Analysis of parameters affecting impact force attenuation during landing in human vertical free fall

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The characteristics of impact forces on the legs during vertical landing of human vertical free fall in different falling conditions were studied to reveal the parameters which take part in the attenuation of these impact forces. The following parameters were investigated: body position during landing, range of flexion of the joints of the legs at impact, usage of ground-roll immediately after impact and softness of the ground.

The results indicate that joint movements and muscle action play a major role in reducing peak forces during landing. This emphasizes the importance of adequate training to improve the pre-programmed non-reflex muscle action, necessary in the early phase of impact.

Introduction

Exposure of the lower limbs to unusually high forces, as occurs during landing from free fall may under certain conditions lead to one of the two following consequences:

1. Occurrence of fractures in bones at sites and frequency as reported by Orava *et al* (1978), or rupture of tendons at their insertions, as reported by Welsh (1979). These are caused by disproportion between load and strength of the musculo-skeletal tissues, which do not have enough time to adapt to increasing mechanical stresses.
2. If bones have enough time to adapt to increasing loads, conditions for joint degeneration may present. A mechanism for that has been proposed by Radin *et al* (1972), according to which impulsive loading leads to trabecular micro-fracture. The process of bone remodelling results in stiffening of cancellous bone, which becomes a poorer shock absorber. The outcome is exposure of articular cartilage to increased stress, followed by cartilage breakdown and joint degeneration.

Both consequences listed above can be avoided by decreasing the impulsive forces transmitted to the body, which can be done by adequate shock absorption. According to Radin (1974), there exist two types of mechanisms of shock absorption: one passive and one active.

In the passive mechanism, shock attenuation is achieved by bone and soft tissues. Radin *et al* (1970) suggested that trabecular bone has a major role in this mechanism, explaining that attenuation of dynamic peak forces is a function of ability to deflect under these forces. As compared to synovial fluid and articular cartilage, bone deforms a little, but because it exists in much larger amounts, bone acts altogether in a more meaningful way. Fracture of the trabeculae is the most effective way to absorb energy. Moreover, as has been shown by Ducheyne *et al* (1977), the process of fracture of the trabeculae is localized and gradual, allowing only minimal rupture during shock loading, which is restored in the living bone unless multiple overloading occurs, in which case progressive collapse of the trabecular structure may occur.

The active mechanism is far more significant than the passive one and it essentially relies on lengthening of muscles under tension, accompanied by joint motion. This mechanism can be easily demonstrated in walking on the heels, with the ankles and knees extended and the muscles tight, causing a jolt in each step, mainly because the active mechanism cannot function. A more extreme example was given by Smith (1953), who showed that a free fall of an 80 kg man from one metre height with a 'knee-locked situation' is likely to produce severe damage of the head and the neck of the femur or will push the head of the femur through the acetabulum. It follows then that the main part of kinetic energy acquired by the body is dissipated by the muscles. It should however be remembered that the active mechanisms are under control of the reflexive neuro-muscular system and that there is a measurable time for this reflexive action. Radin (1974) mentioned a time of 75 ms, while McMahon and Greene (1979) quoted a time of 100 ms required for reflex activity from the otoliths to activate antigravity muscles in man, concluding that unexpected falls of less than about 5 cm are normally unaccompanied by reflex accommodation. In expected falls, on the other hand, it is possible to 'pre-program' the muscle action and joint motion, even when the reflex system cannot participate. In this case preparation of the subject is necessary, increasing the importance of training and experience. This point was considered by McMahon and Greene (1979), who divided the supported period of one leg in human running into two parts: the first quarter, in which non-reflex control of the limb position is preprogrammed and is modelled as a rack and pinion; and the remaining three-quarters, in which local control of the muscle stiffness exists, and is modelled as a damped spring.

Ideally, both shock-absorbing mechanisms complement each other. When deformation starts in the passive mechanisms, then according to Finlay and Repo (1979) a neurological feedback system senses the resulting increased force and so brings muscles into play before the forces have time to reach destructive levels. It should, however, be kept in mind that reflex mechanisms can be fatigued, as may happen with individuals performing repetitive tasks (Radin, 1974).

The ability to pre-program muscle action and joint motion therefore has major importance when reflex activity has not yet appeared. To make use of this ability, the parameters related to it should be investigated. In this work we have studied the characteristics of impact

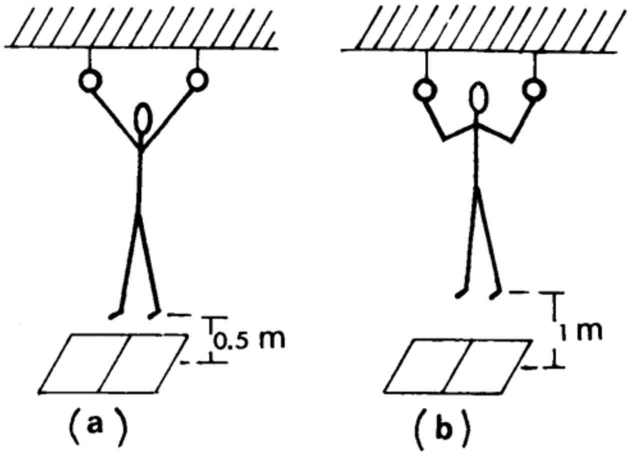

Fig. 1. Schematic illustration of subject at hang before free fall
(a) arms straight for lower height (0.5 m)
(b) arms flexed at elbow for higher height (1 m)

forces on the legs during vertical landing in different falling conditions, to reveal the parameters that take part in the attenuation of these impact forces. The following parameters were investigated: body position during landing, range of joint flexion at impact, usage of ground-roll immediately after impact (sometimes done by parachutists) and softness of the ground. The contribution of these parameters in reducing damage to the body during landing is considered.

Materials and methods

Five right-footed instructors of physical training aged twenty to twenty-two years, two females and three males participated in the experiment. Each subject performed a number of free falls from two different heights, from a suspended position on rings hung from the ceiling, as shown schematically in Fig. 1. The lower height, corresponding to 0.5 m was from suspension with the arms straight (Fig. 1a), and the higher one, approximately 1 m, with the arms flexed (Fig. 1b). From the lower height, three landings were on the balls of the feet and three with the feet flat. At least four jumps were from the greater height, with all landings done on the balls of the feet, two of which however were followed by a lateral ground-roll (the latter performed by three of the subjects). Soft landing was done by two subjects by using 5 cm thickness foam rubber sheets placed on ground.

The forces on the feet during landing were measured by two Kistler multi-component piezoelectric platforms types Z4305 mounted co-laterally on a concrete foundation, with their top plates levelled with the floor. The vertical force signals were channelled through a Kistler amplifying unit type 9803sp to a multi-channel Bell and Howell UV recorder. During recording the paper speed in the recorder was set at 20 cm/s for the lesser height jumps and at 40 cm/s for the greater height jumps. The output of the force acting on the left foot of each subject was also connected to a Tektronix 5115 storage oscilloscope, and displayed on its screen in all experiments. The vertical force traces obtained from the force plates were digitized by a GP-3 digitizer (SAC, Southport, Conn.) and the data fed through a GP-3 interface model 1358 to a Data General Nova minicomputer. Considering that impact with the ground normally lasted less than 0.5 s, after which the forces between ground and feet stabilized, digitization was performed over the standard period of 0.7 s from the onset of each trace. With an interval time of 3 ms, this gave a total number of 250 points for each trace. The numerical analysis included time-integration of the forces, yielding the impulse on each foot during impact, total impulse on the body and impulsing symmetry between both feet. The total impulse provided a measure to characterize and compare falls in different conditions. Timings and magnitudes of the peaks in the force curves were directly measured from the traces.

Kinematics of the motion was studied by 50 frames/s video tape recording. The system used was an IVC-800 video tape recorder, with slow motion and stop motion capability, to which three Sony TV cameras model AV were connected through a Sony special effect generator (SEGII). One camera was placed in the left lateral side of the subject from which the body motion was recorded from its initial rest state at hang position to its final rest state after landing. Markers attached over the left ankle (external maleolus), knee (lateral condyle), hips (greater trochanter) and iliac crest were used for this purpose. The second camera was focused on the trace of the left foot force during impact, which was displayed on the scope screen. Superposition of the pictures from both these cameras on the monitor's screen was achieved by mixing them with the SEG. In this way time synchronization between the measured force on the foot and the position of the body segments during impact was provided, as demonstrated in Fig. 2. The third camera was zoomed on the left foot at impact and its image inserted in the top right-hand corner of the monitor's screen by using the corner insert facility of the SEG. An electronic timer incorporated in the recording system and displayed on the upper part of the screen monitor served for frame identification and reference.

From frame-by-frame replay of the recordings, the following variables were extracted (see Fig. 3): angles of the hip, knee and ankle joints of the left leg as well as the position of a marker attached on the left iliac crest. The corresponding force values were taken from the recorder traces after visual comparison with the force trace obtained in each frame.

Results

Typical curves corresponding to landing from 0.5 m on the balls of the feet are shown in Fig. 4. The following curves are presented on this figure: vertical force on the left foot, angles of ankle, knee and hip (as defined in Fig. 3) of the left side of the body and the height of the marker attached above the iliac crest (which is close to the centre of gravity of the body). The onset of the force curve is followed by two peaks of special importance due to their high intensity. More peaks appear at a later stage of the curve, but their intensities are much lower than these of the first two. The force intensity in all landings will be characterized by the first two peaks of the force curve. The curves of the angles of the joints reflect the amount of flexion of these joints and they all show stabilization of the angles after impact. The iliac crest marker curve shows the trajectory of a point on the body, close to the centre of gravity.

The effect of landing with the feet flat is demonstrated in Fig. 5. This landing can be compared with landing on

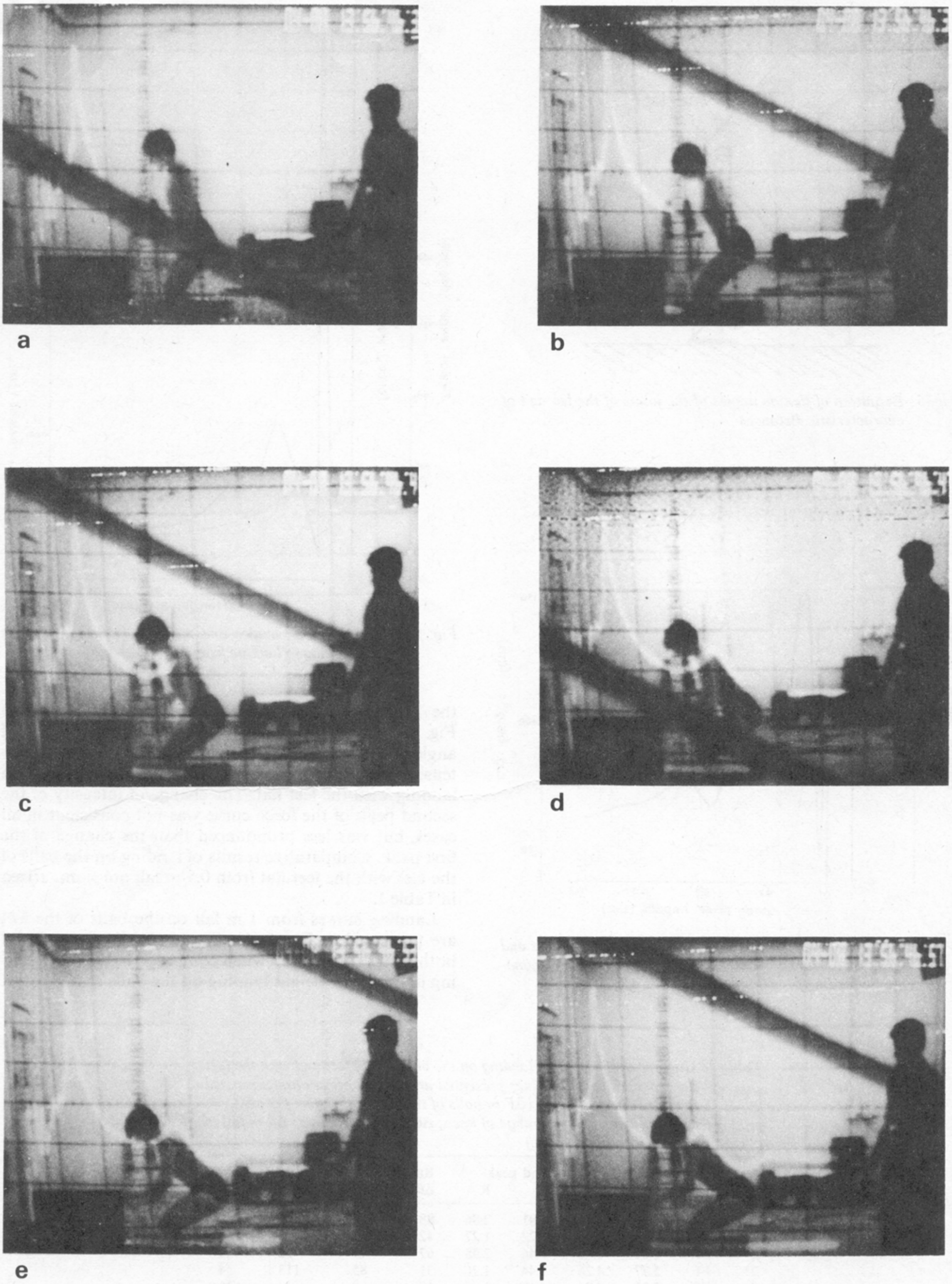

Fig. 2. Frame-by-frame demonstration of superposition of force curve and corresponding body image to achieve synchronization between force and flexion angle of the joints (the corner insert of left foot and ankle is not shown in this series)

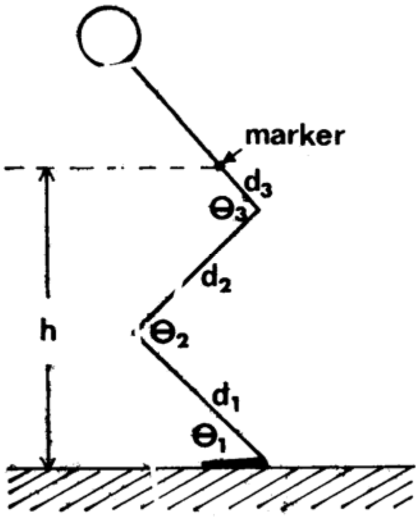

Fig. 3. Definition of flexion angles of the joints of the leg and of characteristic distances

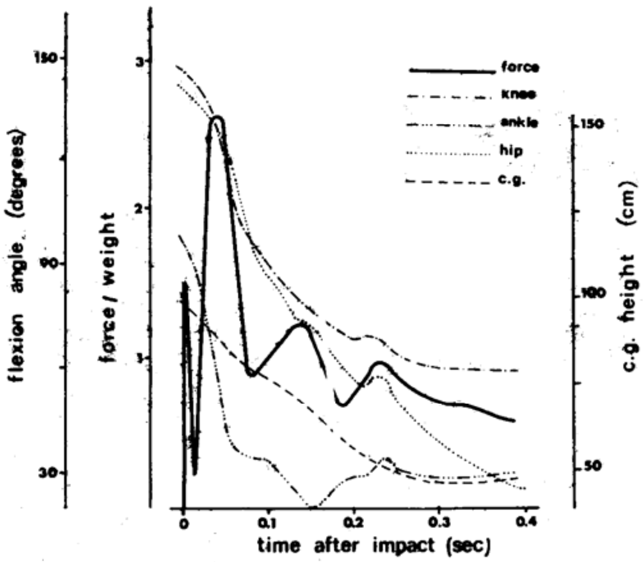

Fig. 4. Typical curves of force angles of flexion of the joints and position of marker (near c.g.) for the left side, corresponding to landing from 0.5 m on the balls of the feet

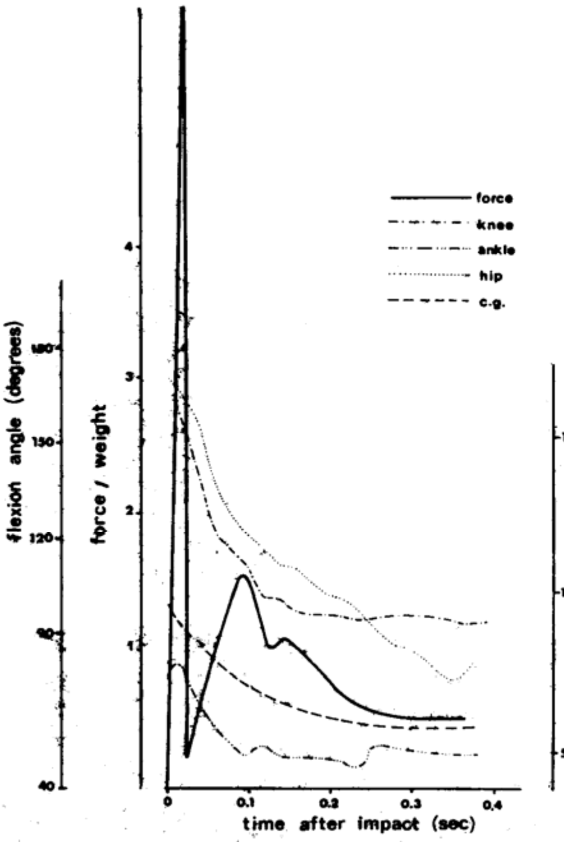

Fig. 5. Curves of force, angles and marker position (near c.g.) corresponding to landing from 0.5 m with the feet flat

the balls of the feet from the same height, shown in Fig. 4. Apart from the obvious difference in the initial angle of the ankle joint, a significant increase in the intensity of the first peak of the force curve appears when landing with the feet flat. The change of intensity of the second peak of the force curve was not consistent in all cases, but was less pronounced than the change of the first peak. Comparative results of landing on the balls of the feet with the feet flat from 0.5 m fall are summarized in Table 1.

Landing curves from 1 m fall on the balls of the feet are presented in Fig. 6. An increase in the intensity of both peaks can be seen when compared to those appearing in the lower height landing on the balls of the feet. It

Table 1. Comparative results of landing on the balls of the feet and with the feet flat from 0.5 m fall. Values presented are means. Forces are expressed in units of body weight. (BF = balls of the feet; FF = feet flat; $\Delta\theta_1$ = range of ankle, $\Delta\theta_2$ = range of knee; $\Delta\theta_3$ = range of hip; Δh = fall of marker located near c.g.)

Subject		1st peak L	1st peak R	2nd peak L	2nd peak R	Range of joint angles $\Delta\theta_1$	$\Delta\theta_2$	$\Delta\theta_3$	Δh, cm
1	BF	1.69	2.89	1.91	2.86	88	68	103	42
1*	FF	3.68	4.23	1.22	1.27	42	72	107	35
2	BF	1.67	1.70	3.46	2.98	67	79	98	33
2*	FF	5.77	4.26	1.44	1.26	31	85	113	34
3	BF	2.22	2.75	4.17	1.75	60	80	30	36
3*	FF	5.28	1.72	1.25	5.30	40	75	70	35
4	BF	2.15	2.70	1.69	2.09	60	80	30	35
4*	FF	6.15	6.18	1.54	1.87	45	65	45	10

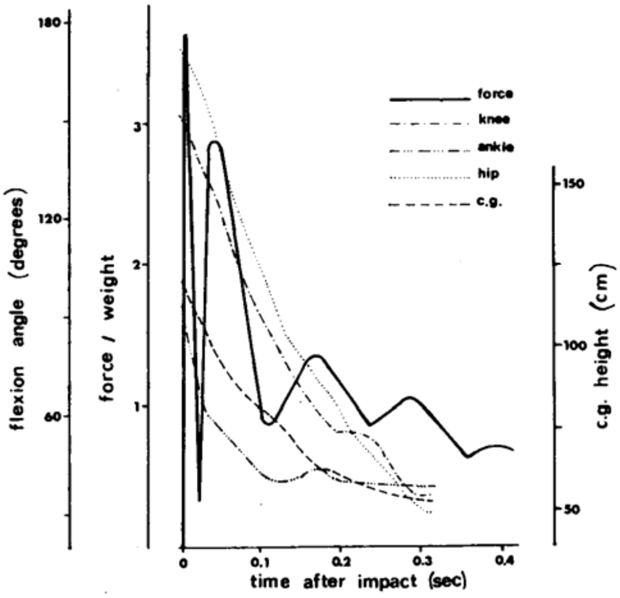

Fig. 6. Landing curves for 1 m fall on the balls of the feet

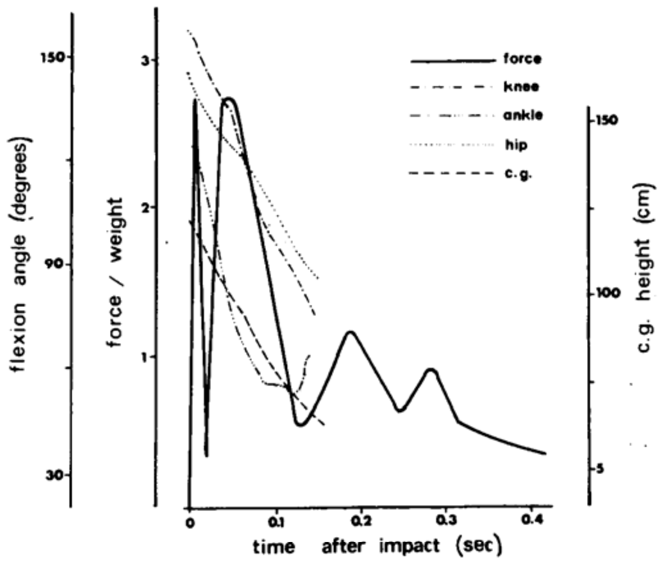

Fig. 7. Landing curves for 1 m fall with the feet flat and with lateral ground-roll succeeding impact

is interesting, however, to note that the first peak of a 1.0 m fall on the balls of the feet was smaller than the peaks of a 0.5 m fall with the feet flat. The range of joint flexion was generally bigger from the higher jump as compared to the lower one. A summary of the results obtained for landing from 1 m is presented in Table 2.

The effect of lateral ground-roll succeeding impact with the ground with the feet flat is demonstrated in Fig. 7. Both peak forces were decreased in the cases where ground-roll was performed. A summary of the results obtained in this case is presented in Table 3. The range of joint angles was limited to the position where the ground-roll started, since from that position onwards the measuring error became significant.

Table 4 summarizes the results obtained for soft landing, with foam-rubber sheets placed on the floor. The values of both peak forces decreased considerably demonstrating the damping effect of cushioning.

Numerical integrations of the force curves corresponding to landings from the same heights of each subject yielded values with maximum difference of 4 per cent. Noting that part of the impulse differences can be attributed to slight differences in body positions, it can be assumed that the dynamic error was less than 4 per cent.

Finally, the error of angle measurement was estimated by comparing the measured height of the iliac crest marker with the calculated height from the following relation (see Fig. 3):

$$h = d_1 \sin \theta_1 + d_2 \sin (\theta_2 - \theta_1) + d_3 \sin (\theta_3 - \theta_2 + \theta_1)$$

in which d_1, d_2 and d_3 are measurable distances for each subject. A maximum deviation of 2.5 per cent was ob-

Table 2. Summary of results obtained for landing from 1 m on the balls of the feet. Values presented are means, forces are expressed in units of body weight

Subject	1st peak L	1st peak R	2nd peak L	2nd peak R	Range of joint angles $\Delta\theta_1$	$\Delta\theta_2$	$\Delta\theta_3$	Δh, cm
1	5.60	4.70	3.88	4.51	85	93	141	78
2	3.67	2.98	2.83	3.07	83	110	134	72
3	4.31	4.08	2.92	2.69	75	80	80	65
4	3.69	4.75	4.62	5.55	70	65	80	65
5	3.73	3.28	3.13	3.98	60	110	135	80

Table 3. Summary of results for lateral ground-roll succeeding impact with the ground. Values presented are means, forces are expressed in units of body weight

Subject	Height	1st peak L	1st peak R	2nd peak L	2nd peak R	Range of joint angles $\Delta\theta_1$	$\Delta\theta_2$	$\Delta\theta_3$
1	0.5	2.22	3.52	2.10	2.2	60	80	90
1	1	4.56	4.49	3.63	3.69	78	101	94
2	0.5	1.37	1.57	1.98	2.19	50	95	70
2	1	3.67	2.98	2.83	3.07	83	110	134
3	1	2.23	3.88	2.18	2.43	75	83	67

Table 4. Summary of results obtained for soft landing with foam-rubber sheets placed on the floor. Values presented are means, forces expressed in units of body weight

Subject	1st peak L	1st peak R	2nd peak L	2nd peak R
1	1.35	2.28	1.55	1.87
2	2.08	1.86	0.70	0.73

tained between the measured and calculated height of the marker.

Discussion

The results obtained in this study emphasize the role of joint motion and muscle action in reducing peak forces during landing. Correct action of the joints and muscles can achieve this effect even before the reflex activity has come into action. This required, however, adequate preparation of the subject. Greene and McMahon (1979) have mentioned two factors affecting the stiffness of the leg: knee flexion angle and magnitude of the force acting on the leg. Stiffness of the leg decreases with increasing knee flexion and is maximal in a fully extended knee; it also increases with increasing load on the leg. During landing impact, high forces are imposed on the leg, with the result that if a low stiffness of the leg is desired (to reduce impact forces) a relatively high initial flexion angle of the joints is required. This, however, would limit considerably the range of flexion in these joints afterwards, to effectively absorb the energy by the muscles spanning the joints. This latter point was quantitatively studied by Thys (1978), who evaluated the amount of elastic energy which could be stored and re-used in human hopping. His main conclusion was that the dissipated energy in muscles increased when the amplitudes of joint movements were bigger. Bosco and Komi (1979) also commented on utilization of stored elastic energy, stating that this depended on the shortness in latency between the stretch and shortening phases of the muscles. We are interested in dissipation of energy, therefore the bigger the range of joint angle, the better.

The conclusions reached at above for the knee joint can be extended to the hip and ankle joints. Initial extension in the ankle joint, as it appears during landing on the balls of the feet, allows furthermore to increase the flexion range of this joint during impact.

Simultaneous use of all joints is probably the basis for reducing peak forces at impact. In this case, however, proper co-ordination between the different joints is important. Luthanen and Komi (1978) showed that the efficiency of individual tasks decreases with increasing the task complexity, as may be presented in multi-joint action. The reason for this is related to timing in multi-joint motion which affects the movement synchronization of the various segments of the leg. If the acceleration maxima of the different segments are exactly in phase, the overall performance is at maximum. Luthanen and Komi (1978) suggested that the level of co-ordination depends on the training stage of the subject. Training should reduce the time difference between the segmental acceleration maxima to zero. It should however be remembered that repetitive performance of a task may fatigue the active mechanisms of the human body, as has been pointed out by Radin (1974).

The pre-programmed non-reflex muscle action during the early phase of impact has proved important in peak force attenuation. As mentioned earlier, this procedure can be trained to and controlled by the subject. Although this study has quantitatively shown the participation of each of the joints of the leg at impact, it did not provide information on the individual activity of the various muscles of the leg. For that purpose electromyographic (EMG) measurements should be incorporated and synchronized with the kinematics and dynamics already studied during landing impact.

Analysis of the landings, succeeded by lateral ground-rolls indicated a decrease in peak forces during impact. This effect is somewhat surprising since ground-roll usually started after the appearance of the first two peaks. It can, however, be interpreted in two ways. Firstly, it may be suggested that the very intention to perform the ground-roll serves to prepare the subject, therefore to pre-program his muscle activity and flexion of his joints. Secondly, the ground-roll actually increases the range of motion of the joints and body, which continues to move during the ground-roll. In this way, only part of the linear kinetic energy which the body has before impact is eliminated during impact, while the other part is transformed into angular kinetic energy.

The attenuation of peak forces achieved in soft landing on foam-rubber sheets can be interpreted by the decrease in stiffness of the impact medium. Another contributing factor is obviously contact time. In a study on human running, McMahon and Greene (1979) demonstrated that a more compliant track increased the contact time of the foot with the ground. For a given linear impulse time increase results in decrease of the impulsive forces present.

Conclusions

1. Landing on the balls of the feet considerably decreased peak forces as compared to landing with the feet flat. The first peak significantly decreased in all cases whereas the change of the second peak intensity was not always consistent but was generally less pronounced than that of the first.
2. The pre-programmed non-reflex muscle action and multi-joint motion during the early phase of impact has proved important in peak attenuation. This procedure should be trained to and controlled by the subject to achieve simultaneous use of all joints and co-ordination between the various segments of the leg. Better attenuation is also a result of increasing the flexion range of the joints of the leg.
3. Ground-roll succeeding impact caused a decrease in landing peak forces. Due to the appearance of ground-roll later than the force peaks, its effect could only be indirect. It was suggested that the actual reason for force attenuation was related to the better preparation of the subject as a result of the intention to roll and to the increased range of body and joint motion in landing during the ground-roll.
4. Peak force attenuation in soft landing indicates the role of the quality of the ground and of the shoes worn by the subject. The attenuation effect was attributed to both decrease in stiffness and increase in contact time during impact.

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References

Bosco, C. and **Komi, P. V.** (1979) Mechanical characteristics and fiber composition of human leg extensor muscles. *Eur. J. Appl. Physiol.*, **41**, 275–284.

Ducheyne, P., Heymans, L., Martens, M., Aernoudt, E., De Meester, P. and **Mulier, J. C.** (1977) The mechanical behaviour of intracondylar cancellous bone of the femur at different loading rates. *J. Biomech.*, **10**, 747–762.

Finlay, J. B. and **Repo, R. U.** (1979) Energy absorbing ability of articular cartilage during impact. *Med. Biol. Engng Comput.*, **17**, 397–403.

Greene, P. R. and **McMahon, T. A.** (1979) Reflex stiffness of man's anti-gravity muscles during knee bends while carrying extra weights. *J. Biomech.*, **12**, 881–891.

Luthanen, P. and **Komi, P. V.** (1978) Segmental contribution to forces in vertical jump. *Eur. J. Appl. Physiol.*, **38**, 181–188.

McMahon, T. A. and **Greene, P. R.** (1979) The influence of track compliance on running. *J. Biomech.*, **12**, 893–904.

Orava, S., Puranen, J. and **Ala-Ketola, L.** (1978) Stress fractures caused by physical exercise. *Acta Orthop. Scand.*, **49**, 19–29.

Radin, E. L., Paul, I. L. and **Lowy, M.** (1970) A comparison of the dynamic force transmitting properties of subchondral bone and articular cartilage. *J. Bone Jt Surg.*, **52A**, No. 3, 444–456.

Radin, E. L., Paul, I. L. and **Rose, R. M.** (1972) Role of mechanical factors in pathogenesis of primary osteoarthritis. *Lancet*, 519–521.

Radin, E. L. (1974) Nature of mechanical factors causing degeneration of joints, in the hip. *Proc. of the 2nd open scientific meeting of the Hip Society 1974*, C. V. Mosby, St. Louis, pp 76–81.

Smith, L. D. (1953) Hip fractures. The role of muscle contraction or intrinsic forces in the causation of fractures of the femoral neck. *J. Bone Jt Surg.*, **35A**, 367–383.

Thys, H. (1978) Evaluations indirecte de l'energie elastique utilisée dans l'impulsions des sauts. *Schweizerischen Zeitschrift für Sportmedizin*, **4**, 169–177.

Welsh, P. (1979) Stress syndrome in athletes. *New Zealand Medical Journal*, **89**, 223–225.