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Effect of foot fixation on muscle activation and power output during the Wingate test in cyclists

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Abstract

This study examined the effect of simulated foot fixation on neuromuscular activation and power output during a 30-second isokinetic Wingate-type test in trained cyclists. Sixteen well-trained male cyclists and triathletes completed two randomized crossover trials under distinct pedal–foot interface conditions: fixed-foot, using standard athletic shoes secured to flat pedals with elastic bandage to restrict vertical movement, and free-foot, using standard shoes without fixation. Surface Electromyography (sEMG) was recorded from the vastus lateralis, vastus medialis, rectus femoris, biceps femoris, and tibialis anterior to quantify muscle activation, while an isokinetic ergometer continuously measured mechanical power. Compared with the free-foot condition, simulated fixation produced significantly higher peak power (1365.3 ± 153.5 W *vs*. 1299.0 ± 150.7 W, $p = 0.016$, $d = 0.68$) and mean power (900.6 ± 88.4 W *vs*. 838.3 ± 98.3 W, $p < 0.001$, $d = 1.35$), but also a greater fatigue index (47.8 ± 8.0 % *vs*. 37.2 ± 8.5 %, $p < 0.001$, $d = 1.25$). Integrated EMG (iEMG) values increased across all monitored muscles, particularly in the rectus femoris, biceps femoris, and tibialis anterior, indicating altered neuromuscular coordination and enhanced activation of biarticular and stabilizing muscles. These findings suggest that mechanical stabilization of the foot–pedal interface enhances torque transmission and short-term power generation but concurrently accelerates fatigue development due to elevated neuromuscular and metabolic demands. The results emphasize a trade-off between instantaneous performance gains and fatigue progression,

relevant for interpreting Wingate-type tests and optimizing sprint-specific cycling training and equipment configurations.

Key words: foot fixation, cycling biomechanics, surface EMG, power output, fatigue index, isokinetic Wingate test.

Efficient cycling performance depends on the coordinated activation of lower-limb muscles, particularly under high-intensity or supramaximal conditions where rapid force generation and fatigue resistance are critical. While the biomechanical and physiological determinants of cycling efficiency are well established, the influence of the shoe–pedal interface on neuromuscular activation during maximal efforts remains insufficiently explored. Clip-in pedal systems mechanically couple the shoe and pedal, allowing continuous force application throughout the crank cycle.[1] This enhanced coupling theoretically improves torque transmission and mechanical efficiency, yet its neuromuscular consequences under anaerobic conditions are not fully understood.

Surface Electromyography (sEMG) studies have demonstrated that external constraints, cadence, and workload significantly influence muscle recruitment timing and intermuscular coordination during cycling.[2,3] The Vastus Lateralis (VL), Vastus Medialis (VM), and Rectus Femoris (RF) are key contributors to the downstroke phase, while the Biceps Femoris (BF) and Tibialis Anterior (TA) are more active during the upstroke and transition. Modifying the mechanical coupling between foot and pedal could therefore alter these activation strategies. At submaximal workloads, increased pull-up action can improve pedaling effectiveness but may reduce gross efficiency because of elevated co-activation and metabolic cost.[4-6] These neuromechanical trade-offs may be amplified at supramaximal intensities, where both power output and fatigue rates are maximized.

The Wingate anaerobic test provides a robust model for examining these interactions. As a standardized 30-second all-out protocol, it quantifies peak and mean power, fatigue dynamics, and anaerobic capacity while eliciting maximal muscle activation.[6,7] Integrating EMG measurements within this framework enables simultaneous assessment of mechanical and neuromuscular responses.

Therefore, the purpose of this study was to investigate the acute effects of foot fixation on power output and sEMG activity during a 30-second Wingate test in trained cyclists. It was hypothesized that, compared with free-foot pedaling, the fixed-foot condition would increase activation of primary knee extensors and enhance peak and mean power, but also accelerate fatigue. These findings may advance understanding of how mechanical coupling influences performance and fatigue during maximal cycling and support evidence-based approaches to training and testing design.

The primary research question of this study was whether simulated foot fixation alters mechanical power output and lower-limb neuromuscular activation during a 30-second Wingate test in trained cyclists.

We hypothesized that simulated foot fixation would result in higher peak and mean power output, accompanied by increased neuromuscular activation, particularly in muscles involved in pedal stabilization and upstroke phases, compared with free foot pedaling.

Materials and Methods

Participants

Seventeen well-trained male cyclists and triathletes (mean ± SD: age = 23.2 ± 1.75 years, height = 183.0 ± 4.02 cm, body mass = 76.2 ± 4.58 kg, BMI = 22.8 ± 2.43 $kg \cdot m^{-2}$) volunteered to participate in this study.

The inclusion criteria were: i) male sex, ii) age between 18 and 30 years, iii) a minimum of five years of systematic endurance training experience, iv) current participation in competitive cycling or triathlon at the national level, and v) familiarity with high-intensity ergometer testing protocols.

Participants were excluded if they i) reported any current or chronic musculoskeletal pain, ii) had sustained an acute musculoskeletal injury within the previous six months requiring medical intervention or training interruption longer than 14 days, iii) presented with a diagnosed cardiovascular or neurological disorder, or iv) reported any condition that could affect lower-limb function, power production, or neuromuscular activity during cycling performance.

Each subject was familiarized with laboratory ergometer testing procedures prior to data collection to minimize learning effects. The study was approved by the Ethics Committee of the Faculty of Physical

Education and Sport (Protocol No. 14/2024) and was conducted in accordance with the ethical principles of the Declaration of Helsinki (1964), as revised in 2013. In line with the recommendations of Winter and Fowler,[8] all subsequent mechanical and performance variables in this study were quantified using standardized SI-based definitions to ensure methodological consistency and comparability across participants.

Although seventeen participants were initially recruited, one participant was excluded from all analyses due to excessive surface EMG noise and signal saturation affecting more than 40% of the recording. Consequently, data from sixteen participants were included in the final statistical analysis.

The sample size was determined based on previous EMG-based Wingate and high-intensity cycling studies reporting medium to large within-subject effect sizes for changes in power output and muscle activation (Cohen's d ≈ 0.6–1.2). For paired experimental designs with an alpha level of 0.05, a sample of 12–15 participants is generally sufficient to achieve statistical power above 80% for detecting such effects. Given the homogeneous training status of the present cohort and the within-subject crossover design, the final sample of sixteen participants was considered adequate to detect practically meaningful differences between conditions.

Experimental design

All testing was conducted under standardized environmental conditions (temperature 22–23 °C, relative humidity 45–55%) in the Human Performance Laboratory of the Faculty of Physical Education and Sport, Comenius University in Bratislava, between May and July 2025. Although based on the principles of the classical Wingate anaerobic test, the use of an isokinetic ergometer ensured constant cadence and thus represented a modified Wingate-type protocol. A randomized, counterbalanced, crossover design was employed to minimize order and carry-over effects. Each participant completed two experimental trials under distinct pedal–foot interface conditions: i) fixed-foot, using standard athletic shoes on flat pedals fixed with elastic bandage to maintain identical conditions for all tested athletes, and ii) free-foot, using standard athletic shoes on flat pedals without mechanical fixation. The trials were separated by a minimum of seven days to allow full recovery and to prevent residual fatigue. Prior to each test, participants performed a standardized 5-minute warm-up at 90 rpm on the ergometer at a self-selected submaximal load, followed by a brief rest interval. The main trial consisted of a 30-second Wingate anaerobic test performed at a target cadence of 110 rpm, with participants instructed to reach maximal effort from the onset and maintain it throughout the test duration. Power output was recorded continuously, and all verbal cues and environmental stimuli were standardized across sessions.

Muscle activity (sEMG)

Surface EMG (sEMG) was used to quantify the activation of the dominant-leg VL, VM, RF, BF, and TA muscles during the 30-s Wingate test. Measurements were obtained using a Trigno Wireless EMG System (Delsys Inc., Delsys Europe, Manchester, UK), which provides high-fidelity, low-noise signal acquisition suitable for dynamic exercise conditions. Electrode placement followed the recommendations of the Surface EMG for Non-Invasive Assessment of Muscles (SENIAM) project.[11] Before electrode application, the skin was shaved, lightly abraded, and cleansed with isopropyl alcohol to reduce impedance, electrode sites were outlined with a surgical marker to ensure consistent placement across sessions.

Raw EMG signals were sampled at 2000 Hz and processed offline using EMGworks Analysis software (Delsys Inc.). The signals were band-pass filtered (20–450 Hz, 4th-order zero-lag Butterworth) to eliminate motion artifacts and high-frequency noise, full-wave rectified, and smoothed using a Root Mean Square (RMS) algorithm with a 0.125 s window and 50% overlap. The resulting amplitude-time integral was expressed as integrated EMG (iEMG, μV·s), computed over the entire 30-s effort for each muscle. The iEMG metric was selected as it provides a robust index of total neuromuscular activation, reflecting both activation magnitude and duration during maximal, non-steady-state exercise. However, it is important to note that surface EMG amplitude reflects the neural drive to the muscle rather than muscle force per se, and can be influenced by factors such as electrode placement, tissue filtering, and muscle fiber conduction velocity.[9-11]

Power output assessment

Mechanical performance was assessed using an isokinetic cycle ergometer (Physio Studio, Slovakia) developed by Hamar *et al.*,[20] a fully computerized, feedback-controlled system designed to maintain a constant pedaling cadence irrespective of the force applied by the participant. The ergometer features an electronically regulated braking mechanism that continuously adjusts resistive torque: any increase in crank angular velocity above the target cadence automatically triggers a proportional rise in resistance, whereas slight reductions prompt an immediate release of braking force. This closed-loop control ensures a virtually stable rotational frequency throughout the test, allowing the athlete to concentrate exclusively on maximal power generation without interference from cadence fluctuations. In practice, transient deviations in rotational speed are imperceptible to the participant due to the system's rapid feedback response.

Prior to testing, the ergometer was individually adjusted for each participant. Saddle height was set so that the knee joint angle at bottom dead centre was approximately 25–35°, measured with a handheld goniometer while the participant sat in the saddle with the crank aligned vertically. Crank length was fixed at 170 mm for all participants. These settings were recorded at the first visit and replicated identically in both experimental conditions and across sessions. This procedure was adopted to minimize the influence of individual differences in lower-limb length and saddle height on pedaling mechanics and power output.

The ergometer simultaneously records effective (tangential) crank forces and computes instantaneous and mean power outputs across the entire pedal cycle. The integrated data-acquisition module enables time- and crank-angle-resolved analyses of force and power, allowing detailed evaluation of pedaling technique, inter-limb asymmetries, and determination of the subject's individual optimal cadence - the cadence at which maximal mechanical power is achieved. This instrument is routinely used in both functional diagnostics and sports-performance assessment, providing reliable quantification of lower-limb strength and anaerobic power capacities.

From the ergometer's output, three primary performance indices were extracted: peak power (W), defined as the highest instantaneous value achieved during the 30-s Wingate effort, mean power (W), representing the average power maintained across the test duration, and the fatigue index (%), calculated as the relative decline from maximal to minimal power within the 30-s period. These metrics collectively describe maximal anaerobic capacity and fatigue dynamics of the lower-limb musculature under supramaximal loading. All power data were automatically recorded by the ergometer's proprietary software (Ergocycle 1.751, Physio Studio) and subsequently verified using independent spreadsheet-based calculations to ensure analytical consistency across trials.

Limitations

Although saddle height and crank length were standardized within each participant and replicated across conditions, we did not systematically analyse the influence of inter-individual differences in lower-limb length or joint geometry on the observed responses. It is therefore possible that subtle anthropometric variations contributed to between-subject variability in power and EMG outcomes. A further methodological limitation is that surface EMG does not allow direct quantification of muscle force. The present iEMG values should therefore be interpreted as relative indices of neuromuscular activation and coordination between conditions, not as absolute measures of contractile output. In addition, EMG amplitudes were not normalized to a reference contraction, and potential signal crosstalk between adjacent muscles cannot be fully excluded, which may influence the absolute magnitude of recorded iEMG values.

Statistical analysis

All statistical analyses were performed in SPSS (version 23.0, IBM Corp., Armonk, NY, USA). Data distribution normality was verified using the Shapiro–Wilk test, confirming the suitability of parametric analyses. Inter-individual variability was additionally evaluated using Coefficients of Variation (CV%), which ranged between 8–12% for power-related variables and 10–15% for iEMG measures, values consistent with previous reports describing natural variability in cycling performance. To examine within-subject differences between pedal conditions (fixed-foot vs. free-foot), paired-sample t-tests were applied to all dependent variables, including integrated EMG (iEMG) amplitudes and performance indices (peak power, mean power, and fatigue index).

Effect sizes (Cohen's *d*) were interpreted according to the scale proposed by Hopkins,[12] defined as trivial (0–0.2), small (0.2–0.6), moderate (0.6–1.2), large (1.2–2.0), and very large (>2.0). Descriptive statistics (mean ± SD, minimum, maximum, and coefficient of variation) were computed for each variable to characterize central tendency and dispersion. The level of statistical significance was set at $p < 0.05$. All data visualizations and normality inspections were additionally verified by boxplot and Q–Q plot analyses to identify potential outliers or deviations from homoscedasticity.

Results

Performance outcomes consistently favored the fixed-foot condition. Peak power (Table 1) was significantly higher with fixation (1365.3 ± 153.5 W) compared with the free-foot condition (1299.0 ± 150.7 W), representing an average increase of approximately 66 W (Figure 1). Mean power followed a similar trend, being greater with fixation (900.6 ± 88.4 W) than without (838.3 ± 98.3 W), corresponding to an improvement of roughly 62 W (Figure 2). In contrast, the fatigue index was markedly higher when the feet were attached (47.8 ± 8.0 %) relative to the free-foot condition (37.2 ± 8.5 %), indicating a steeper decline in power output across the 30-s effort (Figure 3).

Paired sample t-tests confirmed statistically significant differences between conditions for all three performance variables (peak power: $t(15) = 2.71$, $p = 0.016$, $d = 0.68$, mean power: $t(15) = 5.39$, $p <$

0.001, d = 1.35, fatigue index: t (15) = 4.98, $p < 0.001$, d = 1.25). Expressed as relative changes, fixation enhanced peak power by approximately 5.1 %, mean power by 7.4 %, and increased fatigue index by 28.5 % compared with free foot pedaling. Collectively, these results indicate that foot fixation promotes higher instantaneous and sustained power generation, although with an affiliated acceleration of fatigue. The observed effect sizes ranged from moderate (*d* = 0.68) to large (*d* = 1.35 and *d* = 1.25), indicating practically meaningful performance differences according to Hopkins.[12]The neuromuscular activation analysis revealed systematically greater overall EMG activity in the fixed-foot condition. The integrated EMG (iEMG) summed across all monitored muscles increased from 551.6 ± 106.0 µV·s in the free-foot condition to 632.0 ± 104.0 µV·s with fixation, representing an average rise of 14.6 % (Figure 4). Examination of individual muscle responses showed the most pronounced elevations in activation within the rectus femoris (+25.9 %, 141.8 ± 57.0 *vs* 112.6 ± 47.6 µV·s), biceps femoris (+50.5 %, 101.3 ± 26.1 *vs* 67.3 ± 20.9 µV·s), and tibialis anterior (+71.2 %, 120.7 ± 40.1 *vs* 70.5 ± 29.1 µV·s). Smaller increases were observed in the vastus medialis (+0.9 %, 184.3 ± 75.4 *vs* 182.6 ± 47.6 µV·s), while the vastus lateralis remained effectively unchanged (−0.5 %, 115.2 ± 29.4 *vs* 115.8 ± 27.1 µV·s). These findings suggest that fixation primarily amplifies the contribution of biarticular and stabilizing muscles responsible for smooth force transfer during upstroke and transition phases (Figure 5), rather than uniformly increasing activation across all extensors.

The inter-individual coefficients of variation (8–12% for power indices; 10–15% for iEMG) confirmed consistent within-group responses, reflecting stable measurement reliability and homogeneity typical for trained cyclists.

Discussion

The present study demonstrates that foot fixation during supramaximal cycling enhances both peak and mean power output, while concurrently increasing the rate of fatigue across the 30-second Wingate test. The magnitude of these effects ranged from moderate to large, according to Hopkins'[12] thresholds for effect size interpretation, indicating practically meaningful performance differences between pedal conditions. These findings indicate that mechanical stabilization provided by fixed foot on the pedals promotes more effective force transmission to the crank, enabling greater torque generation during the downstroke and transition phases. This interpretation aligns with the theoretical and computational models of Neptune and Herzog,[1] who showed that alterations in task mechanics directly modulate the coordination of lower-limb muscles and the distribution of power around the crank cycle. The moderate effect size observed for peak power (d = 0.68) supports that this enhancement, while statistically significant, is of moderate practical relevance, whereas the large effects seen for mean power (d = 1.35) and fatigue index (d = 1.25) reflect a more substantial physiological impact.

Our electromyographic data, specifically the increased total iEMG and the marked elevations in rectus femoris, biceps femoris and tibialis anterior, support the notion that fixation modifies neuromuscular recruitment strategies and overall activation patterns. Given that surface EMG provides an indirect index of muscle activation rather than a direct measure of contractile force, these findings should be interpreted as changes in neural drive and coordination rather than quantitative estimates of muscle force production.[9-11] Similar patterns have been reported in controlled pedaling studies, where mechanical constraints such as foot fixation or altered workload reorganize intermuscular coordination patterns.[2,3,10] The enhanced activation of the quadriceps group observed here is consistent with its primary contribution to downstroke power, while the concomitant activation of the biceps femoris and tibialis anterior likely reflects greater engagement during the upstroke and transition phases to maintain smooth pedal force application.

At submaximal workloads, previous work has demonstrated that emphasizing the “pull-up” component of the pedal cycle can improve the index of effectiveness but may reduce gross mechanical efficiency.[4,5] Comprehensive reviews further indicate that pedaling effectiveness tends to increase with power output but declines with cadence, primarily due to shifting mechanical constraints and motor control strategies.[2] Under supramaximal Wingate conditions, the increased torque capacity enabled by fixation likely accentuates these trends—enhancing both peak and sustained power but simultaneously accelerating fatigue-related declines in output. The observed elevation in the fatigue index is consistent with classical physiological models describing rapid phosphocreatine depletion, accumulation of inorganic phosphate, and hydrogen ions during repeated high-intensity contractions.[13,14]

At the joint-mechanical level, our results also parallel the findings of Martin and Brown,[15] who reported that as fatigue develops during maximal cycling, power contribution progressively shifts from the knee extensors to the hip extensors, suggesting compensatory redistribution of effort. Such dynamic adaptations may partially explain the steep decline in power observed toward the end of the Wingate test in the fixed-foot condition.

Foot–pedal interface mechanics likely play an additional role in mediating these effects. Clipless pedal systems increase the rigidity of the foot–shoe–pedal coupling and modify plantar pressure distribution,[2] thereby improving mechanical stability and reducing energy losses through shear deformation at the contact point. While Mornieux *et al.*[5] found limited kinematic differences between flat and clipless pedals under steady-state conditions, the present findings indicate that, during supramaximal efforts, mechanical fixation provides a distinct performance advantage that is offset by a greater fatigue rate.

In addition to findings from traditional cycling biomechanics, relevant insights emerge from research on Functional Electrical Stimulation (FES) cycling, which—although conducted in neurologically impaired populations—provides important evidence regarding the role of mechanical stability in modulating muscle activation patterns. Arnin *et al.*,[16] who analyzed EMG profiles during FES-driven cycling at the Cybathlon 2016, demonstrated that increasing the mechanical stability of the foot–pedal interface markedly alters intra-cycle activation timing and reduces variability in EMG amplitude. Their work showed that stable foot–pedal coupling facilitates more predictable neuromuscular coordination and smoother force transmission, even under highly constrained activation conditions. These findings align with our observations that simulated foot fixation increases iEMG in both prime movers and stabilizing muscles, suggesting that reductions in unwanted vertical or mediolateral foot motion enhance the consistency of force application throughout the crank cycle.

Further support for this interpretation is provided by Coste *et al.*,[17] who compared performance strategies among FES pilots competing in the inaugural Cybathlon. Their analysis highlighted that secure foot fixation was a critical determinant of power output, technical efficiency, and overall race performance, as insufficient stabilization led to asynchronous propulsion, ineffective force transfer, and increased metabolic cost. Although FES cycling differs fundamentally from voluntary supramaximal efforts, the mechanical principles described by Coste and colleagues corroborate the broader implication of the present study: optimizing the foot–pedal interface directly influences torque production capacity and neuromuscular organization. The parallels between FES-based and voluntary cycling models reinforce the conclusion that improved stability enhances mechanical effectiveness but may concurrently amplify fatigue due to heightened neuromuscular demands.

Several methodological considerations strengthen the interpretation of these results. The randomized crossover design minimized interindividual variability and order effects, while the within-subject comparison allowed direct assessment of the mechanical and neuromuscular consequences of foot fixation. Nonetheless, the interpretation of surface EMG amplitude data must consider physiological factors such as muscle fiber conduction velocity and metabolic fatigue, which influence signal magnitude.[11] To further delineate the mechanical mechanisms underlying these observations, future research should integrate instrumented pedals and crank-based force sensors with high-density EMG to distinguish between effective and ineffective force components across crank phases.[18,9] Additionally, experimental manipulations of cadence, seat-tube angle, and crank length could elucidate how geometric and kinematic factors influence the relationship between fixation, torque application, and fatigue.[14,16]

Finally, extending the present approach to different Wingate durations or intermittent protocols[19] may clarify the temporal evolution of neuromuscular fatigue and its dependence on mechanical constraints. In practical terms, the large effects for mean power and fatigue index indicate that mechanical fixation meaningfully alters power-generation dynamics, whereas the moderate effect on peak power suggests that its influence is less pronounced at the onset of the sprint. Collectively, these findings highlight the complex interplay between equipment configuration, neuromuscular control, and metabolic demand during maximal cycling, emphasizing that mechanical fixation enhances instantaneous performance but accelerates the onset of fatigue through intensified muscular and metabolic loading.

Conclusions

In summary, simulated foot fixation during 30-s supramaximal cycling significantly enhances both peak and mean power output, reflecting improved mechanical force transfer and more coordinated neuromuscular activation. However, these performance gains come at the cost of faster fatigue development, highlighting a practical trade-off between enhanced power generation and reduced fatigue resistance during short-duration supramaximal cycling. This finding has direct implications for the design and interpretation of laboratory testing and sprint-specific training.

From an applied perspective, simulated fixation representing a controlled analogue of clip-in pedal systems appears advantageous for athletes targeting explosive or high intensity efforts, where brief increases in power outweigh the cost of faster fatigue onset. Conversely, for endurance-oriented cycling or repeated sprint scenarios, the elevated fatigue index observed under simulated fixation should be carefully considered when optimizing pedaling technique and equipment configuration.

It is important to note that these findings pertain specifically to short-duration supramaximal cycling performed under simulated foot fixation and should not be directly generalized to prolonged or submaximal cycling conditions, where different neuromuscular and metabolic dynamics may prevail. Future investigations integrating force-pedal sensors, high-density EMG, and metabolic measurements are warranted to clarify the underlying neuromechanical mechanisms and to determine whether training adaptations can mitigate the fatigue effects associated with mechanical fixation. Collectively, these results contribute to a more comprehensive understanding of how foot-pedal interface mechanics modulate performance and fatigue dynamics during maximal cycling exercise.

List of Abbreviations

BF, *Biceps femoris*

BMI, Body mass index

CV%, Coefficient of Variation

d, Cohen's *d* (effect size)

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EMG, Electromyography

Hz, Hertz

ICMJE, International Committee of Medical Journal Editors

iEMG, Integrated electromyography

$kg \cdot m^{-2}$, Kilogram per square meter

$\mu V \cdot s$, Microvolt-second

p, Probability value (statistical significance)

RMS, Root mean square

RF, *Rectus femoris*

sEMG, Surface electromyography

SENIAM, Surface EMG for Non-Invasive Assessment of Muscles (project)

TA, *Tibialis anterior*

VL, *Vastus lateralis*

VM, *Vastus medialis*

VEGA, Scientific Grant Agency of the Ministry of Education, Science, Research and Sport of the Slovak Republic

W, Watt

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Contributions

Matúš Putala, conceptualization, data interpretation, manuscript writing, and final approval of the submitted version; Viktor Oliva, methodological supervision, laboratory coordination, validation of physiological data, and critical revision of the manuscript; Simon Brunovský, participant coordination, data collection, and assistance with ergometer testing procedures; Gabiel Buzgo, data collection, study design, electromyography setup and signal processing, statistical analysis, and figure preparation. All authors have read and approved the final version of the manuscript and agree with its submission to the journal. The authors meet the criteria for authorship as defined by the International Committee of Medical Journal Editors (ICMJE).

Conflict of interest

The authors declare no commercial or financial relationships that could be construed as a potential conflict of interest. No author has any personal or institutional interest in the materials, equipment, or software used in this research.

Ethical approval

The study was approved by the Ethics Committee of the Faculty of Physical Education and Sport (Protocol No. 14/2024) and was conducted in accordance with the ethical principles of the Declaration of Helsinki (1964), as revised in 2013. Informed consent was obtained from all subjects involved in the study prior to participation.

Availability of data and materials

The datasets generated and analyzed during the current study are available from the corresponding author upon reasonable request.

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References

1. Neptune RR, Herzog W. Adaptation of muscle coordination to altered task mechanics during steady-state cycling. J Biomech 2000;33:165-72.
2. Dorel S, Couturier A, Hug F. Intra-session repeatability of lower limb muscles activation pattern during pedaling. J Electromyogr Kinesiol 2008;18:857-65.
3. Hug F, Dorel S. Electromyographic analysis of pedaling: a review. J Electromyogr Kinesiol 2009;19:182-98.
4. Korff T, Romer LM, Mayhew I, Martin JC. Effect of pedaling technique on mechanical effectiveness and efficiency in cyclists. Med Sci Sports Exerc 2007;39:991-5.
5. Mornieux G, Stapelfeldt B, Gollhofer A, Belli A. Effects of pedal type and pull-up action during cycling. Int J Sports Med 2008;29:817-22.
6. Bini RR, Hume PA, Croft JL, Kilding AE. Pedal force effectiveness in cycling: a review of constraints and training effects. Sports Biomech 2013;12:343-58.
7. Inbar O, Bar-Or O, Skinner JS. The Wingate anaerobic test. Champaign (IL): Human Kinetics; 1996.
8. Winter EM, Fowler N. Exercise defined and quantified according to the Système International d'Unités. J Sports Sci 2009;27:447-60.
9. Enoka RM, Duchateau J. Muscle fatigue: what, why and how it influences muscle function. J Physiol 2008;586:11-23.
10. Hermens HJ, Freriks B, Disselhorst-Klug C, Rau G. Development of recommendations for SEMG sensors and sensor-placement procedures. J Electromyogr Kinesiol 2000;10:361-74.
11. Stewart DG, Macaluso A, De Vito G. Muscle fiber conduction velocity during a 30-s Wingate test. J Electromyogr Kinesiol 2011;21:1072-7.
12. Hopkins WG. A scale of magnitudes for effect statistics. Sports Sci 2002;6:1-7.
13. Gaitanos GC, Williams C, Boobis LH, Brooks S. Human muscle metabolism during intermittent maximal exercise. J Appl Physiol 1993;75:712-9.
14. Sahlin K. Muscle energetics during explosive activities and potential effects of nutrition and training. Sports Med 2014;44:167-73.
15. Martin JC, Brown NAT. Joint-specific power production and fatigue during maximal cycling. J Biomech 2009;42:474-9.

16. Arnin J, Yamsa-Ard T, Triponyuwasin P, Wongsawat Y. Development of practical functional electrical stimulation cycling systems based on an electromyography study of the Cybathlon 2016. Eur J Transl Myol 2017;27:7111.
17. Coste CA, Bergeron V, Berkelmans R, et al. Comparison of strategies and performance of functional electrical stimulation cycling in spinal cord injury pilots for competition in the first ever CYBATHLON. Eur J Transl Myol 2017;27:7219.
18. Sanderson DJ, Black A. The effect of prolonged cycling on pedal forces. Med Sci Sports Exerc 2003;35:1607-13.
19. Driss T, Vandewalle H. The measurement of maximal (anaerobic) power output on a cycle ergometer: a critical review. Sports Med 2013;43:647-69.
20. Hamar, D., Gazovic, O., & Schickhofer, P.. Isokinetic cycle ergometer--application in sports medicine. Wiener medizinische Wochenschrift, 1995;145 22, 606-10.

Table 1. Performance outcomes under fixed-foot and free-foot conditions.

Variable	Fixed-foot (Mean ± SD)	Free-foot (Mean ± SD)	p-value
Peak Power (W)	1365.3 ± 153.5	1299.0 ± 150.7	0.016
Mean Power (W)	900.6 ± 88.4	838.3 ± 98.3	<0.001
Fatigue Index (%)	47.8 ± 8.0	37.2 ± 8.5	<0.001

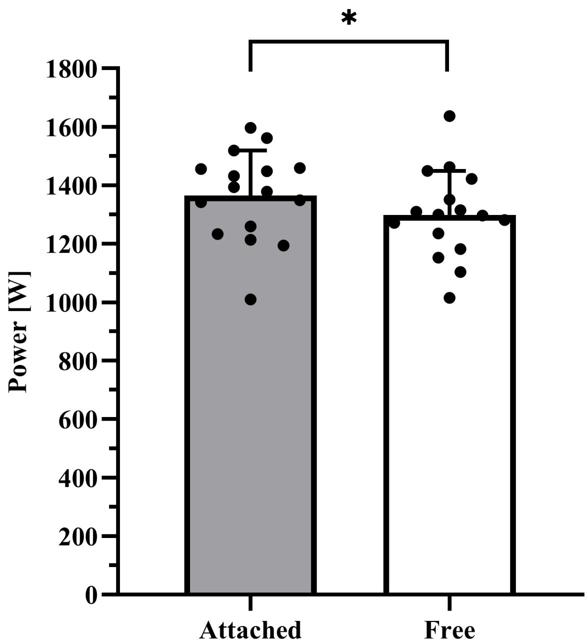

Figure 1. Maximal power during Wingate test (fixed *vs*. free-foot).

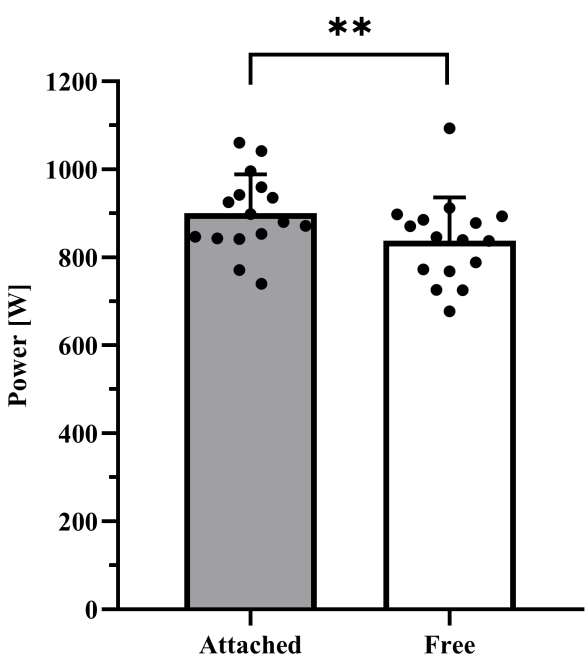

Figure 2. Mean power during Wingate test (fixed *vs*. free-foot).

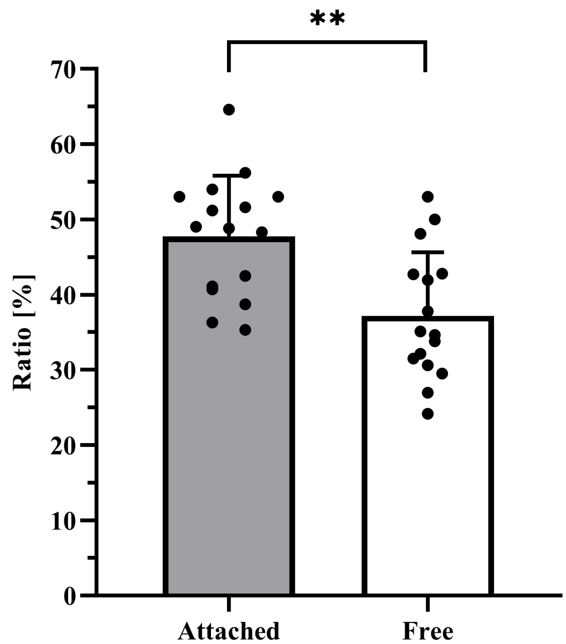

Figure 3. Fatigue index during Wingate test (fixed *vs*. free-foot).

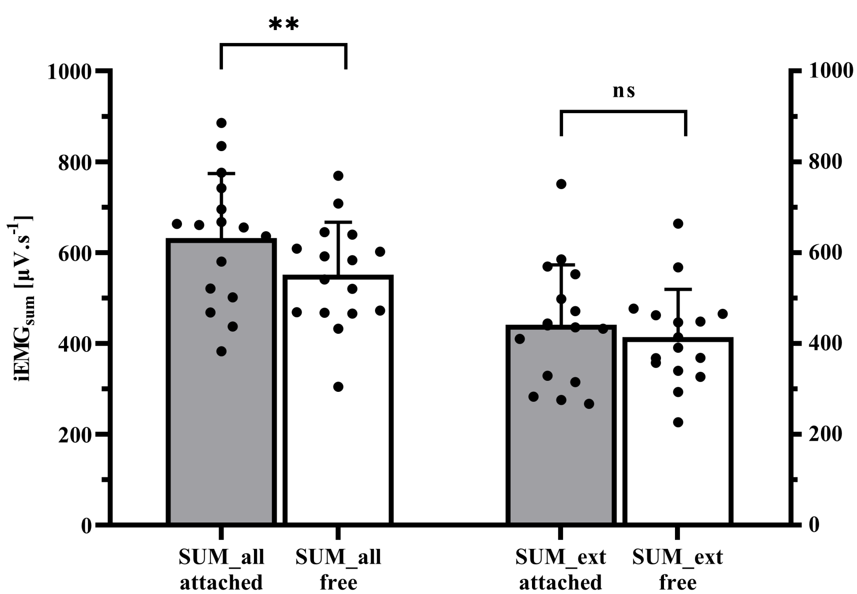

Figure 4. Integrated EMG across all muscles (fixed *vs*. free-foot).

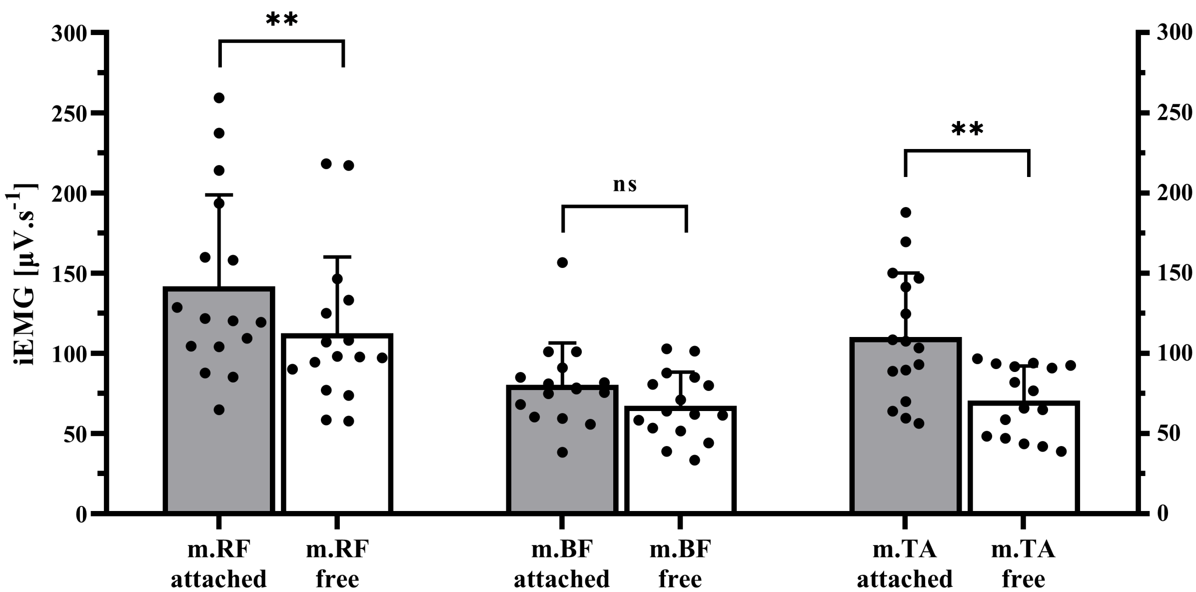

Figure 5. Integrated EMG by individual muscles (fixed *vs*. free-foot).