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Lower limb loading during knee up in step aerobics: a pilot study

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Summary

Study aim: Step aerobics is a form of aerobic power distinguished from other types of aerobic exercise by its use of an elevated platform. The purpose of this study was to examine how the aerobic exercise "knee up" affects kinematic and kinetic parameters and, above all, the length of the muscle contractions.
Material and methods: The study analysed ten female fitness instructors with at least six years of experience. The task consisted in the knee up move performed using a 15 cm step and music with the beat frequency of 148 BPM. Kinematic and kinetic parameters were recorded using the Vicon system synchronized with two Kistler force plates. OpenSim software was used for calculation of the length of involved muscles.
Results: Ranges in angles and torques suggest that the location that is the most prone to injuries and overtraining is the knee joint, followed by the hip and ankle joints. Greater values of the vertical component of ground reaction forces were observed during stepping down, which suggests greater load to the joints. The greatest work in the move analysed in this study was performed by the sartorius muscle and the tensor fasciae latae muscle.
Conclusions: Despite the benefits that have been demonstrated when step classes are structured correctly and adapted to the participants, further research is needed concerning biomechanical load, exercise prescription, and injury prevention.

Key words: Force platform – Torque – Muscle length

Introduction

Sports biomechanics involves the study of recreational activity and requires laboratory representation of the real conditions of practice. Aerobics is one of the oldest forms of physical exercise, while one of its most recent modifications, termed step aerobics, has become a very popular type of physical activity. Analysis of a six-week step aerobics training programme revealed positive tendencies of changes in the index for evaluation of bone mineral density (BMD) [10, 24]. Therefore, this type of exercise can be used for osteoporosis prevention. However, around 76% of instructors and around 43% of examined participants of aerobics classes suffered from injuries connected with participation in this sport [23]. The study showed that around 60% of all injuries concerned lower limbs and 40% low back pain. The increase in the load in these locations may be caused, among other things, by poor movement technique [13, 24].

Several components of a step aerobics routine can be manipulated to change the intensity and loading of step workouts: bench height, stepping cadence and step movements. Bench heights usually range from 15 to 30 cm, and normal stepping cadences vary from 30 to 32 cycles/min (120 to 130 BPM). Different stepping patterns such as the basic step, knee up, straddle jump, and lunges are used to vary the intensity of the workout and the muscle groups involved in the step exercise [24]. Extrinsic factors, such as incorrect movements during sports activities, and anatomical factors, such as malalignment of the lower limbs or flexibility deficits and ligamentous laxity, are concomitant causes. The knee joint is most frequently affected by functional overload syndromes in adolescents as it is engaged in almost all sports activities [18, 22]. Therefore, coaches of step aerobics should be aware of the consequences of the increase in the height of the platform or music beat frequency, especially in the beginner groups. Santos Rocha [24] suggests that most of the injuries in step aerobics are caused by an inappropriate (excessively

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high) music beat frequency, complex choreography, an excessive number of high-impact steps and improper step height. It was demonstrated that the highest number of injuries is recorded among instructors who have more than 5 classes a week [24]. The faster the music is, the shorter is the time to perform the move, the higher is the muscular activity and, consequently, the greater are the differences between the muscle maximal and minimal length. So, the most frequent problem of overtraining is the atrophy of the medial heads of the quadriceps femoris muscle and the dominance of the lateral and intermediate head, with their work reinforced by the tensor fasciae latae muscle. This leads to the dominance of forces that draw the knee cap in the lateral direction, and the excessive lateral pressure syndrome [2, 5]. The knee cap is a type of block which reinforces the work of the thigh muscles. Its role is to reduce the friction resistance during knee extension, but it is also a load-bearing bone, with the pressure on the thigh that amounts to 30% of body weight during vigorous walking on a flat surface. An even greater load is observed for movement on an uneven surface: stepping down a 35 cm step causes that 3.5 times the body weight is transferred to the knee cap. Therefore, if the knee cap does not move on its proper course, it is prematurely worn down and chondromalacia patellae is observed, that is, softening of the cartilage in the knee joint [5].

The study of ground reaction forces and joint torques helps understand the magnitude and pattern of loading experienced by the body while in contact with the ground. Therefore, this study focused on the analysis of kinematic and kinetic parameters and, above all, the length of the muscle contractions of one of the basic step aerobics moves termed knee up. The OpenSim software [30, 31] was used for the purpose. The software allows for comprehensive visualisation of human movements, solving the inverse kinematics and inverse dynamics problem, simulation, and evaluation of joint reaction forces and forces of individual muscles that are difficult to measure in vivo.

Material and methods

The study analysed ten female fitness instructors with at least six years of experience (height = 165 ± 1.2 cm, mass = 68 ± 1.5 kg). The study was conducted according to the ethical guidelines and principles of the Declaration of Helsinki. The task was to perform the knee up move repeated 8 times without rest, alternately with the right and left leg on a standard 15 cm step. The rate of stepping up and down was forced by music with the beat frequency of 148 BPM (beats per minute). Analysis was carried out for the fifth step (the left foot stood on the platform and the right one was lifted up), considered as the most relaxed. Description of the step:

- Initial position – basic position facing the step, arms down, feet together, knees slightly bent.
- Step with the left leg moving forward and to the right. Stepping the foot on the step, starting from the heel.
- Raising the right knee only to the height of the hip.
- Stepping the right foot down on the ground.
- Stepping down with the left foot to the initial position on the floor [25].

First, anthropometric measurements were taken for each participant. Next, 34 spherical markers were placed at anatomical landmarks according to the biomechanical model PlugInGait standards available within the motion capture system (Vicon Motion Systems Ltd, Oxford, UK). Two force plates (Kistler Holding AG, Winterthur, Switzerland) were used to determine ground reaction force (GRF) data at a sampling rate of 100 Hz. One plate was embedded in the floor. On the other one, on the sensors, on the steel legs 15 cm high a wooden platform was mounted. The legs were fixed using a screw. This setting guaranteed the impossibility of movement of the step during exercise. The GRF data were filtered using a 4th-order low-pass Butterworth filter with a cut-off frequency of 10 Hz in order to smooth the force-time curves and eliminate the interfering noise caused by the vibration of the step structure [3]. For analysis purposes, one ascending and one descending cycle were selected.

A motion capture system, consisting of six infra-red cameras, was employed to collect kinematic data at a sampling rate of 60 Hz. The force plates were synchronized with the motion capture system. Both systems were calibrated according to the manufacturers' recommendations before the trials were conducted. The data from the platforms and the Vicon system were uploaded into the OpenSim software. Analysis and simulation of the knee up move was based on a 10-segment human body model with 23 degrees of freedom and 54 muscles (24 per leg, 6 muscles controlling body trunk movement with respect to the pelvis). In the OpenSim software, each musculotendinous unit is modelled as a Hill-type contractile element in series with the tendon. Driving each musculotendinous unit is a first-order differential equation which couples neutral excitation to muscle activation [2]. The values of maximum isometric strength, optimal fibre length, pennation angle and tendon slack length for each actuator are based on data reported by Delp [9]. Where possible, the action of a muscle group is reported as a single line of force. Many of the smaller muscles which originate on the tibia and insert on the foot were grouped into one. The inverse kinematics problem was solved in order to find the angles that best represent the movement performed by the participants. The inverse dynamics problem was used to evaluate muscle torques in the time domain. The data were corrected by the RRA algorithm (Residual Reduction Algorithm) in order to obtain dynamic coherence. The CMC algorithm (Computed Muscle Control) was used to compute the length of muscles during movement. Moreover, in order to examine the ranges of motion and loading of lower limbs the extreme values of kinematic and kinetic parameters were taken into account.

Results

Figure 1 presents average and 10 changes in angular profiles in the hip, knee and ankle joints during performance of the knee up move.

Analysis of joint angles shows that the greatest angular changes occur for the front leg (the right leg for all studied instructors). In the study group, the average difference between maximal and minimal values was 96.6 ± 1.5 deg for the knee joint, 90.2 ± 2.3 deg for the hip joint and 33.1 ± 2.1 deg for the ankle joint. For the left leg, the differences were much lower: 50.6 ± 0.5 deg, 64.9 ± 1.4 deg and 34.2 ± 0.9 deg for the hip joint, knee joint and ankle joint, respectively.

The mechanical load to the human motor system is associated with the frequency and value of internal and external forces that act on the human body. Figure 2 presents diagrams of the average vertical, anterior-posterior and medial-lateral components of ground reaction forces for the right and left leg, respectively.

On the vertical component of the L-GRF curve, the first peak F1 represents the first contact of the foot with the step, F2 represents transfer of the entire weight to the left leg, and F3 represents the beginning of stepping down. On the R-GRF curve, peak F4 represents the first contact of the foot with the floor, F5 represents transfer of the entire weight to the right leg and F6 represents the beginning of stepping up. Both curves are similar to those observed for the analysis of the kinetics of human gait [11]. However, it was noticeable that in the study group the average value of the F4 peak was 50% greater than the analogous F1 peak. Furthermore, other average values of peaks were greater for the leg placed on the step by 31% and 1% for F2 and F3, respectively. The maximum peaks for anteroposterior and mediolateral components of GRF are approximately a factor of 4 less as compared to the vertical component.

Apart from the solution of the inverse kinematics problem, the OpenSim software also allowed the inverse dynamics problem to be solved. Table 1 contains average maximal and minimal values of muscle torques standardized for body mass for the right and left lower limb, respectively. Negative values indicate flexion of the leg and dorsal flexion for the ankle joint. Positive values reflect extension of the leg and plantar flexion for the ankle joint.

Analysis of the results (Table 1) reveals that the greatest changes in torques are observed for the hip joints, and the lowest for the ankle joints. The mean range of torques in the hip joint, knee joint and ankle joint for the front leg is greater than the mean range of the corresponding torques for the joints of the support leg by 66%, 24% and 62%, respectively.

After performing specific simulations using the OpenSim software, the changes in the length of 54 muscles were obtained during the motion analysis (Table 2).

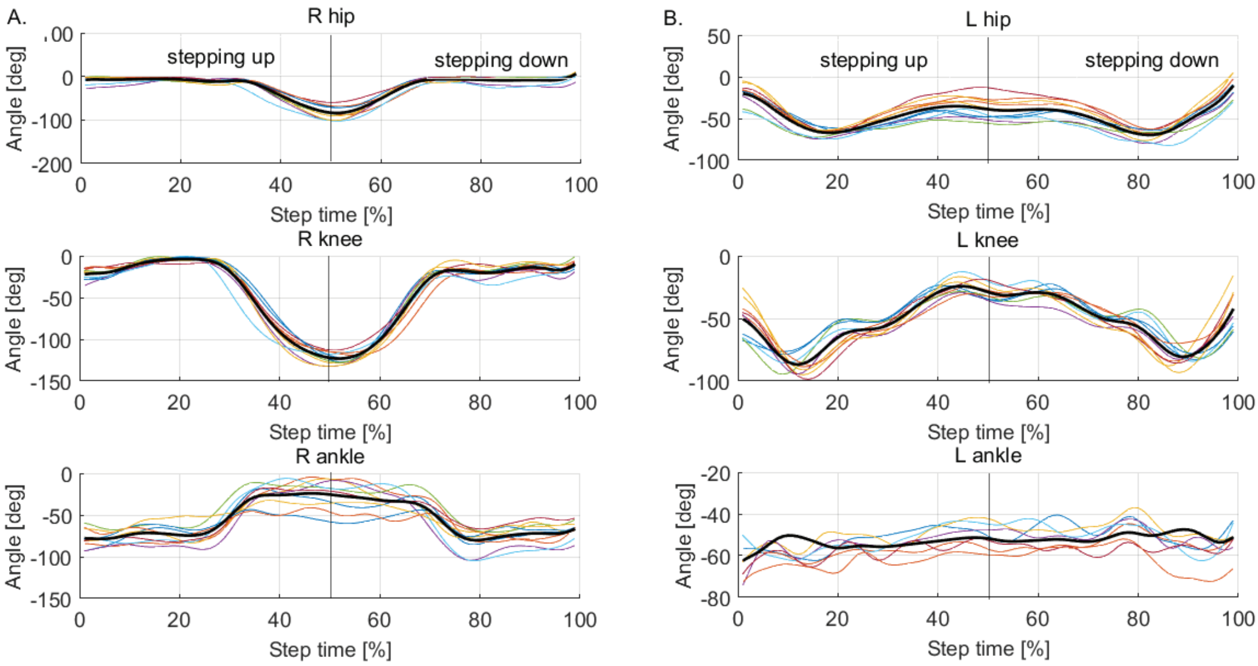

Fig. 1. Average and 10 changes in angular values in joints: A. The right leg lifted up; B. The left leg placed on the step in the phase of stepping up and down the platform

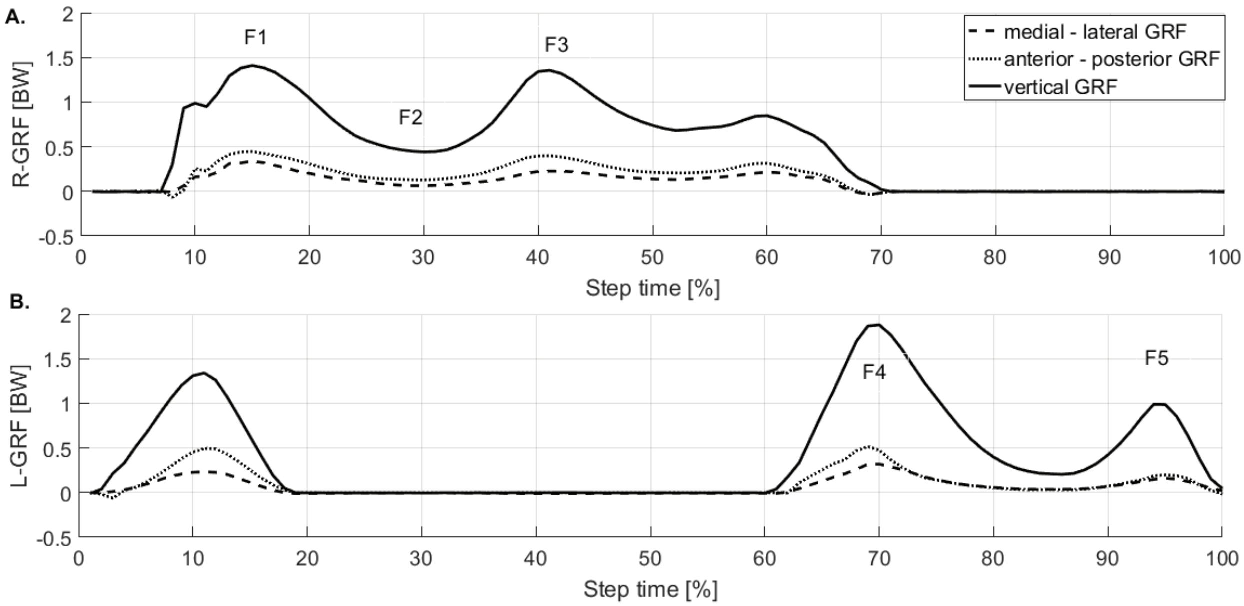

Fig. 2. Average changes in the vertical, anterior – posterior and medial – lateral components of ground reaction forces, for the right (R-GRF) and left (L-GRF) lower limb in the phases of A. stepping up and B. stepping down; F1 to F6: extreme values

Table 1. Average minimum and maximum values of muscle torques (mean ± SD). Negative values denote flexion; positive values denote extension

	Right hip joint	Left hip joint	Right knee joint	Left knee joint	Right ankle joint	Left ankle joint
Min [Nm/kg]	–5.20 ± 0.21	–2.11 ± 0.16	–0.42 ± 0.15	–0.40 ± 0.21	–1.12 ± 0.14	–0.24 ± 0.09
Max [Nm/kg]	2.80 ± 0.17	2.69 ± 1.01	2.97 ± 0.31	2.32 ± 0.12	1.21 ± 0.20	1.19 ± 0.21

The most pronounced changes in the length were found for the flexors of the right hip: sartorius muscle (0.12 m), tensor fasciae latae and gluteus maximus (0.09 m and 0.08 m). Similar tendencies were observed for the muscles of the left leg. The sartorius muscle and tensor fasciae latae changed their length by 0.08 m, whereas the length of the gluteus maximus changed by 0.05 m. The smallest differences were found for the tibialis posterior muscle of the right and left leg.

Discussion

This study was conducted for the purpose of examining kinematic and kinetic parameters during the knee raise move of female instructors. Therefore, it can represent an invaluable help for other athletes in general understanding of the movement and changes in the values of various body parameters. In the study there participated only ten female fitness instructors, who performed the exercise at the same step height at one music frequency. However, in many studies [19, 21, 27, 32, 33] that address the issue of muscles properties' simulation, or analysis of step aerobics movement [15, 16, 24, 26], the number of 10 participants is sufficient.

The average maximum values of ground reaction forces were: 1.4 BW for the stepping up phase and 1.7 BW for the stepping down phase (Figure 2). These values are close to those reported by Machado et al. [15], Santos Rocha and Veloso [26] for the music beat frequency of 140 BPM (1.5/1.6 BW) and close to the values presented during stairs walking. The maximum values of GRF for stair ascent were found to be between 1.2 and 1.7 BW, and between 1.4 and 2 BW for stair descent [6, 28]. Moreover, the biomechanical analysis showed that the vertical component of ground reaction forces during the knee raise is greater than in walking and lower than in running. During running, the maximal vertical component is around 3 BW, whereas in walking, this value is 1.2 BW [7, 14]. Also, in this paper, we reported that the maximum peaks for anteroposterior and mediolateral components of GRF during knee up are approximately a factor of 4 less than for the vertical component, which is 1.5 less than during walking. In the literature, there is no detailed description of these two components of ground reaction force during aerobic movements. Biomechanical research [12, 16, 24] has

Table 2. Average maximum and minimum length of 27 muscles (mean ± SD)

Muscle	Left lower limb		Right lower limb	
	Max length [m]	Min length [m]	Max length [m]	Min length [m]
Adductor magnus	0.25 ± 0.10	0.21 ± 0.08	0.25 ± 0.08	0.20 ± 0.02
Biceps Femoris Long Head	0.48 ± 0.20	0.42 ± 0.09	0.46 ± 0.14	0.41 ± 0.11
Biceps Femoris Short Head	0.23 ± 0.01	0.20 ± 0.02	0.23 ± 0.08	0.19 ± 0.02
Gluteus maximus 1	0.25 ± 0.11	0.22 ± 0.09	0.25 ± 0.02	0.20 ± 0.07
Gluteus maximus 2	0.27 ± 0.12	0.23 ± 0.03	0.27 ± 0.02	0.21 ± 0.02
Gluteus maximus 3	0.32 ± 0.10	0.27 ± 0.12	0.32 ± 0.11	0.24 ± 0.08
Gluteus medius 1	0.13 ± 0.03	0.11 ± 0.11	0.13 ± 0.03	0.11 ± 0.02
Gluteus medius 2	0.16 ± 0.04	0.14 ± 0.08	0.16 ± 0.05	0.13 ± 0.06
Gluteus medius 3	0.14 ± 0.05	0.11 ± 0.09	0.15 ± 0.01	0.10 ± 0.12
Gracilis	0.46 ± 0.12	0.42 ± 0.19	0.44 ± 0.09	0.40 ± 0.03
Iliacus	0.19 ± 0.08	0.15 ± 0.06	0.20 ± 0.08	0.13 ± 0.09
Gastrocnemius Medial Head	0.41 ± 0.06	0.39 ± 0.06	0.41 ± 0.04	0.39 ± 0.13
Pectineus	0.10 ± 0.04	0.08 ± 0.01	0.11 ± 0.09	0.07 ± 0.02
Piriformis	0.15 ± 0.07	0.13 ± 0.03	0.16 ± 0.02	0.12 ± 0.05
Psoas	0.25 ± 0.03	0.20 ± 0.01	0.26 ± 0.04	0.19 ± 0.02
Quadratus Femoris	0.10 ± 0.01	0.06 ± 0.04	0.11 ± 0.06	0.05 ± 0.01
Rectus Femoris	0.41 ± 0.10	0.36 ± 0.02	0.42 ± 0.01	0.40 ± 0.02
Sartorius	0.53 ± 0.11	0.45 ± 0.11	0.55 ± 0.02	0.43 ± 0.04
Soleus	0.28 ± 0.04	0.25 ± 0.02	0.29 ± 0.01	0.26 ± 0.08
Tensor Fascia Lata	0.52 ± 0.08	0.44 ± 0.03	0.53 ± 0.05	0.44 ± 0.10
Tibialis Anterior	0.30 ± 0.13	0.28 ± 0.12	0.29 ± 0.03	0.27 ± 0.01
Tibialis Posterior	0.32 ± 0.14	0.32 ± 0.13	0.32 ± 0.10	0.32 ± 0.09
Vastus Intermedius	0.23 ± 0.02	0.19 ± 0.07	0.25 ± 0.13	0.19 ± 0.14
Erector Spinae	0.16 ± 0.04	0.15 ± 0.07	0.17 ± 0.07	0.16 ± 0.03
External Obliques	0.26 ± 0.10	0.24 ± 0.05	0.27 ± 0.05	0.26 ± 0.11
Gemelli	0.07 ± 0.03	0.05 ± 0.04	0.08 ± 0.04	0.05 ± 0.02
Internal Obliques	0.19 ± 0.07	0.16 ± 0.05	0.21 ± 0.09	0.02 ± 0.01

shown that the ground reaction forces experienced during bench stepping are directly related to the step height and the type of manoeuvre [12, 15]. An increase in step height significantly reduces the loading rate and the peak force acting on the subjects' joints. Maybury and Waterfield [16] used step benches with step heights of 15, 20 and 25 cm and 120 BPM music tempo. They found statistically significant differences for peak impact force between the 15 and 20 cm and 15 and 20 cm heights, but not between 20 and 25 cm. No significant differences were found in any other parameters. Moreover, according to the results of Fujarczuk et al. [12] there is no significant influence of music tempo on the peak GRF in the step exercise. Therefore, the studies support the present advice that participants should use low step heights. For this reason, in the present work we decided to analyse only one aerobic movement at the most common height of step at one music tempo.

Analysis of muscle torques provides a more in depth insight into lower limb loading during knee rise. Comparison of peak torques for the knee up move to their corresponding values during walking, both forward and backward [4], shows that the torques for extension in the hip joint and knee joint are 449% and 781%, respectively, greater than for walking, whereas for the ankle joint, peak torque is 25% greater than its corresponding value during walking. In the flexion movement, the torques for the hip joint and ankle joint are greater for the knee up move by 631% and 124%, respectively. Furthermore, in the knee joint, the torque values recorded for walking were greater by 71%. Therefore, when opting for activity on the step it is worth knowing that overload syndromes are caused by repetitive micro trauma and they involve bone, cartilage, bursae, muscles and tendons [25]. The degree of muscle contraction depends on its length, with longer fibres enabling greater contraction. Maximal extension of the muscle is 1.6 times its natural length and maximal contraction of the muscle amounts to 0.5 of this value. The muscle works most effectively and has the greatest strength if its length is as close to its natural length as possible. Longer and weaker muscles are used to improve movement speed, while shorter muscles contract in a smaller space. Therefore, they have greater efficiency and are used for activities that need strength [2, 33]. A review of the literature has shown that few studies [1, 8, 17, 20, 29] go beyond a description of muscle length changing during a crouching gait, and no references were found in sport application and above all in knee up move analysis.

The study described in this paper showed that the greatest changes in the muscles length were found for the flexors of the right hip: sartorius muscle (0.12 m), tensor fasciae latae and gluteus maximus (0.09 m and 0.08 m). Similar tendencies are observed for the muscles of the left leg. The sartorius muscle and tensor fasciae latae changed their length by 0.08 m, whereas the length of the gluteus maximus changed by 0.05 m. The smallest differences were found for the tibialis posterior muscle of the right and left leg. Therefore, these muscles perform the greatest work. It is primarily related to the kinematics of the analysed motion. In the study group, the largest range of motion was in the sequence in the right knee (96.6 ± 1.5 deg), hip (90.2 ± 2.3 deg) and ankle (33.1 ± 2.1 deg) joint. For the left leg, the differences were much smaller: 50.6 ± 0.5 deg, 64.9 ± 1.4 deg and 34.2 ± 0.9 deg for the hip joint, knee joint and ankle joint, respectively. Therefore, customers of fitness clubs often have to undergo physiotherapy, since they exercise irregularly but intensively, which leads to ankle instability and knee joint pain. In this case, an insignificant difference in the exercise technique means substantial changes in the distribution of stresses in the knee joint. Therefore, it is necessary to ensure the smooth, harmonious work of the knee stabilizers that dynamically support the meniscus and the ligament system.

In conclusion: Ranges of angles and torques suggest that the knee joint is the most prone to injuries and overtraining. Greater values of the vertical component of ground reaction forces are observed during stepping down, which suggests greater load to the joints of the lower limbs. The most pronounced changes in the length for the knee up move were observed for the sartorius muscle and the tensor fasciae latae. The present study adds to the limited amount of published information about lower limb loading during knee up in step aerobics exercise and increases the knowledge base on the change in length of individual muscles. A still open and interesting problem is the analysis of the muscle force contribution during various step heights and/or various music frequencies.

References

1. Agarwal-Harding K.J., M.H. Schwartz, S.L. Delp (2010) Variation of hamstrings lengths and velocities with walking speed. *J. Biomech.*, 43(8): 1522-1526. DOI: http://dx.doi.org/10.1016/j.jbiomech.2010.01.008.
2. Barton C.J., P. Levinger, H.B. Menz, K.E. Webster (2009) Kinematic gait characteristics associated with patellofemoral pain syndrome: a systematic review. *Gait Posture*, 30(4): 405-16. DOI: 10.1016/j.gaitpost.2009.07.109.
3. Biomechanics, I.S.o., Signal Processing Software. 2016, International Society of Biomechanics.
4. Błażkiewicz M. (2013) Muscle force distribution during forward and backward locomotion. *Acta Bioeng. Biomech.*, 15(3): 3-9.
5. Bolgla L.A., T.R. Malone, B.R. Umberger, T.L. Uhl (2008) Hip strength and hip and knee kinematics during stair descent in females with and without patellofemoral pain syndrome. *J. Orthop. Sports Phys. Ther.*, 38(1): 12-8. DOI: 10.2519/jospt.2008.2462.
6. Christina K.A. P.R. Cavanagh (2002) Ground reaction forces and frictional demands during stair descent: effects of age and illumination. *Gait Posture*, 15(2): 153-158. DOI: http://dx.doi.org/10.1016/S0966-6362(01)00164-3.
7. Damavandi M., P.C. Dixon, D.J. Pearsall (2012) Ground reaction force adaptations during cross-slope walking and running. *Hum. Mov. Sci.*, 31(1): 182-189. DOI: http://dx.doi.org/10.1016/j.humov.2011.06.004.
8. Damiano D.L., E. Laws, D.V. Carmines, M.F. Abel (2006) Relationship of spasticity to knee angular velocity and motion during gait in cerebral palsy. *Gait Posture*, 23(1): 1-8. DOI: http://dx.doi.org/10.1016/j.gaitpost.2004.10.007.
9. Delp S.L., F.C. Anderson, A.S. Arnold, P. Loan, A. Habib, C.T. John, E. Guendelman, D.G. Thelen (2007) OpenSim: open-source software to create and analyze dynamic simulations of movement. *IEEE Trans. Biomed. Eng.*, 54(11): 1940-1950. DOI: 10.1109/TBME.2007.901024.
10. Dong L.W., L.C. Ping, C. Hui (2004) A study of aerobic dance on health conditions and related sickness of university female students. In: Pre-Olympic Congress Thessaloniki. Thessaloniki, pp. 6-11.
11. Forczek W., T. Ruchlewicz, A. Gawęda (2014) Kinematic gait analysis of a young man after amputation of the toes. *Biomed. Hum. Kinet.*, 6(1): 2080-2234. DOI: 10.2478/bhk-2014-0007.
12. Fujarczuk K., S. Winiarski, A. Rutkowska-Kucharska (2006) Ground reaction forces in step aerobics. *Acta Bioeng. Biomech.*, 8(2): 111-118.

13. Gröpel P., M. Wegner, J. Schüler (2016) Achievement motive and sport participation. *Psychol. Sport Exerc.*, 27: 93-100. DOI: http://dx.doi.org/10.1016/j.psychsport.2016.08.007.
14. Lopes A.D., L.C. Hespanhol Junior, S.S. Yeung, L.O. Costa (2012) What are the main running-related musculoskeletal injuries? A Systematic Review. *Sports Med.*, 42(10): 891-905. DOI: 10.2165/11631170-000000000-00000.
15. Machado M., H. Santos, F. Babtista, A.P. Veloso, Biomechanical effects of step exercise program of elderly women, in The 7th Annual Congress of the ECSS, P.0286/188, Editor. 2002: Athens.
16. Maybury M. J. Waterfield (1997) An investigation into the relation between step height and ground reaction forces in step exercise: a pilot study. *Br. J. Sports Med.*, 31: 109-113.
17. Moreau N.G., M.J. Falvo, D.L. Damiano (2012) Rapid force generation is impaired in cerebral palsy and is related to decreased muscle size and functional mobility. *Gait Posture*, 35(1): 154-158. DOI: http://dx.doi.org/10.1016/j.gaitpost.2011.08.027.
18. Pecina M., I. Bojanic, M. Haspl (2001) [Overuse injury syndromes of the knee]. *Arch. Hig. Rada. Toksikol.*, 52(4): 429-439.
19. Pizzolato C., M. Reggiani, L. Modenese, D.G. Lloyd (2014) Real-time estimation of lower limb joint angles through inverse kinematics during walking using a scaled OpenSim model. *J. Sci. Med. Sport*, 18(Supplement 1): e142. DOI: http://dx.doi.org/10.1016/j.jsams.2014.11.143.
20. Rabey K.N., D.J. Green, A.B. Taylor, D.R. Begun, B.G. Richmond, S.C. McFarlin (2015) Locomotor activity influences muscle architecture and bone growth but not muscle attachment site morphology. *J. Hum. Evol.*, 78: 91-102. DOI: http://dx.doi.org/10.1016/j.jhevol.2014.10.010.
21. Reinbolt J.A., A. Seth, S.L. Delp (2011) Simulation of human movement: applications using OpenSim. *Procedia IUTAM*, 2: 186-198. DOI: http://dx.doi.org/10.1016/j.piutam.2011.04.019.
22. Relph N. L. Herrington (2016) The effects of knee direction, physical activity and age on knee joint position sense. *The Knee*, 23(3): 393-398. DOI: http://dx.doi.org/10.1016/j.knee.2016.02.018.
23. Richie D.H., S.F. Kelso, P.A. Bellucci (1985) Aerobic dance injuries: a retrospective study of instructors and participants. Phys. Sports Med., 13(2): 130-140. DOI: http://dx.doi.org/10.1080/00913847.1985.11708751.
24. Santos-Rocha R.A., Assessment of Biomechanical Loading in Musculoskeletal Structures of the Lower Limb. Biomechanical Model of Step Exercise., in Faculty of Human Kinetics. 2006, Technical University of Lisbon, Portugal.
25. Santos-Rocha R.A., C.S. Oliveira, A.P. Veloso (2006) Osteogenic index of step exercise depending on choreographic movements, session duration, and stepping rate. *Br. J. Sports Med.*, 40(10): 860-866; discussion 866. DOI: 10.1136/bjsm.2006.029413.
26. Santos Rocha R.A.V., A. (2005) Plantar Pressure and Peak Ground Reaction Forces in Step Exercise. In: H. Rodrigues, M. Cerrolaza, M. Doblaré (eds.) Proceedings of the ICCB – II International Conference on Computational Bioengineering. Vol 2, Lisbon, IST Press, pp. 885-894.
27. Seth A., M. Sherman, J.A. Reinbolt, S.L. Delp (2011) OpenSim: a musculoskeletal modeling and simulation framework for in silico investigations and exchange. *Procedia IUTAM*, 2: 212-232. DOI: http://dx.doi.org/10.1016/j.piutam.2011.04.021.
28. Stacoff A., C. Diezi, G. Luder, E. Stussi, I.A. Kramers-de Quervain (2005) Ground reaction forces on stairs: effects of stair inclination and age. *Gait Posture*, 21(1): 24-38. DOI: 10.1016/j.gaitpost.2003.11.003.
29. Steele K.M., M.M. van der Krogt, M.H. Schwartz, S.L. Delp (2012) How much muscle strength is required to walk in a crouch gait? *J. Biomech.*, 45(15): 2564-2569. DOI: http://dx.doi.org/10.1016/j.jbiomech.2012.07.028.
30. Thelen D.G., F.C. Anderson (2006) Using computed muscle control to generate forward dynamic simulations of human walking from experimental data. *J. Biomech.*, 39(6): 1107-1115. DOI: 10.1016/j.jbiomech.2005.02.010.
31. Thelen D.G., F.C. Anderson, S.L. Delp (2003) Generating dynamic simulations of movement using computed muscle control. *J. Biomech.*, 36(3): 321-328.
32. Van der Krogt M., A. Seth, K. Steele, L. Bar-On, K. Desloovere, J. Harlaar, S. Delp (2013) A model of muscle spasticity in opensim. *Gait Posture*, 38(Supplement 1): S16. DOI: http://dx.doi.org/10.1016/j.gaitpost.2013.07.039.
33. Van der Krogt M.M., C.A.M. Doorenbosch, J. Harlaar (2007) Muscle length and lengthening velocity in voluntary crouch gait. *Gait Posture*, 26(4): 532-538. DOI: http://dx.doi.org/10.1016/j.gaitpost.2006.11.208.

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