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Characteristics of walking techniques with different pelvic height and pelvic rotation: effects on muscle activation and energy consumption

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Abstract

Purpose

Previous studies have proven that modifications in the natural walking technique alter muscle activation and energy consumption. This research aims to determine the differences in muscle activation, energy consumption, kinematic characteristics, perceived muscular exertion and perceived cardio-respiratory fatigue between natural and modified walking techniques with altered pelvic height and rotation.

Methods

Nine physically active, non-injured males walked on a treadmill. Modified walking techniques assumed maintenance of constant pelvic height and application of maximal pelvic rotation. Walking speed was subtransit – 0.4 km/h less than the transit. Sampled variables were average normalized maximal activation during contact and swing phase relativized to maximal voluntary activation, average submaximal oxygen consumption relativized to body mass and subtransit speed, average step length and frequency, rating of perceived muscular exertion and perceived cardio-respiratory fatigue. Muscle activation, energy consumption, and kinematic characteristics were assessed throughout each walking session. Perceived muscular exertion and perceived cardio-respiratory fatigue were evaluated post-session. Electromyographic activity was assessed for rectus femoris, gluteus maximus, vastus medialis, biceps femoris, tibialis anterior and gastrocnemius lateralis.

Results

The most significant changes in muscle activation were observed during the contact phase. A decrease in pelvic height increased muscle activation of rectus femoris, vastus medialis, and gastrocnemius lateralis. An increase in pelvic rotation increased muscle activation of all monitored muscles except gluteus maximus. Both modifications increased energy consumption, perceived muscular exertion and perceived cardio-respiratory fatigue, and altered kinematic characteristics.

Conclusions

Modifications in pelvic height and rotation at the same walking speed alter muscle activation, energy consumption, kinematic characteristics, perceived exertion and fatigue.

Keywords

Electromyography, movement kinematics, muscular exertion.

1. Introduction

Walking is a fundamental human movement that enables the performance of various motor tasks. It represents a complex and cyclic movement that engages a large amount of musculature. Walking can be performed in different ways, according to different trajectories [29], speeds [7], combinations of step length and frequency [9], footwear characteristics [25], terrain slopes [23] and characteristics [38], as well as with different external load (added weight) [16] and internal load (rhythm and type of breathing) [30]. Also, walking can be performed with different goals, such as the achievement of a sports result [28], regulation of body composition [29], and rehabilitation [26].

Walking consists of synchronized movements of the contralateral limbs throughout two-step cycles. The foot alternately represents the open and closed end of the kinetic chain that passes through the swing and the contact phase [1]. The contact phase lasts longer than the swing phase, and always one foot is in contact with the ground [37]. Extremity movements are performed mainly in the sagittal plane but also in the frontal (pelvic height decline on the side of the swing leg) and transverse (pelvic and spinal column rotation) plane [37]. Movement amplitudes of the lower limbs in the sagittal and transverse planes affect the step length, while the step length, together with the step frequency, affects the walking speed. The same speed can be achieved by different combinations of step length and frequency [9], while different step lengths and frequencies can be achieved by different walking techniques, such as when walking with a minimized center of mass vertical movement [27] or when walking with a greater knee flexion during the stance phase [40].

Energy consumption is one of the important aspects of walking that is modifiable by changing the natural walking technique [20,32]. The first classic study about factors affecting energy consumption during walking defined movement as the path of the body's center of gravity through space with the trajectory that requires the least energy consumption [17]. This first study described in detail the movement of the body's center of gravity in the transverse plane and along the vertical axis throughout a two-step cycle, that is, the reduction of energy consumption when such movements are present in a small amplitude displacements, through "smooth" sinusoidal trajectories. Six primary factors that reduce vertical oscillations of the body's center of gravity have been described: rotation, tilt, and lateral movement of the pelvis, flexion of the knee joint during the contact phase, and co-action of the foot and knee joint. From the analysis of the changes within those six factors, it was concluded that the increase in the amplitude displacements of the body's center of gravity (when walking in a "rigid" way without "amortization") increases energy consumption [4,17].

Thereafter, several studies were conducted to disprove the theory that "reducing the vertical oscillation of the body's center of gravity reduces energy consumption" [8,9,19,24,17,40]. The first classic study described changes within the six above-mentioned factors and their effect on energy consumption from pathological to normal walking, but not from normal walking to walking with a straight trajectory of the body's center of gravity [17]. Led by the initial theoretical assumption, the "opposite" conclusion was drawn compared to the first classic study. In the first classic study, it was stated that a person chooses a way of movement that reduces the vertical oscillation of the body's center of gravity, but not a way of movement that will completely neutralize it [17]. From this statement, it can be assumed that the forced maintenance of the body's center of gravity at a constant height could increase energy consumption compared to the natural walking technique.

Previous studies have proven that modifications in the natural walking technique alter muscle activation and energy consumption, that is, changes in the step length and step frequency on the same walking speed from the ones that are naturally selected increase and muscle activation [32] and energy consumption [9,20,32,33]. Also, the walking technique with a lowered body's center of gravity [3,6,21,40] and its reduced vertical oscillation [8,9,24,27] produces increased energy consumption. The authors of this paper are not aware of any study that dealt with the effects of the increased pelvic rotation in the transverse plane, but it would be reasonable to assume that it would further increase energy consumption, given that one of the factors for reducing the vertical oscillations of the body's center of gravity is the pelvic rotation [17]. Since numerous studies have examined changes within the natural walking technique, while differences between various walking techniques are an insufficiently and non-comprehensively researched area, the goal of this research is to examine the changes within various aspects of movement with respect to different walking technique modifications – pelvic height and pelvic rotation.

This research aims to determine the differences in muscle activation, energy consumption, kinematic characteristics, perceived muscular exertion, and cardio-respiratory fatigue between different walking techniques in a healthy and physically active population. These differences in walking techniques are achieved through changes in (1) pelvic height, with the requirement for the reduction of vertical oscillation of the body's center of gravity, and (2) pelvic rotation in the transverse plane. The assumption is that decreased pelvic height and increased pelvic rotation during walking significantly increase muscle activation, energy consumption, and perceived exertion and fatigue, along with modifications in kinematic characteristics.

2. Materials and Methods

2.1. Participants

The sample of subjects consisted of nine physically active males (28.6±4.3 years, body height 1.81±0.06 m, body mass 76.1±4.9 kg, and BMI 23.3±1.7 kg/m^2), without injuries or damage to the locomotor apparatus. The subjects were told not to consume food and liquid for at least one hour before the testing session and not to engage in any intense physical activity for at least 24 hours before the testing session. Each subject signed a written consent form confirming their voluntary participation in the experiment.

2.2. Experimental protocol

The experiment was carried out in the laboratory of "PROFEX – Academy of Healthy Living" in Belgrade, in two experimental days, with an interval of three days for each subject. On the first experimental day, basic anthropometric and morphological measurements (BH, BM, and BMI), body composition, energy expenditure during rest (VO_{2rest}), and measurement of pelvic height (PH) in an upright and lowered position were done. After the warm-up procedure, the transition speed from walking to running ($V_{transit}$) was determined. Lastly, subjects were familiarized with different walking techniques used in the study. On the second experimental day, after the warm-up procedure and a short familiarization with walking techniques, maximal voluntary activation (MVA) was tested. After that, the main part of the experiment was performed – sampling the variables of electromyographic activity, energy consumption, kinematic characteristics, and the rating of perceived muscular exertion and cardio-respiratory fatigue during walking with different techniques.

Walking was carried out in the following order to reduce the potential negative effects of local muscle fatigue on the quality of subsequent walking sessions: (1) the natural walking technique, (2) walking with reduced PH, (3) walking with an increased PR in the transverse plane, (4) walking with reduced PH and increased PR in the transverse plane.

PH (while walking with reduced PH) was relativized according to the angle in the knee joint measured with the goniometer, to take into account individual anthropometric characteristics (Appendix A). The height from the flat surface to the *spina iliaca anterior superior* (the closest projection of the body's center of gravity) was measured in an upright and in a semi-squat position, with the knee joint angle of 120°. Reduced PH was 9.4±1.1% (range – 7.8-11.0%) lower than PH in an upright position. While walking with reduced PH, the pelvis was rotated naturally and the requirement was also to reduce the vertical oscillation of the body's center of gravity, which increases significantly with potential decreases in step frequency with this technique modification, if not voluntarily controlled [35]. While walking with an increased PR, the pelvic rotation (PR) in the transverse plane was of maximal amplitude. The shoulders and the trunk were not rotated along with the pelvis but were maintained relatively fixed. An illustration of these walking techniques is presented in Figure 1.

Figure 1. Walking technique with reduced pelvic height and pelvic height monitoring mechanism with laser device (left) and walking technique with increased pelvic rotation in the transverse plane (right).

PH was monitored utilizing a horizontal laser by a *Bosch PCL20* device (Figure 1). The marked spot on the *spina iliaca anterior superior* was meant to match the height of the laser during walking. The device was placed on a tripod next to the treadmill and the laser height was adjusted according to individual anthropometric characteristics (treadmill height and sole height were also included – subjects wore the same shoes as on the first experimental day). The PR in the transverse plane was monitored visually. All technique deviations were corrected using verbal and tactile instructions during walking.

Basic anthropometric and morphological measurements consisted of the measurement of BH [m], BM [kg], and BMI [kg/m^2]. BH was measured using a digital altimeter (*BSM170, Arab Engineers*), while BM, BMI, and body composition were measured using a bioelectrical impedance (*InBody770* – software version 3.0.0.1). The body composition measurement procedure was done according to the manufacturer's recommendations.

VO_{2rest} [ml/min], for the relativization of the oxygen consumption (VO_2) during walking, was measured using a gas exchange analyzer (*Fitmate PRO, Cosmed*). The measurement was carried out in a relaxed lying position on the back, and with spontaneous breathing depth and frequency, during 12 minutes. The average value was taken for 5 minutes during which the subject's breathing depth and frequency and VO_2 were the least variable.

The warm-up procedure for $V_{transit}$ testing consisted of walking on a treadmill with progressively increasing the speed for 5 minutes, and exercises for dinamic stretching and toning the musculature. $V_{transit}$ testing was carried out on the treadmill of the Italian company *COSMED* (*h/p/cosmos*) using an incremental protocol, which was slightly modified in comparison to some previous studies [12,14,15,18]. The subjects were told that the goal was not to determine the maximal walking speed, but the speed at which walking is more comfortable movement than running. The subjects walked naturally while maintaining a grip on the treadmill handle in front of them. They did not have the information about the movement speed, which eliminated autosuggestion and previous experience factors. Each walking sequence lasted 30 seconds. The initial walking speed was 5 km/h. The speed increased by 0.4 km/h or by 0.2 km/h depending on the proximity of $V_{transit}$. In the second case, each sequence was followed by 30 seconds of walking at the initial speed (5 km/h), to reduce the potential negative effects of local muscle fatigue [13] on walking efficiency during subsequently increased speeds. Additionally, subjects provided information about the subjective feel of $V_{transit}$ proximity on a scale from 1 to 10 to match the subjective and objective indicators of $V_{transit}$ prior to determination. The objective indicator of $V_{transit}$ was the presence of a flight phase. After the first $V_{transit}$ determination, an additional walking sequence was performed at the same speed. If subjects confirmed that the speed was not $V_{transit}$, it was continued with the same protocol of increasing the walking speed until $V_{transit}$ was finally determined. The obtained $V_{transit}$ in this sample of subjects was on average 7.6±0.3 km/h, which is a higher value compared to the values obtained in some previous studies [12,14,15] (Appendix B).

MVA testing (and later monitoring during walking) was done for the following 6 muscles on the dominant side of the body, representing movements performed in the sagittal plane at the hip, knee, and ankle joint: *m. rectus femoris* (RF), *m. gluteus maximus* (GM), *m. vastus medialis*

(VM), *m. biceps femoris – caput longum* (BF), *m. tibialis anterior* (TA) and *m. gastrocnemius lateralis* (GL). The MVA testing was carried out according to the recommendations from the literature [2]. The muscles were tested under isometric conditions. The subjects were instructed to gradually increase their voluntary effort reaching the maximum after 2-3 seconds, then to maintain the maximum for 5 seconds, and after that to gradually decrease the voluntary effort over 2-3 seconds. Two repetitions were done for each muscle. The resting duration between repetitions was 30 seconds, while between testing the muscles was 60 seconds.

Electromyographic activity and energy consumption were obtained on separate occasions (on two experimental days). Subjects walked on a treadmill, causing the largest difference in body's center of gravity vertical oscillation between natural and modified walking technique with maximal PR addition, since it is proven that PR is less while walking on treadmill compared to overground [34]. Walking speed was subtransit ($V_{subtransit}$), i.e. 0.4 km/h less than $V_{transit}$, producing maximal potential body's center of gravity vertical oscillation that is to be controlled and reduced with PH technique modification, since it is proven that with an increase in walking speed, body's center of gravity vertical oscillation also increases [35]. Electromyographic activity was monitored during 30 seconds of walking. Right before the acquisition, 15 seconds were devoted to stabilizing the walking technique. The rest between walking sessions was passive standing for 30 seconds. The electrode positioning and fixation were identical to those during MVA testing, as it followed immediately after it. Energy consumption was monitored during 2 minutes of walking. Just before the acquisition, 60 seconds were devoted to reaching a stable state of VO_2. The rest between walking sessions was passive stretching of the lower extremity muscles for 3 minutes. The subjects did not get any instructions about the type and rhythm of respiration during walking. Immediately after walking sessions, subjects provided information about the rating of perceived muscular exertion ($RPE_{exertion}$), when electromyographic activity was monitored, and the rating of perceived cardio-respiratory fatigue ($RPE_{fatigue}$), when energy consumption was monitored.

2.3. *Electrode placement and EMG signal acquisitioning, processing and normalization*

Electrodes were placed approximately in the middle of the muscle belly (at the greatest palpated tightness and density of muscle fibers during a voluntary contraction), away from tendinous areas and edges of the muscle, in the direction of the muscle fibers and oriented from distal to the proximal attachment. The electrodes were fixed using a short adhesive tape, as well as a long adhesive tape wrapped around the corresponding body segment.

Electromyographic activity was monitored through *Delsys Trigno* telemetry sensors, during the contact phase (CP) and the swing phase (SP). The activity was separated into CP and SP, using two pressure sensors synchronized with the software and muscle signals via a suitable *Delsys* adapter. Both sensors were placed inside the subject's shoes (one under the heel and the other under the forefoot). CP was defined as the period between heel contact and full-weight unloading of the forefoot, and SP as the reverse of CP. A pressure value of 20% of the maximal value was set as the beginning of heel contact.

The electrodes had a preset band-pass filter from 20 to 450 [Hz] to eliminate movement artifacts and non-physiological sounds. The sampling frequency was set to 1926 Hz. The signal during walking was processed in the *EMGworks* software (*Delsys* – software version 4.8.0) and included the following procedures presented in Figure 2. In the final processing step, the relativized muscle activation and standard deviation were shown in linear form throughout the entire two-step cycle. Outliers (standard deviation within 10 adjacent two-step cycles above 60% of the mean value) were excluded from further analysis. The same procedure (up to the signal smoothing) was done when processing the raw signal obtained from MVA testing. The repetition with the highest maximal activation was taken for further analysis. The minimum cut-off duration of the signal for all the processing was 2.5 seconds.

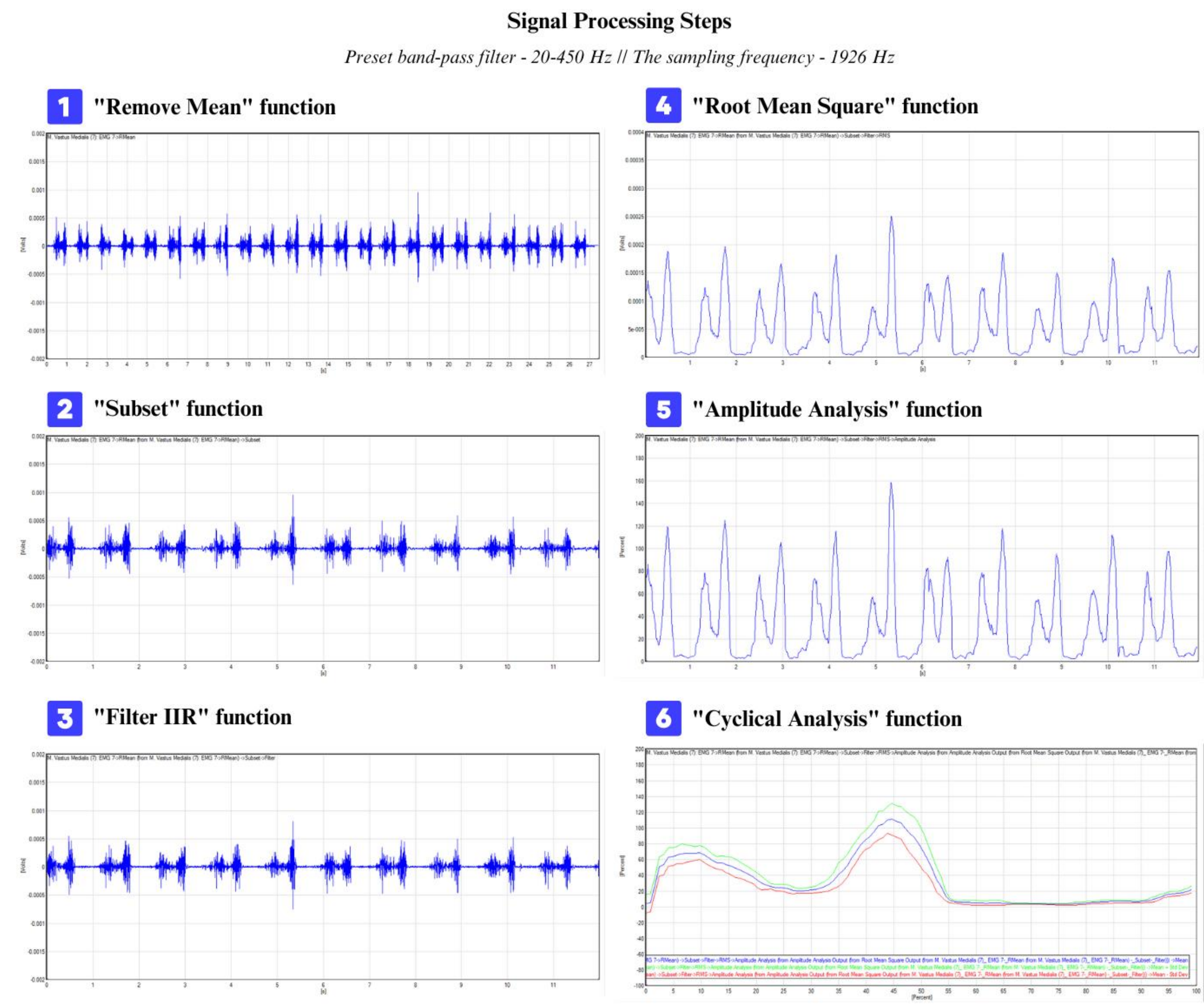

Figure 2. EMG signal processing steps for the EMG signal during walking from one subject.
(1) Aligning the isoelectric line of the raw EMG signal; (2) Cutting-off ten adjacent signals; (3) Filtering the signal with a second-order high-pass Butterworth class filter (filtering frequency 1-20 Hz); (4) Rectification and Smoothing of the signal with Root Mean Square algorithm (window length 0.1s, window overlap 0.09s); (5) Normalizing the signal to MVA (window length 0.1s, window overlap 0.09s); (6) Cyclical Analysis of the signal to pressure sensors (contact phase – the period between heel contact and full-weight unloading of the forefoot; swing phase - the period between full-weight unloading of the forefoot and heel contact; the beginning of heel contact – pressure value of 20% of the maximal value).

2.4. Variables

1. Electromyographic activity – The average normalized maximal activation (%EMG_{max}) relativized to MVA [%], separately for CP and SP. The average value was taken for 10 adjacent two-step cycles.
2. Energy expenditure – The average NET submaximal oxygen consumption ($VO_{2submax}$) relativized to BM and $V_{subtransit}$ [NET ml/kg/m] (Appendix C). The average value was taken for 2 minutes of monitoring energy consumption.
3. Kinematic characteristics – Average step length (SL) [m] and average step frequency (SF) [step/s] (Appendix D). The average values were taken for 10 adjacent two-step cycles.

4. Rating of perceived muscular exertion and cardio-respiratory fatigue – The rating scale of $RPE_{exertion}$ and $RPE_{fatigue}$ was from 1 to 10, where a score of 10 represented perceived exertion or fatigue if the walking speed was maximal.

2.5. Statistical analysis

The resulting $\%EMG_{max}$ and $VO_{2submax}$ datasets first underwent Outlier analysis. Outlier analysis was not applied for kinematic variables, since these were individual movement parameters according to the given instructions on PH and PR. Each variable is presented through its mean (Mean) and standard deviation (SD). Parametric statistics were applied for $\%EMG_{max}$ separately during CP and SP, $VO_{2submax}$, SL, and SF – factorial analysis of variance (ANOVA) with repeated measures to determine the influence of changes within the PH and the PR factors, and their interaction effect. In case of statistically significant interaction effect, individual post-hoc T-tests for dependent samples with Holm-Bonferroni correction were applied. Non-parametric statistics were applied for $RPE_{exertion}$ and $RPE_{fatigue}$ since the data do not belong to a rationale scale but represent ratings – Friedman's two-way ANOVA and individual post-hoc Wilcoxon matched-pairs signed-ranks tests with Holm-Bonferroni correction to determine the differences between walking techniques. The alpha confidence level was preset at <0.05. The data were processed in the *IBM SPSS Statistics* application program (version 20.0.0).

3. Results

Tables 1-3 show the results of factorial ANOVA with repeated measures on the variables of electromyographic activity, energy consumption, and kinematic characteristics of walking. Table 4 shows the results of Friedman's two-way ANOVA on the rating of perceived muscular exertion and cardio-respiratory fatigue.

Table 1. Influence of pelvic height factor, pelvic rotation factor, and their interaction effect on the variable of electromyographic activity (N = 9).
RF – m. rectus femoris; GM – m. gluteus maximus; VM – m. vastus medialis; BF – m. biceps femoris - caput longum; TA – m. tibialis anterior; GL – m. gastrocnemius lateralis; (CP) – contact phase; (SP) – swing phase. Presented are F-value, Df, Df error and Partial η^2. Statistical significance is marked with gray color. The arrow (↑) indicates that the value increases and the arrow (↓) indicates that the value decreases (↓), all under the influence of a change in the walking technique within the given factor. The abbreviation after the symbol indicates that the change refers to the upright (up), lowered (low), with rotation (wr), or without rotation (wor) walking technique. The alpha confidence level was preset at <0.05.

	Pelvic height	**Pelvic rotation**	**Pel.height x Pel.rotation**
$RF_{(CP)}$	$p \leq 0.01$ (↑) $F_{(1,8)} = 15.09$ $Partial\ \eta^2 = 0.662$	$p \leq 0.01$ (↑) $F_{(1,8)} = 10.52$ $Partial\ \eta^2 = 0.576$	$p = 0.30$ $F_{(1,8)} = 1.25$ $Partial\ \eta^2 = 0.115$
$GM_{(CP)}$	$p = 0.40$	$p = 0.77$	$p = 0.59$

	$F_{(1,8)} = 0.78$ Partial $\eta^2 = 0.089$	$F_{(1,8)} = 0.09$ Partial $\eta^2 = 0.012$	$F_{(1,8)} = 0.31$ Partial $\eta^2 = 0.038$
VM(CP)	$p \leq 0.01$ (↑) $F_{(1,8)} = 10.23$ Partial $\eta^2 = 0.561$	$p \leq 0.001$ (↑) $F_{(1,8)} = 22.85$ Partial $\eta^2 = 0.741$	$p = 0.11$ $F_{(1,8)} = 3.21$ Partial $\eta^2 = 0.286$
BF(CP)	$p = 0.37$ $F_{(1,8)} = 0.90$ Partial $\eta^2 = 0.101$	$p \leq 0.05$ (↑) $F_{(1,8)} = 6.20$ Partial $\eta^2 = 0.437$	$p = 0.56$ $F_{(1,8)} = 0.37$ Partial $\eta^2 = 0.045$
TA(CP)	$p = 0.16$ $F_{(1,8)} = 2.43$ Partial $\eta^2 = 0.233$	$p \leq 0.05$ (↑) $F_{(1,8)} = 6.09$ Partial $\eta^2 = 0.432$	$p = 0.68$ $F_{(1,8)} = 0.19$ Partial $\eta^2 = 0.023$
GL(CP)	$p \leq 0.05$ (↑wor) $F_{(1,8)} = 21.91$ Partial $\eta^2 = 0.732$	$p \leq 0.05$ (↑up) $F_{(1,8)} = 14.98$ Partial $\eta^2 = 0.652$	$p \leq 0.05$ $F_{(1,8)} = 6.09$ Partial $\eta^2 = 0.432$
RF(SP)	$p = 0.30$ $F_{(1,8)} = 1.20$ Partial $\eta^2 = 0.131$	$p = 0.95$ $F_{(1,8)} = 0.004$ Partial $\eta^2 = 0.0005$	$p \leq 0.05$ $F_{(1,8)} = 6.04$ Partial $\eta^2 = 0.430$
GM(SP)	$p = 0.14$ $F_{(1,8)} = 2.63$ Partial $\eta^2 = 0.247$	$p = 0.63$ $F_{(1,8)} = 0.25$ Partial $\eta^2 = 0.030$	$p = 0.98$ $F_{(1,8)} = 0.001$ Partial $\eta^2 = 0.00008$
VM(SP)	$p = 0.20$ $F_{(1,8)} = 1.95$ Partial $\eta^2 = 0.196$	$p = 0.19$ $F_{(1,8)} = 2.02$ Partial $\eta^2 = 0.202$	$p = 0.87$ $F_{(1,8)} = 0.03$ Partial $\eta^2 = 0.003$
BF(SP)	$p = 0.26$ $F_{(1,8)} = 1.44$ Partial $\eta^2 = 0.152$	$p = 0.49$ $F_{(1,8)} = 0.53$ Partial $\eta^2 = 0.062$	$p = 0.90$ $F_{(1,8)} = 0.02$ Partial $\eta^2 = 0.002$
TA(SP)	$p = 0.38$ $F_{(1,8)} = 0.88$ Partial $\eta^2 = 0.099$	$p = 0.79$ $F_{(1,8)} = 0.07$ Partial $\eta^2 = 0.009$	$p = 0.89$ $F_{(1,8)} = 0.02$ Partial $\eta^2 = 0.003$
GL(SP)	$p = 0.22$ $F_{(1,8)} = 1.74$ Partial $\eta^2 = 0.179$	$p = 0.43$ $F_{(1,8)} = 0.68$ Partial $\eta^2 = 0.079$	$p = 0.43$ $F_{(1,8)} = 0.68$ Partial $\eta^2 = 0.078$

Table 2. Influence of pelvic height factor, pelvic rotation factor, and their interaction effect on the variable of energy consumption (N = 9).

$VO_{2submax}$ – submaximal oxygen consumption. Presented are F-value, Df, Df error and Partial η^2. Statistical significance is marked with gray color. The arrow (↑) indicates that the value increases and the arrow (↓) indicates that the value decreases (↓), all under the influence of a change in the walking technique within the given factor. The abbreviation after the symbol indicates that the change refers to the upright (up), lowered (low), with rotation (wr), or without rotation (wor) walking technique. The alpha confidence level was preset at <0.05.

	Pelvic height	Pelvic rotation	Pel.height x Pel.rotation
$VO2_{submax}$	$p \leq 0.00001$ (↑) $F_{(1,8)} = 127.37$ Partial $\eta^2 = 0.941$	$p \leq 0.0001$ (↑) $F_{(1,8)} = 46.72$ Partial $\eta^2 = 0.854$	$p = 0.72$ $F_{(1,8)} = 0.13$ Partial $\eta^2 = 0.016$

Table 3. Influence of pelvic height factor, pelvic rotation factor, and their interaction effect on kinematic characteristics of walking (N = 9).

SL – step length; SF – step frequency. Presented are F-value, Df, Df error and Partial η^2. Statistical significance is marked with gray color. The arrow (↑) indicates that the value increases and the arrow (↓) indicates that the value decreases (↓), all under the influence of a change in the walking technique within the given factor. The abbreviation after the symbol indicates that the change refers to the upright (up), lowered (low), with rotation (wr), or without rotation (wor) walking technique. The alpha confidence level was preset at <0.05.

	Pelvic height	Pelvic rotation	Pel.height x Pel.rotation
SL	$p \leq 0.001$ (↑)	$p \leq 0.001$ (↑)	$p \leq 0.05$
	$F_{(1,8)} = 43.61$	$F_{(1,8)} = 33.25$	$F_{(1,8)} = 5.81$
	Partial $\eta^2 = 0.845$	Partial $\eta^2 = 0.806$	Partial $\eta^2 = 0.421$
SF	$p \leq 0.0001$ (↓)	$p \leq 0.001$ (↓)	$p = 0.18$
	$F_{(1,8)} = 47.19$	$F_{(1,8)} = 31.53$	$F_{(1,8)} = 2.16$
	Partial $\eta^2 = 0.855$	Partial $\eta^2 = 0.798$	Partial $\eta^2 = 0.213$

Table 4. Results of Friedman's two-way ANOVA on the rating of perceived muscular exertion and cardio-respiratory fatigue (N = 9).

$RPE_{exertion}$ – rating of perceived muscular exertion; $RPE_{fatigue}$ – rating of perceived cardio-respiratory fatigue. Pretented are χ^2 value and Df. Statistical significance is marked with gray color. The alpha confidence level was preset at <0.05.

$RPE_{exertion}$	$p \leq 0.001$
	$\chi^2(3) = 19.70$
$RPE_{fatigue}$	$p \leq 0.001$
	$\chi^2(3) = 21.07$

A statistically significant influence of changes within the PH and the PR factors on the increase in $\%EMG_{max}$ of RF during CP was observed (p≤0.01). During SP a statistically significant interaction effect was observed (p≤0.05). Individual post-hoc T-tests for dependent samples with Holm-Bonferroni correction did not reveal a statistically significant source of interaction effect (p>0.05). GM remained insensitive to changes under the influence of the changes within the PH and the PR factors, neither during CP nor during SP (p>0.05). A statistically significant influence of changes within the PH and the PR factors on the increase in $\%EMG_{max}$ of VM during CP was observed (p≤0.01), but not during SP (p>0.05).

The changes within the PR factor had a statistically significant influence on the increase in $\%EMG_{max}$ of BF and TA during CP (p≤0.05), but not the changes within the PH factor (p>0.05). During SP, no statistically significant influence of any factor on the increase in $\%EMG_{max}$ of these muscles was observed (p>0.05). During CP, a statistically significant interaction effect was observed on the change in $\%EMG_{max}$ of GL (p≤0.05). Individual post-hoc T-tests for dependent samples with Holm-Bonferroni correction revealed a statistically significant influence of decreasing PH on the increase in $\%EMG_{max}$ when PR was natural (p≤0.05), and a statistically significant effect of increasing PR on the increase in %EMG_{max} when PH is usual ($p \leq 0.05$). During SP, no statistically significant influence of the PH and the PR factors on the change in %EMG_{max} of GL was observed ($p > 0.05$).

A statistically significant influence of changes within the PH and the PR factors on the increase in $VO_{2submax}$ ($p \leq 0.01$) and SL ($p \leq 0.01$), and on the decrease in SF ($p \leq 0.01$) was observed. Given that a statistically significant interaction effect on SL was observed ($p \leq 0.05$), individual post-hoc T-tests for dependent samples with Holm-Bonferroni correction revealed statistically significant differences between all comparisons ($p \leq 0.01$).

Results of Friedman's two-way ANOVA showed statistically significant differences in $RPE_{exertion}$ and $RPE_{fatigue}$ between walking techniques. Individual Wilcoxon matched-pairs signed-ranks tests with Holm-Bonferroni correction for the $RPE_{exertion}$ showed statistically significant differences between the natural walking technique and the walking technique with reduced PH ($p \leq 0.05$), as well as between the walking technique with an increased PR and the walking technique with reduced PH and increased PR ($p \leq 0.05$). In the remaining comparisons, no statistically significant differences were observed ($p > 0.05$). Individual Wilcoxon matched-pairs signed-ranks tests with Holm-Bonferroni correction for the $RPE_{fatigue}$ showed statistically significant differences between all comparisons ($p \leq 0.05$).

Figure 3 shows descriptive statistics and post-hoc analysis for the variable of electromyographic activity during contact phase.

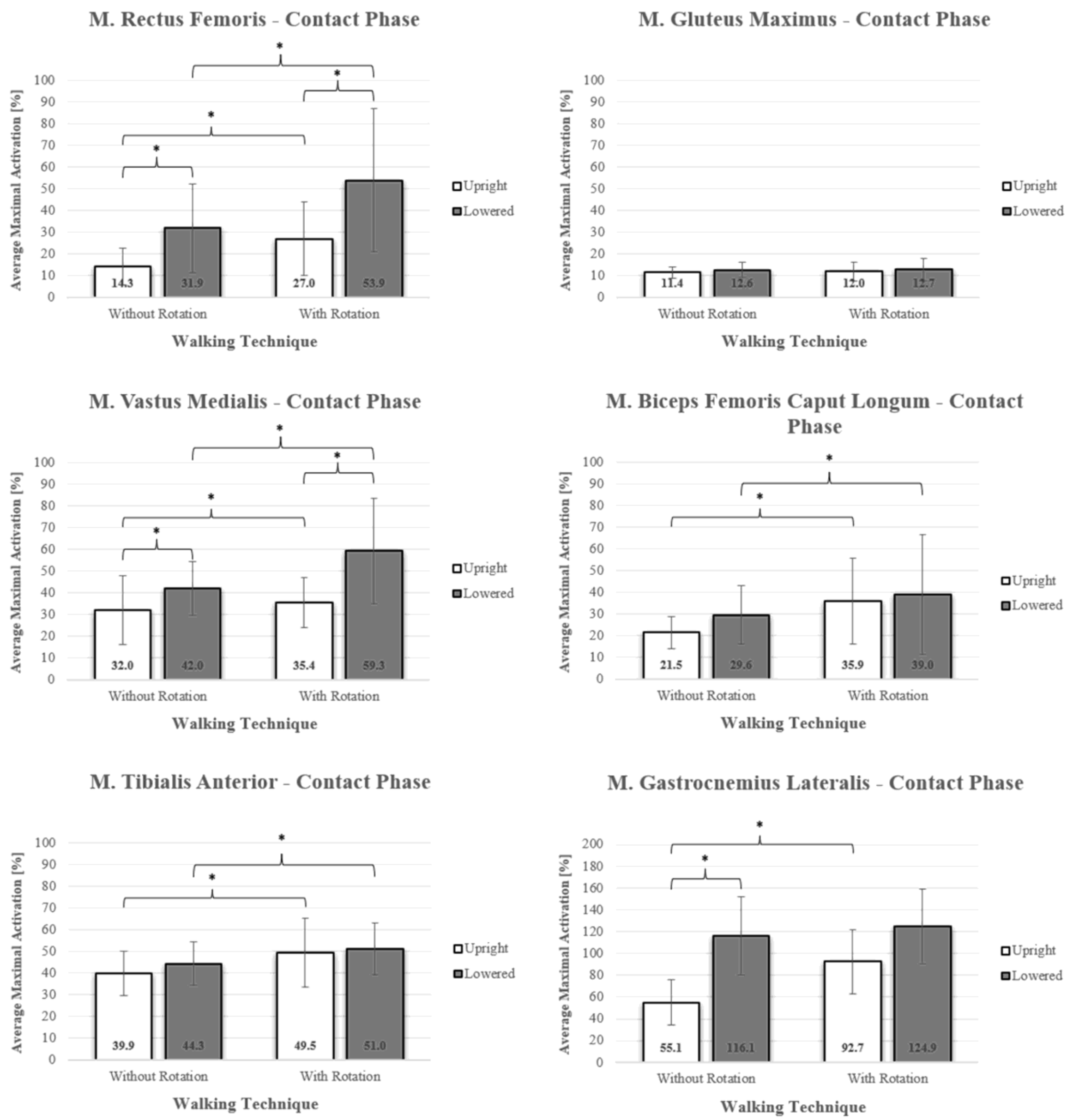

Figure 3. Descriptive statistics (Mean and SD) and results of post-hoc T-tests for dependent samples with Holm-Bonferroni correction for the average normalized maximal activation for all monitored muscles (*m. rectus femoris*, *m. vastus medialis*, *m. tibialis anterior*, *m. gluteus maximus*, *m. biceps femoris – caput longum*, *m. gastrocnemius lateralis*), during the contact phase (N = 9). Due to the supramaximal activation of m. gastrocnemius lateralis in the swing phase, the Y scale on that graph is increased to 200%.

Upright – walking technique with natural height of the pelvis; Lowered – walking technique with reduced pelvis height; Without rotation – walking technique with natural pelvic rotation in the transverse plane; With rotation – walking technique with increased pelvic rotation in the transverse plane. The symbol (*) indicates statistical significance at the <0.05 level.

Figure 4 shows descriptive statistics and post-hoc analysis for the variable of electromyographic activity during swing phase.

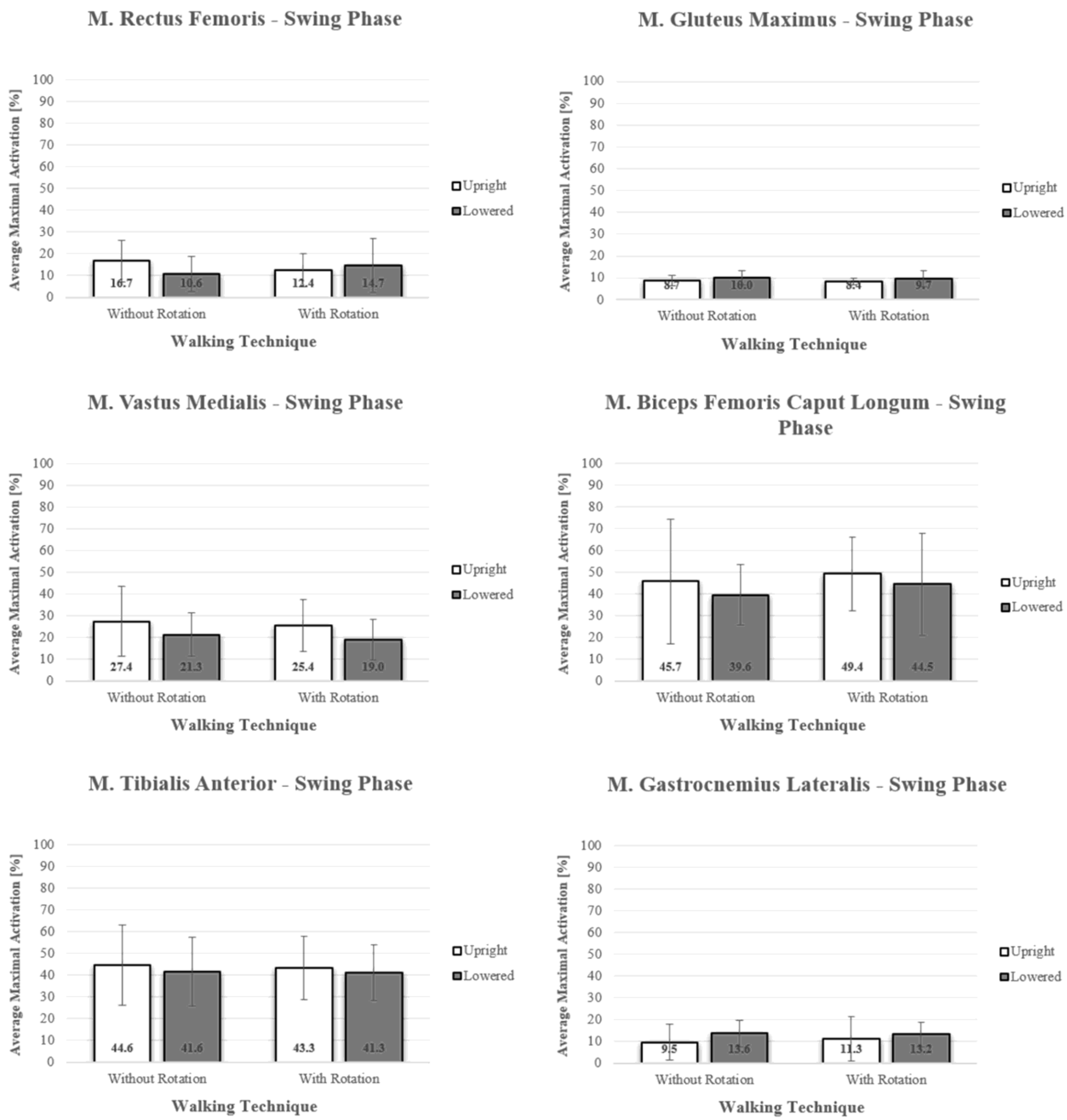

Figure 4. Descriptive statistics (Mean and SD) and results of post-hoc T-tests for dependent samples with Holm-Bonferroni correction for the average normalized maximal activation for all monitored muscles (*m. rectus femoris*, *m. vastus medialis*, *m. tibialis anterior*, *m. gluteus maximus*, *m. biceps femoris – caput longum*, *m. gastrocnemius lateralis*), during the swing phase (N = 9).
Upright – walking technique with natural height of the pelvis; Lowered – walking technique with reduced pelvis height; Without rotation – walking technique with natural pelvic rotation in the transverse plane; With rotation – walking technique with increased pelvic rotation in the transverse plane. The symbol (*) indicates statistical significance at the <0.05 level.

Figure 5 shows descriptive statistics and post-hoc analysis for the variable of energy consumption.

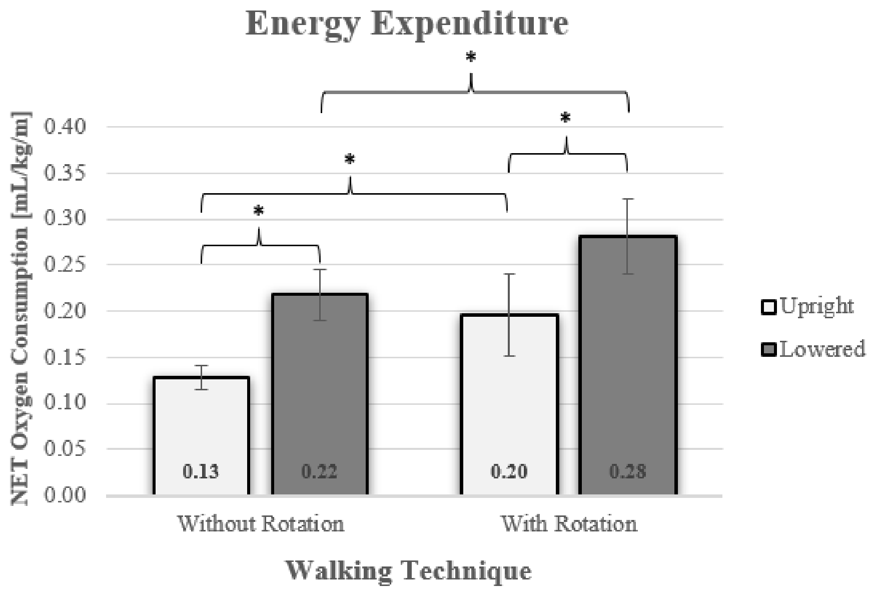

Figure 5. Descriptive statistics (Mean and SD) and results of post-hoc T-tests for dependent samples with Holm-Bonferroni correction for the average NET submaximal oxygen consumption (N = 9).
Upright – walking technique with natural height of the pelvis; Lowered – walking technique with reduced pelvis height; Without rotation – walking technique with natural pelvic rotation in the transverse plane; With rotation – walking technique with increased pelvic rotation in the transverse plane. The symbol (*) indicates statistical significance at the <0.05 level.

Figure 6 shows descriptive statistics and post-hoc analysis for kinematic characteristics of walking.

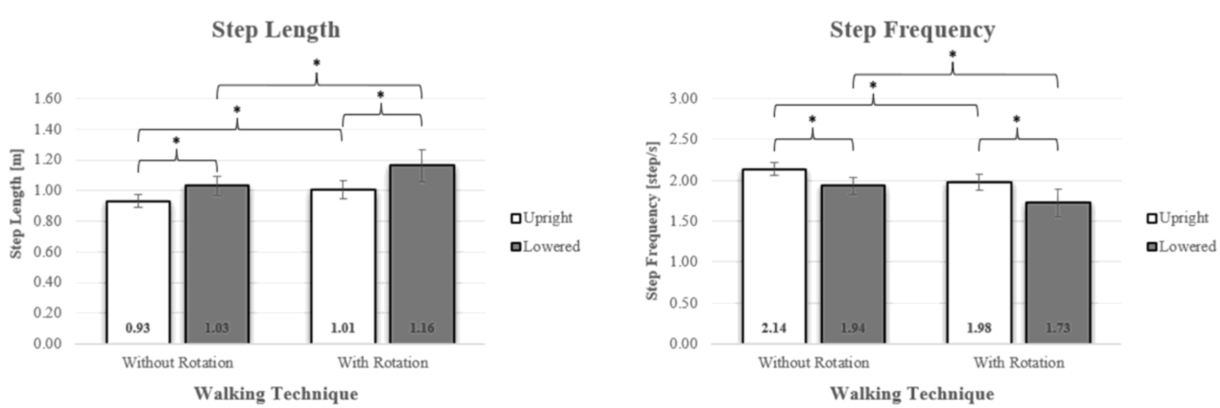

Figure 6. Descriptive statistics (Mean and SD) and results of post-hoc T-tests for dependent samples with Holm-Bonferroni correction for the step length and the step frequency (N = 9).
Upright – walking technique with natural height of the pelvis; Lowered – walking technique with reduced pelvis height; Without rotation – walking technique with natural pelvic rotation in the transverse plane; With rotation – walking technique with increased pelvic rotation in the transverse plane. The symbol (*) indicates statistical significance at the <0.05 level.

Figure 7 shows descriptive statistics and post-hoc analysis for the rating of perceived muscular exertion and cardio-respiratory fatigue.

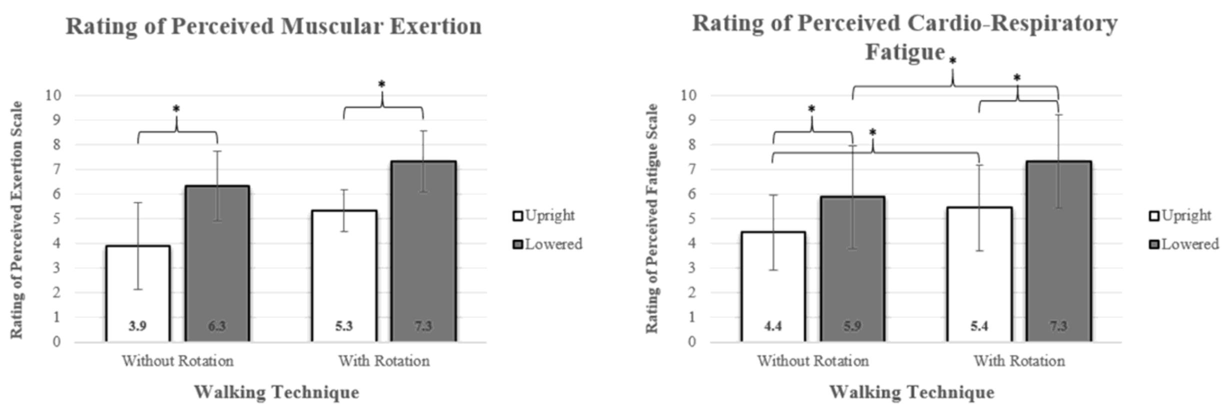

Figure 7. Descriptive statistics (Mean and SD) and results of post-hoc Wilcoxon matched-pairs signed-ranks tests with Holm-Bonferroni correction for the rating of perceived muscular exertion and cardio-respiratory fatigue (N = 9).

Upright – walking technique with natural height of the pelvis; Lowered – walking technique with reduced pelvis height; Without rotation – walking technique with natural pelvic rotation in the transverse plane; With rotation – walking technique with increased pelvic rotation in the transverse plane. The symbol (*) indicates statistical significance at the <0.05 level.

4. Discussion

This research aims to determine the differences in muscle activation, energy consumption, kinematic characteristics, perceived muscular exertion and cardio-respiratory fatigue between different walking techniques in a healthy and physically active population achieved by the changes within (1) pelvic height with the requirement for the reduction of vertical oscillation of the body's center of gravity and (2) pelvic rotation in the transverse plane. By the initial assumption, decreased pelvic height and increased pelvic rotation during walking significantly increased muscle activation during the contact phase in most lower extremities muscles. Also, such changes in walking technique affected an increase in energy consumption and perceived exertion and fatigue, along with modifications in kinematic characteristics.

Knowing that the knee and hip extensors during CP overcome the largest amount of the external load [22,39], it was already expected that RF and VM were sensitive to changes within the walking technique (Table 1). A decrease in PH and an increase in PR significantly increased their $\%EMG_{max}$ (Figure 3). Step lengthening achieved by modified walking techniques (Table 3, Figure 6) plays an important role in this phenomenon since it has been proven that SL, compared to SF, has a greater effect on the increase in the knee joint moment of force [22]. On the other hand, GM was insensitive to changes within the walking technique during CP showing relatively lower $\%EMG_{max}$ values (Table 1, Figure 3). A possible explanation is that GM and other gluteal muscles play a role in maintaining pelvic stability and that these changes in walking technique do not place additional load on maintaining this stability. In other words, GM remains active in a relatively smaller and insensitive amount throughout the entire contact and swing phase, which is necessary for maintaining pelvic stability, independent from walking technique, whether natural or modified. According to the obtained $\%EMG_{max}$ of GM, RF, and VM, it is noticeable that the subjects achieved the particular movement strategy with a greater loading of the knee joint extensors, compared to the possible occurrence of a greater loading of the hip joint extensors by some people [39].

It was observed that only increased PR significantly increased the $\%EMG_{max}$ of TA and BF during CP (Table 1, Figure 3). It is already known that TA controls foot lowering through eccentric contraction from the beginning of CP [11,39]. Furthermore, the primary role of BF is the knee joint flexion, which is also assumed to control foot lowering by controlling the knee joint flexion. On the other hand, a decrease in PH did not significantly change $\%EMG_{max}$, neither of TA nor BF, likely due to the inability to express the basic role in these semiflexed positions in the lower limb joints.

A significant interaction effect was observed on the change in $\%EMG_{max}$ of GL during CP (Table 1). Namely, a significant increase in his $\%EMG_{max}$ due to reduced PH was observed when PR was natural and increased PR when PH was normal (Figure 3). Figure 3 shows supramaximal activation of GL during walking with reduced PH, which is a natural phenomenon in some muscles in the conditions of dynamic contraction [31]. Namely, it is assumed that in the first-mentioned technique, the supramaximal activation occurs because the walking is dominantly performed on the forefoot after the leg passes the vertical projection of the body's center of gravity. In the second-mentioned technique, due to a more extended leg position in the knee joint, propulsion is transferred primarily to the ankle joint, where the more emphasized "pushing" of the foot against the ground occurs, which is already assumed as the function of GL in the later periods of CP [11,39].

The factor of PH and PR did not significantly change the $\%EMG_{max}$ of any monitored muscle during SP (Table 1, Figure 4). Unlike SP when only the weight and inertia of individual body segments are to be overcome, during CP the weight and inertia of the whole body need to be overcome. This explains the low sensitivity of muscle activation to significantly change during the SP with relatively small changes in the inertial load of individual body segments when modifying walking technique. Furthermore, qualitative relationship between the degree of muscle exertion during CP and energy consumption of movement can be observed. Knowing that the greatest changes in %EMG_{max} occurred during CP, it is clear why activities during CP, compared to SP, are more responsible for a greater contribution to total energy consumption [36].

A significant increase in energy consumption as a consequence of decreasing PH and increasing PR (Table 2, Figure 5) can be explained by the modification of the natural and reflex activity and the mechanical conditions of the movement (Table 3). In other words, to attain the same $V_{subtransit}$, SL significantly increases, while SF significantly decreases along with modification of walking technique (Figure 6), which agrees with the results of previous studies [9,20,32,33]. Also, the increased co-activation of antagonistic muscles, and the increased engagement of cognitive structures when the motor task is not natural, may partly contribute to the increase in energy consumption [10,27].

The non-parametric analysis (Table 4) showed that both $RPE_{exertion}$ and $RPE_{fatigue}$ significantly differed between walking techniques (Figure 7). Reducing PH had a greater effect than increasing PR, while applying both changes in walking technique caused the most intense $RPE_{exertion}$ and $RPE_{fatigue}$.

Increasing PR significantly increased %EMG_{max} during CP of almost all monitored muscles, except for GM and GL when PH was reduced, with a more moderate increase in energy consumption and changes in SL and SF relative to decreasing PH. Decreasing PH significantly increased %EMG_{max} of primary movement executors during CP – in RF, VM, and GL when PR was natural, but with a greater increase in energy consumption, changes in SL and SF, and a more intensive %EMG_{max} of the primary extensors in the knee joint – RF and VM, relative to increasing PR. Furthermore, an isolated increase in PR led to a greater increase in %EMG_{max} of TA, BF, and GL during CP, relative to an isolated decrease in PH, with the reverse trend change in %EMG_{max} of RF and VM. All these results lead to the assumption that when walking with an isolated increase in PR, load on the knee joint reduces compared to walking with an isolated decrease in PH, which is important for people who achieve the particular movement strategy with a greater loading and torque production relative to the knee extensors and knee joint [39]. Also, an isolated increase in PR produced similar step lengthening, an increase in energy expenditure, and a less intensive $RPE_{exertion}$ and $RPE_{fatigue}$, compared to an isolated decrease in PH.

Considering all the results, the initial assumption is confirmed – decreased pelvic height and increased pelvic rotation during walking significantly increase muscle activation, energy consumption, perceived exertion and fatigue, and modifications in kinematic characteristics in a healthy and physically active population. Also, it is assumed that the walking technique with an isolated increase in PR in the transverse plane is safer for the locomotor apparatus and can be performed longer while producing similar movement outcomes. To examine the long-term effects of the application of these walking techniques, it is necessary to conduct a longitudinal study. Furthermore, regarding low statistical power of the analysis because of the small sample of subjects, it is recommended to conduct this research on a larger sample of subjects. Since the results of this research could be applied only to healthy and physically active population, they can be attempted to be generalized by conducting the research on a population with general movement disabilities, as there are many different walking modalities, both normal and impaired ones.

5. Conclusions

The results of this research showed the greatest changes in electromyographic activity during CP with the modification of walking technique, while during SP no significant effect of any factor on the change of $\%EMG_{max}$ of any monitored muscle was observed. An increase in PR was shown to be a significant factor for the increase in $\%EMG_{max}$ in almost all monitored muscles during CP (except for GM and GL when PH was reduced). The reduction of PH significantly increased the $\%EMG_{max}$ during CP of RF, VM, and GL when PR was natural. Energy consumption increased significantly, and SL and SF changed inversely and proportionally with the modification of walking technique. Furthermore, both factors significantly increased $RPE_{exertion}$ and $RPE_{fatigue}$. Reducing PH had a greater effect than increasing PR, while applying both changes in walking technique caused the most intense $RPE_{exertion}$ and $RPE_{fatigue}$. In general, decreased pelvic height and increased pelvic rotation during walking significantly increase muscle activation, energy consumption, and perceived exertion and fatigue, along with modifications in kinematic characteristics in a healthy and physically active population. It is assumed that the walking technique with an isolated increase in PR in the transverse plane is safer for the locomotor apparatus and can be performed longer while producing similar movement outcomes. Conducting longitudinal research is recommended to examine the long-term effects of the application of these walking techniques. Furthermore, the results of this research should be attempted to be generalized by repeating it on a larger sample and different populations.

Appendix A. Supporting information

The relativization of the PH during walking, according to the measured angle in the knee joint, is justified by the fact that different proportions between the lengths of the upper and lower leg affect the expression of a different moment of force relative to the hip and knee joint (if PH reduction is the same in different subjects). By introducing angular measures, the height is relativized according to individual anthropometric characteristics, which can be seen from the range of percentage reduction of PH (7.8-11.0%).

Appendix B. Supporting information

The mean $V_{transit}$ value of 7.6 km/h can be explained from three aspects: (1) the subjects maintained a grip on the treadmill handle in front of them while walking, which enabled a slightly increased PR in the transverse plane (step lengthening), (2) the incremental test protocol for $V_{transit}$ determination contained sequences of interval rest – 30 seconds of walking at an initial speed (5 km/h), (3) the subjects did not have the information about the movement speed.

Appendix C. Supporting information

Energy consumption during walking was expressed as the NET $VO_{2submax}$ value [ml/kg/m] because the subjects had an individually determined $V_{subtransit}$ and because of the likely different requirements for balance, which is why the NET value during walking was subtracted from the GROSS value during lying down. First, the NET VO_{2rest} [ml/min] was subtracted from the GROSS $VO_{2submax}$ in the unit of time [ml/min], then the obtained NET value was divided by 60 seconds [s], thus obtaining the value expressed in seconds [ml/s]. Then, the value was divided by the individually determined $V_{subtransit}$ [m/s], thus obtaining the value in the unit of distance [ml/m]. Finally, the value was relativized by dividing with BM [ml/kg/m].

Appendix D. Supporting information

First, the elapsed time [s] required to achieve 10 adjacent two-step cycles was sampled through pressure sensors. Then, the total reached distance [m] was calculated as the multiplication of $V_{subtransit}$ [m/s] and elapsed time [s]. The average SL [m] was calculated as the quotient of the total reached distance [m] and the number of steps, while the average SF [step/s] was calculated as the quotient of the number of steps and the elapsed time [s].

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