NEW TECHNOLOGIES IN TRAUMATOLOGY AND ORTHOPEDICS

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VIDEO-ANALYSIS OF THE EFFECT OF DIFFERENT TYPES OF ADAPTED SHOES ON KNEE ADDUCTION MOMENT

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Background. The effect of different footwear profiles on knee adduction moment have not been fully studied.
Methods. Fifteen healthy volunteer subjects, age 25.3 (±2.73), undertook a series of gait laboratory trials with adapted shoes. Kinematic and kinetic data were collect using 16 Oqus 3+ cameras and the walking speed was controlled using timing gates. High street shoes were adapted to include five different heel heights (varying from a 1.5 cm to 5.5 cm heels), two heel profile conditions (curved and semi-curved heels), three varying apex angles (10, 15, and 20 degrees), and barefoot and 3CR footwear conditions. The baseline shoe had no heel curve, a heel height of 3.5cm, an apex position of 62.5% of the shoe length, an apex angle of 15 deg, and a rigid forepart of the shoe.
Results. The shoe with 5.5 cm heel height significantly increased the mean knee adduction moment during 50%–100% of the stance phase compared to the 1.5 cm heel ($p = 0.008$). The high heel shoe also significantly increased knee adduction impulse (area under the curve) versus the 1.5, 2.5, and 3.5 cm heels, and the 10° toe angle and barefoot condition. Ten degrees of toe angle reduced mean knee adduction moment during 0%–50% of the stance phase versus 20° and significantly reduced mean knee adduction moment during the late stance phase versus 15° and 20° toe angle footwear conditions. Walking with the curved heel for the healthy subjects increased mean knee adduction moment during 0%–50% of the stance phase compared to the heel without curvature ($p < 0.0009$).
Conclusion. Further study is required to investigate those changes in patients with high risk of knee osteoarthritis.

Keywords: knee osteoarthritis, knee adduction moment, video analysis, kinetics, footwear profiles.

ПРИМЕНЕНИЕ ВИДЕОАНАЛИЗА С ЦЕЛЬЮ ОПРЕДЕЛЕНИЯ ОПТИМАЛЬНОЙ КОНСТРУКЦИИ ОБУВИ ДЛЯ СНИЖЕНИЯ АДДУКЦИОННОГО МОМЕНТА КОЛЕННОГО СУСТАВА

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Введение. Для повышения эффективности лечения и реабилитации пациентов, страдающих остеоартрозом, часто рекомендуется использовать ортопедические стельки. Лечебный эффект в этом случае достигается за счет уменьшения аддукционного момента, воздействующего на биомеханическую ось нижней конечности, благодаря особой конструкции обуви, параметры которой определяются с учетом биомеханических параметров локомоции пациента. Однако к настоящему времени в научных источниках отсутствует достаточно убедительная информация о том, как влияют различные параметры конструкции обуви на нагрузку коленного сустава.
Методология: видеоанализ ходьбы при использовании обуви с различной высотой каблука, уровнем переката носка и пятки.
Результаты. В исследовании принимали участие 15 здоровых мужчин. Исследование показало, что аддукционный момент возрос при ходьбе в обуви с каблуком 5,5 см, а также с увеличенным перекатом пятки и углом носка $p < 0,05$. Ортопедическая обувь 3CR, которая включает все вышеупомянутые факторы, в значительной степени повысила суммарный медиальный момент и аддукционный момент колена $p < 0,05$.

Заключение: использование неправильно подобранной обуви и ортопедических средств в течение длительного времени может существенно способствовать развитию остеоартроза коленного сустава.

Ключевые слова: остеоартроз коленного сустава, аддукционный момент, видеоанализ, кинетика, конструкция обуви.

Introduction

Osteoarthrosis of the knee joint is one of the most common orthopedic diseases in the elderly and represents a great medical and social problem [1]. Approximately 9.6% of men and 18% of women suffer from osteoarthrosis of the knee joint between 50–60 years of age and older [2]. This disease leads to a disorder in the functionality of the musculoskeletal system, reducing the patient's mobility by an average of 80%, due to severe pain while walking. The first symptoms of osteoarthrosis of the knee joint often appear at the age of 38–40 years [3]. Currently, an effective method for treating this disease has not yet been found [4].

The results of scientific research have shown that the medial part of the knee joint develops complications 5 times more than the lateral one [5-7].

The factors for risk and progression of osteoarthrosis of the knee joint include: age, female sex, obesity, high physical activity, sedentary lifestyle (weak leg muscles), high bone weight index and their density, traumas, hormonal treatment, vitamin D deficiency, smoking, and intense sports [2, 8]. There is a hypothesis that the development of osteoarthrosis of the knee joint is almost always directly related to an increase in the load on the joint, for example with abnormal anatomy (acquired or congenital), with intense sports, traumas, and obesity [9]. Nevertheless, the exact cause for the development of osteoarthrosis of the knee joint remains unclear.

One of the most pronounced biomechanical changes with osteoarthrosis is an increase in the adduction (medial) moment of the knee joint [7, 10-12]. An increase in the value of the medial moment during walking contributes to an increase in the strength of muscle generation, which is necessary to overcome the load and can significantly affect the total load on the knee joint. A meta-analysis of 1,078 articles showed a positive correlation with the development of osteoarthrosis and increased knee load for 12–72 months in 452 patients who participated in the studies [13]. These studies had certain limitations, which could affect the outcome of the results.

Nevertheless, to reduce the rate of progression of osteoarthrosis and to reduce the need for surgical interventions, it is recommended to reduce the medial moment [14]. A decrease in the adduction moment can be achieved through physical rehabilitation and muscle training, but studies on the participation of 233 patients did not confirm this hypothesis [15]. Significant changes in the magnitude of the medial moment were achieved through the use of various orthopedic products [14, 16-18].

At the National Institute of Clinical Excellence (NICE UK), advice on the use of orthopedic insoles and special footwear is included in the recommendations for the rehabilitation of the knee joint and reducing the risk of developing the disease [19]. For example, orthopedists use correcting insoles to correct the initial valgus curvature of the ankle joint, leading to reduced knee joint adduction and can be used to reduce rolling of the mass center of the foot (Fig. 1) [16, 17].

To date, the "gold standard" for accurate clinical diagnosis of musculoskeletal disorders is the use of video analysis methods together with force plates [20]. This method enables one to determine accurately the kinematic changes in the joints and calculate their loads.

With the help of these systems, researchers have attempted to create orthopedic insoles, which can reduce joint pain during walking by reducing the adduction moment. The results of these studies have been highly variable; therefore, an individual approach to each patient is recommended. The design of shoes can also lead to changes in the biomechanical parameters during walking due to various characteristics (thickness of the sole, angle of rolling, the length of the metatarsal zone rolling, the height of a heel, and stiffness) [21-25]. The information on how the structural features of footwear affect the adduction moment of the knee joint is represented by a very small number of publications.

The aim of the scientific research is to study the influence of the structural features of footwear on the kinetic and kinematic parameters of the knee joint. This information can be used for further

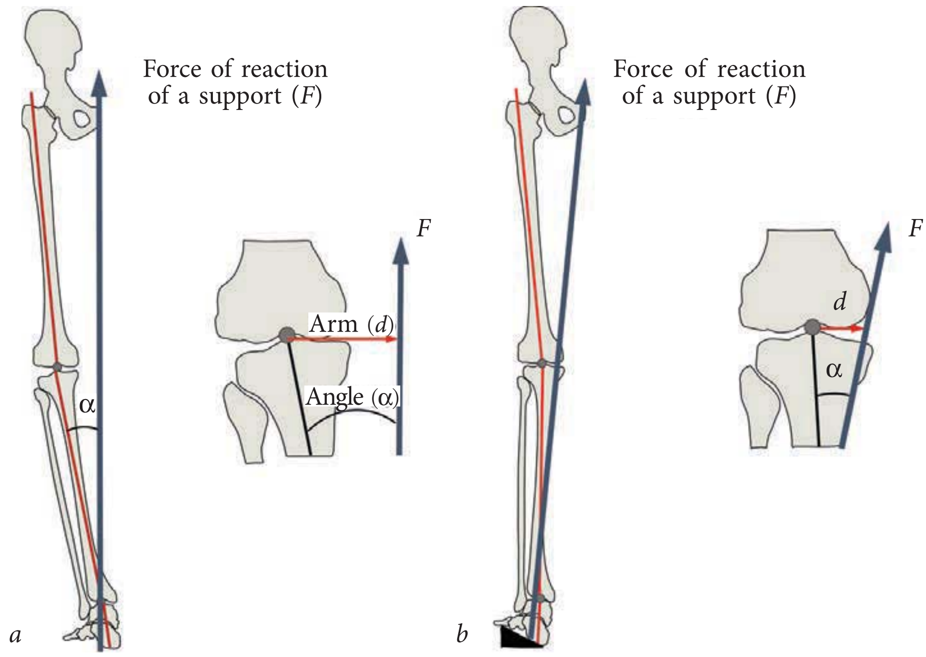

Fig. 1. Direction of the reaction force of the support relative to the knee joint for calculation of the adduction moment ($M = F \cdot d$): *a* - varus curvature of the knee joint; *b* - decrease in the varus, due to the use of an orthopedic insole

analyses to determine possible factors for the development of osteoarthrosis and the development of optimal footwear that helps to reduce the load on the knee joint.

Methodology

The study was conducted in the clinical and biomechanical laboratory of Salford University (University of Salford, UK). All patients voluntarily signed the informed consent to participate in the study. The ethical justification for the experimental protocol developed and its safety for the subjects were approved by the ethics committee of the University (experimental protocol ETHICS APPLICATION HSCR12/04 - An investigation into the relationship between footwear features and lower limb muscle action and activity, April 2012) [26]. The general view of the laboratory and the features of the location of the system components for video analysis of human movement are shown in Fig. 2.

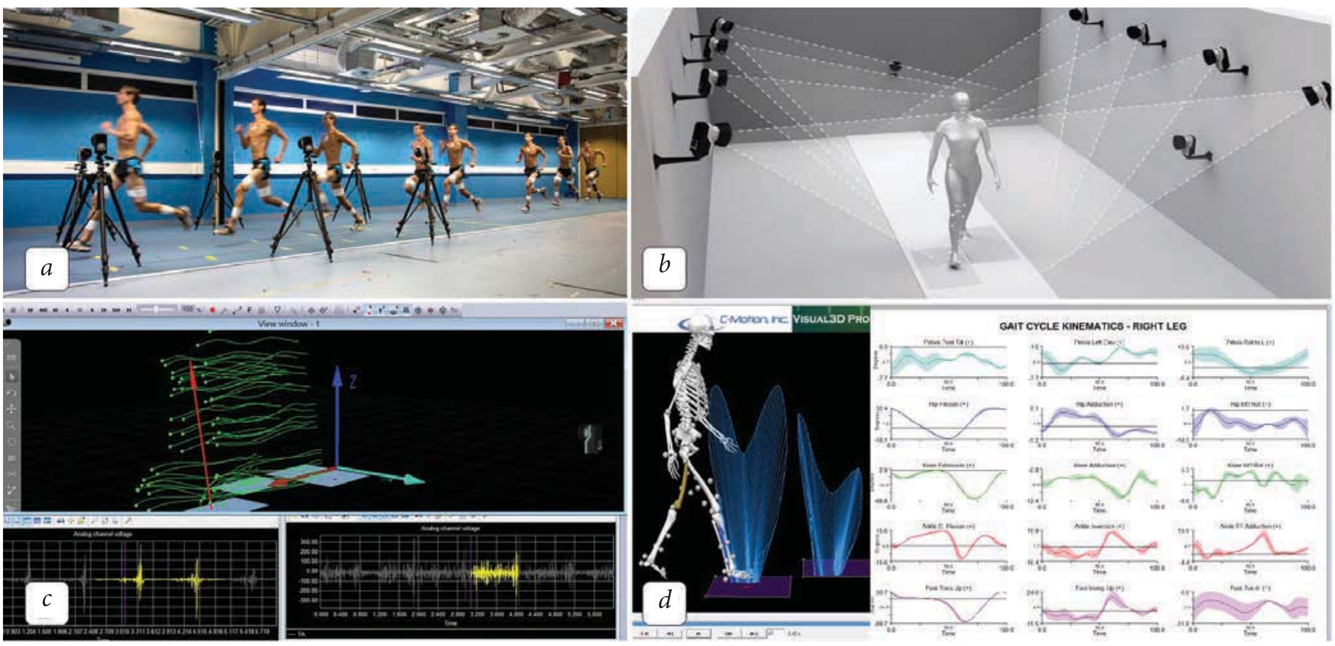

Fig. 2. An example of using the video analysis system: *a* is a general view of the video analysis laboratory; *b* is the layout of the arrangement of cameras and force plates; *c* is the Qualisys (Sweden) program window for recording real-time movement; *d* is Visual3D program window (C-Motion, USA) with a human model and the examination results

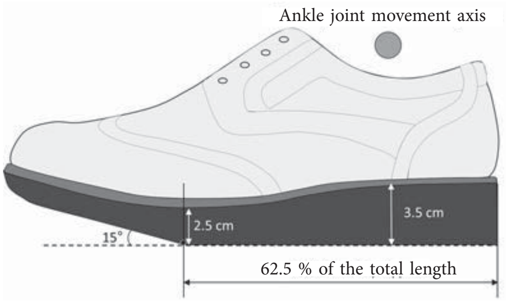

Fig. 3. Design of the control footwear

The study involved 15 men aged 25.3 ± 2.73 years with a body weight of 71.3 ± 8.5 kg, a height of 1.74 ± 0.06 m and a foot size corresponding to the shoe size 8 in the European system. The kinematic data were recorded in three planes using 16 high-speed infrared cameras of Qualisys (Sweden) OQUS™ 3+ with a set frequency of 100 Hz. The recording of the kinetic data was performed using four force plates with the frequency of measurement of 1000 Hz (AMTI, Watertown, MA, USA, model BP600400).

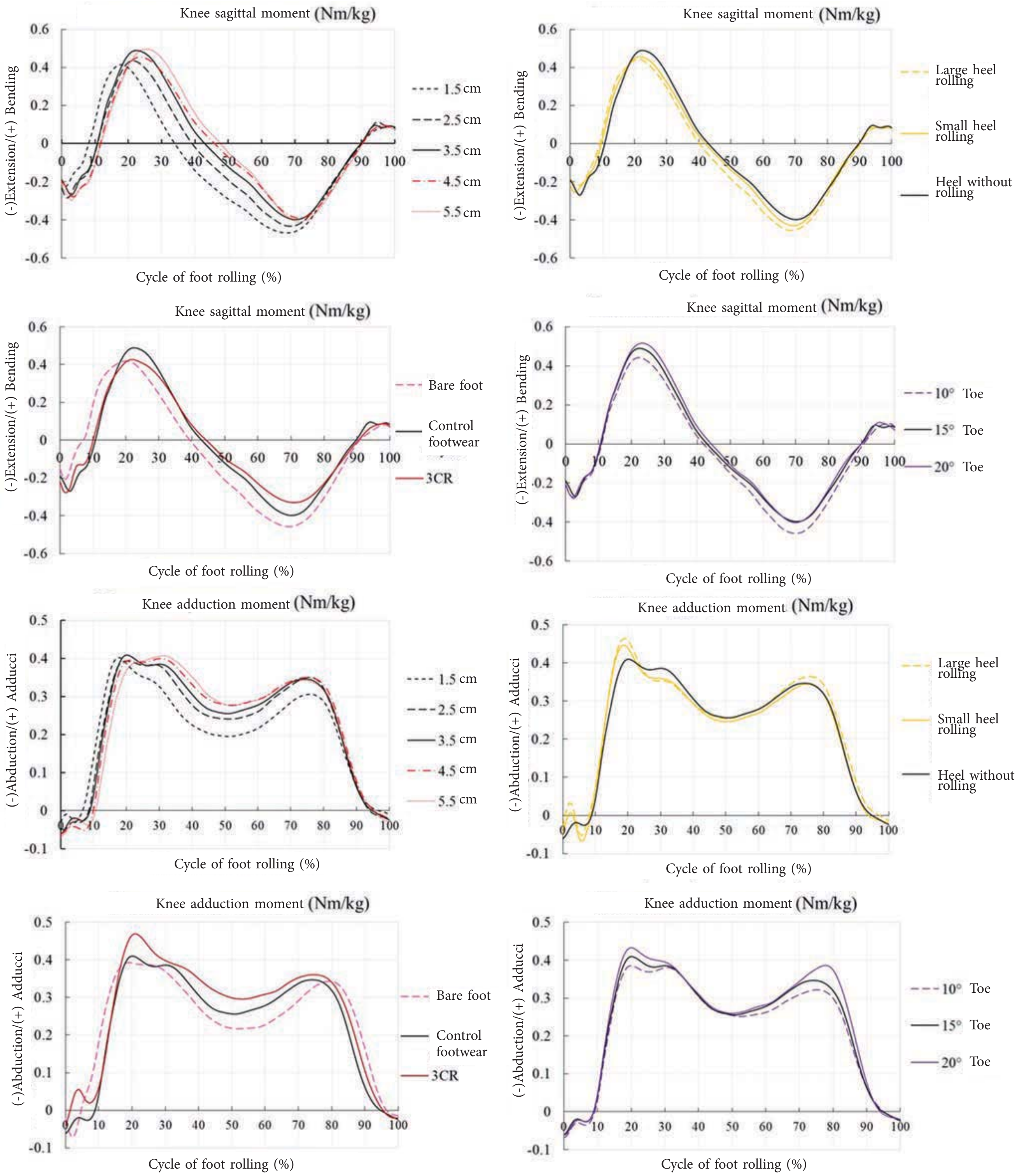

Fig. 4. Results of knee joint moment in sagittal and frontal planes for various designs of footwear

In the process of biomechanical research, five pairs of shoes with different heel heights (1.5, 2.5, 3.5, 4.5, and 5.5 cm), three pairs of shoes with different toe angles (10°, 15°, and 20°), three pairs of shoes with different levels of heel rolling, the orthopedic footwear 3CR [27], and walking without shoes were tested. The control shoes had the following characteristics: the distance from the heel to the metatarsal area was 62.5% of the total length, the height of the heel was 3.5 cm, and the angle of the toe rise was 15° (Figure 3).

Walking speed was set at 5 km/h ± 2.5% to reduce changes in the characteristics of the musculoskeletal system [28, 29]. The data were processed in a Visual3D software package (C-Motion, USA). In processing, the kinematic data were filtered by a fourth-order low-frequency Butterworth filter with a cutoff frequency of 12 Hz. The kinetic data were filtered at a frequency of 25 Hz. The results obtained were analyzed in accordance with international standards of biomechanics of movements.

The pulse count (the graph curve area) of the energy and the moment of the ankle joint was calculated using the trapezoidal formula to calculate the definite integrals [30]. The statistical data were processed in the IBM SPSS statistics V.23 program (IBM, USA) by the single-factor analysis of variance (ANOVA) with Bonferroni correction. The level of statistical significance was established as $p < 0.05$.

Results

Table 1 and Figure 4 show differences in the biomechanical indexes of the knee joint for different footwear designs, which were different from each other in the heel height, in the angles of toe rolling, in the heel rolling levels, in the use of orthopedic footwear 3CR (UK), and in the use of walking without shoes.

Table 1

Results of the knee joint moment for various footwear designs: the number outside the brackets is the mean value for 15 patients, in the brackets is the mean square deviation ($\mu(\pm$ SD))

Biomechanical parameters	Heel height		
Statistically significant when in the index	1	2	3
Moment of the knee joint (Nm/kg)	1.5 cm (1)	3.5 cm (2)	5.5 cm (3)
Max. bending moment	0.431 (0.078) $^{2,\,3}$	0.501 (0.074) ^{1}	0.516 (0.064) ^{1}
Max. extension moment	–0.488 (0.05)	–0.446 (0.043)	-0.468 (0.05)
Max. moment of adduction between (0%–50% of the step cycle)	0.423 (0.029)	0.44 (0.03)	0.444 (0.033)
Max. moment of adduction between (50%–100 % of the step cycle)	0.327 (0.028) $^{2,\,3}$	0.379 (0.029) ^{1}	0.382 (0.029) ^{1}
Impulse of adduction of the whole step cycle	14.16 (0.89) $^{2,\,3}$	16.72 (0.98) ^{1}	17.21 (1.05) ^{1}
Impulse of abduction of the whole step cycle	–0.27 (0.11) $^{2,\,3}$	–0.38 (0.13) $^{1,\,3}$	-0.48 (0.14) ^{1,2}
Moment of the knee joint (Nm/kg)	10° toe (1)	15° toe (2)	20° toe (3)
Max. bending moment–	0.456 (0.052) ^{3}	0.501 (0.074)	0.531 (0.068) ^{1}
Max. extension moment	–0.499 (0.046)	–0.446 (0.043)	-0.452 (0.04)
Max. moment of adduction between (0%–50% of the step cycle)	0.428 (0.026) ^{3}	0.44 (0.03)	0.465 (0.039) ^{1}
Max. moment of adduction between (50%–100% of the step cycle)	0.358 (0.029) $^{2,\,3}$	0.379 (0.029) ^{1}	0.405 (0.024) ^{1}
Impulse of adduction of the whole step cycle	15.99 (0.85) ^{3}	16.72 (0.98)	17.54 (1.0)
Impulse of abduction of the whole step cycle	–0.45 (0.12)	–0.38 (0.13)	-0.42 (0.14)
Moment of the knee joint (Nm/kg)	Heel without rolling (1)	Small heel rolling (2)	Large heel rolling (3)
Max. bending moment	0.501 (0.074)	0.472 (0.066)	0.464 (0.068)
Max. extension moment	–0.446 (0.043)	–0.468 (0.044)	-0.481 (0.042)
Max. moment of adduction between (0%–50% of the step cycle)	0.44 (0.03) ^{3}	0.471 (0.034)	0.481 (0.034) ^{1}
Max. moment of adduction between (50%–100% of the step cycle)	0.379 (0.029)	0.374 (0.029)	0.39 (0.031)
Impulse of adduction of the whole step cycle	16.72 (0.98)	16.27 (1.04)	16.93 (1.03)
Impulse of abduction of the whole step cycle	–0.38 (0.13)	–0.42 (0.11)	–0.3 (0.11)

Moment of the knee joint (Nm/kg)	Bare foot (1)	Control footwear (2)	Orthopedic footwear 3CR (3)
Max. bending moment	0.429 (0.046)	0.501 (0.074)	0.451 (0.075)
Max. extension moment	–0.479 (0.042) [3]	–0.446 (0.043)	–0.396 (0.051) [1]
Max. moment of adduction between (0%–50% of the step cycle)	0.433 (0.032) [3]	0.44 (0.03)	0.495 (0.031) [1]
Max. moment of adduction between (50–100 % of the step cycle)	0.365 (0.027)	0.379 (0.029)	0.387 (0.026)
Impulse of adduction of the whole step cycle	15.82 (1.12) [3]	16.72 (0.98) [3]	18.13 (0.86) [1,2]
Impulse of abduction of the whole step cycle	–0.21 (0.07) [2]	–0.38 (0.13) [1, 3]	–0.22 (0.08) [2]
[1–3] Shows the statistical significance between the tested shoe designs. $p < 0.05$			

The analysis of the study results showed that increasing the heel height led to the moment of bending and the maximum adduction moment between 50%–100% of the step cycles. Heel height also significantly increased the impulse of the adduction moment (the area of the adduction graph). The increase in the angle of rolling of the footwear toe also led to a significant increase in the moment both in the sagittal and frontal planes, $p < 0.05$. The results of increasing the value of the adduction moment in the first half of the foot rolling cycle when walking in shoes with a large heel roll should be particularly emphasized. The use of orthopedic footwear 3CR, used to rehabilitate patients with intermittent claudication, greatly increased the adduction moment of the knee joint in comparison with other shoe designs $p < 0.05$.

Discussion

This article focuses on the results of increasing the knee joint adduction moment for various shoe designs.

In the previous study, the authors showed that the height of the heel significantly increases the turnout of the foot [31]. Based on these data, it was suggested that the height of the heel also changes the trajectory of the distribution of the center of mass of the foot and increases the moment arm of the ankle joint, leading to an increase in the knee joint varus when walking. Ultimately, all this contributes to an increase in the maximum values of the medial moment of the knee joint, which may be an unfavorable factor for people with high risk of development of osteoarthrosis of the knee joint.

In addition, data showed that an increase in the shoe toe rolling leads to an increased plate-flexion (bending) of the ankle joint, resulting in significant kinematic and kinetic changes in the parameters of the knee joint in the interval of the foot rolling cycle between 50%–100% [26]. In this case, the increase in the rolling of the shoe toe to 20° leads to an increase in maximum values of the adduction moment in comparison with the angle of 10°, $p = 0.01$. Most sports running footwear have an increased toe angle, so they should not be recommended for use in people with knee joint injuries.

Data also showed that increased heel rolling led to an increase in the adduction moment in the first half of the foot rolling cycle, and changes in the biomechanics of the ankle and knee joints [26].

Orthopedic footwear 3CR has a small heel rolling, which resulted in an increase in the maximum knee adduction moment in the first half of the foot rolling cycle. This footwear has an increased heel (5.5 cm) and a toe angle rolling of approximately 18°, leading to a considerable degree of increase in the total area of the adduction moment (impulse) of the knee joint when compared to walking without shoes and a control pair of shoes (angle of the toe is 15°, the heel height is 3.5 cm).

The study was conducted on men who did not have musculoskeletal disorders, in order to reduce the influence of additional factors on the results obtained. Additional work should be done with the involvement of patients with pathologies of the knee joint.

Previous studies have shown that the use of the orthopedic products reduced the value of the adduction moment, but there was not always a reduction in knee pain. This suggests that there are other factors affecting the load of the knee joint. Therefore, the next stage of the studies is planned to use 3D simulation of the walking process in various shoes, using the OpenSim program (Stanford

University), which allows one to calculate the total load on a joint, consisting of the forces of gravity and the work of the muscle-tendon complexes.

Conclusion

This study showed that the use of orthopedic footwear of various designs contributes to a significant increase or decrease in the load of the medial part of the knee joint. Such design features as an increased heel height, an increased angle of toe rolling, and also even a slight change in the footwear heel rolling, led to a marked increase in the load on the medial part of the knee joint. Therefore, when selecting orthopedic footwear for people with various functional disorders of the musculoskeletal system, the design features of the footwear should be considered.

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Conflict of interest. The authors declare no conflict of interest.

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