Original Article

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Changes in Muscle Activity of Selected Muscles during Walking in High-Heeled Shoes on a Treadmill

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Abstract: The aim of the study was to investigate how walking in high heels on a treadmill affects the changes in timing of activation in selected lower limb muscles, pelvic muscles, and upper trunk muscles in young women. Surface electromyography (EMG) was used for data collection of selected muscles on the right half of the body: m. pectoralis major, m. trapezius pars transversa, m. obliquus abdominis externus, m. erector spinae in the lumbar spine, m. gluteus medius, m. gluteus maximus, m. rectus femoris, m. biceps femoris-caput longum. The research group consisted of 30 women (age: 24.2 ± 2.3 years; weight: 57 ± 3.2kg; height: 1.65 ± 0.04m). Statistical significance in changes in timing was confirmed for seven of the measured muscles. The timing of the m. trapezius muscle was not statistically affected by changes in walking in high heels (HH), but HH and treadmill walking increased the intensity of muscle contraction in all monitored muscles. Walking on a treadmill in flat shoes (FS) similarly increases the intensity of muscle contraction. This work expands the theoretical knowledge of bipedal locomotion in HH on a treadmill.

Keywords: high heels, gait cycle, muscle activity, treadmill

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INTRODUCTION

Walking in high heels significantly alters the natural gait cycle, leading to shorter steps, reduced walking speed, and changes in the timing of lower limb muscle activation compared to walking in flat shoes [1-5]. According to the theory of muscle synergies [6,7], high heels not only affect the lower body but also disrupt the coordination of upper body muscles [8]. Muscle synergies are strategies employed by the central nervous system (CNS) to manage complex movements more efficiently [9,10], and analyzing these synergies offers insight into motor and postural control during dynamic activities like walking [11].

While treadmill walking has provided valuable insights into human locomotion, many studies focus on natural walking conditions, where participants move freely over ground surfaces [12,13]. The unique combination of wearing high heels (HH) and walking on a treadmill presents a particularly challenging scenario, altering muscle activity patterns and the timing of muscle activation and deactivation. The treadmill environment itself modifies the gait cycle [2], and walking in HH further increases the intensity of muscle work, as high heels create a more unstable position for the body [14].

In this study, we measured the intensity of muscle contractions using EMG microvolt readings. We focused on specific muscles involved in controlling the upper limbs and trunk, which play a key role in the cross-pattern walking movement. These muscles form part of chains running from the feet to the head along the body's front and back [10], and were chosen for their critical role in bipedal locomotion. Despite the abundance of research on walking in high heels on flat surfaces, there is a notable gap in the literature regarding muscle activity during treadmill walking in high heels [15,16]. Our study aims to explore how the body and CNS adapt to the combined challenges of high heels and treadmill walking, focusing on changes in muscle timing and coordination under these unstable conditions.

The main objective of the study was to determine and elucidate how walking in high heels (HH) on a treadmill alters timing of selected muscle groups (m. pectoralis major, m. trapezius pars transversa, m. obliquus abdominis externus, m. erector spinae in the lumbar spine, m. gluteus medius, m. gluteus maximus, m. rectus femoris, m. biceps femoris-caput longum), compared to walking in flat shoes (FS). Another goal was to investigate how the intensity of muscle contraction of the monitored muscles changes during walking in HH on a treadmill, as there is a scarcity of studies addressing similar topics. We set two research questions: Does walking on a treadmill in HH change the timing of all muscle groups studied compared to walking in FS and will muscle contraction intensity increase during treadmill walking in HH compared to FS walking?

MATERIALS AND METHODS

Participants

The sample consisted of 30 healthy women (age: 24.2 ± 2.3 years; weight: 57 ± 3.2 kg; height: 1.65 ± 0.04 m). All participants were categorized as less experienced wearers of high heels, wearing high heels less than 4 times a week and for less than 4 hours in the past year [17]. We examined a total of 40 participants, and we excluded 10 females from the sample who had a history of lower limb injury or flat feet. The participants did not have faulty posture, scoliosis, psychological problems, were healthy without any respiratory or cardiological problems. To maintain maximum objectivity, we provided the participants with identical measurement conditions and footwear. All participants were given the same type of high-heeled shoes with a heel height of 7 cm and flat shoes with a heel size of 2cm. The selection of casual flat shoes followed the same procedure. The chosen sizes for high heels and casual flat shoes were 37, 38, and 39 (EU) (Figure 1). The study was approved by the Institutional Ethics Committee, and all participants read and completed informed consent forms.

Measurement protocol

Data collection during walking in both types of footwear on the treadmill for all participants took place on the same day, indoors, without any disruptive external environmental influences.

Participants walked on the treadmill (ENERGETICS PW 880 treadmill (Woelflistrasse 2, CH-3006 Bern, Switzerland) at a 0% incline. Fixed walking speed v=3.6 km/h (1 m/s). The walking speed was uniformly set for all participants on the treadmill.

Before the actual measurement of electromyographic muscle activity during walking, each participant was allowed to walk freely on the treadmill for 10 minutes without the devices being activated. Following a 10-minute period of walking without data collection, the treadmill was stopped, and the EMG device was activated (BIOMONITOR ME6000, Mega Electronica Ltd., Finland). The participant then proceeded to walk on the treadmill for 40 step cycles. Upon completion of the measurements, the electrodes were removed and disposed of as biological waste. This procedure was consistently applied to all participants.

The muscle electrical activity was recorded using the portable electromyograph BIOMONITOR ME6000 (Mega Electronica Ltd., Finland). This electromyograph allows for the creation of a 16-channel EMG recording. However, four input channels were utilized to connect an accelerometer (Mega Electronica Ltd., Finland) for capturing signals representing angular acceleration, which is advantageous for subsequent segmentation of the recording. This portable measuring device was always positioned behind the respondent's waist and did not restrict her movement. The input circuits achieve a consistent interference suppression factor typically of 110 dB. The sampling frequency for digitization was set to 1 kHz in each channel, with a converter resolution of 14 bits. We set the device sampling frequency to 1000 Hz per channel. Motor units and their electrical potentials were recorded using self-adhesive approved hydrogel electrodes MEDICO LEAD-LOK with a diameter of 2.7 cm made of Ag/AgCl (Medico Electrodes Int., India, ISO 13485:2003). We focused on muscle groups outside the shin area, we were interested in muscle groups in the upper parts of the body on the right half of the body unilaterally. Selected muscles on the right half of the body: m. pectoralis major, m. trapezius pars transversa, m. obliquus abdominis externus, m. erector spinae in the lumbar spine, m. gluteus medius, m. gluteus maximus, m. rectus femoris, m. biceps femoris-caput longum. When we analyzed the average EMG signal of all respondents for the m. erector spinae muscle, we observed a 2-fold increase and decrease in activity in a single step cycle.

The skin at the site of electrode attachment was shaved with a disposable razor and degreased using a skin degreasing solution.

A B

Figure 1. Footwear used in the study, high-heeled shoes (A), casual flat shoes (B).

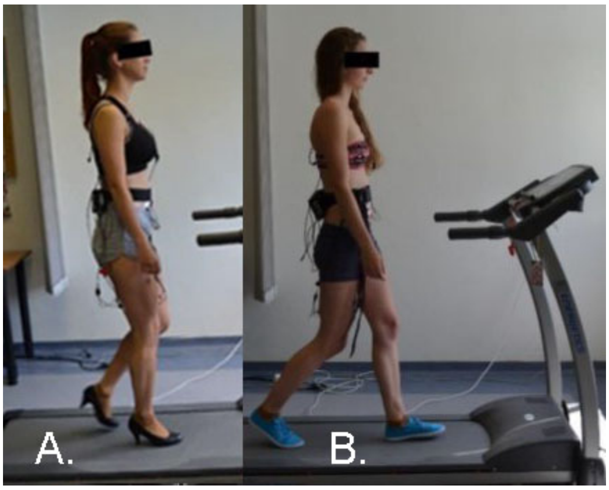

Figure 2. Measured walking on the treadmill in high heels (A) and in flat shoes (B).

The electrodes for EMG analysis were localized on the bodies of the participants according to the SENIAM (Surface EMG for non-invasive assessment of muscles) and ISEK (International Society of Electrophysiology and Kinesiology) standards, which define the principles of electromyographic measurement. The referenced standards specify electrode placement as the positioning of two bipolar sites along the muscle, aligned with the line between two anatomical landmarks in the direction of the muscle fibers, spaced 2 cm apart. The electrode position was always aligned with the direction of the muscle fibers so that their connection was at the point of greatest muscle tension during the selected movement. In this way, the most suitable locations for electrode placement on the muscles selected for measurement were determined, and the electrodes were applied by a trained physiotherapist with over 5 years of experience in the field. The same procedure was followed for all participants. From the obtained data we analyzed, onset, termination, and timing of muscle activity, intensity, and duration of contraction of the monitored muscles. Each participant underwent measurement on the treadmill with surface EMG activated in both types of footwear for a duration of 40 gait cycles, counted for one leg. Data evaluation and algorithmic processing were performed using MegaWin 3.0 software (Mega Electronics Ltd., Finland) and Matlab 2013a (MathWorks, Inc., USA) with an automatic method of motion cycle segmentation.

The analyzed EMG signal did not include the first and last gait cycles during walking in both types of shoes. Participants were required to maintain a forward gaze while walking, with head movement restricted. The upper limbs were positioned loosely alongside the body and moved naturally in coordination with the stepping cycle. Participants were instructed not to hold onto the handrail during the procedure. We evaluated changes in muscle activity during walking in high-heeled shoes (Figure 2 A) and flat shoes (Figure 2 B).

Analysis of the EMG signal

We utilized thresholding detection and the triangular method, which, instead of the raw EMG signal, utilize what's called an envelope, which is the average EMG signal from all measured movement cycles to assess muscle activity. The raw digital electromyographic signal was rectified and converted to absolute values in each channel, thus obtaining the envelope of the EMG curve by filtering the absolute value of the EMG signal with a low-pass filter, specifically, a FIR filter with a length of 1501 coefficients, a cutoff frequency of 4.14 Hz, and an imperceptible transition band with an attenuation of 55 dB for signals with a sampling frequency of 1000 Hz. Individual movement cycles were then marked on the envelopes of the EMG curves. We used a threshold detection method on the envelope of the EMG signal to detect the onset and termination of muscle activity [18,19]. The mean envelope of the EMG signal is determined as its absolute value filtered by a low pass filter with a cutoff frequency of 20 Hz. Part of the detection of the beginnings and ends of muscle activity in this method is the determination of an appropriate threshold value. The detection threshold is determined based on the local extremes of the signal envelope as (max-min) x 0.25 +min. Start detection uses a threshold determined from the minimum envelope value of the previous EMG activation, and end detection uses a threshold determined from the minimum envelope value of the envelope following the EMG activity. The exact positions of the local extrema are determined from the mean envelope analysis that precedes the actual detection of EMG activations. In each movement cycle, a local maximum is found on the envelope of the EMG signal, and sections of the envelope exceeding 25% of this maximum were designated as muscle activity. The triangle method for detecting the activation and deactivation of muscle activity is particularly effective for continuous signals. For the mean envelope of the EMG signal, the maximum was identified within an interval of ±10% of the cycle from the position of the global maximum of the mean envelope, while the minimum was determined in a similar manner. The onset of muscle activity was then marked using the triangle method, identifying the point below the line connecting the minimum and maximum, forming a triangle with the largest possible area. The hypotenuse of this triangle spans from the local minimum of the mean envelope of the movement cycle to the local maximum of the EMG signal. Similarly, the termination of muscle activity was determined by identifying the minimum following the maximum. Muscle activity intervals for each movement cycle were then graphically represented and averaged [20].

We focused on the assessment of muscle contraction expressed in %. The reference muscle was the m. rectus femoris, whose activity accounted for 100%, i.e. one step from the first part of the standing phase to the last (from heel contact with the ground to toe-off of the same foot from the ground). Based on this, we evaluated the other muscles, i.e., in what percentage of the step cycle muscle activity occurred, muscle inactivity and from this we derived the length of muscle contraction.

Statistical analysis

The Matlab 2013a (MathWorks, Inc., USA) program was used to create the average position of the onsets and terminations of muscle activity for each muscle, depending on the activity of the quadriceps femoris-rectus femoris muscle. This muscle was chosen as pivotal, theoretically exhibiting predominant activity immediately from the beginning of the observed movement. Based on these inputs, a matrix of the onsets of measured muscle groups was created for one period, and the obtained values were rounded to whole numbers and expressed as % of the movement cycle. We analyzed the average length of muscle contraction in all measured movement cycles expressed in percentage and the average intensity of muscle contraction expressed in microvolts. The consent was in accordance with the provisions of the Helsinki Declaration and other applicable data protection legislation. Numerical data were processed using the STATISTICA 13.3 program (Hamburg, Germany). Data were subjected to normality testing using the Shapiro-Wilk test. For data with intact normality, paired t-tests for dependent groups were chosen, and for data that were not normally distributed, the Wilcoxon signed-rank test was used. The statistical, practical, and substantive significance of the statistical test was assessed using the effect size coefficient based on its estimate. The most used equation is known as Cohen's "*d* " coefficient. The Matlab program was used to create the average position of the onsets and terminations of muscle activity for each muscle.

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RESULTS

The comparison of gait dependence in both types of footwear on the treadmill revealed a statistically significant change in timing ($p<0.0001$) in activation, deactivation, and contraction duration in seven out of eight monitored muscle groups, with an average effect size coefficient (Cohen's d coefficient) of 0.70, indicating a moderate dependence of the tested groups. However, the trapezius muscle did not exhibit statistically significant changes in timing (see Figure 3, 4, and Table 1).

The m. pectoralis major exhibits a monophasic activity during walking on both types of footwear on the treadmill, but we observed different onsets and declines in EMG muscle activity. While walking on the treadmill in FS, we recorded a similar average onset and decline in muscle activity, with activity beginning at 27% of the gait cycle and reaching its peak at 44% of the movement. Walking on the treadmill resulted in a delayed increase in activity. The onset occurred at 31% of the movement cycle, and the absolute maximum muscle activity was reached at 50% of the movement. The average contraction duration was 18%. Although the treadmill shortened the muscle contraction duration in both types of footwear, the muscle contraction increased more steeply and declined similarly steeply, which we attributed to the less stable walking on the treadmill. The treadmill caused an earlier onset of m. trapezius activity during walking in FS by 8%, reaching a local maximum at 69% of the movement cycle. At the same time, the treadmill shortened the muscle contraction duration by an average of 11%. In HH, we observed the onset of muscle activity while walking on the treadmill at 53% of the movement, with local peaks differing on average by 10%. On the treadmill, the peak of activity occurs at 77%, with HH simultaneously prolonging muscle contraction compared to FS by 5%. The m. obliquus abdominis externus exhibited a single local maximum in all tested situations. Walking on the treadmill changed the onset of muscle activity, with the onset occurring 9% earlier during walking in HH compared to FS. The muscle contraction duration was longer in HH on the treadmill than in FS by 6%. Similarly to all tested muscles, the treadmill caused higher muscle contraction intensity compared to walking in FS.

In the analysis of the EMG signal of the m. erector spinae, we observed two local maxima in both tested situations. On the treadmill during walking in HH, muscle activity occurred in the working cycle earlier than in FS by 2%, but the contraction duration in the first muscle activation in HH was significantly longer by 10%. In the second muscle activation in the same working cycle, the situation was reversed, with an earlier onset of muscle activity while walking in FS by 3%. Similarly, the contraction duration was longer in FS compared to HH by an average of 5%. During the analysis of the EMG signal of the m. gluteus medius during the gait cycle in both FS and HH, we observed a bimodal activity pattern. On the treadmill while walking in both FS and HH, there was a difference of 10% in the onset and end of contraction during the first muscle activation, with activity occurring later in HH. However, the contraction duration only increased by 2% in HH. The second muscle activation started earlier in HH by 4%, with a significantly extended contraction duration by 11%. Therefore, the treadmill extended muscle activity. The m. gluteus maximus exhibited a unimodal EMG activity pattern in all tested situations. Results on the treadmill showed an earlier onset of muscle activity in HH by 11%, with approximately the same decrease in activity and minimum with a difference of 1%, but with a significant extension of contraction by 18% compared to FS. By assessing the activity of the m. rectus femoris, we observed a unimodal activity pattern. Walking on the treadmill resulted in an earlier onset of activity in HH by 2%. The same percentage difference was noted in the contraction duration in favor of HH. Muscle deactivation occurred 4% earlier in FS. Local maximum and microvolt values were again higher in HH. The m. biceps femoris activated earlier on the treadmill than in FS by 2%, with more varied values for deactivation and muscle contraction length. Deactivation occurred in FS at 21% and in high-heeled shoes at 29%. On the treadmill, the percentage of muscle activity during movement increased by 9%.

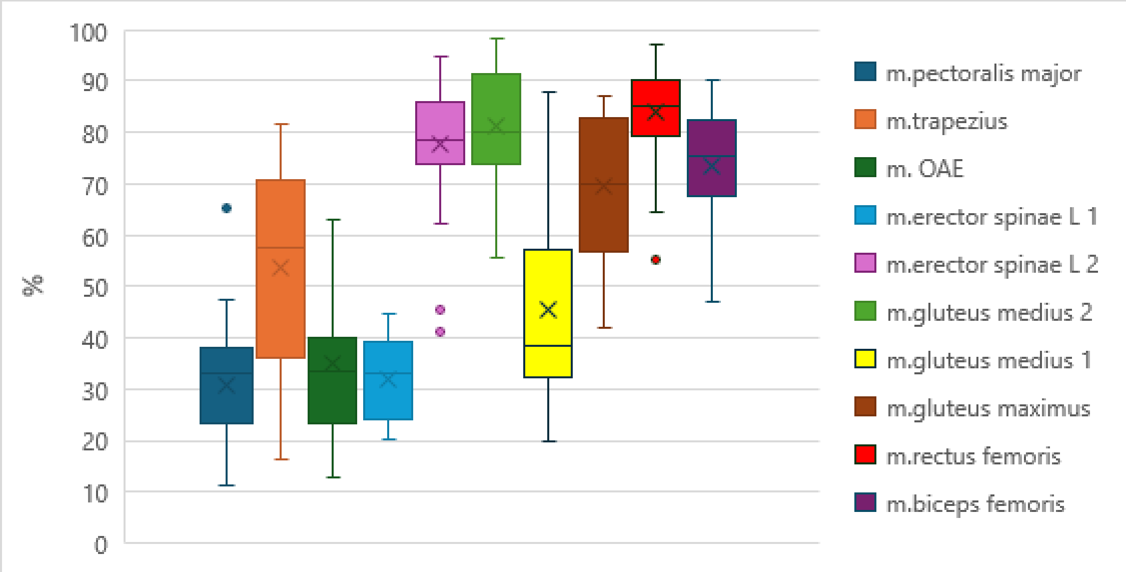

Figure 3. Graphic representation of average activation values of monitored muscles during treadmill walking in HH. Key: m. OAE – external oblique abdominis muscle; m. erector spinae L 1 – first activation of the erector spinae muscle at the lumbar level; m. erector spinae L 2 – second activation of the erector spinae muscle at the lumbar level; m. gluteus medius 1 – first activation of the gluteus medius muscle; m. gluteus medius 2 – second activation of the gluteus medius muscle.

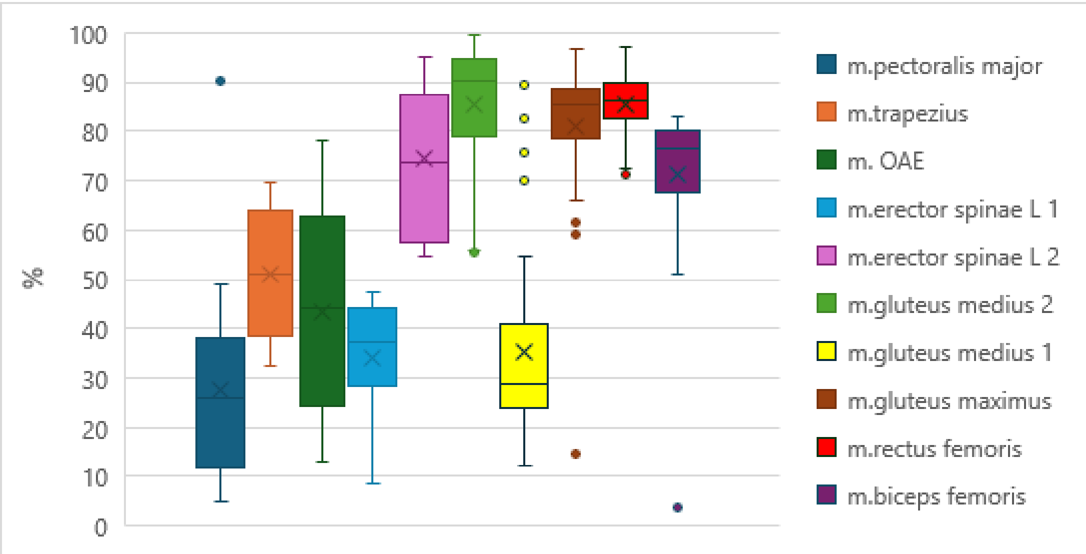

Figure 4. Graphic representation of the average activation values of monitored muscles during treadmill walking in FS. Key: m. OAE – external oblique abdominis muscle; m. erector spinae L 1 – first activation of the erector spinae muscle at the lumbar level; m. erector spinae L 2 – second activation of the erector spinae muscle at the lumbar level; m. gluteus medius 1 – first activation of the gluteus medius muscle; m. gluteus medius 2 – second activation of the gluteus medius muscle.

Table 1. Results of the average muscle activity of the studied muscles when walking in HH and in FS on the treadmill of the whole research set.

Muscle	Activity	HH [%]	SD	FS[%]	SD	p	*d*
m. Pectoralis major	Activation	30.83	3.95	27.51	4.57	< 0.0001	0.52
	Deactivation	49.63	3.8	44.29	5.4	< 0.0001	0.29
	Length of contraction	18.79	3.11	16.78	4.12	< 0.0001	0.42
m. Trapezius, pars transversa	Activation	53.64	3.96	50.96	4.32	0.23	0.69
	Deactivation	77.29	3.81	69.16	3.49	0.80	0.57
	Length of contraction	26.23	2.98	21.38	3.57	0.60	0.42
m. Obliquus abdominis externus	Activation	34.85	4.12	43.32	4.37	< 0.0001	0.5
	Deactivation	65.54	3.6	62.0	3.69	< 0.0001	0.62
	Length of contraction	30.05	3.84	24.42	3.16	< 0.0001	0.73
m. Erector spinae- I. activation	Activation	31.8	3.74	33.9	4.59	< 0.0001	0.77
	Deactivation	61.55	3.83	55.93	4.41	< 0.0001	0.86
	Length of contraction	28.29	2.99	22.03	3.54	< 0.0001	0.51
m. Erector spinae- II. activation	Activation	77.8	3.74	74.64	4.49	< 0.0001	0.68
	Deactivation	11.55	3.83	14.11	4.65	< 0.0001	0.94
	Length of contraction	28.29	3.54	33.66	4.21	< 0.0001	0.73
m. Gluteus medius- I. activation	Activation	45.5	3.77	35.23	4.46	< 0.0001	0.79
	Deactivation	61.25	3.78	51.25	4.57	< 0.0001	0.60
	Length of contraction	24.24	3.12	22.39	4.28	< 0.0001	0.88
m. Gluteus medius- II. activation	Activation	81.11	3.52	83.56	3.5	< 0.0001	0.62
	Deactivation	27.33	3.61	17.0	3.65	< 0.0001	0.54
	Length of contraction	47.52	3.45	36.64	3.88	< 0.0001	0.49
m. Gluteus maximus	Activation	69.58	3.69	80.72	4.06	< 0.0001	1.32
	Deactivation	26.34	4.02	27.32	4.58	< 0.0001	1.07
	Length of contraction	58.99	3.19	40.63	4.33	< 0.0001	0.64
m. Rectus femoris	Activation	83.83	4.31	85.34	4.6	< 0.0001	0.99
	Deactivation	26.15	3.9	22.06	4.62	< 0.0001	1.29
	Length of contraction	38.02	2.75	36.9	4.18	< 0.0001	1.08
m. Biceps femoris	Activation	73.48	3.97	71.32	4.15	< 0.0001	0.5
	Deactivation	28.55	3.65	20.21	4.13	< 0.0001	0.68
	Length of contraction	53.54	3.11	44.99	4.52	< 0.0001	0.43

HH: high-heeled shoes; FS: flat shoes

The timing and coordination of the tested muscles differed due to changes in footwear and alterations in the terrain on which the participants walked, as outlined in Table 1. Specifically, we observed differences in the duration of muscle contraction. Walking in HH prolonged the muscle contraction of all monitored muscles compared to FS. Average and maximum values of muscle contraction intensity expressed in microvolts achieved during treadmill walking by all participants are presented in Table 2. In the table, we see that the highest and average microvolt values occurred during walking in HH, where we anticipate higher activity of muscle groups due to the unstable position of the entire body caused, on one hand, by HH, and on the other hand, by walking on the treadmill itself.

These results suggest, with confirmation of statistical significance, that the intensity of activity of the tested muscles increases with the influence of HH. While walking in HH, the intensity of muscle workload was higher compared to walking in FS, although the difference was not statistically confirmed in the m. pectoralis major and m. trapezius. Nevertheless, we assume that prolonged use of HH has a negative impact on the female population's organism and may promote the development of reflex changes, which may lead, in cases of inadequate compensation, to the development of structural changes.

Table 2. Maximum (μV max) and average (μV average) microvolt values of each muscle group during walking on the treadmill of all probands.

Muscle	HH			FS			*p*	*d*
	μVmax	SD	μV average	μVmax	SD	μV average		
m. pectoralis major	125	10.2	62	85	12.91	52	0.756	0.79
m. trapezius, pars transversa	226	8.75	55	161	11.22	50	0.082	0.55
m. obliquus abdominis externus	97	9.41	49	44	11.35	21	0.045	0.82
m. erector spinae- I. activation	145	11.23	78	98	9.87	63	0.029	0.77
m. erector spinae- II. activation	237	10.9	80	99	10.01	57	0.049	0.92
m. gluteus medius- I. activation	255	7.98	92	124	8.96	77	0.005	0.95
m. gluteus medius- II. activation	298	8.8	88	102	9.33	57	0.043	1.24
m. gluteus maximus	302	12.52	85	162	10.45	60	0.023	0.65
m. rectus femoris	285	10.63	72	144	10.85	58	0.019	0.88
m. biceps femoris	355	7.12	93	132	9.22	82	0.035	0.94

FS (flat shoes): flat-soled shoes; HH (high heels): high-heeled shoes; μVmax: maximum microvolts; μVaverage: average microvolts.

DISCUSSION

The publication focuses on the evaluation and analysis of the coordination of selected muscles on the ventral and dorsal sides of the trunk while walking in HH and FS on a treadmill. We monitored muscle groups of the lower extremities, pelvic area, and upper trunk. Our goal was to determine the changes in the activity of these muscles when the heel of the foot is elevated, based on the synergistic functions of muscle groups under constant control of the CNS [6,7,10,11]. Through analysis and evaluation of the collected data, we concluded that HH alters muscle coordination based on synergistic chains not only in the lower extremities and pelvis but also in the upper body, as confirmed by the resulting statistically significant values of monitored muscles at the onset of their activation, end of activation, and contraction duration. Walking in HH on a treadmill does not significantly affect the muscle m. trapezius pars transversa. We hypothesize that this muscle does not significantly contribute to the correction and movement of the upper extremities during walking, which is intriguing because HH and the treadmill represent unstable and uncomfortable environments for walking, and balancing using the upper extremities is necessary.

While walking on the treadmill in both types of footwear (HH and FS), our study also observed a change in gait patterns. Treadmill walking typically reduces stride length by an average of 4%, increases step cadence by an average of 6%, and widens step width by 22% [4,9]. The results of our study demonstrate a statistically significant difference in the timing of the onset of muscle activity. We observed an earlier onset of EMG activity in HH for seven of the monitored muscles. The muscle m. trapezius did not show a statistically significant change. However, this muscle exhibited higher intensity compared to walking on level ground [16]. Thus, the muscle actively participates in the gait pattern but is distant from the walking effector to the extent that its involvement is not statistically significant in this case. Even though walking in HH shortens the gait cycle similarly to a treadmill, we observed a longer muscle contraction by an average of 10% compared to FS. We assume this is due to the height of the heel itself, where the foot contacts the ground earlier than during walking in FS, and due to the long-term wearing of HH, which leads to biomechanical adaptive changes in the human body [15]. By measuring and evaluating the intensity of muscle contraction of the monitored muscles based on microvolt values, we recorded higher values during walking in HH.

The most significant differences were observed in the m. erector spinae and m. obliquus abdominis externus. Since these muscles are part of the postural system and their synergy with other muscles ensures body stability, we assume that their intensity increased on the treadmill in HH to maintain a vertical body position. The most significant differences in microvolt values were found in the m. quadriceps femoris. HH caused more intense activity of paravertebral muscles in the lumbar region during treadmill walking compared to FS. Similar results regarding muscle activity were also found by authors, who observed earlier activation of the m. erector spinae in the lumbar region in HH [15]. Therefore, results consistent with our study suggest that wearing high-heeled shoes alters muscle coordination, increases muscle activity, and may lead to musculoskeletal damage with long-term submaximal load.

The highest microvolt values were consistently achieved by the tested muscles while walking in HH on the treadmill. The likely cause is the extremely unstable position of the entire body. When comparing microvolt values among different muscles, we find that muscle groups responsible for knee and pelvic stability worked most intensively during walking. Long-term wearing HH leads to health issues and the progression of degenerative knee diseases [1,3]. HH forces muscle groups in the pelvic area (m. gluteus medius et maximus, m. erector spinae, m. obliquus abdominis externus) to higher activity levels. These muscles maintain pelvic stability during natural walking in a horizontal position, but changes in heel height or terrain alter the intensity of their work, similar to [5,24]. Just as with the knee joint, we anticipate that prolonged wearing of HH and increased contraction of these muscles will lead to pain, especially in the lumbar region. As a prevention of degenerative disease of the knee joint or muscle groups involved in the stabilization of the knee and pelvis, we recommend the application of kinesiological tape to the most endangered muscle groups such as the m. rectus femoris [25].

These findings underscore the profound impact that footwear choice can exert on muscle activation patterns during locomotion. The observed alterations in timing parameters suggest a nuanced interplay between footwear characteristics and biomechanical responses, highlighting the importance of considering footwear design in optimizing gait mechanics and preventing potential musculoskeletal issues. Based on the results, we recommend that the choice of footwear for walking should have an appropriate profile, length and width, which has a preventive effect in the long term, these results are confirmed by [26,27]. High-heeled shoe wearers are further advised to increase their theoretical awareness of the possible complications caused by excessive wearing of these shoes and to prioritize their health over societal pressure and attractiveness [28].

Further research into the specific mechanisms underlying these observed changes is warranted, as it could provide valuable insights into optimizing footwear design for various gait patterns and individual biomechanical profiles. Additionally, longitudinal studies exploring the long-term effects of different footwear choices on muscle activation patterns and gait mechanics would contribute to our understanding of how footwear influences human locomotion over time.

Limitations of study

One limitation of this study may be the uniform walking speed, which was set the same for all participants, regardless of their individual anthropometric characteristics, as well as the participants' limited experience with walking in high-heeled shoes on a treadmill. Another limitation could be the use of only one type of high-heeled shoe. To draw more comprehensive conclusions and better understand changes in muscle coordination, shoes with varying heel heights and widths could be tested. The study also did not account for participant fatigue during the measurements. However, the task itself was not physically demanding, and fatigue was minimal and almost undetectable. Another limitation is the sample size; to achieve broader generalizability, groups of one hundred or more participants would be required. The results of this study are valid for the female population with the characteristics described in the research sample chapter. An unavoidable and accepted limitation is the individuality of each participant's gait pattern and emotional state during the measurements.

CONCLUSION

In general, we can say that walking in high-heeled shoes on a treadmill at a constant speed of 3.6 km/h alters the coordination, timing of onset and cessation of activity of monitored muscles active during walking, not only in the lower extremities but also in regions distant from the foot. This fact was confirmed in our study, where these results are supported by statistically significant values. Changes in heel height, similar to the treadmill, result in higher intensity of muscle contraction during walking of the tested muscles. We assume this is due to the extremely unstable position of the entire body caused by walking in high heels on a treadmill. The results indicate that walking in high heels is harmful to upper body parts due to muscle synergies occurring from the foot to higher body parts. From the results, we can conclude that walking in regular shoes on a treadmill is also stressful for the body. Our study does not have a clear practical purpose; it is rather aimed at expanding theoretical knowledge about walking in high heels under certain specific conditions.

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