step off a raised platform. The temporal features of the vertical GRF curves in each of these studies were similar to the present results and showed that the major impact phase ended in about 90 ms and, during this time, two vertical force peaks were observed at approximately 15 and 53 ms. Also, the increased knee flexion and lower vertical GRFs presently observed for the soft landings are in agreement with previously reported results (9,24). One difference between the present results and those in the literature was that the maximum vertical GRF peak magnitudes were about 25% less than in previous studies (11,12,26). This difference may have been due to differences in jumping/landing height, the relative stiffness of the landings, or the footwear characteristics.

The temporal relationships of impact phase events were nearly identical between landing conditions for all subjects, which is in agreement with Ozguven and Berme (19). Based upon the observed moment and power curves, it was apparent that a general movement strategy was used to perform both soft and stiff landings from a moderately high fall of 59 cm. Upon contact with the floor, the subjects first controlled knee flexion and ankle plantarflexion while actively flexing the hip, rotating the trunk forward and downward. Subsequently, hip flexion was arrested, stopping the trunk's motion while knee flexion and ankle plantarflexion continued to be reduced and then stopped. Mizrahi and Susak (16) proposed that all segments should be accelerated simultaneously to reduce peak impact forces. The present results showing velocity reductions at the knee and ankle followed by a reduction at the hip indicated that subjects skilled in landing do not perform as these researchers suggested. The functional role of the flexor moment at the hip may have been to accelerate the trunk forward so that the mass of the upper body was closer horizontally to the knee joint. This action would reduce the moment arm between the weight of the trunk and the knee and thus reduce this external flexing moment at the knee and the load on the knee extensors. Additionally, an extensor moment at the hip at this time might be partially produced by the hamstrings, which would reduce the initial net extensor moment at the knee and interfere with the landing performance.

Fukuda (10) and Bobbert et al. (3,4) have reported joint moment and muscle power patterns in the lower extremity during landing. Fukuda's (10) results differ markedly from the present moment and power curves in two ways. His moment curves at the hip and knee were extensor dominant throughout the landing, and the magnitude of the moment and power curves were much greater than those presently observed. This last difference was especially surprising since the falling height was only 20 cm in Fukuda's study. Fukuda (10) estimated the external landing forces using only kinematic data, and this method may have produced erroneously large values. We recalculated several moment curves using artificially increased GRFs and found that the moments were larger in magnitude and shifted toward extensor dominance.

The moment curves from 60 cm falls in Bobbert et al. (4) had initial phases of hip flexor and knee extensor dominance lasting about 25 ms, which was similar to the present results. The absence of a subsequent flexor moment about the knee in Bobbert et al.'s study may be due to the fact that the landing was followed by a jump, whereas the present landing was not. The present flexor moment at the knee (at 25 ms) may have been related to the large extensor moment at the hip and tension produced in the hamstrings to reduce hip flexion velocity. The maximum moment at the knee and maximum power at the hip and knee during the landing phase in Bobbert et al. (4) were similar to the values presently observed.

The highest muscle moment and power values were observed at the hip joint in both landing techniques. This result was probably due to a combination of two factors. First, the trunk was positively accelerated by the initial flexor moment at the hip and therefore required greater effort to negatively accelerate. Second, the hip muscles must control the angular kinematics of the entire upper body (head-arms-trunk), which contains 68% of the body mass (27). In contrast, the knee and ankle joint muscles assist each other and are assisted by the hip muscles in controlling segmental rotations in the lower extremity. This function was evident in the fact that the knee angular impulse actually decreased from soft to stiff landings even though the subjects were instructed to reduce knee flexion in the stiff landing. Instead, the angular impulse sample means at the hip and ankle were increased, providing greater knee extension support by limiting thigh and leg rotation.

The ratio of the muscular work parameter values at each joint to the summated work values across the three joints can be used to identify the relative contributions of each muscle group to the landing performances. The summated muscular work values were -2.37 and -2.00 $J \cdot kg^{-1}$ for soft and stiff landings. The relative joint contributions to these totals were similar between conditions and were 25, 37, and 37% in soft and 20, 31, and 50% in stiff landings for the hip, knee, and ankle, respectively. These results indicated the ankle plantarflexors and knee extensors were the muscle groups primarily responsible for reducing the body's kinetic energy, followed by the hip extensors. Also, as landing stiffness increased, the relative contribution of the ankle plantarflexors increased while those of the hip and knee extensors decreased. Although the hip had the largest extensor moment values, much of the work done by these muscles was used to counteract the accelerated trunk motion owing to the early flexor moment at this joint. The net contribution of the hip muscles to the reduction of the body's kinetic energy was therefore reduced.

The summated work values indicated the muscular system absorbed 19% more kinetic energy in the soft landing. Specifically, although the ankle plantarflexors absorbed 12% less energy in the soft landing, the hip and knee extensors each absorbed about 50% more energy in this condition, more than counteracting the reduced work at the ankle. Since each subject had the same amount of kinetic energy between conditions and the muscles dissipated more in the soft landing, the remaining tissues, most notably the skeletal system, absorbed more energy and were exposed to greater stress in the stiff landing condition. This result was in agreement with Zatsiorsky and Prilutsky (29), who showed a greater muscular contribution to energy dissipation in softer landings.

This study showed how the kinetic energy acquired during a vertical fall was dissipated through a complex series of extensor and flexor moments in the lower extremity, which worked both eccentrically and concentrically during the floor contact phase. The importance of presetting the joint moments during the descent phase for the successful performance of either soft or stiff landings was also identified. Specifically, the stiffer landing was accomplished by increasing the net extensor moment at the hip and flexor moment at the knee prior to floor contact to produce a more erect body posture and a flexed knee position at initial floor contact. Finally, the relative muscular contribution to energy dissipation was identified and showed an increased contribution to soft landings compared with stiff landings and an increase in the contribution of the ankle plantarflexors as landing stiffness increased.

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