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Sex differences in lower landing kinematics through neuromuscular fatigue

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1. Introduction

Athletes in volleyball, basketball, and soccer have rapid deceleration of the lower extremity, such as landing from a jump [1, 2]. The anterior cruciate ligament injury has been reported to happen to women about three times more often than men in soccer and basketball [3]. Recently, Ford et al [1, 4] and Kernozek et al [1, 5] reported that women demonstrated significantly increased frontal-plane motion of the knee when landing compared with men. Subjects with poor hip abductor strength may demonstrate decreased proximal control of the hip, which then may result in inferior knee kinematics [1, 6]. High levels of strength and muscle strength balance between antagonistic muscle groups may be used to protect the knee during jumping and landing activities [7]. It is known that under loading conditions women are more fatigue-resistant than men, due to slow muscle fibre and relative larger muscle cross-sectional area in females slow muscle fibre [8]. Female hormones reduce mechanic sensitivity of bone tissue and its osteogenic response during mechanic landing [9]. Furthermore, females squat less than males during jump [10]. Jumping as a natural stretch-shortening cycle (SSC) muscle action, which is known to enhance muscle output in the final shortening phase (push-off), is compared with the pure shortening action alone [11].

The relationship between fatigue and antagonistic muscle group strength during prolonged activities suggests that it may play a vital role in neuromuscular control of the knee. Also anterior cruciate ligament injuries have happened more often in females than males from a jump through fatigue that suggests playing attention to that problem. Thus, the aim of the study was to evaluate sex differences in landing from a jump in relation to lower extremity landing kinematics through neuromuscular fatigue.

2. Materials and methods

2.1. Subjects

Healthy and physically active females ($n = 10$) with normal menstrual cycle, aged 19–23 years, body weight – 58.2 ± 6.1 kg, height – 168.4 ± 5.6 cm, and healthy and physically active males ($n = 10$), aged 19–23 years, body weight – 78.2 ± 6.1 kg, height – 179.8 ± 5.8 cm, participated in the study. Female participants had not used oral contraceptives for 6 months and they had a regular menstrual cycle. All subjects were physically active and had not been involved in any jumping or leg strength training programs during the last years. Also, subjects did not have pelvic, hip, knee and ankle surgery or injury to the same joints. Each subject read and signed a written informed consent form consistent with the principles outlined in the Declaration of Helsinki. Ethical approval was obtained from Kaunas Regional Biomedical Research Ethics Committee (Report Number BE-2-24).

2.2. Surface electromyography measurements

Bipolar Ag-AgCl surface electrodes were used for electromyography (EMG) recordings (silver bar electrodes, diameter 10 mm, centre-to-centre distance 20 mm) of the long head of the vastus lateralis and biceps femoris (DataLog type no. P3X8 USB, Biometrics Ltd, Gwent, UK). The skin at the electrode site was shaved and cleaned with alcohol wipes. The electrodes were placed half way on a line between ischial tuberosity and fibula head. The ground electrode was positioned on the patella of the same leg. EMG signals were recorded by amplifiers (gain 1000) with signal measurement using a third order filter (18dB / octave) bandwidth of 20 – 460 Hz [12]. The analogue signal was sampled and converted to digital form at sampling frequency of 1 kHz. The EMG signal was telemetered to a receiver that contained a differential amplifier with an input impedance of 10 MΩ, input noise level was less than 5 μV and the common mode rejection ratio was higher than 96 dB [12]. Before the recordings of EMG, we set 3V for channel sensivity, 4600 mV for excitation output [12]. Electromyography files were stored on the memory card and copied to PC biometrics Datalog (version 5.03; Biometrics Ltd, Gwent, UK) for data processing and analysis.

2.3. Electrogoniometer measurements

The twin axis electrogoniometer (DataLog type no. P3X8 USB, Biometrics Ltd, Gwent, UK) was used to quantify hip joint flexion and extension (SG 150), knee joint flexion and extension (SG 150), ankle joint flexion and extension (SG 110) angles. The electrogoniometer is comprised of optical fibres to measure motion, a fixed end-block and a telescopic end-block [13]. Mechanical signals from the measuring element in the end-blocks were converted into a digital signal by a data log acquisition unit which connected the electrogoniometer to a display unit. A frequency rate of approximately 200 Hz had been previously used for measuring hip, knee and ankle joints movements in functional activities. By moving the telescopic end-block clockwise towards the fixed end-block, joint angles were recorded as positive values. By moving the telescopic end-block anticlockwise to the fixed end-block, negative values of angles were recorded. The fixed end-block was adhered to the template at a known position of 0° using double adhesive tape. The telescopic end-block was then moved to a desired angle [13]. The telescopic end-block for hip joint flexion and extension (SG 150) was attached in parallel with the hip joint on a half way to iliacus and gracilis. The telescopic end-block for knee joint flexion and extension (SG 150) was attached in parallel with the knee joint on a half way to vastus lateralis fascia and to the external side of the shin. The telescopic end-block for ankle joint flexion and extension (SG 110) was attached in parallel with the ankle joint on a half way to shin fascia and to retinaculum flexorum. Calibrations were performed every five degrees within the range of 0°–180° in random order and each angle was measured 10 times to establish consistency of measurement [13]. The angle reading outputs of the electrogoniometer in both directions were calibrated using this procedure [13]. The differences between electrogoniometer angles and the reference angles were recorded [13]. Using this validation procedure, the electrogoniometer was shown to have a measurement error of 0.04° [13].

2.4. Experimental protocol

After 10 – 15 min of non-intensive warming-up (slow pedaling velorgometer, with the heart rate of 120 – 130 b / min), 100 drop jumps were started on a contact mat (New Test, Finland), 30 s interval between each jumps. Subjects stood on 75 cm stage, stretched the right leg forward and performed drop jumps on a contact mat. The jump height (H) was calculated using the formula [10]:

$$h = \frac{g \times t_p^{\ 2}}{8} = 1.22625 \times t_p^{\ 2}, \qquad (1)$$

where h - jumping height (m), g - acceleration due to gra–vity (9.81 m/s^2), t_p - flight time (s).

Before jumping the electrodes and telescopic end-blokes were attached and scoreless values were set. Jumping H, EMG signal and angles of electrogoniometer were measured in every drop jump. During one drop jump we calculated EMG of vastus lateralis, biceps femoris, and peak values of hip, knee and ankle angles. When participant got to the braking phase (the beginning of drop jump to the peak knee joint angle), the phase was named T1 phase; push-off phase (peak knee joint angle to the end of jump) was named T2 phase. EMG values were analyzed by rms (root mean square).

The fatigue index (FI) of jumping H was calculated:

$$FI = \frac{(H\ before\ exercise\ -\ H\ after\ exercise)}{H\ before\ exercise} \times 100, \qquad (2)$$

where H before exercise – average of 3 first H values and H after exercise – average of 3 last H values.

The one-way analyses of variance (ANOVA) for repeated measuring were used to determine the effect of rms of EMG vastus lateralis and biceps femoris properties before and after exercise. Descriptive data are presented as means ± standard deviations (SD). The level of significance was set at $P < 0.05$. Aiming at evaluating the relationship between changes in different indicators of jumping height and EMG values before and after exercise, Pearson's correlation coefficient (R) was established. Based on on alpha level of 0.05, sample size ($n = 10$), SD and average level before and after exercise, power of the test was calculated for all indicators. In all cases power of the tests was more than 80%.

3. Research results

Male jump H values were higher than the female ones ($P < 0.05$; Fig. 1), and they depended on the number of jumps and sex ($P < 0.05$).

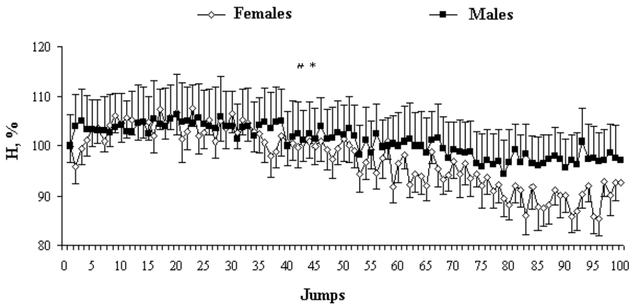

Fig. 1 Percentage values of female and male 100 jump height

Note. # – female and male results compared ($P < 0.05$);
* – compared to the initial value ($P < 0.05$).

There was a significant relationship comparing female and male jump H and body mass during 100 jumps ($P < 0.05$), which depended on the number of jumps and sex ($P < 0.05$; Fig. 2).

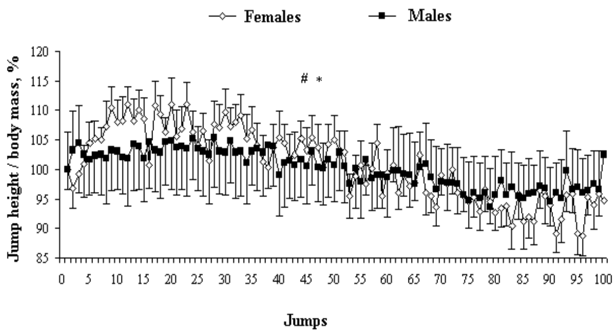

Fig. 2 Female and male jump height and body mass relation during 100 jumps

Note. # – female and male results compared ($P < 0.05$);
* – compared to the initial value ($P < 0.05$).

The fatigue index of H significant differ compare with males (4.92 ± 0.46%) and females (6.67 ± 0.50%).

There was significant relationship between changes in females' H and rmsEMG vastus lateralis (T1 phase, $R = 0.77$, $P < 0.05$) and rmsEMG biceps femoris (T1 phase, $R = 0.78$, $P < 0.05$). Also there was a significant relationship between changes in males' H and rmsEMG vastus lateralis (T1 phase, $R = 0.78$, $P < 0.05$) and rmsEMG biceps femoris (T1 phase, $R = 0.75$, $P < 0.05$). There was a reverse significant relationship between changes in females H and rmsEMG vastus lateralis (T2 phase, $R = -0.75$, $P < 0.05$) and rmsEMG biceps femoris (T2 phase, $R = -0.77$, $P < 0.05$) during the last ten drop jumps. Also there was a reverse significant relationship between changes in males' H and rmsEMG vastus lateralis (T2 phase, $R = -0.68$, $P < 0.05$). There was a significant relationship between changes in males' H and rmsEMG biceps femoris (T2 phase, $R = 0.79$, $P < 0.05$) during the last ten drop jumps.

Female rmsEMG vastus lateralis and biceps femoris values decreased in T1 and T2 phases during 1–10 and 90–100 jumps ($P < 0.05$; Fig. 3). Male rmsEMG vastus lateralis and biceps femoris values increased during 1–10 and 90–100 jumps ($P < 0.05$; Fig. 3).

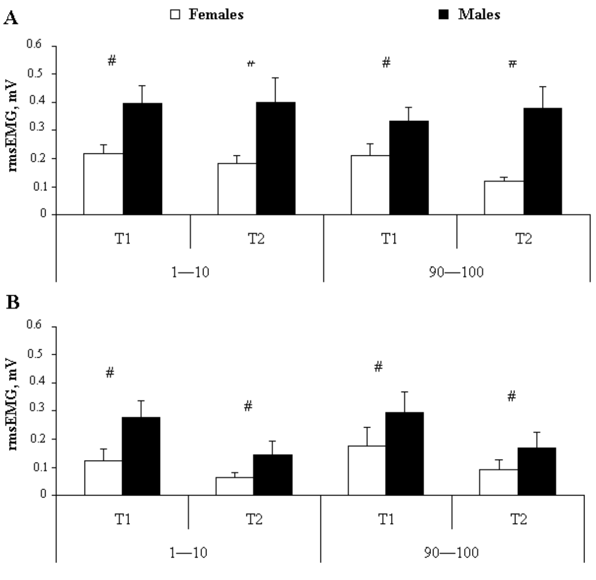

Fig. 3 Female and male rmsEMG vastus lateralis (A) and biceps femoris (B) values at T1 and T2 phases during 1–10 and 90–100 jumps.
Note. # – female and male results compared ($P < 0.05$).

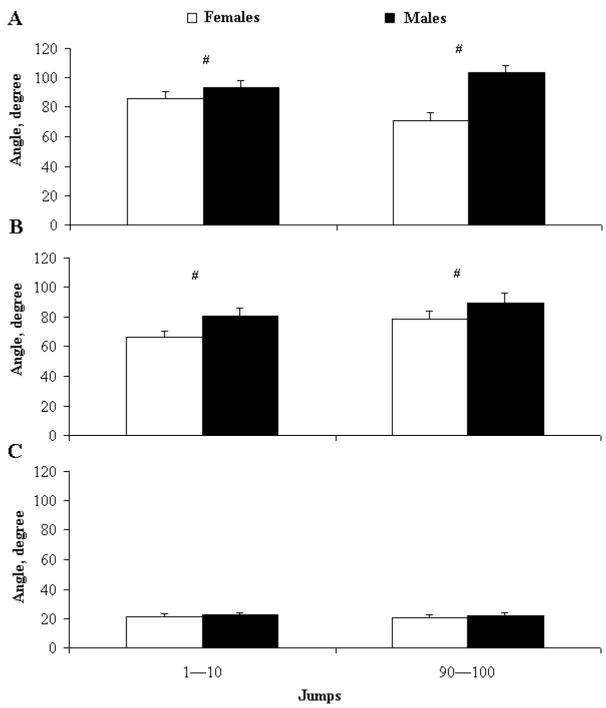

Fig. 4 Peak angles of female and male knee (A), hip (B) and ankle (C) joints during 1–10 and 90–100 jumps
Note. # – female and male results compared ($P < 0.05$).

Females squatted less in knee than males during 1–10 (females 85.81°, males 93.52°) and 90–100 jumps (females 70.43°, males 103.80°) ($P < 0.05$; Fig. 4). A statistically significant difference was found comparing female and male peak hip angle values during 1–10 and 90–100 jumps ($P < 0.05$; Fig. 4).

There was a significant relationship decrease in females' rmsEMG vastus lateralis and biceps femoris fatigue index in T1 and T2 phase compared to that of males changes during 1–10 and 90–100 jumps.

4. Discussion

The aim of the study was to evaluate sex differences in landing from a jump in relation to lower extremity landing activities through neuromuscular fatigue. The higher indices of female jump height decrease and fatigue were ascertained comparing the results during 100 jumps. Women demonstrated lower peak hip, knee joint angles and lower rmsEMG values in T2 phase when landing from a drop jump. The correlations between H and drop jump kinematics were generally higher for women than men.

Study results revealed that when performing 100 jumps, female jump height decreased more than that of males. It is assumed that female jump height decreases more than the male one because of the prevailing slow fibre in female quadriceps femoris [14]. The assumption may be made that because of slow fibre prevailing in muscle [15] female jump height was lower than that of males. For the jump height value to be as high as possible, one should attain a higher explosive force which cannot be developed by slow muscle fibre [16]. Because of fast IIA type fibre (fibre with the largest cross-section area) prevailing in male muscle, the highest amount of sodium and potassium ion channels in muscle and thus the rate of muscle fibre conduction increases [17]. However, low fibre prevailing in female muscle is able to increase muscle oxygen capacity better [18]; this causes greater oxygen uptake by muscle when contracting and slower recovery after eccentric-concentric exercise [14]. It is known that lower body mass can develop higher jump height [19]. In the present study the jump height values were compared with body mass, and we observed that females could hold higher jump height than males. Moreover, it is known that muscular mass, maximal voluntary force, maximal jump force and specific tendon elasticity at the moment of jump depend on the characteristics of bone structure [20]. It has been ascertained that the strength of female tibia distal part is lower than the male one [20], whereas inner and side condyles of tibia are thicker in male bone structure than in the female one [21]. An assumption is made that men may keep the angle of squat longer under eccentric-concentric loading during 90-100 jumps than women. Maybe those anatomic differences in bone structure force females to change the nature of jump while choosing jump height as a priority to jump squat angle. Furthermore, it is maintained that neuromuscular loading depends on bone strength [21, 22], and male bone force is greater than that of females [23]. When comparing female and male jump indices, higher values of male EMG were noticed. Studies indicate that male force is higher than female force due to bigger muscle mass [16, 24]; therefore male muscle activity is higher during jumps. The present results suggest that subjects with increased vastus lateralis and biceps femoris EMG values may demonstrate higher peak hip and knee angle as a result of enhanced proximal control of the hip. We did not measure the strength of other lower extremity or core muscle groups, so we cannot be sure if the males' vastus lateralis and biceps femoris EMG values could provide enhanced proximal control of the hip and were able to better use the quadriceps and hamstrings muscles. Bobbert and van Zandwijk [25] reported that the ability of the quadriceps and hamstrings to resist forces when jumping was significantly improved with increased hip muscle activity. The enhancing function of the quadriceps and hamstrings, increasing strength of the hip abductors may improve neuromuscular control of the knee when landing from a jump [26]. This finding is supported by the results of Stanley et al [10] who measured preseason isokinetic hip abductor torque in male collegiate football athletes and the incidence of lower extremity noncontact injuries during the competitive season. The authors reported no differences in vastus lateralis strength between injured and uninjured male athletes, further suggesting that in men, this muscle group may not play a protective role in neuromuscular control of the lower extremity. Female athletes trained by Hewett et al [27] were able to reduce noncontact ACL injuries by maintaining neutral alignment of the center of gravity with the chest above the knees, no excessive side-to-side or forward-backward motion, and a toe-to-heel landing strategy.

In the present study EMG of vastus lateralis and biceps femoris were higher in T1 phase then T2 phase. In the study of Kuitunen et al. [28] high muscle activity of solea muscle in the braking phase (T1 phase) relative to the push-off phase (T2 phase) was associated with high leg stiffness. In contrast to the maximal jumps, this relationship was not observed when comparing the braking phase EMG alone with the leg stiffness [28]. The implication of this finding is that, during in vivo exercise, appropriate coactivation and coordination around the joint may be maintained, despite exercise induced fatigue of an antagonist muscle group [28]. Studies with fatiguing exercise showed significant decrease in the maximum voluntary EMG in males, but not in females, crediting this difference to a higher synchronization of neuromuscular activation [29]. In the study Shannon, wavelet entropy of EMG vastus lateralis values significantly increased in the last ten jumps compared to the first ten ones [30].

5. Conclusions

The higher indices of female jump height decrease and fatigue when comparing results during 100 jumps were ascertained. Women demonstrated lower peak hip, knee joint angles and lower EMG values in T2 phase when landing from a drop jump. Furthermore, correlations between *H*, EMG values and landing kinematics were generally higher for women than men.

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LYTIES POVEIKIS APATINIŲ GALŪNIŲ NUŠOKIMO KINEMATIKOS RODIKLIAMS NUOVARGIO METU

R e z i u m ė

Tyrimo tikslas įvertinti lyties poveikį apatinių galūnių nušokimo kinematikos rodikliams nuovargio metu. Lyginant rezultatus 100 šuolio metu, nustatytas moterų didesnis šuolio aukščio mažėjimas. Tačiau nustatytos mažesnės moterų klubo bei kelio sąnarių kampo reikšmės, taip pat mažesnės EMG reikšmės T2 fazėje. Koreliacijos koeficientai buvo didesni moterų nei vyrų lyginant šuolio aukštį ir EMG mažėjimą tarpusavyje, kaip ir lyginant šuolio kinetikos rodiklius. 100 šuolio metu didesnis EMG vidutinės kvadratinės amplitudės rodiklių mažėjimas buvo pastebėtas tarp vyrų negu tarp moterų.

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SEX DIFFERENCES IN LOWER EXTREMITY LANDING KINEMATICS THROUGH NEUROMUSCULAR FATIGUE

S u m m a r y

Rapid deceleration during sporting activities, such as landing from a jump, has been identified as a common mechanism of acute knee injury. Jumping and landing activities performed during different phases lead to differences in foot strike knee flexion, as well as peak knee and hip loads. The aim of the study was to evaluate sex differences in landing from a jump in relation to lower extremity landing kinematics through neuromuscular fatigue. The higher indices of female jump height decrease and fatigue when comparing results during 100 jumps were ascertained. Women demonstrated lower peak hip, knee joint angles and lower EMG values in T2 phase when landing from a drop jump. Furthermore, correlations between H, EMG values and landing kinematics were generally higher for women than men.

Keywords: EMG, drop jump, kinematic.

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