ELSEVIER

Clinical Biomechanics 15 (2000) 15–20

CLINICAL BIOMECHANICS

www.elsevier.com/locate/clinbiomech

Shock accelerations and attenuation in downhill and level running

J. Mizrahi [a,b,*], O. Verbitsky [b], E. Isakov [c]

[a] Department of Rehabilitation Sciences, The Hong Kong Polytechnic University, Hung Hom, Kowloon, Hong Kong, People's Republic of China
[b] Department of Biomedical Engineering, Technion – Israel Institute of Technology, Haifa 32000, Israel
[c] Loewenstein Rehabilitation Hospital, Raanana 43100, Israel

Received 3 September 1998; accepted 20 April 1999

Abstract

Objective. A study was conducted to investigate the possible effects of fatigue on the heel strike-initiated shock accelerations and on attenuation of these shocks along the body during eccentric muscle contractions.

Design. Level and decline running on a treadmill were used to acquire the experimental data on the foot strike-initiated shock accelerations.

Background. Eccentric contractions of the lower limb muscles in combination with shock generation and propagation during downhill running and muscle fatigue may diminish their ability to dissipate and attenuate loading on the system.

Methods. Fourteen young healthy males ran on a treadmill at a speed exceeding their anaerobic threshold by 5% for 30 min, as follows: (a) level running and (b) downhill running with a decline angle of $-4°$. The foot strike-induced shock accelerations were recorded every five minutes on the tibial tuberosity and sacrum. Fatigue was monitored by means of the respiratory parameters.

Results. The downhill running related with eccentric muscle contractions was associated with increased shock propagation from the tibial tuberosity to the sacrum levels, even though fatigue did not develop.

Conclusions. Shock propagation from the tibial tuberosity to the sacrum is augmented due to the eccentric action of the muscles, without metabolic fatigue development.

Relevance

Eccentric muscle contraction in downhill running reduces the musculoskeletal ability to attenuate the heel strike-induced shock waves. Knowledge about the effect of fatigue on the shock propagation between the shank and the sacrum levels may help in understanding the mechanism of stress fractures and joint damage.

Keywords: Heel strike; Downhill running; Acceleration; Shock waves

1. Introduction

In downhill running, the extensor muscles of the lower limbs eccentrically contract during each stride to decelerate the center of mass after the foot touches the ground [1]. Eccentric muscle contraction during downhill running has been associated with increased mechanical stress [2] and ultrastructural muscle damage [3].

Compared to level running at the same speed, the oxygen uptake in downhill running was reported to be significantly lower [4,5]. Paul et al. [6] speculated that the muscles play a major role in the attenuation and dissipation of the heel strike-induced shock accelerations. A deficient shock attenuation ability of the muscle tissues has been related to an increased incidence of femoral and tibial stress fractures in elite infantry recruits [7]. In the long term, the high impact forces associated with the heel strike may also cause damage to the articular cartilage and lead to the development of osteoarthrosis [8].

The vertical impact peak force was reported to be higher during short-term downhill running than during level running [9,10]. In a recent study, it has been shown that shank and sacrum shock accelerations increased as a result of fatigue in long-term level running, at a speed just exceeding the anaerobic threshold [11].

Previous studies have demonstrated that in increased eccentric activity, e.g. in downhill running or walking or on a bicycle ergometer modified for use in eccentric work, ultrastructural and morphological low-frequency fatigue related damage is caused to the muscle [3,12–15].

* Corresponding author.
E-mail address: rsjm@polyu.edu.hk (J. Mizrahi)

0268-0033/99/$ - see front matter
PII: S0268-0033(99)00033-9

More recently, Lieber et al. [16] reported in experimental animal models that cyto-skeletal disruption occurs within the first 15 min of cyclic eccentric contraction. It can thus be assumed that, due to the structural disruption that results from repetitive intensive eccentric muscle activity, and despite the above-mentioned reduced metabolic cost, a decreased ability of the muscle to attenuate and dissipate the heel strike-induced shock accelerations may be expected. Spectral analysis of the raw signals sheds additional light on the changes occurring between shank and sacrum and between level and downhill running, and has indeed been used to characterize the acceleration signals in running [17,18]. It has been reported that the typical power spectral density (PSD) of the shank acceleration includes two major regions. The first, referred to as the active region, is in the range of 4–6 Hz and corresponds to the low-frequency motion of the leg during stance. The second region, referred to as the impact region, is in the range of 12–20 Hz and is associated with the high-frequency shock during the foot-ground impact [17–19].

Frequency analysis was also used to study vertical transmission of the acceleration signal from the shank upwards along the body. Thus, it has been reported that the range of the median frequency (MDF) of the entire PSD at the sacrum is 7–9 Hz [11], as compared to 11–13 Hz at the shank level [11,20] and to 3–4 Hz at the head level [20]. It has been suggested that the elimination of higher frequencies in higher body positions is associated with the low-pass filter properties of the body. To the knowledge of the present authors, the shank-to-sacrum filtering effect in long-term downhill running as compared to level running and the possible related effects of fatigue have not as yet been studied.

The aim of the present study was thus to characterize long-term downhill in comparison to level running. The following three specific goals were of special interest: (a) examine the time change of the respiratory variables, (b) determine the time effect during the course of running on the shock accelerations both at the tibial tuberosity and sacrum due to the possible development of fatigue, and (c) study the attenuation properties between the shank and sacrum levels. To achieve the last two of the above goals, analyses in both the time and frequency domains were necessary. Knowledge about the effect of fatigue on the shock propagation from the shank to the sacrum levels may help in understanding the mechanism of stress fractures and joint damage.

2. Methods

2.1. Subjects and general procedure

Fourteen male subjects of mean age 24.2 yr (SD, 3.7), mean height 175.5 cm (SD, 5.9), mean body mass 73.2 kg (SD, 8.3) and mean leg length 0.90 m (SD, 0.03) volunteered to participate in this study. All subjects were in an excellent state of health. They all practised calisthenics at least twice a week. No previous histories of muscle weakness, neurological disease or drug therapy were ever recorded. Each subject provided informal consent according to the local ethical committee's guidelines. To assure uniformity of the testing conditions all subjects were provided with the same manufacturer and type of running shoes.

2.2. Respiratory data

Respiratory data (Sensor-Medics 4400, Alpha Technologies, Laguna Hills, CA, USA) were collected during both the preliminary and the two actual running tests. The measuring instrument was calibrated before every test. During running, the subject breathed through a mouthpiece attached to a turbine device. The subject's respired gas was continuously sampled, for breath-by-breath determination of gas exchange and ventilatory variables. The values obtained during exercise for oxygen consumption ($\dot{V}O_2$), minute ventilation ($\dot{V}E$), carbon dioxide production ($\dot{V}CO_2$), end-tidal carbon dioxide pressure ($PETCO_2$), ventilatory equivalent for oxygen ($\dot{V}E/\dot{V}O_2$) and ventilatory equivalent for carbon dioxide ($\dot{V}E/\dot{V}CO_2$) were calculated as an average of the breath-by-breath data during a time span of 30 s.

2.3. Preliminary running test protocol

The running test was performed on a treadmill (Quinton Q55, Seattle, Washington, USA) to allow repetitive monitoring of the data. A few days prior to the experiment, anaerobic threshold was determined for each subject as follows. The subject was exposed to an incremental load on the treadmill by increasing the running speed from an initial 2.22 m/s. The speed increments were of 0.28 m/s every 2 min and were applied until $PETCO_2$ reached the decline phase. The decline phase was confirmed if it persisted for at least one min. AT was determined as the point of initial increase of $\dot{V}E/\dot{V}O_2$ and $\dot{V}E/\dot{V}CO_2$, which just precedes the initial decline of $PETCO_2$ [21,22].

2.4. Actual running test protocols

Thereafter, each subject performed two different tests, which were made on separate days. At least a 1-week interval of time was allowed between the tests to ensure fatigue-free initial conditions. The first test was with the treadmill set at 0° (horizontal) and the second test was with the treadmill set at a decline angle of −4°. The two actual running tests were performed at a speed exceeding the AT level of each subject by 5% and lasted 30 min. Just before the actual running test, every subject performed a warming up during 15 min on the treadmill at his free running velocity in order to familiarize with the treadmill and measuring equipment.

2.5. Heel strike-induced shock waves

Each subject was instrumented with two light-weight (4.2 g) uniaxial accelerometers (Kistler PiezoBeam, type 8634B50, Kistler, Winterthur, Switzerland), connected to a coupler (Kistler Piezotron, type 5122). One was attached on the tibial tuberosity, and the second – on the sacrum. The accelerometers were pressed onto the skin in closest position to the bony prominences of the tibial tuberosity and the sacrum, by means of two elastic belts passed in a horizontal plane around the shank and the waist, respectively. The tensions of the belts were well above the level in which the acceleration trace for a given impact force became insensitive to the accelerometer attachment force, thus ensuring reproducibility of the data [23,24]. The shank accelerometer was aligned with the axis of the tibia to provide the axial component of the tibial acceleration and the accelerometer on the sacrum was oriented along the spine. These accelerometers allowed to acquire the shock accelerations propagated in the longitudinal directions of the tibia and the spine. As earlier reported, such attachment is suitable for faithfully measuring the amplitude of a shock acceleration [11,25,26]. The signals from the accelerometers were collected into a PC at a sampling rate of 1667 Hz per channel. During the test, acceleration data were acquired every 5 minutes for a time span of 20 s. Thus, information on about 30 heel strikes was provided for each time span. The data were off-line processed. The subjects were not aware of when exactly the data were acquired in order to minimize possible gait modifications. Data processing was simplified due to the constant speed of running during the test. Based on the known running speed and the location of first heel strike, the developed software automatically detected the occurrences of the consequent heel strikes and the acceleration maximum amplitudes. Since there was always a possibility of "bad" data (i.e. subject accidentally stumbled or misplaced the foot), all results were shown on the monitor and confirmed by an operator.

2.6. Frequency domain analysis

The stance phase on the accelerometer traces was identified from a pilot experiment in which the subjects ran on a forceplate with an accelerometer attached to the shank. The stance phase of each time-based accelerometer trace was thereafter extracted for Fast Fourier Transform (FFT) analysis and calculation of the PSD, in a similar way as described by Derrick et al. [19]. An eighth order cascaded, 40 Hz low-pass Butterworth filter was used to remove the noise from the accelerometer signals [20]. For each given frequency, the following transfer function (TF) is defined [17]:

$$\mathrm{TF} = 10\log_{10}(\mathrm{PSD}_{\mathrm{sacrum}}/\mathrm{PSD}_{\mathrm{shank}}), \qquad (1)$$

where $\mathrm{PSD}_{\mathrm{sacrum}}$ and $\mathrm{PSD}_{\mathrm{shank}}$ designate the PSD functions for the sacrum and shank, respectively. The TF assumes a positive value if $\mathrm{PSD}_{\mathrm{sacrum}}$ is larger than $\mathrm{PSD}_{\mathrm{shank}}$ (amplified signal) and a negative value if $\mathrm{PSD}_{\mathrm{sacrum}}$ is smaller than $\mathrm{PSD}_{\mathrm{shank}}$ (attenuated signal).

From the PSD versus frequency plots, three frequency measures were used for the statistical analyses: (a) peak frequency of the active, low-frequency region; (b) peak frequency of the impact shock region; and (c) MDF of the entire PSD.

2.7. Statistical analysis

Differences between results were tested (*t*-test) and the level of significance was determined at $P < 0.05$, using conventional statistical packages.

3. Results

3.1. Treadmill running speed and respiratory data

The average running speed for all 14 subjects was 3.53 m/s (SD, 0.19). In level running, $PETCO_2$ consistently decreased as shown in Fig. 1 (top panel). In decline running, $PETCO_2$ did not vary significantly. As for $\dot{V}O_2$, it did not vary significantly in both the level and decline running (Fig. 1, bottom panel). However, in level running $\dot{V}O_2$ was significantly higher compared to that of decline running.

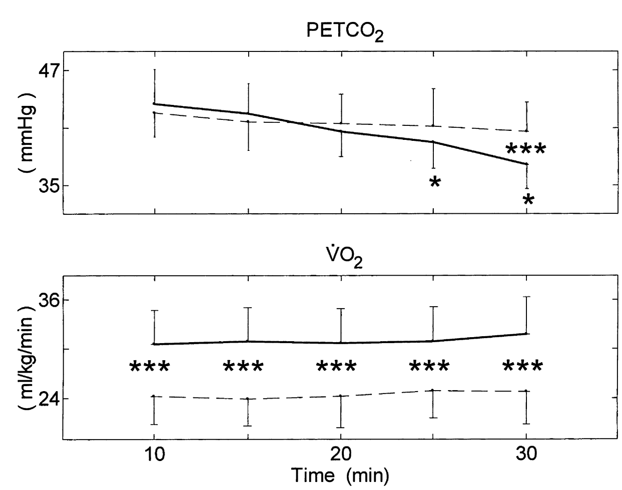

Fig. 1. Average for all the subjects of $PETCO_2$ (top panel) and $\dot{V}O_2$ (bottom panel) (the vertical bars denote one SD) at a running speed, just exceeding the anaerobic threshold (see text). Solid line = level running, dashed line = decline running. * – significantly different from data at the beginning of running in level running ($P < 0.05$). *** – significant difference between level and decline running ($P < 0.05$).

3.2. Shank and sacrum shock acceleration and sacrum/shank attenuation ratio

Variations of the shock acceleration and of the attenuation ratio during the course of running are presented in Fig. 2. On the shank, the shock acceleration increased in level running, but did not vary in decline running (top panel). At the sacrum level the shock acceleration increased in both level and decline running (central panel). The average sacrum shock acceleration was higher in decline compared to level running. The average sacrum/shank shock acceleration ratio decreased in level running (bottom panel), but increased in decline running. The attenuation ratio was significantly higher in decline compared to level running.

3.3. Power spectral density

Fig. 3 shows typical spectra for one subject of the shank and sacrum accelerometer traces during the contact phase of running. In level running, the shank PSD demonstrates in the first minute of running a two-peak pattern (Fig. 3A). The impact peak (12–20 Hz zone) is higher than the active one (4–9 Hz zone). As fatigue develops the peak in the impact region increases and, in parallel, an additional peak within the 25–35 Hz range level becomes noticeable. The sacrum PSD in level running demonstrates a double-peak pattern throughout the running time (Fig. 3C), with the active peak at a higher level than the impact peak.

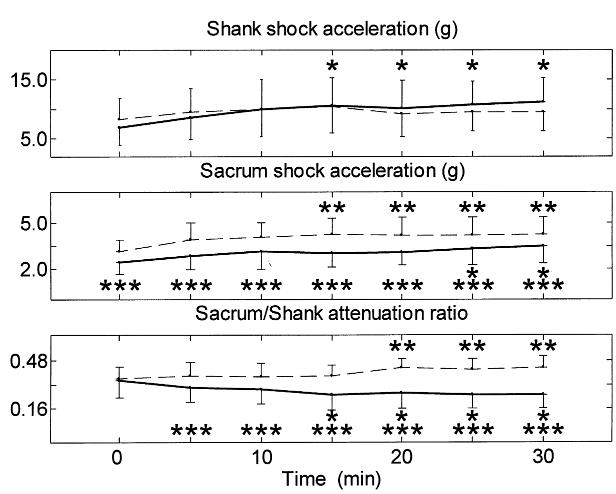

Fig. 2. Average for all subjects of the shank shock acceleration (top panel), sacrum shock acceleration (central panel) and sacrum/shank attenuation ratio (bottom panel) at a running speed, just exceeding the anaerobic threshold (see text). Solid line = level running, dashed line = decline running (the vertical bar denotes one SD). * – significantly different from data at the beginning of running, in level running ($P < 0.05$). ** – significantly different from data at the beginning of running, in decline running ($P < 0.05$). *** – significant difference between level and decline running ($P < 0.05$).

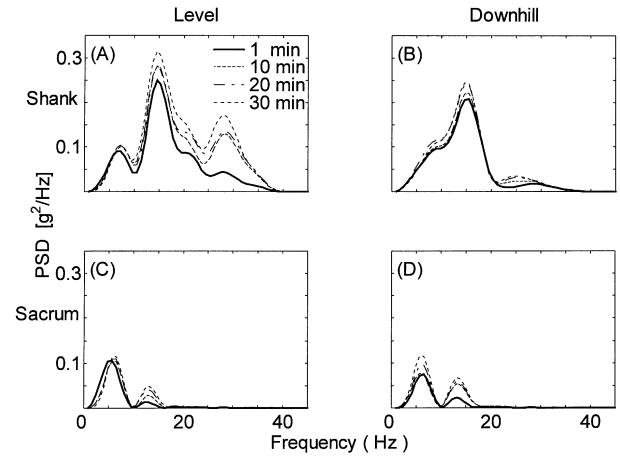

Fig. 3. Power spectral density (PSD) profiles for one typical subject of the accelerometer traces (shank and sacrum) during the stance phase of running. Four time stages of a total of 30 min running at a 3.53 m/s running speed are demonstrated for both level and decline running.

In decline running the shank PSD has a single peak at the impact frequency region, and a shallow inflection zone in the active frequency region (Fig. 3B). The sacrum PSD demonstrates again a double-peak pattern, with a higher peak in the active frequency region. As fatigue develops, the peak magnitudes increase (Fig. 3D).

3.4. Peak frequencies

Summary of the average peak frequencies in the active and impact frequency regions and of the MDF of the entire PSD for all the subjects is presented in Table 1. The statistically significant differences between intensities of the various peaks or MDFs, different stages in the course of running, level and decline running are also indicated in this table.

3.5. Transfer function

The mean TF for all the subjects versus frequency is presented in Fig. 4 for level and decline running, at different stages of running. Frequencies above 5 Hz were attenuated between the shank and sacrum levels (negative TF values). Peak attenuation was noticed at nearly 9 Hz in both level and decline running. The effect of time during running on frequency attenuation was less accentuated in decline running when compared to level running.

4. Discussion

This study highlights the parameters expressing biomechanical differences between decline and level running. From the results obtained in the present study

Table 1
Summary of the average peak frequencies in the active and impact frequency regions and of the median frequency (MDF) of the entire PSD for all the subjects at a running speed, just exceeding the anaerobic threshold

Freq. (Hz)	Running test	Running time (min)						
		1	5	10	15	20	25	30
Shank								
Active peak freq.	Level	6.0 (1.0)	6.5 (1.4)	7.0 (1.4)	7.3 (1.4)	7.3 (1.4)	7.3 (1.4)	7.3 (1.4)
	Decline	9.0 (0.4)*	9.0 (0.4)*	9.0 (0.4)*	9.0 (0.4)*	9.3 (0.4)*	9.3 (0.4)*	9.3 (0.4)*
Impact peak freq.	Level	16.4 (0.7)	16.5 (0.6)	16.5 (0.8)	16.5 (1.0)	16.5 (1.0)	16.5 (1.2)	16.4 (1.1)
	Decline	15.3 (0.6)*	15.2 (0.6)*	15.2 (0.6)*	15.5 (0.8)	15.2 (0.6)*	15.2 (0.6)*	15.1 (0.8)*
MDF	Level	15.0 (0.6)	15.6 (1.0)	15.7 (1.1)	15.9 (1.2)	16.0 (1.0)	16.1 (0.8)**	16.3 (0.7)**
	Decline	14.0 (1.7)	14.8 (1.9)	15.3 (2.0)	15.2 (1.8)	15.0 (1.5)	14.9 (1.3)	14.7 (1.3)*
Sacrum								
Active peak freq.	Level	5.3 (0.4)	5.7 (0.3)	5.7 (0.3)***	6.0 (0.6)	5.7 (0.3)***	5.7 (0.3)***	5.7 (0.3)***
	Decline	5.3 (0.4)***	5.5 (0.4)***	5.3 (0.5)***	5.7 (0.9)***	5.5 (0.4)***	5.5 (0.4)***	5.5 (0.4)***
Impact peak freq.	Level	12.5 (0.4)***	12.5 (0.4)***	12.5 (0.4)***	12.5 (0.4)***	12.5 (0.4)***	12.5 (0.4)***	12.5 (0.4)***
	Decline	12.5 (0.4)***	12.5 (0.4)***	12.5 (0.4)***	12.5 (0.4)***	12.5 (0.4)***	12.5 (0.4)***	12.5 (0.4)***
MDF	Level	7.5 (1.0)	7.3 (0.5)	7.1 (0.7)	7.2 (0.3)	7.4 (0.4)	7.6 (0.1)	7.8 (0.2)
	Decline	7.0 (0.5)	7.1 (0.6)	7.1 (0.3)	7.5 (0.8)	7.3 (0.5)	7.5 (0.6)	7.8 (0.5)**

* Significantly different ($P < 0.05$) when compared to level running, other conditions remaining the same.
** Significantly different ($P < 0.05$) when compared to beginning of running.
*** Significantly different ($P < 0.05$) when compared to shank level, other conditions remaining the same (Mean (SD), $n = 14$).

(Fig. 1), it is concluded that metabolic fatigue developed only in level running. The shank and sacrum shock accelerations increased during level running due to the development of metabolic fatigue (Fig. 2), in agreement with previously published results [11]. However, during decline running, when metabolic fatigue was not attained, a significantly increased shock acceleration was found at the sacrum level, while the shock acceleration at the tibial tuberosity level did not change.

In the frequency domain in decline running, there was an increase in the MDF in the sacrum at the end compared to the beginning of the running (Table 1). This shift towards higher frequencies was not noticed in the shank.

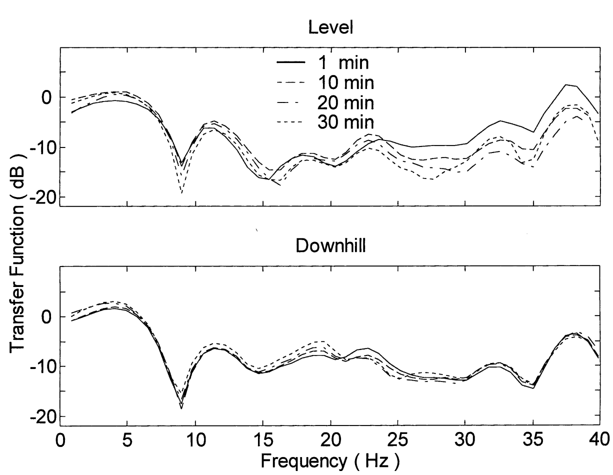

Fig. 4. Mean transfer functions between shank and sacrum accelerometer signals for all subjects running at a running speed, just exceeding the anaerobic threshold (see text for definition).

This phenomenon was additional to the above-mentioned increase in shock, indicating that during the course of downhill running there is a reduction in the attenuation of the high-frequency components of acceleration.

When comparing the sacrum to the shank, the average frequencies of the PSD peaks in the two main frequency regions were found lower in the former (Table 1), demonstrating a low-pass filtering effect associated with impact attenuation along the skeleton [18]. In level running, the difference between sacrum and shank was more pronounced in the impact frequency region (during the entire running period) as compared to the active region (only in the late stages of running). In decline running, the low-pass filtering effect took place in both the impact and active regions throughout the entire running time.

Of special interest was to compare the sacrum/shank acceleration attenuation ratio, as presented in Fig. 2. In level running this ratio decreased significantly with the development of metabolic fatigue while in decline running, it significantly increased. The TF, expressing the attenuation between shank and sacrum, demonstrated a major peak around 9 Hz (Fig. 4). In level running this peak was found to increase over the course of running, hence improving the attenuation between shank and sacrum. In decline running the opposite was noticed demonstrating, in the frequency domain, that the attenuation properties were more inferior in downhill compared to level running.

It has been reported that downhill running was associated with an increased mechanical stress [2] and muscular damage due to eccentric muscular contractions [3]. It is also well known that one of the major functions of lower limb muscle tissues is the dissipation of shock loadings during human locomotion [6,27,28]. Milgrom [7] reported that an increased shock loading on the bone results in an enhanced risk of stress fractures, due to inability of the muscle tissues to effectively dissipate and attenuate the heel strike-induced shock loads. It can therefore be hypothesized that the increased shock transmission from the tibial tuberosity to the sacrum level is associated with an increased muscular damage and, consequently, an inability of the lower limb muscle tissues to dissipate and attenuate shock wave propagation. Interestingly downhill running, which involves eccentric contraction, was found to be associated with increased shock propagation from the tibial tuberosity level to the sacrum levels without the development of metabolic fatigue.

Of the many biomechanical factors influencing performance in distance running two have been identified to be particularly important: (1) impact shock transmission along the body, reported to provoke bone injury and joint damage, and (2) stride rate at a given running speed, reported to affect running economy. In this study, only the first of the two factors was examined. A more complete comparison between downhill and level running should thus include stride parameters as well as kinematical factors, such as ankle, knee and hip angles and hip excursion. These should enable us to obtain additional information on leg stiffness properties in downhill compared to level running, and should be included in future studies.

Acknowledgements

This study was supported by the Israel Ministry of Health, the Segal Foundation and the Henri Gutwirth Promotion of Research Fund.

References

[1] Walmsley B, Hudgson JA, Burke RE. Force produced by medial gastrocnemius and soleus muscles during locomotion in freely moving cats. L Nerophysiol 1978;41:1203–16.

[2] Iversen JR, Macmahon TA. Running on an incline. J Biomech Eng 1992;114:435–41.

[3] Nurenberg CJ, Stray-Gundersen J, Fleckenstein JL, Gonyea WJ, Peshock RM. MR Imaging-guided muscle biopsy for correlation of increased signal intensity with ultrastructural change and delayed-onset muscle soreness after exercise. Radiology 1992;184:865–9.

[4] Byrnes WC, Clarkson PM, White JS, Hsieh SS, Frykman PN, Maughan RJ. Delayed onset muscle soreness following repeated bouts of downhill running. J Appl Physiol 1985;59:710–5.

[5] Liefeldt G, Noakes TD, Dennis SC. Oxygen delivery does not limit peak running speed during incremental downhill running to exhaustion. Eur J Appl Physiol 1992;64:493–6.

[6] Paul IL, Munro MB, Abernethy SSR, Radin EL, Rose RM. Musculo-skeletal shock absorption: Relative contribution of bone and sort tissue at various frequencies. J Biomech 1978;11:237–9.

[7] Milgrom C. The Israeli elite infantry recruit: a model for understanding the biomechanics of stress fractures. J R Coll Surg Edinb 1989;34 Suppl 6:S18–S21.

[8] Radin EL, Parker HG, Pugh GV, Steinberg RS, Paul IL, Rose RM. Response of joints to impact loading. J Biomech 1973;6:51–7.

[9] Hamill CL, Clarke TE, Frederick EC, Goodyear LJ, Goodyear ET. Effect of grade running on the kinematics and impact force (Abstract). Med Sci Sports Exerc 1984;16:S185.

[10] Dick RW, Cavanagh PR. A comparison of ground reaction forces (GRF) during level and downhill running at similar speeds (Abstract). Med Sci Sports Exerc 1987;19:S12.

[11] Voloshin A, Mizrahi J, Verbitsky O, Isakov E. Dynamic loading on the human musculoskeletal system – effect of fatigue. Clinical Biomech 1998;13:515–20.

[12] Fridén J, Sjöström M, Ekblom B. Myofibriallar damage following intence eccentric exercise in man. Int J Sports Med 1983;4:170–6.

[13] Fridén J, Lieber RL. Structural and mechanical basis of exercise-induced muscle injury. Med Sci Sports Exerc 1992;24:521–30.

[14] Sargeant AJ, Dolan P. Human muscle function following prolonged eccentric exercise. Eur J Appl Physiol 1987;56:704–11.

[15] Balnave CD, Thompson MW. Effect of training on eccentric exercise-induced muscle damage. J Appl Physiol 1993;75:15445–551.

[16] Lieber RL, Thornell LA, Friden J. Muscle cytoskeletal disruption occurs within the first 15 min of cyclic eccentric contraction. J Appl Physiol 1996;80:278–84.

[17] Shorthen MR, Winslow DS. Spectral analysis of impact shock during running. Int J Sport Biomech 1992;8:288–304.

[18] Hamill J, Derrick TR, Holt KG. Shock attenuation and stride frequency during running. Hum Mov Sci 1995;14:45–60.

[19] Derrick TR, Hamill J, Caldwell GE. Energy absorption of impacts during running at various stride lengths. Med. Sci. Sports Exerc 1998;30:128–35.

[20] Mahar AT, Derrick TR, Hamill J, Caldwell GE. Impact shock and attenuation during in-line skating. Med Sci Sports Exerc 1997;29:1069–75.

[21] Wasserman K, Whipp BJ, Koyal SN, Beaver WL. Anaerobic threshold and respiratory gas exchange during exercise. J Appl Physiol 1973;35:236–43.

[22] Whipp BJ. Dynamics of pulmonary gas exchange. Circulation 1987;76 Suppl 6:VI-18–VI-27.

[23] Mizrahi J, Susak Z. In vivo elasic and damping response of the human leg to impact forces. ASME J Biomech Engrg 1982;104:63–6.

[24] Streitman A, Pugh J. The response of the lower extremity to impact forces. I. Design of an economical low frequency recording system for physiologic waveforms. Bulletin of the Hospital for Joint Diseases 1978;39(1):63–73.

[25] Mizrahi J, Voloshin A, Russek D, Verbitsky O, Isakov E. The influence of fatigue on EMG and impact acceleration in running. Basic and Appl Myology 1997;7:119–26.

[26] Verbitsky O, Mizrahi J, Voloshin A, Treiger J, Isakov E. Shock transmission and fatigue in human running. J Appl Biomech 1998;14:301–11.

[27] Radin EL. Role of muscles in protecting athletes from injury. Acta Med Scand Suppl 1986;711:143–7.

[28] Jefferson RJ, Collins JJ, Whittle MW, Radin EL, O'Connor JJ. The role of the quadriceps in controlling impulsive forces around heel strike. Proc Inst Mech Engrs 1990;204:21–8.