UNIVERSITA' DEGLI STUDI DI VERONA

DEPARTMENT OF

Neurosciences, Biomedicine and Movement Sciences

DOCTORAL PROGRAM IN

Neurosciences, Psychological and Psychiatric Sciences, and Movement Sciences

Cycle / year: XXXVIII °/2022

TITLE OF THE DOCTORAL THESIS

Biomechanical and Physiological Analysis for the Prescription and Control of Training Programs Using an E-Bike.

S.S.D. M-EDF-01/B

Coordinator: Prof. Gajofatto Alberto

Signature ______________________

Tutor: Prof. Bortolan Lorenzo

Signature ______________________

Tutor: Prof. Pellegrini Barbara

Signature ______________________

Doctoral Student: Dottoressa Todesco Beatrice

Signature

La borsa di dottorato è stata cofinanziata con le risorse del PNRR:

• per il DM 351 nell'ambito della Missione 4 ("Istruzione e ricerca") – Componente 1 ("Potenziamento dell'offerta dei servizi di istruzione: dagli asili nido all'Università"), Investimento 3.4. ("Didattica e competenze universitarie avanzate") e Investimento 4.1 ("Estensione del numero di dottorati di ricerca e dottorati innovativi per la pubblica amministrazione e il patrimonio culturale") - progetto M4C1 –Inv. 3.4 e progetto M4C1 – Inv. 4.1

• per il DM 352, nell'ambito della Missione 4 ("Istruzione e Ricerca") – Componente 2 ("Dalla Ricerca all'Impresa"), Investimento 3.3 ("Introduzione di dottorati innovativi che rispondono ai fabbisogni di innovazione delle imprese e promuovono l'assunzione dei ricercatori da parte delle imprese") – progetto M4C2 Investimento 3.3

Sommario

ABSTRACT

This PhD project consists of two main studies aimed at providing a detailed analysis of the interaction between the cyclist and the electric bicycle from multiple perspectives. E-bikes have become a rapidly growing phenomenon in recent years. They provide pedalling assistance to the rider, offering numerous advantages. Among the most common benefits is the reduction of barriers typically associated with the use of conventional bicycles, such as limited time availability, distance to be covered, and sweating, all of which influence the potential of e-bikes as a viable means of transportation. Moreover, through their assistance, e-bikes make physical activity accessible even to individuals with low fitness levels, while still eliciting an effort sufficient to contribute to improvements in the rider's fitness.

The literature already includes numerous studies reporting the health and performance benefits of e-bikes and describing the behaviour of key physiological parameters during their use. Nevertheless, there appear to be relatively few studies that examine muscular activity and delineate the effects of changes in assistance level and pedalling cadence. Furthermore, with regard to field-based studies, to the best of our knowledge, no research has yet investigated the quality and characteristics of e-bike vibrations on different terrains and the implications these may have on riding comfort and bike handling.

Study one

Purpose: the aim of the study is to identify the effects of the use of different levels of assistance of the electric bike and different cadences on metabolic and neuromuscular responses.

Methods: thirteen recreational cyclists were tested in 5-minute trials at the external power output matching their individual second threshold, comparing no assistance (A0) and two levels of assistance (the lower one- A1 and the highest- A3) at two different pedalling cadences (60 - 90 RPMs). During each trials data of gas exchange parameters and muscular activation were collected and mean values were calculated.

Results: the physiological parameters showed an effect of both cadence and levels of assistance and the presence of interaction between the two factors. Significantly lower values were observed at 60 RPM compared to 90 RPM under motor-assisted conditions. This may be due not only to neuromuscular factors but also to the motor delivering more

power at 60 RPM than at 90 RPM, as measured. In the RMS values of electromyography (EMG) signal, there where an effect of the assistance on the muscles of the upper leg and an effect of the cadence in the muscles of the lower leg. No interaction between factors was found.

Conclusion: based on the results of the study, we can conclude that rider physiological effort and the choice of optimal cadence when riding an e-bike may differ from what is already known for conventional bicycles.

Study two

Purpose: the aim of the study is analyzing the vibrations transmitted to both the bicycle frame and the rider during riding with different type of bike (E-bike and conventional bike-MTB) on different types of terrain.

Methods: triaxial accelerometers were mounted on the handlebar, on the top tube beneath the seat, and on the rider at the T4 and L3 spinal levels. The accelerometer data were processed to compute both total acceleration and the low-frequency component (<5 Hz) using Butterworth filtering. Eight recreational cyclists performed controlled uphill and downhill rides over asphalt, low-difficulty off-road, and high-difficulty off-road tracks.

Results: the MTB exhibited higher frame vibrations compared to the e-bike, while no significant differences were found between electric assistance levels (HIGH vs. ECO) during uphill riding.

Conclusion: overall, the results suggest that terrain complexity and bike mass distribution have a more pronounced impact on vibration exposure than motor assistance settings.

Overall, these two studies explored aspects of E-bike use that remain relatively under-investigated. Based on the physiological parameters examined in the first study, we can conclude that riding an E-bike appears to promote the use of lower cadences, owing both to the additional power provided by the motor and to the inherently lower oxygen consumption associated with low-cadence pedalling. In contrast, the electromyographic parameters did not display a uniform pattern. Specifically, increases in cadence were associated with greater activation of the muscles in the lower leg, whereas increases in motor assistance predominantly reduced muscular activation in the thigh muscles. With regard to the study conducted on vibrations, the e-bike generated lower vibration levels than the conventional bicycle during downhill sections. This is most likely attributable to differences in total weight and its distribution. In contrast, during uphill segments, the various levels of motor assistance did not appear to affect vibration magnitude.

PUBBLICATION AND SUBMITTED PAPERS

Submitted

Study one

Sport science for health, Springer Nature

Minor revision (on 13/03/2026)

PHYSIOLOGICAL, MUSCULAR AND PERCEPTUAL PARAMETERS WHILE RIDING AN E-BIKE AT DIFFERENT LEVELS OF ASSISTANCE AND DIFFERENT PEDALLING CADENCE.

Todesco Beatrice [1,3], Pellegrini Barbara [1,2], Decet Marco[1], Schena Federico [1,3] , Bortolan Lorenzo [1,2].

[1] CeRiSM Sport Mountain and Health Research Centre, University of Verona, Rovereto, Italy;

[2] Department of Engineering for Innovation Medicine, University of Verona, Verona, Italy;

[3] Department of Neurosciences, Biomedicine and Movement Sciences, University of Verona, Verona, Italy.

Submitted

Study two

IEE STAR sport technology and research

COMPARATIVE ANALYSIS OF VIBRATION EXPOSURE ON E-MOUNTAIN BIKES AND CONVENTIONAL MOUNTAIN BIKES ACROSS VARIED TERRAINS.

Todesco Beatrice [1,3], Pellegrini Barbara [1,2], Schena Federico [1,3] , Bortolan Lorenzo [1,2].

[1] CeRiSM Sport Mountain and Health Research Centre, University of Verona, Rovereto, Italy;

[2] Department of Engineering for Innovation Medicine, University of Verona, Verona, Italy;

[3] Department of Neurosciences, Biomedicine and Movement Sciences, University of Verona, Verona, Italy.

POSTER PRESENTATION

SISMES 2023

EVALUATION OF METABOLIC AND MECHANICAL PARAMETERS IN DIFFERENT PHASES OF THE MENSTRUAL CYCLE IN YOUNG FEMALE CYCLIST. A PILOT STUDY.

Todesco Beatrice[1 3], Savoldelli Aldo[1 4], Fornasiero Alessandro [1 2], Decet Marco[1 3], Pellegrini Barbara[1 2], Schena Federico [1 3].

[1] CeRiSM Sport Mountain and Health Research Centre, University of Verona, Rovereto, Italy;

[2] Department of Engineering for Innovation Medicine, University of Verona, Verona, Italy;

[3] Department of Neurosciences, Biomedicine and Movement Sciences, University of Verona, Verona, Italy;

[4] Department of Cellular, Computational and Integrative Biology, University of Trento, Trento, Italy.

SISMES 2024

EFFECTS OF CADENCE, LEVELS OF ASSISTANCE AND THEIR INTERACTION ON MUSCOLAR ACTIVATION WHILE RIDING AN E-BIKE. A PILOT STUDY.

Todesco Beatrice[1 3], Decet Marco[1 3], Pellegrini Barbara[1 2], Bortolan Lorenzo [1 2], Schena Federico [1 3].

[1] CeRiSM Sport Mountain and Health Research Centre, University of Verona, Rovereto, Italy;

[2] Department of Engineering for Innovation Medicine, University of Verona, Verona, Italy;

[3] Department of Neurosciences, Biomedicine and Movement Sciences, University of Verona, Verona, Italy.

ICSS 2025

UPPER LIMB MUSCLE ACTIVATION IN ELITE SKI-MOUNTAINEERS DURING SIMULATED SPRINTS ON SNOW

Beatrice Todesco [1,3], Bortolan Lorenzo [1, 2], Nina Verdel [4], Matej Majerič [4], Hans-Christer Holmberg[5]*, Matej Supej [4], Pellegrini Barbara [1, 2]*

[1] CeRiSM Sport Mountain and Health Research Centre, University of Verona, Rovereto, Italy;

[2] Department of Engineering for Innovation Medicine, University of Verona, Verona, Italy;

[3] Department of Neurosciences, Biomedicine and Movement Sciences, University of Verona, Verona, Italy;

[4] Faculty of Sport, University of Ljubljana, Ljubljana, Slovenia;

[5] Department of Health Sciences, Luleå University of Technology, Luleå, Sweden

ECSS 2025

EFFECTS OF CADENCE, LEVELS OF ASSISTANCE, AND THEIR INTERACTION ON OXYGEN UPTAKE WHILE RIDING AN E-BIKE. WHAT'S THE ROLE OF FREE CHOSEN CADENCE?

Todesco Beatrice[1 3], Decet Marco[1 3], Pellegrini Barbara[1 2], Bortolan Lorenzo [1 2], Schena Federico [1 3].

[1] CeRiSM Sport Mountain and Health Research Centre, University of Verona, Rovereto, Italy;

[2] Department of Engineering for Innovation Medicine, University of Verona, Verona, Italy;

[3] Department of Neurosciences, Biomedicine and Movement Sciences, University of Verona, Verona, Italy.

ABBREVIATIONS

AO	No E-bike assistance – Study one
A1	Lowest level of E-bike assistance – Study one
A3	Highest level of E-bike assistance – Study one
ANOVA	Analysis of variance
AR1	Covariance structure for repeated measures: autoregressive
ASP	Paved asphalt surface
BDC	Bottom dead center
BMI	Body mass index
BF	Biceps femoris
ECO	Lowest level of E-bike assistance – Study two
EMG	Electromyography
En. Exp	Energy Expenditure
EXPW	Maximum level of E-bike assistance – Study two
FCC	Freely chosen cadence
GET	Gas exchange threshold
GL	Gastrocnemius lateralis
GM	Gastrocnemius medialis
GMax	Gluteus maximus
GMed	Gluteus medius
HIGH	High level of E-bike assistance – Study two
HF	High-frequency component of vibration
HR	Heart rate
L3	Lumbar vertebra number 3
$[La^-]$	Blood lactate concentration
LF	Low-frequency component of vibration
LMM	Linear mixed model
OFFR_H	Off-road terrain with high technical complexity
OFFR_L	Off-road terrain with low technical complexity
P_ext_OBLA4	Power corresponding to the second threshold during the second part of the protocol
P_human	Power exerted by the subjects during the first phase of the study protocol

P_pedal	Power collected at the pedal during the third phase of the study protocol
RF	Rectus femoris
RPE	Rate of perceived exertion
SM	Semimembranosus
SOL	Soleus
ST	Semitendinosus
STD	Standard level of E-bike assistance – Study two
T4	Thorax vertebra number 4
TA	Tibialis anterior
TDC	Top dead centre
TS tube	Intersection of the top tube with the sit tube on the bike frame
VE	Ventilation
VL	Vastus lateralis
VM	Vastus medialis
VO_2	Oxygen consumption

1 CHAPTER 1

1.1 GENERAL AIM AND OVERVIEW OF THE THESIS

This thesis presents a comprehensive examination of the most relevant parameters associated with the use of electrically assisted bicycles (e-bikes), with a particular emphasis on aspects related to metabolic responses, neuromuscular engagement, subjective perception, and the analysis of vibrations experienced during riding. The investigation stems from a collaboration with the company designing, producing and selling e-bike, and the overarching objective of the PhD project is to contribute to the design and development of an e-bike whose features and functionalities are as attuned and responsive as possible to the specific needs, preferences, and physical characteristics of the end user.

As a first step, a technical review of the bicycles currently developed and marketed by the partner company was conducted, with the aim of understanding their structural features, mechanical configurations, and intended use cases. Based on this preliminary analysis, a structured research plan was defined to investigate the interaction between rider and e-bike across different contexts of use. The focus was placed not only on evaluating the physiological demands placed on the rider but also on understanding how different terrain types and riding conditions influence the physical and perceptual experience of cycling.

The company in question manufactures a diverse range of e-bike models, encompassing those intended for routine daily commuting as well as high-performance models designed for more demanding off-road applications, including all-mountain and enduro categories. In light of this product diversity, the research was structured around two primary experimental studies.
The first study was conducted in a laboratory environment and focused on the physiological effort required of participants during pedalling. This involved a detailed assessment of physiological parameters and muscular activation under controlled conditions, providing insight into how the e-bike supports or modifies the rider's physical workload.
The second study, was carried out in outdoor settings and placed emphasis on the bike handling in both uphill and downhill scenarios, across a variety of terrains ranging from

paved roads to technically challenging off-road trails. This phase of the research aimed to examine how the e-bike, which differs from traditional bicycles not only for the presence of a propulsive aid but also for specific mechanical feature such as a higher weight, responds to varying riding conditions, and how such variability influences the overall management of the vehicle.

1.2 BACKGROUND

1.2.1 E-BIKE HISTORY AND DEVELOPMENT

Contrary to what many might believe, the concept of the electric bicycle, commonly known today as the e-bike, is not a recent invention. Its origins can be traced back to the late 19th century, a time marked by significant technological and industrial innovation. In 1895, engineer Odegn Bolton was granted a patent for an early prototype of an electric bicycle powered by a battery *(Peine A. et al.,2017)*. This pioneering design can be considered one of the very first formal attempts at what we now call an e-bike. Just two years later, in 1897, another inventor, Hosea W. Libbey, designed a bicycle powered by two electric motors, an innovation that was quite advanced for its time and would be revisited nearly a century later in 1990 by the company Giant Lafree *(Parker A.A et al.,2004)*.

In the 1930s, interest in electric bicycles began to gain traction in the United States. The Lejay Manufacturing Company, based in Minneapolis, filed patents for a model called the "GoBike." This electric bicycle featured a generator taken from a Ford T car, connected directly to the rear wheel. During the same period, activity was also picking up in Europe. In Germany, for example, the Heinzmann company began manufacturing motors specifically for electric bicycles.

During World War II, due to the shortage of large motor vehicles, resulting from massive resource allocation to the war effort, there was a renewed interest in smaller, more efficient means of transportation. As a result, numerous prototypes of electric bicycles were developed during this time. However, many of these designs were eventually absorbed by the rapidly growing motorcycle industry, which dominated the market and pushed electric bicycles into the background.

In post war Europe, particularly in countries like Italy, where the production of aircraft and related technologies was restricted by post-war treaties, many engineers shifted their focus to the development of motorized vehicles. This transition contributed to progress in the electric bicycle sector as well. One particularly noteworthy milestone came in 1946, when inventor Benjamin Bowder unveiled a revolutionary model known as the "Space Lander" *(Peraro, 2010*). This bicycle introduced an innovative technology that allowed for energy regeneration during downhill riding, an idea that anticipated modern regenerative braking systems.

However, the true breakthrough for electric bicycles did not come until 1973, during the first major oil crisis. This global event sparked widespread awareness of the need for alternative, more sustainable forms of transportation. In the United States in particular, the electric bicycle began to be viewed as a viable and eco-friendly urban mobility solution, offering a more affordable and versatile alternative to the bulky and fuel-hungry cars of the time.

The 1980s saw further advancements in the field. In 1982, engineer Egon Gelhard developed one of the first pedal-assist electric bicycles, a concept that would later become central to the industry. Throughout the 1990s, e-bikes gained increasing popularity. A significant innovation came in 1992 when Sinclair Research Ltd. introduced the Zike, a lightweight and portable electric bicycle.

In 1993, the renowned Japanese company Yamaha made a major contribution by launching the power assist bike in Japan. This model achieved significant commercial success and helped solidify public interest in pedal-assist technology. Although the early 2000s saw a temporary decline in the e-bike market, the industry experienced a resurgence starting in 2005, driven by the emergence of lithium-ion batteries. These new batteries allowed for increased range between charges and significantly reduced the overall weight of the bicycles, improving both efficiency and user experience.

The definitive turning point came in 2009, when the German multinational Bosch decided to invest heavily in the e-bike sector, allocating approximately 400 million euros. That same year, Bosch e-Bike Systems was officially established as a dedicated division within the Bosch Group. Drawing on the company's extensive expertise in battery technology, electronics, and sensor systems from the automotive sector, Bosch was able to transfer these capabilities into the electric bicycle industry. From that moment on, the e-bike market has experienced continuous and steady growth in both innovation and global sales, a trend that remains strong to this day (*Centenaro,2022*).

2 CHAPTER 2

2.1 BIOMECHANICS OF CYCLING

Cycling takes various forms, including cycling as a means for fitness, as a tool for rehabilitation, and as a mode of transportation. Moreover, the competitive aspect has expanded to include road cycling, off-road cycling, track cycling, BMX *(Biomechanics of Cycling)*, and e-bikes *(Ruter S. et al.,2023)*. Due to the widespread use of bicycles, one of the main areas of interest is reducing the energetic cost of pedaling and preventing potential injuries.

For this reason, cycling biomechanics aims to study the movements of the body's joint segments and the forces they generate (*Biomechanics of Cycling*).

A substantial portion of the biomechanical analyses conducted on the bicycle is aimed at examining the movement patterns of body segments and joints in relation to potential adjustments, such as saddle height and the saddle-to-handlebar distance.

Additionally, foot position on the pedal also plays an important role. Two positions can be distinguished: anterior, commonly used, and posterior. In the first case, contact with the pedal occurs at the metatarsophalangeal joint; in the second, contact occurs in the middle of the foot *(Fonda et al., 2010)*. Modifying these parameters consequently changes the position and length of the levers (joint segments) that transmit force to the pedal.

Adjustments can also be made by modifying crank length, although small changes in this area do not appear to explain performance improvements, especially at submaximal levels *(Bini RR et al., 2012)*, which is why cyclists typically use cranks of very similar, if not identical, length *(Biomechanics of Cycling)*.

Regarding saddle height, it can be defined in two ways: in the scientific literature, it is defined as the distance between the top of the saddle and the pedal axis at its lowest point. In this case, the recommended height ranges from 96% to 100% of the greater trochanter height *(Price D. et al., 1997; Biomechanics of Cycling)*. In other cases, commonly among cyclists, saddle height is defined as the distance between the top of the saddle and the point where the crank connects to the bottom bracket *(Fonda et al., 2010)*.

2.1.1 The pedalling cycle

The pedaling cycle is divided into phases according to the angular position of the pedal in relation to its 360° movement. Based on the angular position of the crank arm, the forces involved, their orientation, the position of body segments, and muscle activation are described.

Within the entire pedaling cycle, two main points are distinguished: the top dead centre (TDC), located at 0°, and the bottom dead centre (BDC), located at 180° *(Fig. 2.1)*. They are called "dead centers" because at those exact points the crank is mainly aligned with the applied force, so it is not possible to effectively generate torque.

Consequently, the first pedaling phase, called the down stroke phase (from 0° to 180°), and the second phase, called the up stroke phase (from 180° to 360°), are defined. In addition, two transition zones of ±5° around the top dead centre and bottom dead centre are identified (*Fonda et al., 2010*).

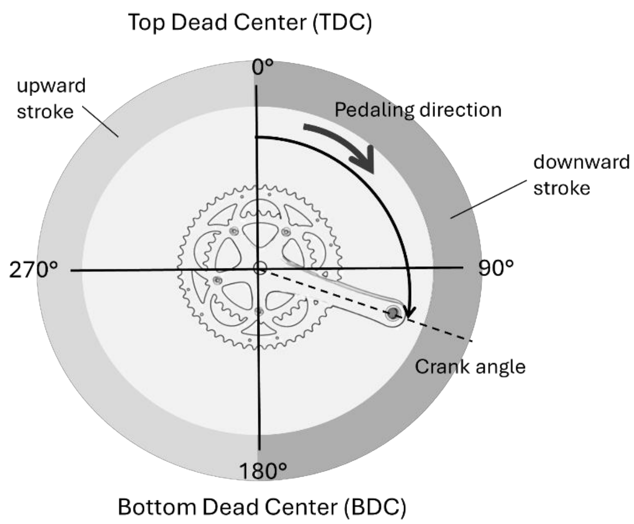


Fig.2.1: Image showing the top dead centre (TDC), the bottom dead centre (BDC) and the two main pedalling phases: downward phase (0-180°) and upward phase (360-0°).

2.1.2 Motion analysis

The motion analysis system aims to extract joint angles, which determine muscle length during the pedalling cycle. In addition, data on velocity and acceleration of the investigated reference points (body segments) are collected.

The system used is video analysis through high-speed cameras, or more innovative systems that replace video, such as the Retul system – www.retul.com.
From a practical perspective, motion analysis provides information about the cyclist's position while pedalling, as well as the effects of fatigue, cadence changes, and power output. Data are extrapolated by averaging values over multiple cycles, which typically start at the top dead centre of crank, usually considering the sagittal plane (although three-dimensional analysis is also possible) *(Biomechanics of cycling)*.
The three main monitored angles are the hip (range of motion 42–44°), the knee (range 73–78°), and the ankle (21–25°). The upper body is also monitored, as it supports trunk position, influences lower limb joint angles, and can vary depending on saddle tilt and hand position on the handlebars. Adjusting these angles can also help address possible onset of lower back pain.

Innovative methods for movement analysis include dynamic three-dimensional modelling of the cyclist combined with additional saddle pressure analysis *(Swart J. et al., 2019)*.

2.1.3 Force analysis

Forces acting on the bicycle

In cycling, movement results from the balance between resistive forces and propulsive forces acting on the rider. The resistive forces are due to air resistance and the rolling resistance of the wheel in contact with the ground. Gravity does not exert any force for perfectly flat riding; it acts as a resistive force when riding uphill and as a propulsive force when riding downhill. Effect of gravity, strongly dependent by the inline, can be modified by modification in bikers and bicycle mass. The drag forces can me decreased the former by reducing the frontal area of the body, specifically, the smaller it is, the lower the air resistance, and by changing the shape coefficient. The rolling resistance and the latter by changing the type of tire and pressure on inflation that can modify, the tire section in contact with the ground. When the tire undergoes greater deformation, rolling resistance increases; likewise, a knobby tire generates more resistance than a smooth one. Furthermore, rolling resistance is also influenced by gravity through the cosine of the slope angle.

Forces exerted on the bicycle

In the case of cycling, there are three points of contact between the rider and the bicycle, through which the acting forces are distributed. These three contact points are the handlebars, the saddle, and the pedals. Of particular interest is the force applied to the pedals, as it generates movement. The force that produce the movement is generated by the force applied by the rider, which can be broken down into a component parallel to the crank arm, called ineffective force, and a component tangential to the crank arm, called effective force. The tangential force, when multiplied by the crank length, produces the torque acting on the bicycle's drivetrain. Consequently, power is defined as torque multiplied by pedalling cadence. These forces fluctuate throughout the pedalling cycle depending on the position of the lower limb and the crank arm *(Fig 2.2)*.

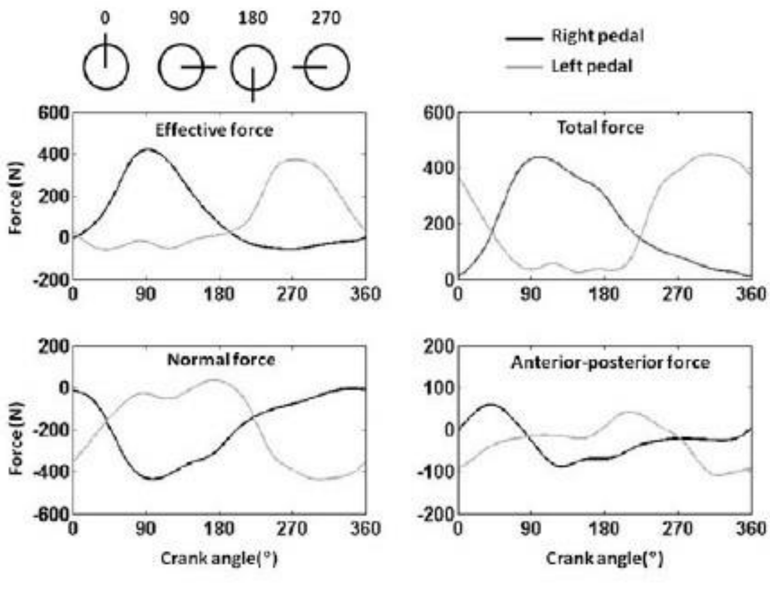


Fig. 2.2: Image from the book Biomechanics of Cycling illustrating the different forces acting on the pedal during the pedalling cycle. 'Effective force' refers to the tangential force applied to the pedal.

These changes consequently lead to different levels of power output due to the varying direction of forces relative to the pedal, among which the propulsive force is the one that generates movement. Literature shows that the propulsive component accounts for 40–60% of the total forces (with its peak occurring between 90° and 180°) and that it is not constant, varying depending on cadence (Laursen PB et al., 2003; Patterson PR et al., 1990), the required power (Zameziati K et al., 2006; Kautz S et al., 1991), the cyclist's state of fatigue (Diefenthaeler F et al., 2007; Dorel S et al., 2009), saddle positioning (Dorel S et al., 2009; Diefenthaeler F et al., 2006), and the cyclist's skills (Candotti CT et al., 2007; Sanderson DJ, 1991). In particular, the more experienced and skilled the cyclist, the greater the effective force component they can apply to the pedal relative to the total force.

The ratio between these two forces is defined as the index of effectiveness *(Equation 2.1) (Biomechanics of Cycling)*.

$$IE = \frac{\int_0^{360} EF\,dt}{\int_0^{360} TF\,dt}$$

Equation 2.1

Equation 2.1: index of effectiveness (IE); impulse of the effective force (EF); impulse of the total pedal force (TF).

Pedalling effectiveness decreases as cadence increases. This occurs because the higher movement frequency leads to a greater number of ineffective isometric contractions and reduced muscular coordination *(Patterson and Moreno, 1990)*. Furthermore, effectiveness is also influenced by power output; specifically, it decreases as power decreases. This happens because it becomes more difficult to direct force effectively onto the pedal *(Patterson and Moreno, 1990).*

In *figure 2.3 (Ettema et al., 2009)*, the variations of force and power within the pedalling cycle are illustrated.

As can be observed from the graphs, total force reaches its maximum between 90° and 180°, power peaks between 90° and 150°, while effective force reaches its maximum between approximately 45° and 100°. Moreover, as can be observed from *Figure 2.4*, the profile of force and power within the pedalling cycle varies with changes in cadence.

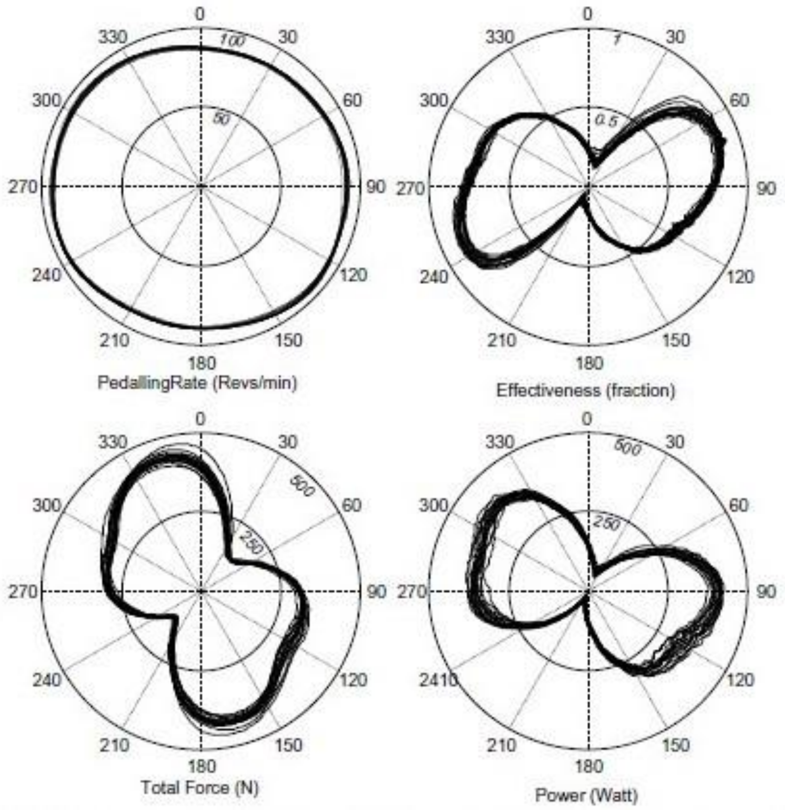


Fig. 2.3: Force and power in the pedalling cycle obtained at a cadence of 90 rpm. The curves are derived from the average values of both limbs over 20 pedalling cycles (Ettema et al., 2009).

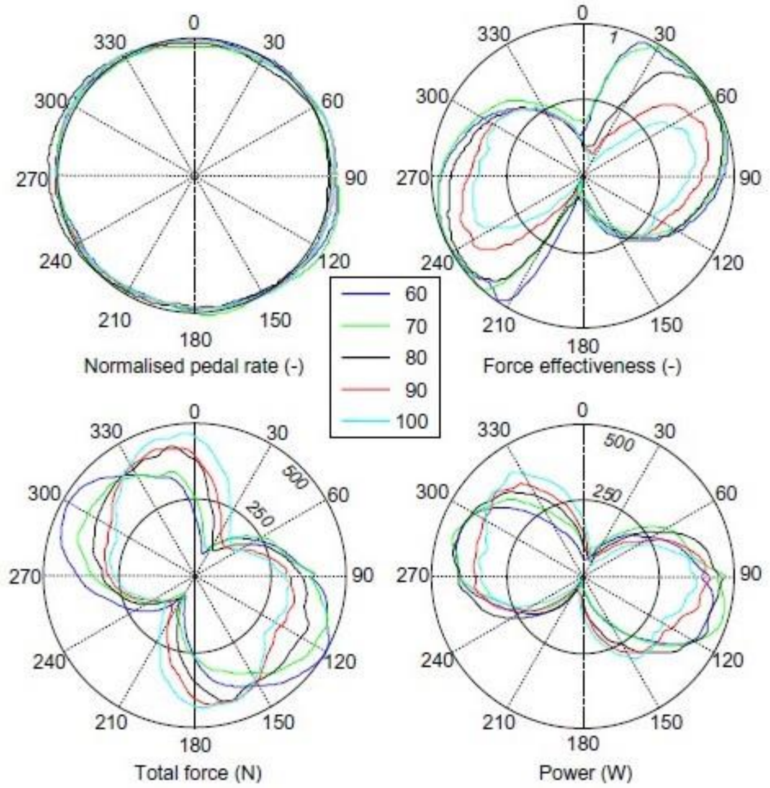


Fig. 2.4: Force and power in the pedalling cycle obtained at different cadences (Ettema et al., 2009). Changes in cadence cause variations in both the shape and the magnitude of force effectiveness.

Forces acting on the joints

The forces acting on the joints are influenced by pedalling technique and the rider's positioning on the saddle *(Biomechanics of Cycling)* and their analysis is important for two main reasons: injury prevention *(Biomechanics of Cycling)* and performance in both elite and amateur athletes *(Aasvold et al., 2019).* It is indeed important to convert as much muscular force as possible into power at the pedals to optimize performance *(Biomechanics of Cycling; Aasvold et al., 2019)*.

The power produced results from the action and coordination of the muscles acting on the hip, knee, and ankle joints *(Aasvold et al., 2019)*.

The forces acting on the joints cannot be measured directly but must be estimated using inverse dynamics models, taking into account the forces applied at the pedal and joint motion in order to determine the relative contribution of each joint to the total power produced *(Biomechanics of Cycling; Aasvold et al., 2019)*.

The literature reports various findings derived from different studies, often using different experimental settings. The two main parameters analysed are different power outputs and different pedalling cadences.

The study by *Aasvold et al., 2019*, is one of the most comprehensive, analysing multiple pedal power outputs and a cadence range from 40 to 100 rpm *(Fig. 2.5)*. The study highlights that an increase in cadence results in a reduction of forces acting on the hip joint and an increase of forces on the knee joint, a phenomenon that occurs only above 60 rpm. In addition, forces acting on the ankle were observed to decrease with increasing cadence, starting from 80 rpm.

Regarding exercise intensity, an effect of power on the hip and knee joints was reported. The force associated with the hip joint was lower at the lowest intensity compared with the two higher intensities studied (moderate: 85% LT, and threshold intensity: LT). No differences were observed between the moderate and threshold intensities.

For the knee joint, as power increased, a decrease in its force contribution was recorded. Again, this difference was observed between the lowest and the moderate/threshold intensities, with no difference between the two higher intensities.

For the ankle joint, no differences were found. The authors report that as power increases, from low to moderate and threshold, the pedalling technique shifts, leading to the changes in joint forces described above. Finally, an analysis was conducted comparing elite and amateur athletes. Results show that the patterns of joint force variation are the same in both groups and reflect the trend observed in the overall sample.

The only difference between amateurs and elite athletes is that the latter exhibit approximately a 10% higher contribution of hip joint force *(Aasvold et al., 2019)*.

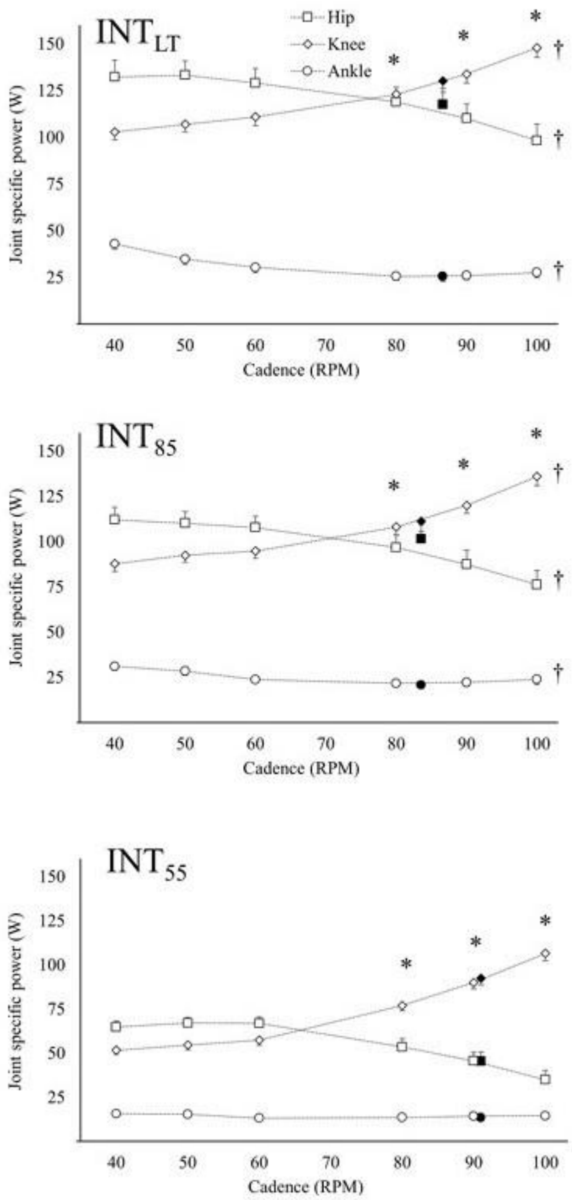


Fig. 2.5: Image from the study by Aasvold et al., 2019. From top to bottom: forces acting on the joints at different cadences with power corresponding to a [BLa¯] of 4 mmol/L; forces acting on the joints at different cadences with power corresponding to 85% of a [BLa¯] of 4 mmol/L; forces acting on the joints at different cadences with power corresponding to 55% of a [BLa¯] of 4 mmol/L.

2.1.4 Electromyographic analysis of pedalling motion

The analysis of muscle activation during the pedalling cycle has been widely studied since 1959 with the work of *Houzt et al*., and many reviews, including those by *Cheryl A. Wozniak Timmer (1991), Fonda B. et al. (2010), and Hug F. et al. (2007)*, report the main concepts and findings on the topic.

The purposes of electromyographic studies in cycling are not limited to athletes and pedalling efficiency, but also include injury prevention, given the large number of people who practice this sport *(Cheryl A. Wozniak Timmer, 1991; Hug F. et al., 2007).* The activation of lower limb muscles allows both the force production and its optimal orientation on the pedals *(Hug et al., 2009).* The aim is identifying the lower limb muscles which are activated and precisely knowing their timing and levels of activation *(Hug et al., 2009)*. In addition is important to know how the coordination strategies adapt to various power output, pedalling rate, body position, shoe-pedal interface, training status and fatigue *(Hug et al., 2009)*.

An initial schematic description of muscle activity during the pedalling cycle was provided by *Cheryl A. Wozniak Timmer (1991)*, who categorized muscle action based on the joint they move. Thus, extensors and flexors of the hip and knee, as well as the muscles responsible for dorsiflexion and plantarflexion of the ankle, were described.

Specifically, hip flexors are active from 300° to 60°, while hip extensors are active from 330° to 210°. Knee flexors activate from 45° and deactivate at 200°, while knee extensors are active from 300° to 105°. Finally, ankle dorsiflexors are active from 270° to 90°, while plantar flexors activate from 30° to 270°.A more detailed analysis of muscle activation was provided by *Fonda et al. (2010)*, who, through the study of *Ryan et al. (1992)*, reported both the activation range and the peak activation point with respect to TDC (top dead centre) and BDC (bottom dead centre). Among the mono-articular muscles most studied are the gluteus maximus (GMax), gluteus medius (GMed), vastus lateralis (VL), vastus medialis (VM), tibialis anterior (TA), soleus (SOL), and iliopsoas (IP). The bi-articular muscles included in the studies were the rectus femoris (RF), semimembranosus (SM), semitendinosus (ST), biceps femoris (BF), gastrocnemius lateralis (GL), and gastrocnemius medialis (GM).

Ryan et al. (1992) reported the following muscle activation *(Fig. 2.6)*: GMax extends the hip and is active from 340° to 130°, with a peak at 80°. VL and VM extend the knee, active from 340° to 270°, with a peak at 30°. RF acts as both a knee extensor and a hip flexor, active from 200° to 110°, with a peak at 20°. SOL stabilizes the talocrural joint from 340° to 270°, with a peak at 90°. GM and GL act as stabilizers of the talocrural joint and as knee flexors. They are active from 350° to 270°, with a peak at 110°. TA also stabilizes and dorsiflexes the talocrural joint, active throughout the pedaling cycle, with a peak at 280°. SM and ST flex the knee, active from 10° to 230°, with a peak at 100°.Finally, BF flexes the knee and extends the hip, active from 350° to 230°, with a peak at 110°.

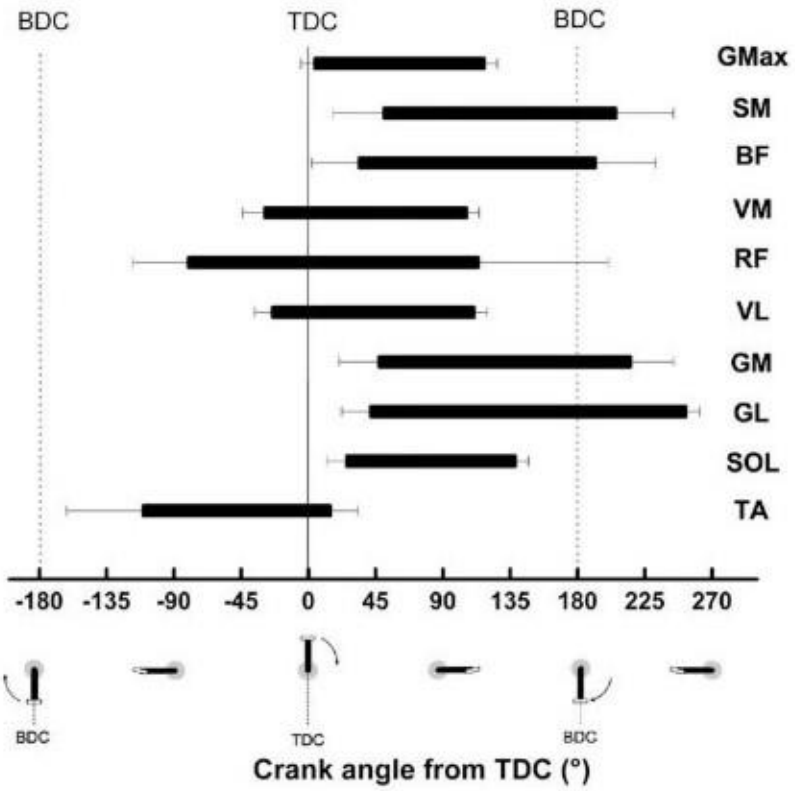


Fig. 2.6: Image taken from the study of *Hug et al., 2009*, illustrating the muscles activation during the pedalling cycle.

As illustrated in the image, multiple muscle co-activations can be observed, whose role extends beyond the mere transmission of force to the articulated segments, contributing also to the stabilization and protection of the joints.

EMG pattern and pedalling rate

Muscle activation is influenced by numerous factors, among which cadence, together with power output, represents one of the primary determinants.
Several studies have investigated this relationship, yet their findings remain inconsistent, making it difficult to establish a unified conclusion regarding the effect of cadence on muscle activation *(Fonda et al., 2010)*. Most studies indicate that the medial and lateral gastrocnemius, together with the semimembranosus, increase their activity with rising cadence. For other muscles, however, the results vary considerably and are often contradictory *(Hug et al., 2009)*.
For instance, *Ericson (1986)* reported an increase in muscle activity with increasing cadence for the gluteus maximus (GMax), semimembranosus (SM), vastus medialis (VM), gastrocnemius medialis (GM), and soleus (SOL). Similarly, *Neptune et al. (1997)* found that the gastrocnemius medialis (GM), biceps femoris (BF), semimembranosus (SM), and vastus medialis (VM) exhibit increased activation with higher cadences. In contrast, the gluteus maximus (GMax) and soleus (SOL) showed a quadratic trend, while the rectus femoris (RF) and tibialis anterior (TA) appeared unaffected by changes in cadence. Moreover, according to *Lucia et al. (2004)*, the activation of the vastus lateralis (VL) and gluteus maximus (GMax) decreases as cadence increases.
These heterogeneous results may be attributed to differences in participants' training status, the cadence ranges analysed, and the power outputs considered *(Hug et al., 2009)*.

In general, according to the review by Ansley and *Cangley (2009)*, the EMG activity of the muscles involved in pedalling exhibits a 'J-shaped' curve as cadence increases (Fig. 2.7).

Author	Power (W)	Cadence (rev · min^{-1})	Nadir (rev · min^{-1})
Baum & Li (2003)	250	60–100	80
Farina et al. (2004)	241	45–120	70[#]
MacIntosh et al. (2000)	100–400	50–120	80 (at 300 W)
Marsh & Martin (1995)	200	50–110	95
Neptune et al. (1997)	250	45–120	90
Sarre et al. (2003)	372	62–114	88[#]
Takaishi et al. (1996)	200–240	50–100	80

[#]No significant difference between cadences.

Fig. 2.7 Image adapted from the study by *Ansley and Cangley (2009)*, which describes a "J-shaped" relationship between EMG activity and pedalling cadence.

Furthermore, when considering the minimum of the cadence–EMG curve, it can be observed that it shifts rightward as power output increases *(MacIntosh et al., 2000; Farina et al., 2004, Artioli 2019-2020) (Fig. 2.8)*.

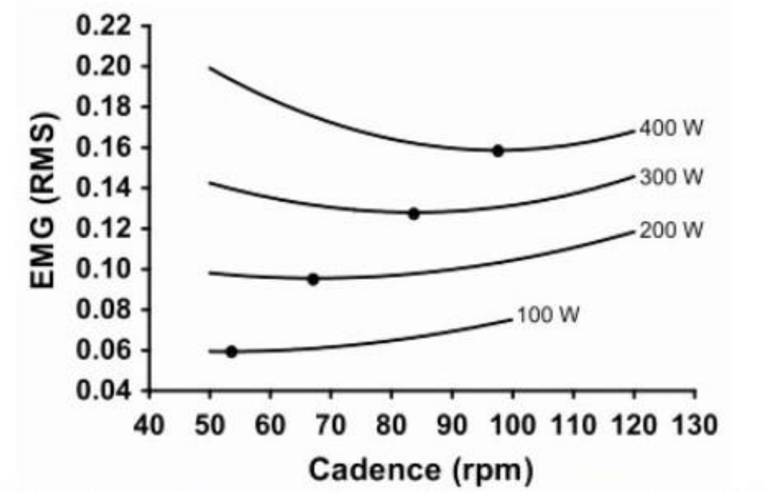


Fig. 2.8 Image adapted from the study by *MacIntosh et al. (2000)*, illustrating the effect of cadence on EMG activity during cycling at different power outputs.

EMG pattern and power output

The power output is a parameter that can influence muscle activation, modifying its magnitude but without greatly affecting the activation pattern.

Based on studies available in the literature, the authors report an increase in the electromyographic activity of the muscles as the required intensity increases. Specifically, *Ericson 1986* reports higher values for the gluteus maximus (GMax), vastus lateralis (VL), rectus femoris (RF), vastus medialis (VM), biceps femoris (BF), and gastrocnemius medialis (GM), highlighting in particular the increased activation of the GMax with rising power output. The study by *Sarre et al. 2003* confirms these results, reporting an increase in the activity of the vastus medialis (VM), vastus lateralis (VL), and rectus femoris (RF). On the other hand, the gastrocnemius muscle does not change its activation when the required power and the increase in power are low *(Jorge and Hull, 1986; Hug et al., 2004)*.

2.2 MECHANICAL DETERMINANTS OF THE ENERGY COST IN CYCLING

Minetti et al. (2001) describe the energy cascade in cycling (Figure 2.9), outlining the components that contribute to the overall metabolic expenditure (E). All components are described below.

Total mechanical work (WTOT): is partitioned into the work of the vehicle with respect to the external environment (*W'BIKE'*) and the one related to the propelling machinery (*WPROPULSOR*).

W'BIKE': it includes the work to overcome air drag (*WD)* and the rolling resistance *(WR)* that include the work against the resistance of the drive chain and gearing. Moreover, the sum of WD and WR makes the mechanical external work *(W EXT)*.

WPROPULSOR is defined by several components: *WEXT*,* that could be included in WEXT, and represents a small residual movement of the body centre of mass due to the alternating moving limbs; *WOTHER* that describes the work of deformation of pedals during the push and the *WINT (*the mechanical internal work, further described).

Metabolic expenditure (*E*) is partitioned into *Wtot*, which is required for movement, and heat, which should be minimized.
The ratio between *Wtot* and *E*, Energy expenditure due to *Wtot* + *heat* is defined as muscular efficiency.
The product of muscular efficiency and transmission efficiency, defined as the product of pedalling efficiency and bicycle efficiency, determines performance efficiency.

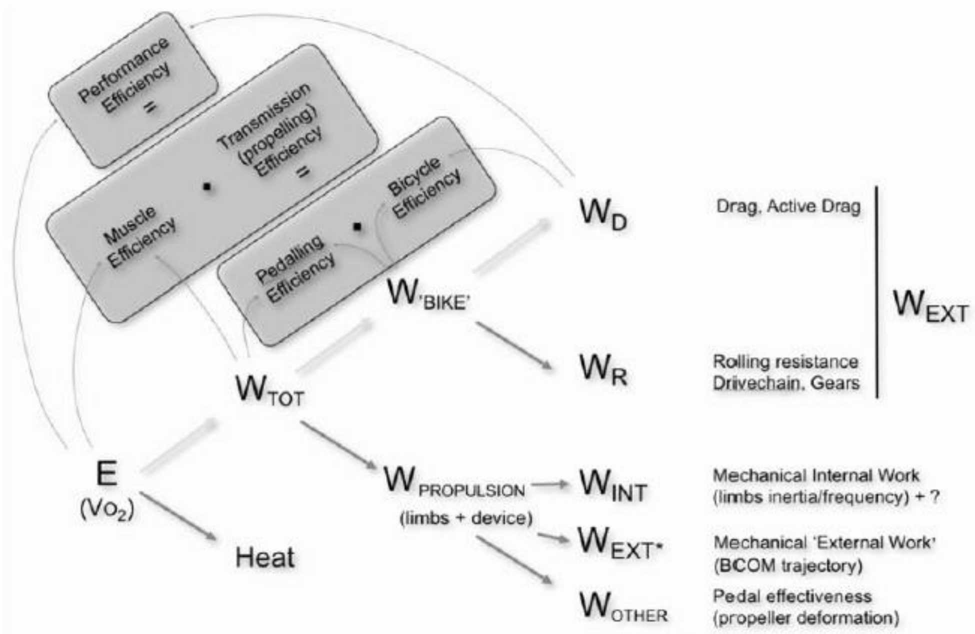


Fig. 2.9 Efficiency cascade in cycling *(Minetti et al., 2001)*.

In this section, it is also important to define the difference between energy cost and energy expenditure. Energy cost was previously defined in relation to cycling, however, in general terms, it represents the amount of energy expended per unit of work (for example, in walking: $J \cdot kg^{-1} \cdot m^{-1}$). In contrast, energy expenditure refers to the total amount of energy consumed over a given time interval (kcal or $kcal \cdot min^{-1}$). These two concepts are mathematically related, in fact, energy cost is defined as the ratio between energy expenditure and the unit of performance. From a physiological perspective, however, an increase in energy expenditure does not necessarily imply an increase in energy cost. For example, if energy expenditure increases but walking speed increases proportionally, the energy cost per meter travelled will remain unchanged.

2.2.1 The internal work

In multi-segment bodies, internal work refers to the kinetic energy changes of individual limbs as they move relative to the body's centre of mass (BCOM). While external work accounts for raising and accelerating the BCOM, it does not include the energy spent to reciprocally move the limbs. According to König's Theorem, total kinetic energy of a multi segments body indeed equals the kinetic energy of the BCOM plus the relative kinetic energy of the segments; the latter constitutes the internal work. This classic approach, introduced by *Fenn 1930* and later adopted by many investigators, provides a reliable estimate of total locomotion work. Internal mechanical work has been defined by *Francescato et al. 1995 and Minetti 2011* as a component of total mechanical work that can be considered. The internal mechanical work is proportional to pedalling cadence and appears to result from an internal viscous component, energy dissipation within the body that leads to a reduction in movement efficiency, rather than from the "kinematic work" associated with changes in the kinetic energy (both linear and rotational) of the lower limbs relative to the centre of mass *(Artioli, 2019–2020)*.

According to *Minetti 2011*, internal work (Wint) for cycling can be estimated using the following equation *(Equation 2.3)*:

$$Wint = q\, f^3 \,/\, v \qquad \textit{Equation 2.3}$$

Where f is the pedalling frequency expressed in Hz, v is the velocity (m/s), and q is a term dependent on the inertial characteristics of the moving limbs.
Consequently, internal mechanical work increases with both the velocity and the movement frequency of the limbs.

In the study by *Minetti et al. 2001*, equation 2.4 is described, according to which internal mechanical power per unit of body mass (W/kg) can be calculated.

$$Wint = 0.153\, f^3 \qquad \textit{Equation 2.4}$$

Where f is expressed in Hz. For example, 60 RPM correspond to 1 Hz, and 120 RPM correspond to 2 Hz. When the frequency value is raised to the third power, multiplied by 0.153, and then by the subject's body mass, the resulting value represents the internal mechanical power expressed in watts (W).

.

2.2.2 The negative work

In addition to internal work, the pedalling cycle may also include a component of negative work arising from the counter-rotational torque during the recovery phase of the pedal stroke. This results in an additional force requirement from the leg in the down stroke phase to counteract the negative torque generated by the rising leg *(Artioli, 2019–2020)*.
Such negative work can be mitigated through the "pulling-up" action performed by the ascending leg, which is possible when the pedal is equipped with a binding that secures the foot to it. It may also be interpreted as a transfer of potential energy between the two legs via the crank arm *(Kautz and Hull, 1993; Sanderson, 1991)*. Negative work increases with pedalling cadence, due to altered intra-cycle activation and deactivation patterns of the hip and knee extensors. Indeed, limb extension tends to persist into the pedal recovery phase, thereby increasing the resistance that the pushing leg must overcome *(Neptune and Herzog, 1999)*.

2.3 THE ROLE OF CADENCE IN CYCLING

Pedaling cadence has been, and continues to be, a topic of particular interest for scientists, coaches, and athletes (*Artioli, 2019–2020).* Cadence selection is an important performance component, as it is one of the few variables that an athlete can dynamically regulate to manage both performance and fatigue *(Ansley & Cangley, 2009; Artioli, 2019–2020)*.

Studies have shown that the cadences chosen by cyclists (FCC – freely chosen cadence) (80–100 RPM) are higher than the metabolically optimal ones (50–70 RPM) *(Brisswalter et al., 2000; Lucia et al., 2001; Marsh & Martin, 1998; Sarre et al., 2003; Artioli, 2019–2020)*. However, the term “preferred cadence” can be misleading, as it does not account for the influence of external factors, including environmental, physical, and competitive conditions *(Artioli, 2019–2020)*. Professional cyclists freely adopt a cadence of approximately 70 RPM on climbs, while their preferred cadence on flat terrain is around 100 RPM *(Lucia et al., 2001).*

The debate regarding cyclists’ cadence selection has traditionally been based on the hypothesis that adopting an energetically optimal cadence allows for greater work output (and therefore speed) at the same energy cost.

Optimal cadence has commonly been defined as the pedalling frequency that minimizes energy cost, muscular stress, and perceived exertion *(Ansley & Cangley, 2009)*.

2.3.1 Pedaling cadence and metabolic work

Energy cost (*C*) is defined as the oxygen consumption (VO_2) required to maintain a given speed: EC = VO2 / v *(Artioli A. 2019-2021)*.

VO_2max represents an individual’s maximal aerobic capacity (Coyle et al., 1991). The lower the oxygen consumption as a percentage of VO_2max (%VO_2max) for a given power output, the lower the energy cost (*EC*) of the exercise.

In the study by Ansley and Cangley (2009), the literature review highlighted that the O_2 response to increasing pedalling cadence follows a J-shaped curve *(Fig. 2.10)*.

Authors	Power (W)	Cadence (rev · min^{-1})	Nadir
Argentin et al. (2006)	246	50–110	67
Boning et al. (1984)	200	40–100	70
Brisswalter et al. (2000)	301	50–110	80
Buchanan & Weltman (1985)	~324	60–120	60
Chen et al. (1999)	180	40–100	60
Coast & Welch (1985)	300	40–120	80
Coast et al. (1986)#	~300	60–120	70
Foss & Hallen (2004)	258	60–120	80
Foss & Hallen (2005)#	312	60–120	75
Hagberg et al. (1981)	82% $\dot{V}O_{2max}$	68–126	72
Hansen et al. (2002a)#	258	61–115	66
Hintzy et al. (1999)	150	40–120	57
Marsh & Martin (1993)	200	50–110	56
Seabury et al. (1977)	196	30–120	58
Takaishi et al. (1996)	220	50–100	65
Takaishi et al. (1998)	200	45–105	60

Fig. 2.10: Immagine ripresa dallo studio di *Ansley e Cangley 2009* in cui si evidenzia l'andamento "J-shaped" del VO_2 al variare della cedenza, mantenendo la stessa potenza.

From the figure by *Coast and Welch 1985* (Fig. 2.11), it can be observed that, as power output decreases, the lowest oxygen consumption values along the curve are attained at lower cadences.

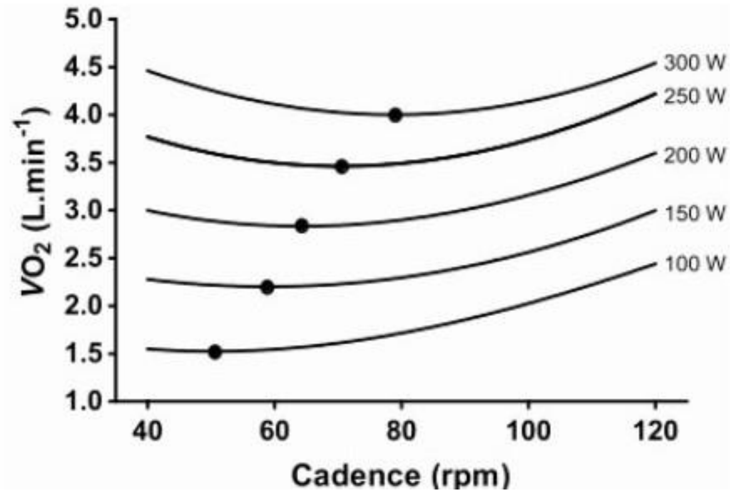


Fig. 2.11 Image adapted from the study by Coast and Welch (1985), illustrating the effect of cadence on VO_2 while cycling at constant power.

Lactate concentration exhibits a pattern similar to that of VO_2. Specifically, at a constant power output of 300 W, the minimum occurs at 70 RPM (Fig. 2.12) *(Brisswalter et al., 2000)*.

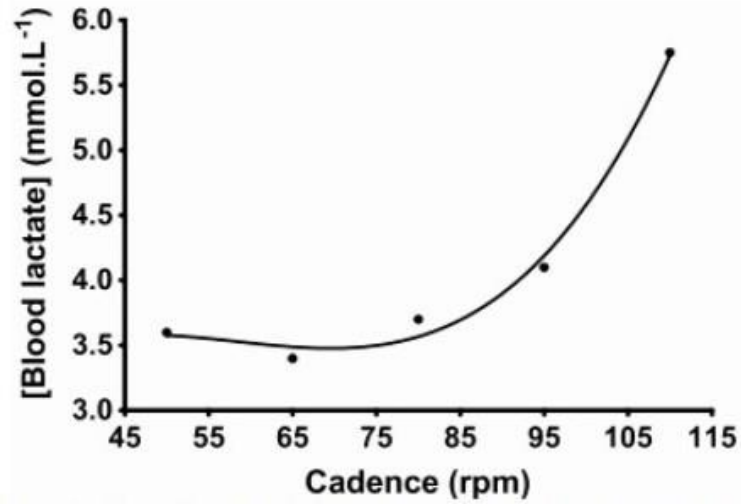


Fig. 2.12 Image adapted from the study by *Brisswalter et al. 2000*, illustrating the lactate curve with a nadir at 70 RPM at constant power output.

Moreover, the ventilation (VE) exhibits a pattern similar to that of VO_2, with the highest values occurring at higher cadences *(Fig. 2.13)*. In the study by *Mitchell et al. 2019*, at elevated cadences, VO_2 exceeds the gas exchange threshold (GET), consequently leading to increased ventilation. This explains why anaerobic metabolism is activated at constant power solely by altering pedalling cadence *(Mitchell et al., 2019)*.

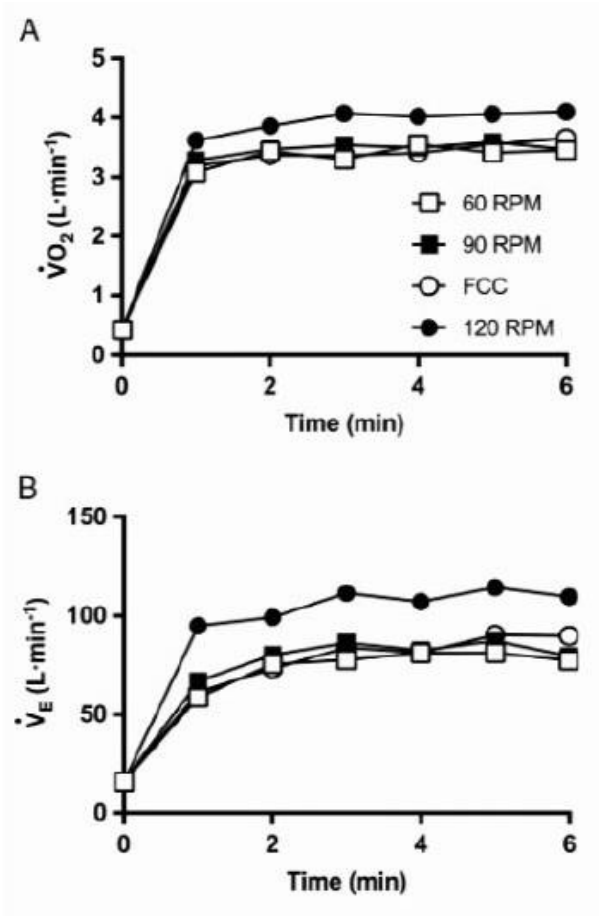


Fig. 2.13 Image adapted from the study by Mitchell et al. (2019), illustrating (a) the VO_2 response and (b) VE at different cadences while maintaining the same power output.

It can therefore be stated that the metabolically optimal cadence ranges between 50 and 70 RPM and increases with rising external power output.

2.3.2 Pedalling cadence and muscular work

Joint work has previously been introduced; in this section, it is addressed as a method for estimating muscular work. The sum of the net joint powers at the ankle, knee, and hip has indeed been proposed as a measure of mechanical work and is closely related to energy expenditure *(de Groot et al., 1995)*.

In the study by *Marsh et al. (2000)*, the curve describing the sum of joint powers is characterized as J-shaped, and the lowest point of the curve, at which muscular work is minimized, shifts to the right as load increases *(Artioli, 2019–2020).*

3 CHAPTER 3

3.1 PHYSICAL ACTIVITY AND PHYSIOLOGICAL ASPECTS WHILE RIDING AN E-BIKE

3.1.1 General background

Electric bicycles are bicycles that are equipped with an electric motor, delivering part of the external power and that require human power input to activate the motor *(Langford B. C. et al., 2017)*. As reported in the chapter 1.2.2, this type of bicycle has emerged in recent years as a form of active transportation and as a means of promoting physical activity *(Langford B. C. et al., 2017; McVicar J. et al., 2016; Rauter S., et al., 2023)*. Recent studies have reported that e-bike sales increased by 145% between 2019 and 2020, surpassing the sales of conventional bicycles *(Rauter et al., 2023; Johnson N., et al., 2023)*.
The World Health Organization recommends 150–300 minutes of moderate activity or 75–150 minutes of vigorous activity per week for health benefits, and electric bicycles, by replacing many commuting trips and beyond, help to achieve the recommended amount. They help overcome typical barriers associated with conventional bicycles, such as lack of time *(Hoare E., et al., 2017)* or the ability to pedal for extended periods *(Bourne J.E. et al., 2020)*.

The physical activity of e-bike users has been analysed in several studies, which report that it requires moderate physical activity (MET > 3) on flat segments and vigorous activity (MET > 6) on uphill segments *(Rauter S., et al., 2023; Sperlich B. et al., 2015)*. Furthermore, it has been specified that even when using maximum pedal-assist support, activity still reaches 3 MET *(Langford B. C. et al., 2017; Louis J. et al., 2012, Alessio et al. 2021, Bersten et al., 2017, Gojanovic et al. 2011, Hansen et al., 2018, Simpson et al, 2009)*. It has also been observed that after four weeks of e-bike use there is an improvement in cardiorespiratory fitness levels, and after an additional four weeks, an increase in power output is observed *(Peterman J.E., 2016)*. More generally, as reported by *Peterman J.E. et al., 2016*, there is an improvement in health markers such as glucose, maximal oxygen uptake, and peak power.
Finally, the use of e-bikes among inactive individuals is described as providing a sense of pleasure and enjoyment *(Peterman J.E., 2016)*.

Maintaining certain levels of physical activity is important not only for fitness but also for training purposes, where the electric bicycle must still provide a certain type of effort. Since 2019, the e-bike has also seen another application in competitions. That year, the first world championship as part of the UCI Mountain Bike World Championships was held. In this competition, athletes had to tackle varied terrain with high-intensity uphill sections *(Rauter et al., 2023)*.

3.1.2 Overview of experimental protocols

The summarized text reviews four empirical studies investigating the physiological and perceptual effects of electric-assisted cycling (e-bike or e-MTB) compared to conventional cycling.
Each study employed controlled, within-subject designs, in which participants completed comparable cycling tasks both with and without electrical assistance.
Across studies, physiological and biomechanical variables were monitored using portable metabolic analysers, electromyography (EMG), heart rate monitoring, GPS tracking, and subjective questionnaires.

Authors & Year	Participants (n)	Participant Characteristics	Experimental Design	Measured Variables	VO_2 / METs / Energy Findings
Sperlich et al. (2012)	8 woman	Low physical activity, no cycling experience	5 assisted/unassisted sprints (0–25 km/h); 9.5 km circuit	Power output, EMG, VO_2, RER, ventilation, HR, lactate, RPE enjoyment rate	VO_2 ↓33%, energy expenditure ↓37%, ventilation ↓49% with e-assist.
Bersten et al. (2017)	6 adults	General cyclists	Two commuting routes (flat/hilly), with and without e-assist	VO_2 (METs), GPS speed/elevation, time of MVPA/VPA	VO_2 ↓ from 10.9 to 8.5 METs (~22% reduction), larger effect on hilly routes.
Hall et al. (2019)	≈33 (88% men)	Experienced mountain bikers	8.85 km MTB course, conventional vs. e-MTB	HR (%MHR), speed, time, perceptions of effort	HR ~10 bpm lower; both conditions within vigorous intensity zone. ↓ participants' perceived effort
Rauter et al. (2023)	6 riders	Experienced MTB riders	4.5 km technical MTB course, both conditions	Speed, cadence, power, VO_2, ventilation, HR, GE	VO_2 ↓14.3%, ventilation ↓21.1%, HR ↓8.5%, energy expenditure lower despite higher speed.

3.1.3 Experimental outcomes and interpretation

Across studies, electrically assisted cycling resulted in significant reductions in cardiorespiratory demands, while still maintaining exercise intensities within the moderate-to-vigorous range.
Key findings include, lower HR and VO_2, higher average speeds, and greater perceived enjoyment. Also in this case, as reported by *McVicar et al., 2022*, the heterogeneity between the participant's groups may have an impact on the outcome assessed.

3.1.4 Cross-study synthesis

All four studies consistently reported lower oxygen consumption and energy expenditure during electrically assisted cycling compared with conventional cycling.

Parameter	**Trend with E-Bike Use**	**Approx. Range of Reduction**
Oxygen consumption (VO_2)	↓	14–33%
Energy expenditure	↓	22–37%
Heart rate	↓	8–29%
Ventilation / Respiratory rate	↓	15–49%
Power output (rider-generated)	↓	19–36%
Completion time	↓ (faster)	13–29%
Perceived exertion	↓	Subjective improvement
Enjoyment / Perception	↑	More positive post-use
Gross efficiency	↓	14% e-bike vs. 18% conventional bike (Rauter 2023)

3.1.5 Conclusion

Electric-assisted cycling significantly reduces metabolic and cardiovascular demands, such as VO_2, energy expenditure, and heart rate, while maintaining intensities sufficient for health benefits. Despite reduced mechanical efficiency, e-bikes enhance accessibility, adherence, and enjoyment, offering a balanced approach between effective exercise and reduced physiological strain. As reported form the study of *McVicar et al, 2022* E-bikes give riders greater control over their levels of exertion and increase the feelings of exercise self-efficacy.

4 CHAPTER 4

4.1 VIBRATIONS AND ACCELERATIONS IN BYCICLE RIDING

4.1.1 Overview of experimental protocols

The studies of *Mcdermid 2015 and Kirkwood 2017* aimed to determine the magnitude of the accelerations to which the athlete and the bicycle are subjected while riding on different types of terrain. Moreover, the study of *Kirkwood et al., 2017* analysed the differences between elite and non-elite cyclists.

Data were collected both in simulated tests and in real competition settings. The ultimate goal is to understand the entities of vibrations and how vibrations may affect the rider's handling and overall performance.

Authors & Year	**Participants (n)**	**Participant Characteristics**	**Experimental Design**	**Measured Variables**
Mcdermid et al. (2015)	7 athletes	National level cyclist	Ramp test on a cycle ergometer; (field test) two climbs with 4.2% of gradient on asphalt and off-road terrain	VO_2, HR, accelerations
Kirkwood et al. (2017)	11 participants	Elite and non-elite group	Test to determine the power at 2 and 4 mmol/L of [BLA$^-$]; peak VO2. Enduro specific test; race event	VO_2 , power, acceleration and load of athletes and bike

4.1.2 Accelerations outcome and responses

Both studies report that on technical and irregular surfaces there is an increase in vibrations and in the load to which the bike and athletes are subjected.

Study	Accelerations/load
Mcdermid et al. (2015)	↑ accelerations during off-road riding for handlebar, left arm, left leg and saddle.
Kirkwood et al. (2017)	↑ more time spent during off-road riding in higher load zones, both for the bike and the riders ↑elite cyclists vs. non-elite cyclists in maintaining the balance ad an optimal performance

4.1.3 Experimental outcomes and interpretation

The increase in vibrations on technical terrains results in a higher demand for the riders for both propulsive and non-propulsive work in order to complete more technical sections of the course. Moreover, another finding emerging from the two studies is that variations in acceleration are detected at the bicycle frame and at the contact points with the body, but not at more distant areas such as the back and head. This suggests that the limbs play a major role in dissipating accelerations, thereby enabling optimal control and performance during riding.

4.1.4 Cross-study synthesis

Parameter	**Trend in technical terrain vs. non technical**
Frame bike accelerations	↑
Riders’ body acceleration: contact points	↑
Riders’ body acceleration: distant areas	=
Cadence	=
Speed	=,↓

4.1.5 Conclusion

The accelerations experienced during cycling influence the physical effort required of the athlete to complete a given course. Evidence from the literature shows that this phenomenon occurs in both national-level athletes and amateur cyclists.

Specifically, the greater the technical complexity of the course, the higher the accelerations to which both the bicycle and the athlete are subjected. Moreover, studies highlight the crucial role of the limbs in attenuating vibrations, thereby preventing their transmission to the trunk and head. This mechanism contributes to maintaining an adequate riding technique and balance.

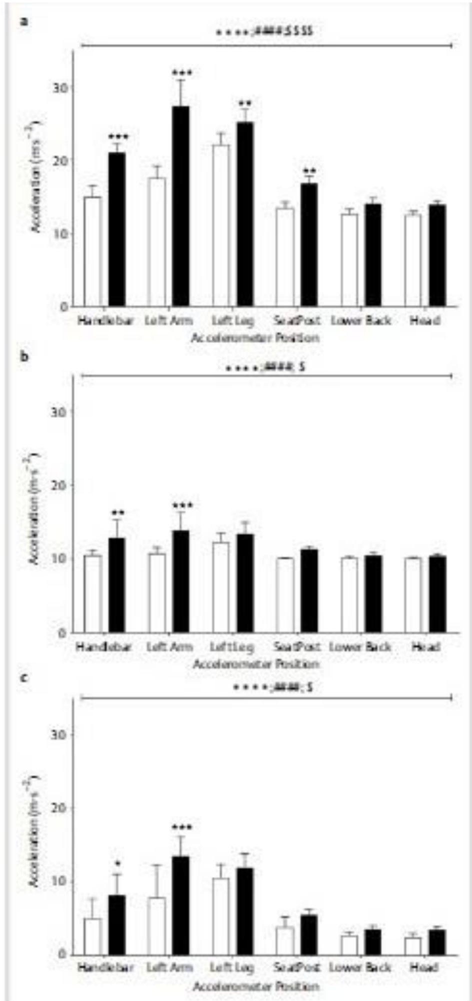


Fig. 1 Mean ± SD amplitude (RMS) for **a** Total; **b** Vertical; and **c** Horizontal components of acceleration over the climb. □ denotes the tar-sealed road condition, while ■ denotes single-track off-road condition. **** ($P<0.0001$) main effect of terrain surface; #### ($P<0.0001$) main effect of acceleration location; $$$$ ($P<0.0001$), $ ($P<0.05$) terrain surface * accelerometer location interaction; Post-hoc analysis differences ★★★ ($P<0.001$), ★★ ($P<0.01$) and ★ ($P<0.05$) when tar-sealed Rd is compared to single-track off-Rd (MTB).

Fig. 4.1 Image taken from the study by *Macdermid et al. (2015)*, showing the comparison of total (a), vertical (b), and horizontal (c) accelerations between road and off-road terrain.

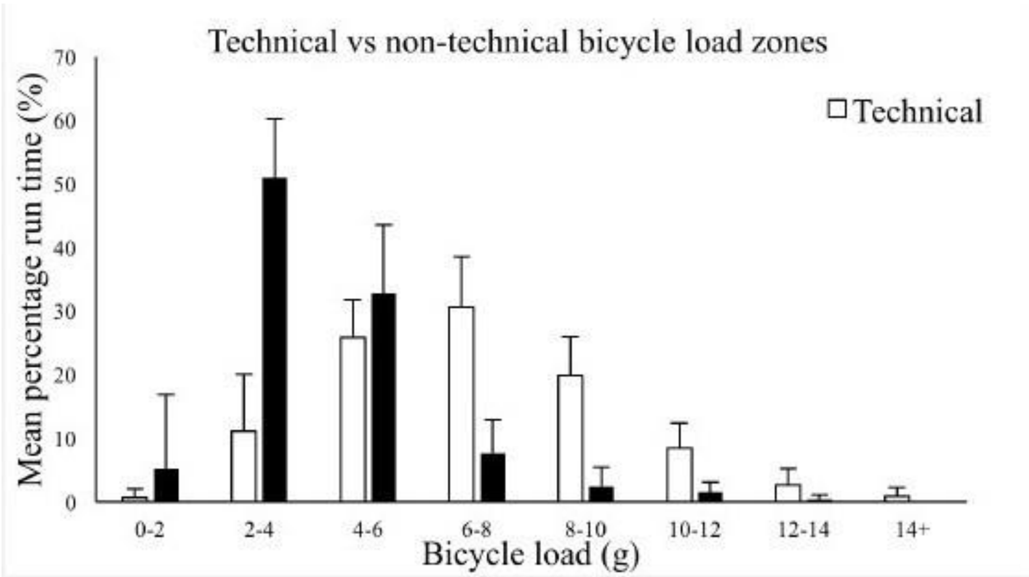


Fig.4.2 Image taken from the study by *Kirkwood L. A. et al. (2017)*, showing the load zones to which the bicycle is subjected on technical and non-technical terrain.

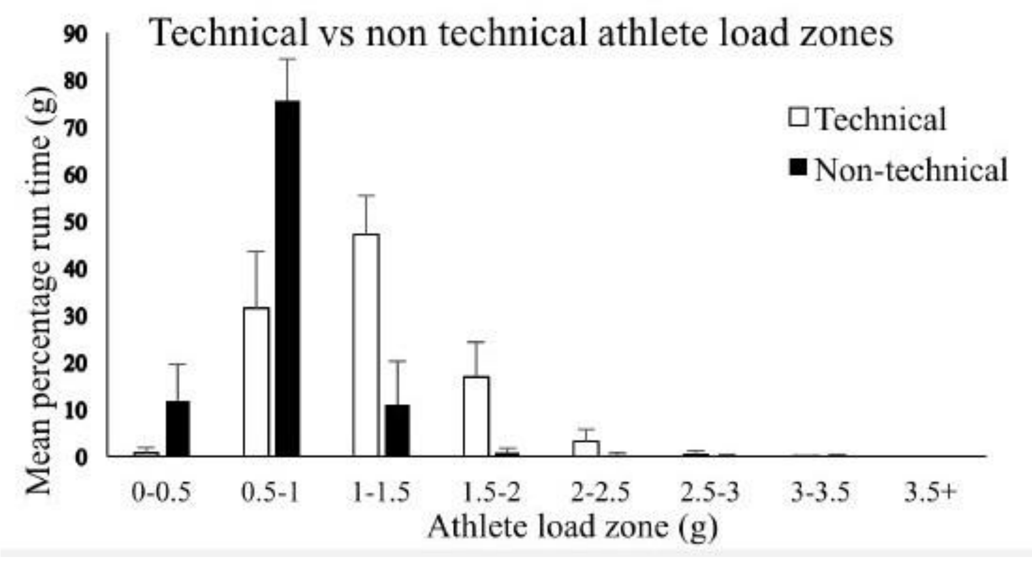


Fig. 4.3 Image taken from the study by Kirkwood L. A. et al. (2017), showing the load zones to which the athlete is subjected on technical and non-technical terrain.

5 CHAPTER 5: STUDY ONE

PHYSIOLOGICAL, MUSCULAR AND PERCEPTUAL PARAMETERS WHILE RIDING AN E-BIKE AT DIFFERENT LEVELS OF ASSISTANCE AND DIFFERENT PEDALLING CADENCE

5.1 ABSTRACT

Purpose: the aim of this study was to identify the effects of the use of different levels of assistance of the electric bike and different cadences on metabolic (oxygen consumption-VO_2, heart rate- Hr, ventilation- VE, blood lactate concentration- [La^-], Energy Expenditure – En. Exp.), perceptual (RPE) and muscle activation by surface (EMG). The second aim was to understand how the e-bike motor operates as a function of cadence and to identify the cadence freely selected by cyclists, in order to better characterize the cyclist–bicycle interaction.

Methods: we tested thirteen recreational cyclists in 5-minute trials at the external power output matching their individual second threshold, comparing no assistance (A0) and two levels of assistance (the lower one- A1 and the highest- A3) at two different pedalling cadences (60 - 90 RPMs). Participants were then asked to pedal at their freely selected cadence across all three assistance levels. Finally, a motor mapping was performed using the anaerobic threshold power of one of the participants as a reference.

Results: the physiological parameters showed an effect of both cadence and levels of assistance and the presence of interaction between the two factors. Significantly lower values were observed at 60 RPM compared to 90 RPM under motor-assisted conditions. This may be due not only to neuromuscular factors but also to the motor delivering more power at 60 RPM than at 90 RPM related to total external power, as measured. In the RMS values of electromyography (EMG) signal, there where an effect of the assistance on the muscles of the upper leg and an effect of the cadence in the muscles of the lower leg. No interaction between factors was found. The preferred pedalling cadence identified was 80 rpm across all assistance levels, whereas the motor operates most efficiently at 60 rpm.

Conclusion: based on the results of this study, we can conclude that rider physiological effort and the choice of optimal cadence when riding an e-bike may differ from what is already known for conventional bicycles. Finally, the motor efficiency could be aligned with the cyclist's freely selected cadence in order to create an optimal cyclist–bicycle system.

Keyword: E-bike; energy expenditure; oxygen consumption, electromyography; cadence management; assistance management.

5.2 INTRODUCTION

As we have seen in chapter three, E-bikes function is to assist cyclists by adding electric motor power to human power. Due to this, the total power to overcome the cycling resistance is distributed between both power sources *(Meyer D. et al., 2017)*.

A trend which is experiencing significant growth, is related to the use of e-bike as a way to be engaged in aerobic and enhance physical activity. In fact, the advent of e-bikes has made it possible to participate in exercise even for those who do not have a high level of fitness *(J Hinder, M Jäger, 2019)* and in ride conditions in which the effort demand is high *(E.G.Avina-Bravo et al.,2021)*. Riding an e-bike has been reported to elicit lower heart rate *(Hall et al., 2019)*, lower oxygen consumption *(Sperlich et al. 2012, Bersten et al. 2017, Rauter et al. 2023)* and lower participants' effort perception *(Hall et al., 2019)*.

In the review of *McVicar et al., 2022* is reported that many of the studies, the participants were advised to cycle at a self-selected pace, only one, over 14 studies, specified the target speeds.

Another aspect is that, to our knowledge, only the study of Sperlich et al., 2012 had investigated the EMG end no articles have monitored the effect of the cadence while riding an e-bike at different levels of assistance. The effect of the cadence is an important aspect to consider for quantifying the exercise because the gear ratio change system on bicycles makes the cadence one of the few variables that the biker can adjust dynamically during a performance to manage performance and fatigue *(Ansley et al., 2009)*.

While cycling a normal bike, cadence choice (80-100 RPM) is higher than the metabolic optimal cadence (50-70 RPM) *(Brisswalter et al., 2000, Lucia et al., 2001; Marsh et al., 1998; Sarre et al., 2003)*.
However, the "optimal" cadence has commonly been defined as that rate of pedaling that leads to minimizing the values of energy cost, muscle stress, and perception of effort *(Ansley et al., 2009)*.

The aim of the study is to identify the effects of the use of different levels of assistance of the electric bike and different cadences on metabolic and neuromuscular responses. To this end, the present study will compare pedaling without assistance and with two levels of assistance at three different cadences, including the freely chosen cadence.

5.3 METHODS

Participants

Thirteen recreational riders (6 females and 7 males, aged 28 ± 3.5 years) participated in the study. Three inclusion criteria were applied: participants were required to be recreational riders, have a body max index (BMI) within the normal range (18.5 – 24.9 kg/m²), and be free from any diagnosed diseases. High-level athletes were excluded because they do not represent the typical target of e-bike users. In contrast, sedentary individuals were excluded as they may not be able to sustain the intensity required by the protocol. The research protocol was approved by the ethical board of the University of Verona (protocol number: CARP# 33/2024). Participants signed informed consent before participating in data collection.

Overall design

The study proposed here is a repeated-measurement experimental design. Measurements were conducted in two different phases divided at least by 24 h of rest, where participants were asked to ride an e-bike (city bike) mounted on a bike trainer.
All participants used the same e-bike, which saddle and handlebar were adjusted for optimal fit based on their personal comfort.

A warm-up of 10 min. at a power of 90 W for men, and 70 W for women, has been proposed before each phase. During the first phase, were asked to perform a Mader Test (5 min. of activity, 2 min. of rest) until reaching a blood lactate [La¯] concentration of 4mmol/ L. For men, the test started at 130 W, followed by increments of 30 W every step, while it started at 100 W with increments of 20 W for women. During the second phase the subjects had to perform in a random order nine different conditions of 5 min. and 2 min. of rest at the power corresponding to blood lactate concentration of 4mmol/ L, modifying the cadence and the levels of the bike assistance (e-bike's assistance levels range from level 1 to level 3). They were asked to pedal at 60, 90 RPMs and with their preferred cadence, without (A0) and with the lowest (A1) and the highest (A3) level of assistance provided by the electric motor. *Figure: 5.1*.

A third testing session was conducted to quantify the actual power output delivered by the bicycle's motor across various assistance levels and pedalling cadences. The participant was instructed to maintain a constant external power output. Power data were simultaneously recorded at the pedal and from the bike trainer.

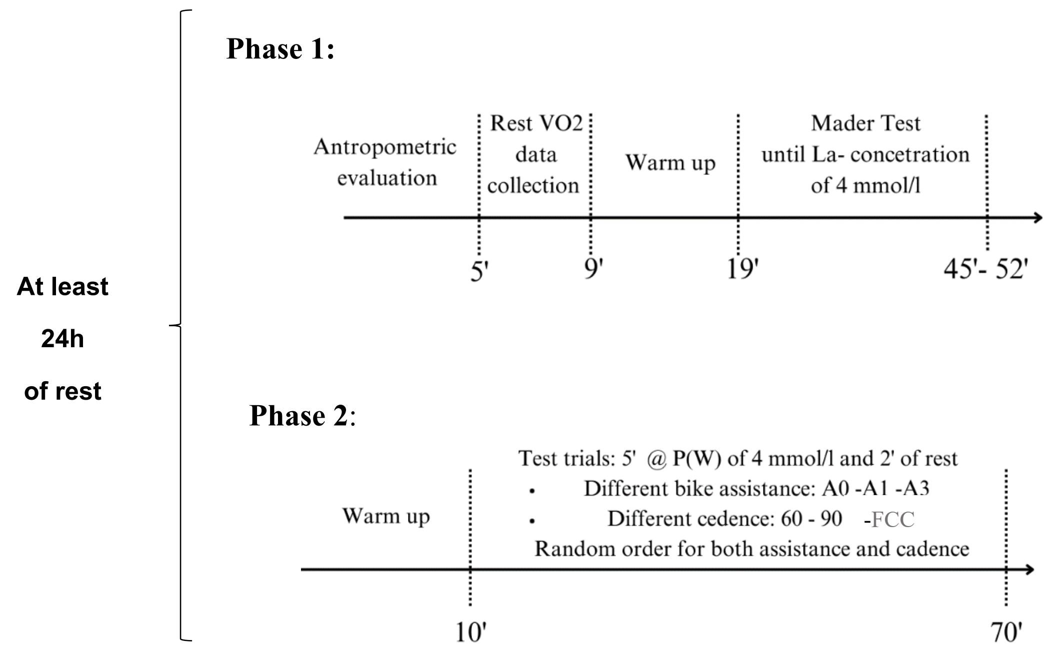


Fig. 5.1: description of the two testing phases.

Measurements

During the first phase, heart rate (HR) and respiratory gas exchange data such as, oxygen consumption (VO_2) and ventilation (VE) were collected both in basal condition and during the exercise.

At the end of each step, a peripheral blood sample was taken from the ear lobe and collected in a 25 μL capillary tube. Moreover, the rate of perceived exertion (RPE) was collected.

As mentioned, the test finished once the subject reached the blood lactate concentration of 4 mmol/L to establish the corresponding power (P_ext_OBLA4).

During the second phase of the experiment *(Fig. 5.2)*, gas exchange data (VO_2, VE), [La^-], RPE, and surface EMG signals were collected.

Eight leg muscles on the right side of the body were involved in the analysis: rectus femoris (RF), vastus medialis (VM), vastus lateralis (VL), biceps femoris (BF), tibialis anterior (TA), soleus (SOL), gastrocnemius medialis (GM) and lateralis (GL). EMG data were collected for 30 s between the second and the third minute of the trials, allowing subjects to standardize their cycling movement. For the FCC condition, only gas exchange variables were analysed.

Fig. 5.2 Set-up of data collection during phase two of the study using the metabolic cart and surface electromyography.

Instruments and materials

Bike and bike trainer equipment

All trials were conducted using the same e-bike (Seven Days Living, Fantic, Dosson di Casier (TV), Italy) mounted on a bike trainer (Kickr Kore, Wahoo, Atlanta, Georgia). The bicycle has an electric motor (Brose T; 36 V, maximum power: 250 W, torque:70 Nm) located at the bottom bracket. The motor assists the rider by providing extra power proportional to the rider's effort. The motor provides three levels of assistance: (ECO) A1, (STD) A2 and (HIGH) A3.

Since it only operates when the bike's speed is above zero and below 25 km/h, detected by sensors and magnets attached respectively to the frame and the wheel, once the bike's wheel was removed, these were positioned on the fixed frame of the trainer and on the flywheel of the trainer. The bike trainer smartphone app (iOS, Wahoo, 6.52.0) was used to set the external power during the trials. The application was additionally used to visually monitor pedalling cadence and to verify that it remained within the prescribed range.

Physiological, Mechanical and Muscular Parameters

Respiratory gas exchange data were measured using a Cosmed Quark CEP system with breath-by-breath analysis, calibrated before each test according to manufacturer instructions. [La^{-}] was measured using a Biosen C-line blood lactate analyser (EKF Diagnostic, GmbH, Magdeburg, Germany).

Participants' perceived exertion was assessed using the Borg CR100 scale after receiving instructions on its proper use.

Muscle activity was monitored using a portable and wireless EMG system (DuePro, OTBioelettronica, Turin, Italy) operating at 2048Hz (Gain 200 V/V, Bandwidth 10 - 500Hz, CMRR > 100 dB, A/D resolution 16 bits). Bipolar pre-gelled disposable electrode pairs (CDE, Spes Medica, Genova, Italy), diameter of 24mm, interelectrode distance ~30 mm, were positioned following guidelines for skin preparation and electrode position previously described (Hermens et al., 2000).

The participant's mechanical power output at the pedal (P_human) was recorded using an instrumented pedal system (Vector 3s, Garmin, Milano, Italy) connected to a sports watch (Forerunner 255, Garmin, Milano, Italy).

Data analysis

Physiological data analysis

The average of VO_2, expressed in ml/min/kg, VE, and HR data, was calculated over the last 30 s. of each testing condition. For [La¯] and RPE data the mean value of every condition was calculated.

Muscular data analysis

EMG signal was recorded for the 30 s of every trial. Raw EMG signals were band-pass filtered (bi-directional, 4^{th}-order, zero lag Butterworth, band-width 20±400 Hz) to attenuate motion artefacts.

All EMG signals were then rectified, and the root mean square (RMS) value was obtained by an overlapping moving window with a length of 125 ms and shift of 1 point.

Pedalling cycles were identified based on the RMS curves of gastrocnemius medialis, using the instant when the muscle becomes inactive as reference point. For each trial, at least 20 non-corrupted by artefacts pedalling cycles were selected for the analysis. The value of RMS was normalized with respect to the averaged EMG amplitude detected during the 60 RPMs-A0 pedalling condition and then expressed as a percentage (EMG%). This condition was selected as the reference (instead of static MVC) because it is theoretically the one that requires higher force values during the pedalling movement. The average value of normalized EMG for each subject and condition was calculated to express the muscle activation over a pedalling cycle.

Mechanical power

During the first session of the test, the trials were conducted without the motor assistance. The value of human power (P_human) corresponding to the intensity that elicited [La¯] exceeding the threshold of 4 mmol/ L was used as the external power setting for trials in the second session (P_ext_OBLA4).

In the third session, the data of power at the pedal (P_pedal) was collected to calculate the real assistance of the motor in every test condition (level of assistance and pedalling cadence) using the formula *(Equation 5.1)*:

$$P_{motor} = P_{ext_{OBLA4}} - P_{pedal}$$

Equation 5.1

Statistical analysis

The effects of assistance level and cadence were analysed using a Linear Mixed Model (LMM). The covariance structure for repeated measures was specified as autoregressive (AR1), and the model was estimated using the Restricted Maximum Likelihood (REML) method.

This was selected to appropriately handle a few missing data in some of the tested parameters. Post hoc pairwise comparisons were adjusted using the Bonferroni correction to control for multiple testing. The effect of three levels of bike assistance on freely chosen cadence was examined using a one-way repeated-measures ANOVA.

5.4 RESULTS

5.4.1 Physiological and perceptual results

The physiological variables, including the RPE, were significantly affected by the levels of assistance with lower values, as expected, as the assistance increases *(Table 5.1).* VO_2 values at 60 RPMs increased by 10.67 ml/min/kg in A0 vs. A1, 8.13 ml/kg/min in A1 vs. A3, 18.8 ml/kg/min in A0 vs. A3; while at 90 RPMs differed by of 5.74 ml/min/kg in A0 vs. A1, 8.33 ml/kg/min in A1vs. A3 and 14.1 ml/kg/min in A0 vs. A3.

HR differences, at 60RPMs, were approximately 42 bpm between A0 and A1, 30 bpm between A1 and A3 and 72 bpm between A0 and A3; while at 90RPMs the differences were about 30 bpm in A0 vs. A1, 35 bpm in A1 vs. A0 and 65bpm in A0 vs. A3.

VE at 60 RPMs differed by 30.5 L/min in A0 vs. A1, 30.86 L/min A1 vs. A3, and 61.36 L/min A0 vs. A3; at 90 RPMs VE differed of 21.13 L/min in A0 vs. A1, 22. 48 L/min in A1 vs A3 and 43.61 L/min in A0 vs. A3; [BLa¯] values, at 60 RPMs, differed by 2.8 mmol/l in A0 vs A3 and 0.3 in A1 vs. A3; at 90 RPMs [BLa¯] varied of 2.48 mmol/L in A0 vs A1, 1.17 mmol/l in A1 vs A3 and 3.65 mmol/l in A0 vs A3.

RPE at 60 RPMs had a difference of 28 pt in A0 vs A1, 17 pt in A1 vs. A3, and 45 pt in A0 vs. A3; at 90 RPMs it differed by 24 pt in A0 vs. A1, 18 pt in A1 vs. A3 and 42pt in A0 vs. A3.

The energy cos values, at 60 RPMs, increased by 3.95 Kcal/min in A0 vs. A1, 3.89 Kcal/min in A1 vs. A3 and 7.84 kcal/min in A0 vs. A3; at 90 RPMs it increased by 2.42 Kcal/min in A0 vs. A1, 4.02 Kcal/min in A0 vs A1 and 6.45 Kcal/min in A0 vs. A3.

There was also a significant effect of the RPMs for all the physiological parameters, especially in A1 and A3 with the highest values at 90 RPMs *(Table 5.1)*.

The VO_2 had a difference of 5.97 (ml/min/kg) in A1 and 5.77 (ml/min/kg) in A3; HR had a difference of 16.17 (bpm) in A1 and 12.68 (bpm) in A3; VE had a difference of 13.01 (l/min) in A1 and 8.3 (l/min) in A3; the [BLa^-] changed only in A1 with 0.81 (mmol/l) of difference and the energy expenditure had a difference of 1.86 (kcal/min) in A1 and 1.72 (kcal/min) in A3. In A0, only the [BLa^-] changed significantly with higher values at 90RPMs (Δ = 2.55 mmol/l).

All the physiological parameters, but not the RPE, were also affected by the interaction of the different levels of assistance and different RPMs (VO_2, HR , VE, Energy expenditure) *(Table 5.1)*. Specifically, the parameters of VO_2, HR and VE showed differences, with the highest values at 90 RPMs as told before, in A1 and A3.

In A0 no effect of cadence. The [BLa^-] showed higher values at 90 RPMs especially in A0 and A1, while the energy expenditure showed an effect of interaction only in A1, also in this case with the highest values at 90 RPMs.

5.4.2 Muscular results

Muscles activation changed significantly across the pedalling cadence and the levels of assistance in two different ways: the muscles of the lower leg were affected by the cadence with higher values at 90RPMs, while the muscles of the upper leg, also including the SOL for the lower leg, were affected by the levels of assistance with lower values in A3 compared to A2 and A1 *(Table 5.1)*.

As far as the effect of pedalling cadence, the differences for TA were about 57% in A0 and 32% in A3, no difference were found in A1.

The SOL increased the activation at higher cadence in A1 and A3 of respectively 32% and 61.28%; GM had a difference of 55% in A0, 61% in A1 and 35% in A3.

Last, GL had a difference of 69% in A0, 44% in A1, and no difference was found in A3.

In the case of the effect of the levels of assistance RF at 60 RPMs showed differences of 30% in A0 vs. A1, 20% in A1 vs A3 and 50% in A0 vs A3; while at 90 RPMs showed a differences of 24 % in A0 vs A1, 15% in A1 vs A3 and 40% in A3 vs. A0.

VM showed at 60 RPMs a difference of 27% in A0 vs. A1, 39 % in A1 vs. A3 and 66% in A0 vs. A3; while at 90 RPMs showed 15% of difference in A1 vs A3 and 40% in A0 vs. A3.

VL at 60 RPMs showed a change of 24% in A0 vs A1, 39% in A1 vs. A3, and 63% in A0 vs. A3; at 90 RPMs, there was a difference only between A0 vs. A3 of 51.03% and A3 vs A1 of 36 %. BF at 60 RPMs had differences in A0 vs A1 of 35%, A1 vs A3 of 20% and 55 % between A0 and A3; at 90 RPMs had a difference of 28% in A0 vs A1, of 25% in A1 vs A3 and 53% in A0 vs. A3.

Also two muscles of the lower leg were affected by the assistance: the SOL showed differences at 60 RPMs only in A1 vs A3 and A3 vs. A0, in both cases of 18%; and at 90 RPMs in A3 vs. A0 of 65.13%. No effects of interaction between different levels of assistance and RPMs were found.

	60RPMs			Effect of assistance for 60RPM		90RPMs			Effect of assistance for 90RPM		Assistance	Cadence	Interaction
	A0	A1	A3			A0	A1	A3					
	Mean (SD)	Mean (SD)	Mean (SD)	P	F	Mean (SD)	Mean (SD)	Mean (SD)	P	F			
RF [a]	100 (0.0)	**69.6 (13.6)*** **	**49.8 (22.6) ***; #**	**<0.001**	20.57	89.5 (23.2)	**65.2 (20.2) ** **	**50.1 (24.1) *** **	**<0.001**	12.74	**<0.001**	.174	.522
VM [a]	100 (0.0)	**73.4 (16.2) *** **	**33.6 () ***; ###**	**<0.001**	51.63	94.8 (28.4)	79.4 (7.3)	**43.2 (14.6) ***; ###**	**<0.001**	32.37	**<0.001**	.329	.231
VL [a]	100 (0.0)	**75.5 (39.8) ** **	**36.7 (11.4) ***; ###**	**<0.001**	40.99	98.6 (32.7)	83.4 (11.3)	**47.6 (11.8) ***; ###**	**<0.001**	27.58	**<0.001**	.103	.378
BF [a]	100 (0.0)	**65.5 (13.8) *** **	**45.3 (25.3) ***; #**	**<0.001**	23.41	98.7 (16.4)	**70.8 (5.7) *** **	**45.6 (24.1) ***; ##**	**<0.001**	21.28	**<0.001**	.669	.776
TA [a]	100 (0.0)	105.7 (74.5)	82.1 (45.7)	0.610	0.50	155.7 (110.6) **§§**	134.0 (66.6)	127.9 (73.1) **§**	0.540	0.62	.602	**<0.001**	.623
SOL [a]	100 (0.0)	**70.6 (25.5)**	**53.0 (16.1) ***; #**	**0.002**	6.99	104.3 (39.0)	102.9 (42.1) **§§**	**74.6 (43.4) #; §**	**0.025**	3.89	.002	**<0.001**	.241
GM [a]	100 (0.0)	85.6 (26.4)	89.5 (48.3)	0.724	0.32	154.6 (59.8) **§§**	146.8 (67.8) **§§§**	124.4 (41.9) **§**	0.275	1.32	.414	**<0.001**	.535
GL [a]	100 (0.0)	93.7 (17.6)	78.2 (43.0)	0.667	0.41	168.7 (106.6) **§§§**	137.7 (67.9) **§§**	**103.1 (58.6) ***	**0.037**	3.45	.110	**<0.001**	.308
VO2 (ml/min/kg)	38.5 (5.8)	**27.85 (5.95) *** **	**19.72 (3.7) ***; ###**	**<0.001**	57.97	39.5 (6.1)	**33.8 (5.1) ***;§§§**	**25.4 (4.8) ***; ###; §§§**	**<0.001**	32.77	**<0.001**	**<0.001**	**.003**
HR (bpm)	163.0 (16.3)	**121.4 (13.6) *** **	**90.5 (15.0) ***; ###**	**<0.001**	102.76	166.5 (14.0)	**137.6 (4.4) ***;§§§**	**103.2 (13.4) ***; ###; §§§**	**<0.001**	77.73	**<0.001**	**<0.001**	**.019**
VE (l/min)	73.6 (16.5)	**43.1 (3.9) *** **	**25.4 (3.1) ***; ###**	**<0.001**	100.86	77.3 (16.5)	**56.1 (6.5) ***;§§§**	**33.7 (4.4) ***; ###;§§§**	**<0.001**	78.58	**<0.001**	**<0.001**	**.011**
[BLa] (mmol/l)	3.82 (1.12)	1.27 (0.42)	**1.01 (0.21) ***; ###**	**<0.001**	58.51	4.56 (1.91) **§§**	**2.08 (0.46) ***;§§**	**0.91 (0.16) ***; ###**	**<0.001**	81.15	**<0.001**	**.001**	**.021**
RPE (CR100 - Borg)	48.4 (10.5)	**20.2 (3.1) *** **	**3.8 (3.4) ***; ##**	**<0.001**	42.45	47.8 (14.0)	**23.6 (11.1) *** **	**5.3(3.8) ***; ###**	**<0.001**	37.83	**<0.001**	.540	.822
En. Exp (kcal/min)	10.61 (2.38)	**6.66 (1.03) *** **	**2.77 (0.78) ***; ###**	**<0.001**	113.21	10.94 (2.54)	**8.52 (1.25) ***;§§§**	**4.49 (0.78) ***; ###**	**<0.001**	80.05	**<0.001**	**<0.001**	**.001**

[a] unit of measurement: %A0_60RPMs; En. Exp = energy expenditure

Tab. 5.1: effects of pedalling cadence (P values, F values) across the levels of assistance in muscular, physiological, and perceptual parameters (Mean ± SD; * statistically different from A0 (P < 0.001 ***; P < 0.01 **, P< 0.05 *); # statistically different from A1 (P< 0.001 ###; P < 0.01 ##, P< 0.05#);§ effect of cadence (90 RPM vs 60) at the same assistance level (§ = p<0.05 §§ = p<0.01 §§§ = p<0.001)

Statistical analysis output on the effects of cadence and assistance

Effect of E-bike's levels of assistance on physiological parameters

	A0			A1			A3			Assistance differences		
	n	Mean (SE)	95% CI	n	Mean (SE)	95% CI	n	Mean (SE)	95% CI	P (a ; b; c*)	Δ (a; b; c*)	F
VO2 (ml/min/kg)	26	39.04 (1.43)	36.09 - 41.99	26	30.83 (1.43)	27.88 - 33.68	26	22.61 (1.43)	19.66 - 25.56	**<0.001 ; <0.001; <0.001**	8.21; 8.22; 16.43	54.04
HR (bpm)	26	165 (4.18)	156.15 - 173.43	26	130 (4.18)	120.86 - 138.16	26	97 (4.18)	88.27 - 105.55	**<0.001 ; <0.001; <0.001**	35; 33; 68	109.6
VE (l/min)	26	70.5 (3.03)	69.16 - 81.83	26	49.69 (3.03)	43.35 - 56.02	26	29.56 (3.03)	23.23 - 35.90	**<0.001 ;< 0.001; <0.001**	20.36; 20.13; 40.94	107.60
[BLa] (mmol/l)	24	4.18 (0.17)	3.83 - 4.56	24	1.67 (0.17)	1.32 - 2.02	24	0.96 (0.17)	0.61 - 1.31	**<0.001; 0.004; <0.001**	2.51; 0.71; 3.22	105
RPE (CR100 - Borg)	24	48.13 (2.70)	42.65 - 50.60	24	21.96 (2.70)	16.48 - 27.43	24	4.60 (2.70)	0.00 - 10.08	**<0.001 ;< 0.001; <0.001**	26.17; 17.36; 43.53	66.22
En. Exp (kcal/min)	26	10.78 (0.48)	9.78 - 11.78	26	7.60 (0.48)	6.56 - 8. 60	26	3.63 (0.48)	2.64 - 4.67	**<0.001 ; <0.001; <0.001**	3.18; 3.97; 7.15	114.7

* a: A0 - A1 ; b: A1 - A3; c: A0 - A3

Table 5.2: effects of E-bike's levels of assistance (P values, F values) in physiological, and perceptual parameters (Mean ± SE).

Effect of RPMs on physiological parameters

	60 RPMs			90 RPMs			RPM differences		
	n	Mean (SE)	95% CI	n	Mean (SE)	95% CI	P	Δ (60 -90)	F
VO2 (ml/min/kg)	39	20.70 (1.24)	26.10 -31.32	39	32.96 (1.24)	30.33 - 35.60	**<0.001**	-12.26	58.17
HR (bpm)	39	125 (3.64)	117.29 - 132.73	39	136 (3.64)	128.10 - 143.52	**<0.001**	-11	44.75
VE (l/min)	39	47.42 (2.68)	41.66 - 53.20	39	55.74 (2.68)	49.97 - 61.53	**<0.001**	-8.32	58.83
[BLa] (mmol/l)	38	2.03 (0.13)	1.77 - 2 .30	38	2.51 (0.13)	2.24 - 2. 77	**<0.001**	-0.48	14.08
RPE (CR100 - Borg)	38	24.17 (2.10)	20.01 -28.33	38	25.63 (2.10)	21.47 -29.78	0.549	-1.46	0.38
En. Exp. (kcal/min)	38	6.68 (0.43)	5.76 - 7.61	38	7.98 (0.43)	7.05 - 8.91	**<0.001**	-1.3	65.58

Table 5.3: effects of RPMs (P values, F values) in physiological, and perceptual parameters (Mean ± SE).

Effect of E-bike's levels of assistance on muscular activation

	A0			A1			A3			Assistance differences		
	n	Mean (SE)	95% CI	n	Mean (SE)	95% CI	n	Mean (SE)	95% CI	P (a ; b; c*)	Δ (a; b; c*)	F
Rectus Femoris (% A0_60RPMs)	26	94.80 (4.50)	85.67 - 103.92	26	67.45 (4.50)	58.33 - 76.58	24	49.98 (4.62)	40.64 - 59.32	**<0.001; 0.010; <0.001**	27.35; 18.17; 45.52	25.39
Vastus medialis (% A0_60RPMs)	26	97.42 (3.33)	90.72 - 104.13	26	76.44 (3.33)	69.74 - 83.15	24	38.45 (3.46)	31.50 - 45.39	**<0.001 ; <0.001; <0.001**	20.98; 37.90; 58.97	77.27
Vastus lateralis (% A0_60RPMs)	26	99.35 (3.74)	91.82 - 106.88	26	79.50 (3.74)	71.97 - 87.03	24	42.20 (3.86)	34.43 - 49.96	**<0.001; <0.001; <0.001**	19.85; 37.3; 57.15	58.37

Biceps femoris (% A0_60RPMs)	26	99.37 (4.72)	89.76 - 108.97	26	68.19 (4.72)	58.59 - 77.80	24	45.52 (4.83)	35.71 - 55.33	**<0.001; <0.001; <0.001**	31.18; 22.67; 53.85	33.42
Tibialis anterior (% A0_60RPMs)	26	127.84 (17.30)	92.63 - 163.06	26	119. 85 (17.30)	84.63 - 155.06	24	105.00 (17.58)	69.28 - 140.72	1.000; 1.000; 0.983	7.99; 14.85 ;22.84	0.511
Soleus (% A0_60RPMs)	26	102.16 (8.60)	84.53 - 119.79	26	91.26 (8.60)	73.63 - 108.89	24	63.81 (8.71)	45.99 - 81.63	0.624; **0.007; 0.003**	10.09; 27.45; 38.35	6.782
Gastrocnemius medialis (% A0_60RPMs)	26	127.31 (10.69)	105.65 - 148.97	26	116.20 (10.69)	94.54 - 137.86	24	106.95 (11.00)	84.71 - 129.18	1.000; 1.000; 0.567	11.11; 31.49; 42.6	0.897
Gastrocnemius lateralis (% A0_60RPMs)	26	134.36 (14.65)	104.72 - 164. 01	26	115.73 (14.65)	86.08 - 145.37	24	90.62 (14.98)	60.35 - 120. 87	0.887; 0.500; 0.114	18.63; 25.11; 74.01	2.29

* a: A0 - A1 ; b: A1 - A3; c: A0 - A3

Table 5.4: effects of E-bike's levels of assistance (P values, F values) in in muscular parameters (Mean ± SE).

Effect of RPMs on muscular activation

	60 RPMs			90 RPMs			RPM differences		
	n	**Mean (SE)**	**95% CI**	**n**	**Mean (SE)**	**95% CI**	**P**	**Δ (60 -90)**	**F**
Rectus Femoris (% A0_60RPMs)	38	73.16 (3.45)	66.15 - 80.16	38	68.33 (3.46)	61.31-75.34	0.174	4.83	1.192
Vastus medialis (% A0_60RPMs)	38	60.02 (3.63)	62.86 -77.70	38	72.53 (3.63)	64.34 - 79.21	0.329	-12.51	0.981
Vastus lateralis (% A0_60RPMs)	38	70.76 (2.91)	64.93 - 76.59	38	76.60 (2.91)	70.77 - 82.43	0.103	0.16	2.78
Biceps femoris (% A0_60RPMs)	38	70.28 (3.63)	62.86 - 77.70	38	71.77 (3.63)	64.34 - 79.21	0.669	-1.49	0.185
Tibialis anterior (% A0_60RPMs)	38	95.96 (13.77)	67.51 - 124.40	38	139.17 (13.79)	110.68 - 167.66	**<0.001**	-43.21	18.55
Soleus (% A0_60RPMs)	38	77.54 (7.03)	62.80 - 92.27	38	93.95 (7.04)	79.13 - 108.70	**<0.001**	-16.41	13.66
Gastrocnemius medialis (% A0_60RPMms)	38	91.70 (8.22)	75.10 - 108.30	38	141.94 (8.23)	125.33 - 158.55	**<0.001**	-50.24	31.7
Gastrocnemius lateralis (% A0_60RPMs)	38	90.63 (11.29)	67.67 - 113.60	38	136.5 (11.31)	113.51 - 159.50	**<0.001**	-45.87	19.37

Table 5.5: effects of RPMs (P values, F values) in muscular parameters (Mean ± SE).

5.4.3 Mechanical analysis results

With the section three of the experiment we assessed the map of the motor behaviour. This procedure allowed us, first, to determine the amount of power delivered by the motor at different cadences for each level of assistance according to the total power request *(table 5.3)*, and second, to assess the differences between the conditions.

We found that in all the e-bike levels of assistance the power delivered by the motor decreases as cadence increases: in A1 at 60 and 90 RPMs the motor delivers respectively the 40% and the 28% of the total external power, while in A3 it delivers the 80% and the 76% of the total external power respectively at 60 and 90 RPMs.
The power delivered by the motor at 60 rpm was then found to be 13% and 5% higher than at 90rpm in A1 and A3, respectively *(Fig. 5.4)*.

	RPM	**P_ PEDAL (w)**	**P_ MOTOR (w)**	**P_MOTOR/P_nom (%)**
A1	50	132.04	112.96	46%
	60	144.84	100.16	41%
	75	152.39	92.61	38%
	90	175.73	69.27	28%
	105	195.45	49.55	20%
A2	50	74.61	170.39	70%
	60	95.48	149.52	61%
	75	117.42	127.58	52%
	90	110.19	134.81	55%
	105	117.71	127.29	52%
A3	50	38.35	206.65	84%
	60	45.63	199.37	81%
	75	70.77	174.23	71%
	90	58.00	187.00	76%
	105	59.35	185.65	76%

Tab. 5.6: Table showing the power output at the pedal by the subject and the power delivered by the motor in absolute values (W) and, for the motor, as a percentage of the nominal power (the power set on the bike trainer corresponding to the second metabolic threshold, which correspond to the total external power).

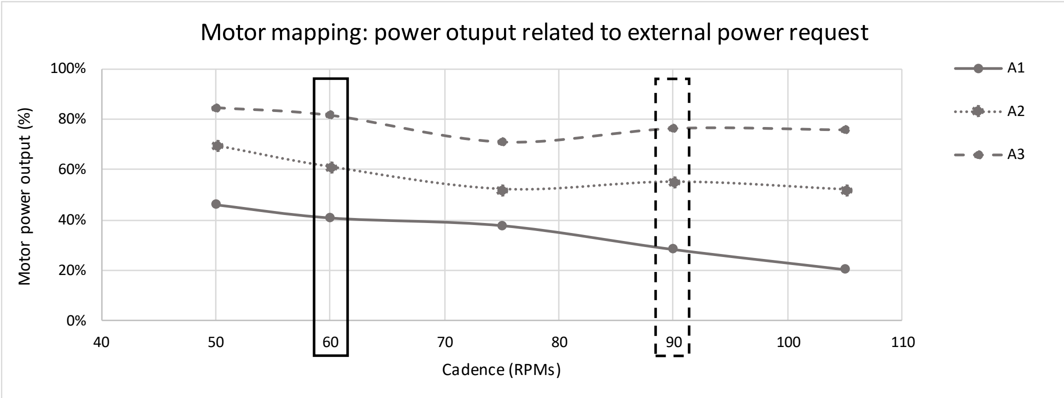


Fig. 5.4: Image showing the power curves generated by the motor at different assistance levels and pedalling cadences.

5.4.4 Vo_2 consumption behaviour and power output across different pedalling condition

Through an analysis conducted to better understand the pattern of oxygen consumption, it was found that VO_2 is higher at 90 RPM compared to 60 RPM. A plausible explanation for this pattern is the greater internal energy required to sustain the cyclic pedalling motion at higher cadences. To investigate this, internal energy was estimated using the model proposed by Rodolfo Minetti (2001). The analysis revealed a higher additional power demand at 90 RPM, accounting for approximately 19% of total power, compared to about 6% at 60 RPM. These results support the interpretation that increased internal energy demand contributes to the elevated oxygen consumption at higher cadences, in line with the findings reported in the literature *(Fig.5.5)*.

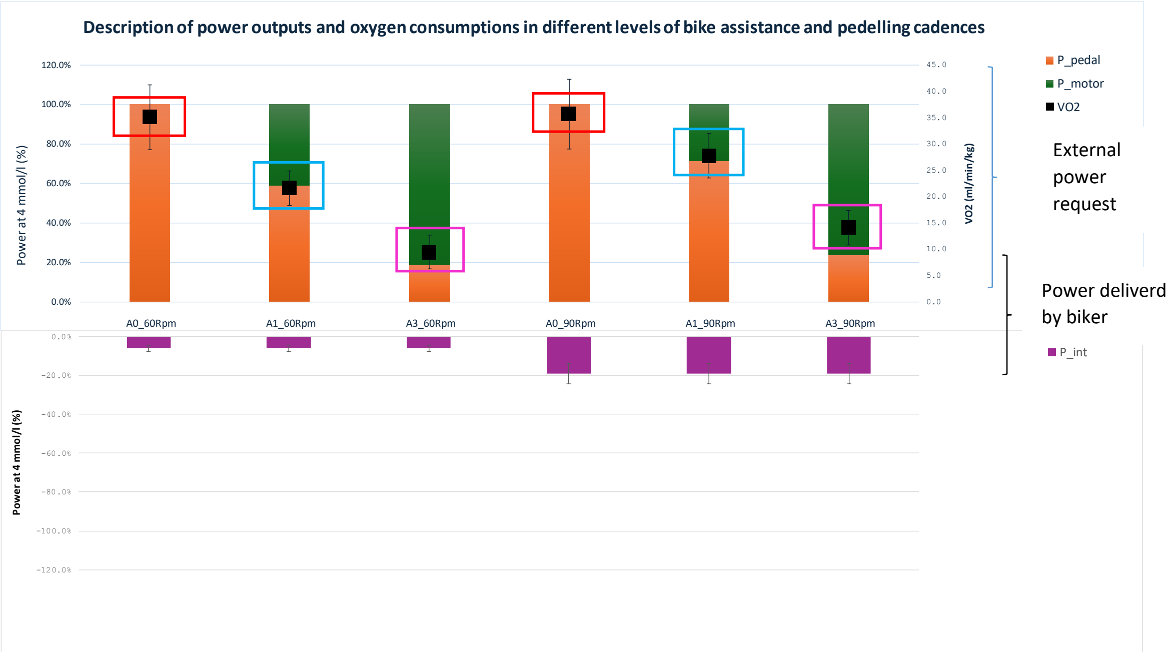


Fig. 5.5: Graph showing the percentage of power delivered by the motor and the cyclist at different assistance levels and cadences, along with the corresponding oxygen consumption. As previously described the same level of assistance results in higher oxygen consumption at 90 RPM compared to 60 RPM (rectangle of the same colour**)**.

5.4.5 Freely chosen cadence

In this study, we also asked participants to pedal at different assistance levels (A0, A1, A3) using their preferred cadence. The main aim was identifying the average cadence typically used on an E-bike and, subsequently, of analysing oxygen consumption by comparing it with the values obtained at 60 and 90 RPM.

As shown in Figure 5.6, the freely chosen cadence tends to increase as assistance decreases. However, the statistical analysis revealed that the effect of bike assistance level on freely chosen cadence was not statistically significant, F (2, 26) = 2.10, p = 0.143. Moreover, as we can see from our data, we discovered that even in the case of the e-bike, the freely chosen cadence (across all three assistance levels) is higher than the metabolically optimal range of 50–70 RPM. The observed values were around 80 Rpm in all three analysed assistances.

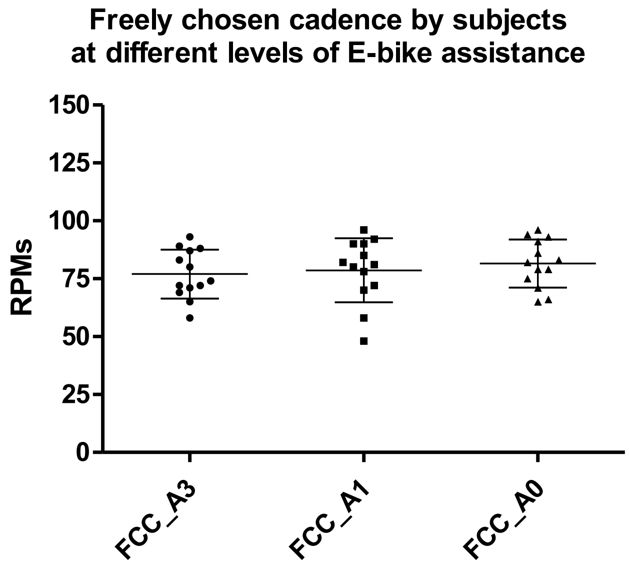


Fig. 5.6: descriptive illustration of the freely chosen pedalling cadences across the three e-bike assistance conditions.

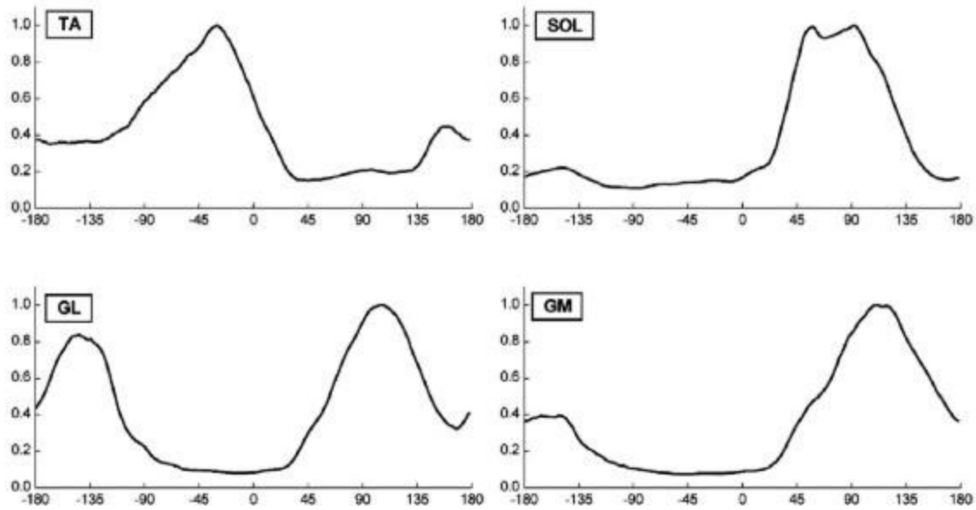
5.4.6 Exploratory finding on muscular activation pattern during pedalling

As described in Chapter Two, muscular activity varies throughout the pedal cycle according to the position of the crank arm relative to top dead centre (TDC) and bottom dead centre (BDC). Muscle activation is therefore characterized by a specific pattern that changes with pedal angle (Figure 5.7). In addition, this pattern is influenced by the power output, pedalling cadence, rider posture, the cyclist's level of fatigue and training, and the type of pedal used (flat, toe-clip, or clipless)

In our study, using a fourth stage of data processing, we applied this method of describing EMG activity in order to understand how motor assistance and different pedalling cadences may influence electromyographic activity, both in terms of muscle activation level and activation timing.
To achieve this, an IMU sensor was mounted at the centre of the crank arm, at the height of the bottom bracket, during data collection. Prior to recording each trial, a subject-specific calibration was performed to identify TDC and BDC. Consequently, during the data analysis phase, TDC was located on the graph to define the beginning of the pedal

cycle. All other time points were subsequently expressed as a percentage of the total pedal cycle. The results are presented in Figure 5.8, which illustrates the pedal-cycle instants as a function of crank angle. In this regard, it should be noted that the angular values were assigned retrospectively; therefore, although pedalling cadence was constant, some angles may not perfectly reflect the actual values.

By comparing the recorded activations with those depicted in Figure 5.7, a good correspondence in activation timing can be observed. In our case, however, differences in timing and activation magnitude emerge as a function of both pedalling cadence and motor assistance. Unfortunately, due to issues encountered during data collection, only one participant could be analysed. As a result, we are unable to draw robust conclusions regarding how these parameters may influence the EMG signal. Nonetheless, for future studies, completing this type of analysis could be of interest, as it may provide missing information into the use of e-bikes.



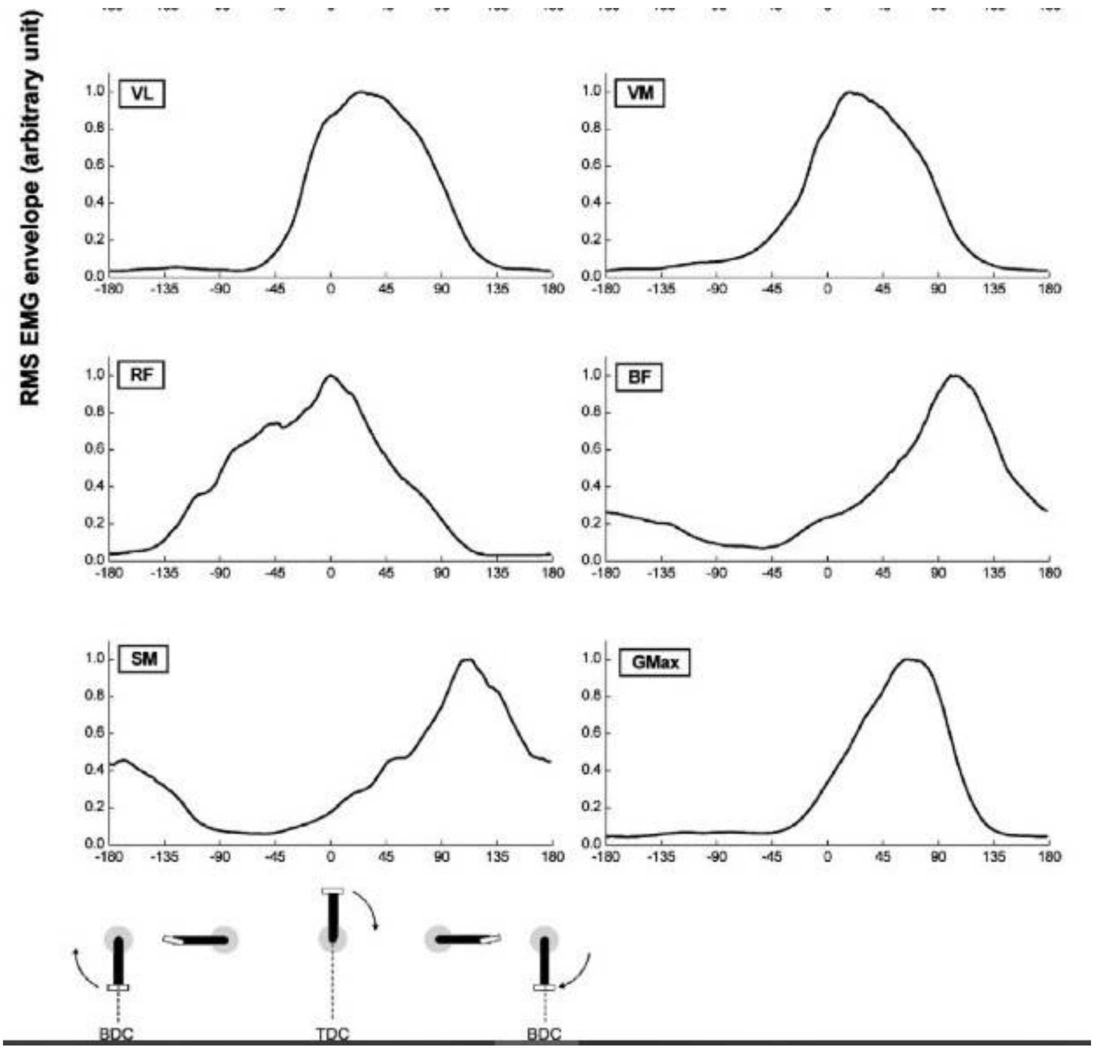
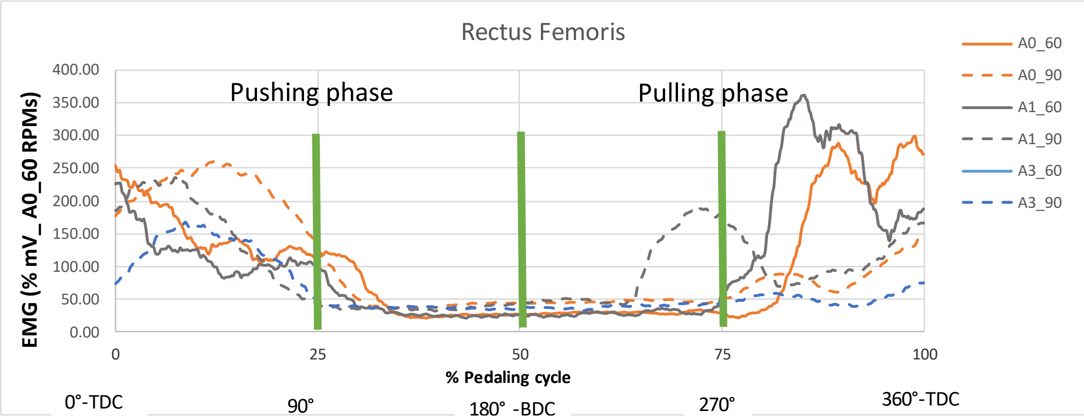
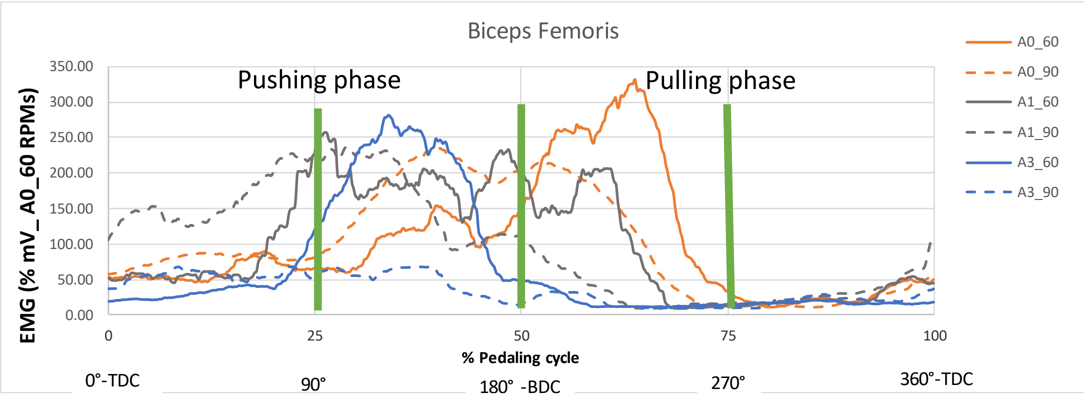


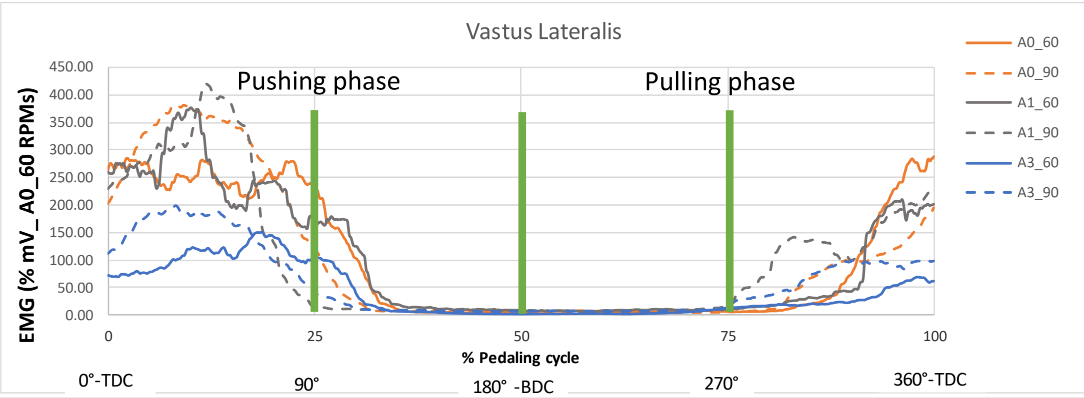
Fig. 5.7: Image taken from the study by *Hug et al. (2009)*, showing the activation pattern of the muscles mainly involved in the pedalling motion according to the position of the crank relative to TDC and BDC.



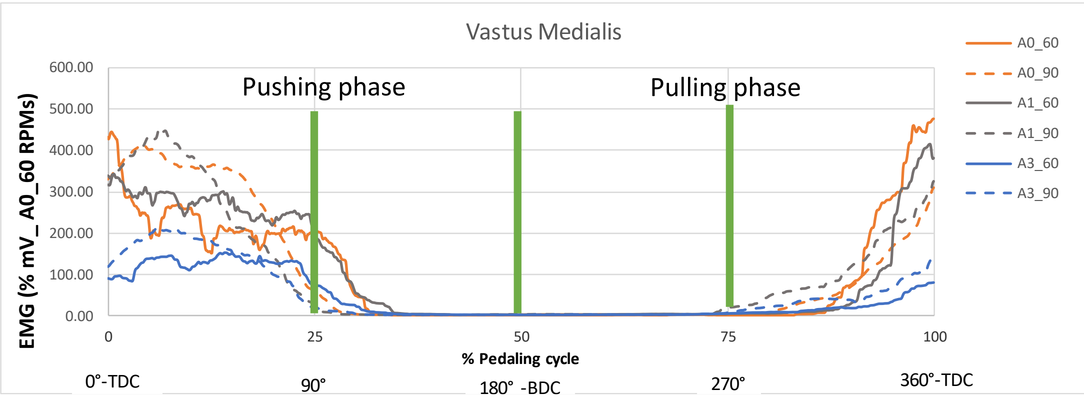
Rectus Femoris
Pushing phase
Pulling phase
EMG (% mV_ A0_60 RPMs)
400.00
350.00
300.00
250.00
200.00
150.00
100.00
50.00
0.00
0
25
50
75
100
% Pedaling cycle
0°-TDC
90°
180° -BDC
270°
360°-TDC
A0_60
A0_90
A1_60
A1_90
A3_60
A3_90



Biceps Femoris
Pushing phase
Pulling phase
EMG (% mV_ A0_60 RPMs)
350.00
300.00
250.00
200.00
150.00
100.00
50.00
0.00
0
25
50
75
100
% Pedaling cycle
0°-TDC
90°
180° -BDC
270°
360°-TDC
A0_60
A0_90
A1_60
A1_90
A3_60
A3_90



Vastus Lateralis
Pushing phase
Pulling phase
EMG (% mV_ A0_60 RPMs)
450.00
400.00
350.00
300.00
250.00
200.00
150.00
100.00
50.00
0.00
0
25
50
75
100
% Pedaling cycle
0°-TDC
90°
180° -BDC
270°
360°-TDC
A0_60
A0_90
A1_60
A1_90
A3_60
A3_90



Vastus Medialis
Pushing phase
Pulling phase
EMG (% mV_ A0_60 RPMs)
600.00
500.00
400.00
300.00
200.00
100.00
0.00
0
25
50
75
100
% Pedaling cycle
0°-TDC
90°
180° -BDC
270°
360°-TDC
A0_60
A0_90
A1_60
A1_90
A3_60
A3_90

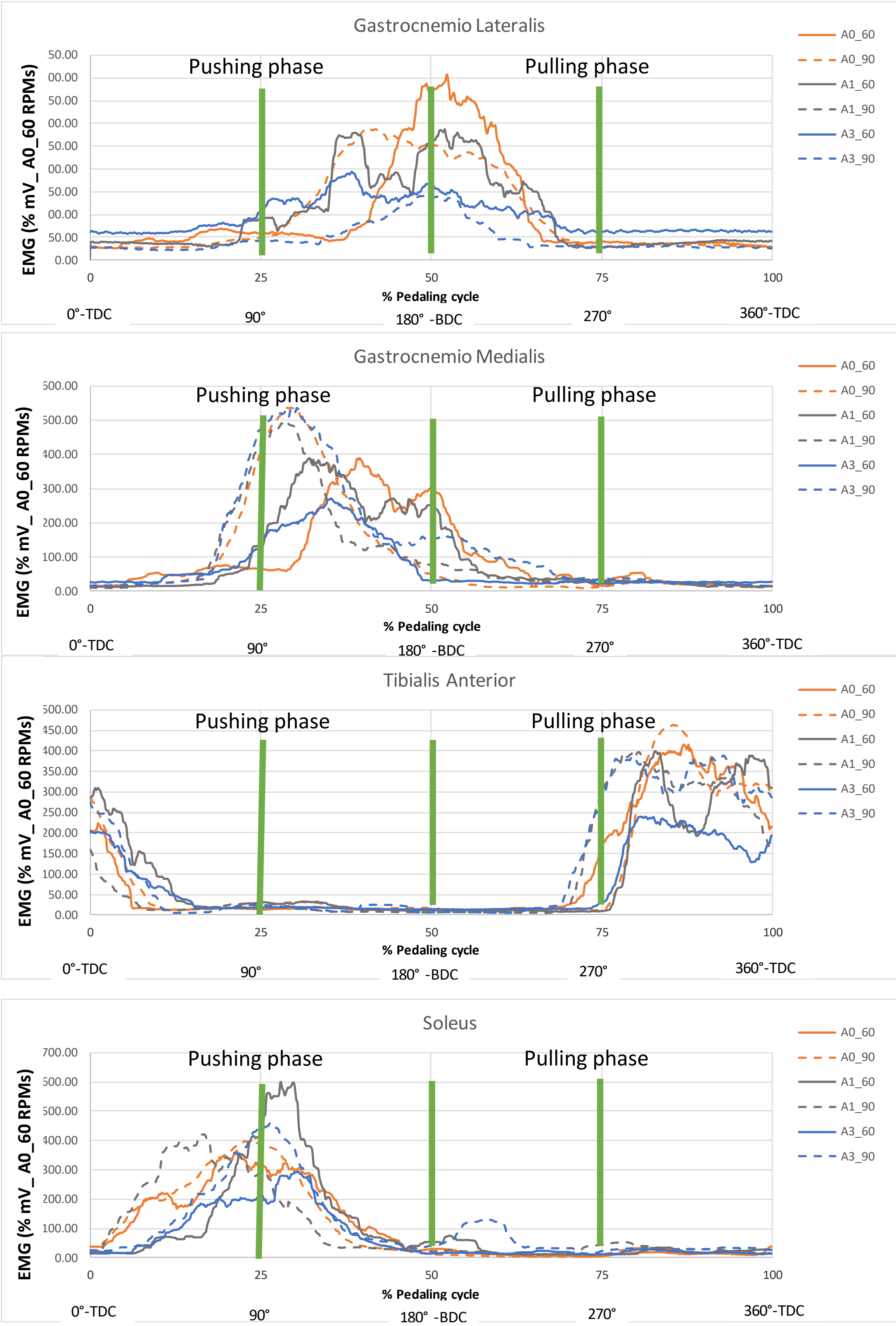


Fig. 5.8: Muscle activation and activation pattern of a subject at three different assistance levels (A0, A1, A3) and two different cadences (60–90 rpm).

5.5 DISCUSSION

The results of this study indicate that both investigated factors, pedalling cadence and e-bike assistance level, significantly influence metabolic responses during e-bike cycling. Specifically, physiological parameters were higher at a cadence of 90 RPM compared to 60 RPM and decreased as the level of motor assistance increased.

For the rate of perceived exertion (RPE) parameter, only an effect of assistance was found, with lower values as the level of assistance increased. These findings are consistent with previous literature. For example, *Hang et al. 1992* reported that pedalling at 90 RPM, compared to 60 RPM, during submaximal exercise intensities (127 W and 166 W), elicited higher ventilator responses, heart rate (HR), and lower exercise efficiency. This factor is influenced by three main elements: the first is due to the fact that an increase in pedalling cadence leads to greater recruitment of muscle mass *(Ericson et al. 1986)*; the second concerns the internal energy used to rotate the legs, which increases as cadence rises *(Masato Tokui et al., 2007)*; and lastly, increasing cadence results in a reduction in pedalling efficiency due to non-productive contractions and reduced muscular coordination *(Patterson and Moreno, 1990)*.

With regard to motor assistance, previous studies comparing E-bike use with conventional cycling have shown that electric assistance is associated with lower values of HR (94%) *(Hall et al., 2019)*, lower values for VO_2 (17% hilly route and 6% flat route) *(Bersten et al., 2017)* and lower participants' effort perceptions *(Hall et al 2019)*. Moreover, in our study, all physiological parameters were influenced by the interaction between cadence and assistance level. VO_2, HR, and ventilation (VE) showed significant cadence effects in conditions A1 and A3, but no cadence effect was observed when the motor was inactive (A0). Blood lactate concentration ($[La^-]$) was affected by cadence in conditions A0 and A1, with no significant effect at the highest assistance level (A3).

Finally, RPE was not influenced by the interaction between cadence and assistance.

On muscular activation, the effect of the pedalling cadence occurred in the muscles of the lower leg (TA, SOL, GM, GL) with higher activation values at the higher RPM condition. This result can be compared for the GM to the study of *Marsh et al., 1995* who reported a substantial increase in GM activation with increasing cadence.

In the present study, no effect of cadence was found for upper leg muscle (RF, VM, VL and BF). This again partially agree with *Marsh et al., 1995* results, as they found changes with linear increase with cadence for VL, a quadratic trend for RF and no effect of cadence for BF. On the other hand, the levels of assistance influence the muscles of the upper leg (RF, VM, VL, BF) showing lower activation values as the assistance increases, as expected. Also the SOL had a significant effect, but only at A3 compared to the lower levels. No effects of interaction between cadence and levels of assistance were found on muscle engagement.

These findings are consistent with those of *Sperlich et al. 2012*, who reported reduced activation in the biceps femoris, vastus lateralis, vastus medialis, and gastrocnemius medialis during e-bike use under assisted conditions across various terrain types.

One interesting point regarding the physiological parameters is that, if the effect of the levels of assistance is obviously referable to of the reduction of the power output request (*Meyer D. et al., 2017*), what is curious to understand is how the motor of the e-bike works.

In fact, as demonstrated by the results of the third part of the study, we found that due to its construction, the motor provides more power at 60 RPMs compared to 90 RPMs in both the levels of assistance investigated. This explain why the physiological parameters at 60 RPMs were lower than those measured at 90 RPM under conditions where the motor was active.

From a practical point of view, cyclists using conventional muscular bicycles, at submaximal intensity, typically select cadences between 80 and 100 RPM, despite the associated increase in VO_2 compared to lower cadences (55–65 RPM), which are known to minimize physiological strain *(Vercruyssen et al., 2008)*. The higher cadences reduce crank forces, net joint moments in the lower limbs, and neuromuscular fatigue.

This indicates that cyclists commonly prefer higher cadences not in spite of the elevated oxygen consumption, but because these cadences offer improved biomechanical efficiency over time, a critical factor in endurance cycling performance *(Vercruyssen et al., 2008)*.

However, in the context of e-bike use, higher pedalling cadences may not represent the most efficient option. This is due to the fact that the power assistance is greater at lower cadences. As a result, lower pedalling rates may be more advantageous during E-bike-assisted cycling. This result is true in the case in the context of our study; however, it should be specified that, although the number of motor manufacturers is limited, it cannot be assumed that all motors operate in the same manner. Indeed, in addition to the power that a motor is capable of delivering, determined by its intrinsic characteristics, it is also

necessary to consider the power actually delivered as regulated by the software, which is modulated on the basis of the power input provided by the user *(Contò C. et al., 2023)*.
Lastly, the freely selected pedalling cadence was found to be 80 rpm across all three assistance levels. However, also if an inter-individual variability was observed, this represents an important factor to consider, as it may suggest the need for motor adaptation to these pedalling cadences, so that the efficiency of both the rider and the motor can be better aligned. The inter-individual variability was observed likely due to the participants' lack of familiarity with using an e-bike, which may have made it difficult for them to accurately identify their optimal cadence.

5.6 CONCLUSION

In the present study we investigated the main physiological parameters, the rate of perceived exertion and the EMG signal while riding an e-bike at fixed external load at three different levels of assistance and two different pedalling cadences, adding the study of the behaviour of the e-bike motor.

First, the assistance influenced all the parameters investigated with lower values as the assistance increased. The physiological parameters were affected by the cadence with higher values as the assistance increased, and, in most cases, there was also an interaction on the values of the RPMs at the different assistance levels.
EMG activity decreased with increasing assistance in upper leg muscles, while lower leg muscles were unaffected by assistance but showed higher activity at higher cadences.
For all the muscles, no interaction between cadence and assistance was found.

To conclude, it is also important to underline the contribution of the e-bike motor to the rider, in fact, varies with the pedalling cadence.
As we found in our study and for our e-bike model, the motor contributed to a higher extent at 60 RPMs compared to 90 RPMs in both levels of assistance investigated, contributing in explain why, in present investigation, the effect of cadence was found only for those conditions where the electric motor was activated. Based on the results of this study, we can conclude that rider physiological effort and the choice of optimal cadence when riding an e-bike may differ from what is already known for conventional bicycles.
Lastly, the freely selected pedalling cadence was found to be higher than the motor's

optimal operating cadence; therefore, the motor could potentially be adjusted to operate at higher cadences in order to improve the efficiency of the human–bicycle system.

6 CHAPTER 6: STUDY TWO

COMPARATIVE ANALYSIS OF VIBRATION EXPOSURE ON E-MOUNTAIN BIKES AND CONVENTIONAL MOUNTAIN BIKES ACROSS VARIED TERRAINS

6.1 ABSTRACT

The increasing popularity of electric mountain bikes has broadened access to off-road cycling, yet their impact on rider vibration exposure remains poorly understood. This study developed and applied a vibration measurement system to record acceleration levels experienced by the bicycle frame and rider. Triaxial accelerometers were mounted on the handlebar, on the top tube beneath the seat, and on the rider at the T4 and L3 spinal levels. The accelerometer data were processed to compute both total acceleration and the low-frequency component (<5 Hz) using Butterworth filtering. The system was used to assess vibration exposure in conventional mountain bike (MTB) and a mountain bike provided with an electric motor (e-bike), across multiple terrains. Eight recreational cyclists performed controlled uphill and downhill rides over asphalt, low-difficulty off-road, and high-difficulty off-road tracks. At the end of each trial, participants were asked to complete three VAS scales assessing their perception of the bicycle-riding experience. The MTB exhibited higher frame vibrations compared to the e-bike, while no significant differences were found between electric assistance levels (HIGH vs. ECO) during uphill riding. These findings suggest that terrain complexity and bike mass distribution have a more pronounced impact on vibration exposure than motor assistance settings. Moreover, the VAS scale results indicate that participants preferred riding the conventional bicycle on downhill segments, whereas no differences were observed on uphill segments. The proposed methodology offers a framework for future studies on rider comfort, equipment optimization, and vibration-related health outcomes in cycling.

Keywords—E-bike, vibration, accelerometer

6.2 INTRODUCTION

The use of road bikes and mountain bikes has always represented a significant component not only in the realm of sports but also in general physical activity. In recent years, the advent of electric bicycles has greatly increased the number of people engaging in these sports for recreational purposes.

The power assistance provided by the electric motor substantially reduces the physical effort required by the rider, allowing even individuals with limited fitness to ride mountain bikes on unpaved and inclined trails at an intensity that is manageable for them *(Hinder J., et al. 2019, Avina-Bravo E. G. et al., 2022)*. However, riding on rough terrain results in the transmission of vibrations from the ground to the bike frame and subsequently to the rider's body higher than when riding road bicycle *(Macdermid P.W., et al. 2015a)*. These vibrations can lead to discomfort, reduced performance *(Titlesatd et al. 2006)*, and potentially higher rate of overuse-related injuries than with road bike *(Visentini P J., et al. 2022)*.

Previous literature has evaluated the magnitude of vibration affecting both the frame and the rider, and has demonstrated its dependence on several factors, such as frame stiffness *(Shah J., et al., 2020)*, wheel diameter, tire volume and inflation pressure *(Mac dermid P.W., et al. 2015b)*, which can influence the transmission of terrain irregularities. Modern bicycles are generally equipped with front suspension, or both front and rear suspension systems (full suspension), which effectively dampen vibrations and reduce the rider's exposure *(Macdermid P.W., et al. 2017)*.

Due to the addition of electric motors and batteries, e-bikes weigh 1.5 to 2 times as much as regular bikes of the same size. To the best of our knowledge, no studies have quantitatively assessed the effects of riding an electric mountain bike compared to a conventional mountain bike.

This work aims to develop an integrated hardware and software system capable of measuring vibrations transmitted to both the bicycle frame and the rider during off-road testing. The system will be used to evaluate vibration characteristics across different bikes, different level of assistance for e-bikes, and riding conditions. Finally, the study also aims to determine participants' preferences for riding an electric or conventional bicycle based on their subjective perceptions.

6.3 SYSTEM DESIGN

Bicycles models and characteristics

To conduct the vibration measurements, a traditional mountain bike and an electric mountain bike were selected, ensuring that the frame size (M), wheel diameter, and tire type were as similar as possible. Both bikes featured 29-inch wheels, aluminum frames, a full suspension setup, with a head tube angle of 66°, and both mounted flat pedals. The two bicycles, however, differed in terms of braking system and front suspension. Specifically, the conventional bike was equipped with Shimano MT400, single piston braking system and a Fox Float 34,140mm front fork, whereas the E-bike featured Sram Level, two piston braking system and a Rockschock Recon Silver, 150mm front fork.

The motor mounted on the e-bike is a Yamaha PW-X3, with the following characteristics: rated voltage: 36 V, rated output: 250 W, weight: 2.75kg, maximal assist speed: 25 km/h, maximal drive torque: 85 Nm. It comes with 5 assist modes: Extra Power (EXPW), High-Performance (HIGH), standard (STD), + Eco (+ECO) and Eco (ECO).

Table 6.1: bicycle characteristics.

	E-bike	**Conventional bike**
Name, brand	XTF 1.5 All Track, Fantic	Orbea, Occam
Year	2023	2021
Weigth	24.9 kg	13.8 kg
Braking system	Sram Level, two-piston	Shimano MT400, single piston
Drive train	12-speed Sram SX	12-speed Shimano Deore
Front shock	Rockschock Recon Silver, 150mm travel	Fox Float 34, 140 mm travel
Rear shock	Rockshock Deluxe Select, 130mm travel	Rockshock Deluxe, 130mm travel

Vibration measuring system

To measure the accelerations experienced by both the bicycle and the rider, four X16-1D devices (Gulf Coast Data Concepts, Waveland, USA) were employed. These devices are equipped with a triaxial 16g accelerometer (Analog Devices ADXL345) and were configured with a sampling rate of 200 Hz. Data was stored internally. All four devices were synchronized with each other according to the procedure described in the manufacturer's manual. The collected data were subsequently processed using Matlab (MATLAB R2020b, The MathWorks, Natick, MA, USA) to extract the relevant time windows.

Two accelerometers were placed on the bike frame, one at the center of the handlebar and the other on the top tube close to the intersection with the seat tube (TS tube). Two additional accelerometers were attached to the rider's body, firmly secured with medical tape to the skin over the thorax vertebra (T4) and the lumbar region, above the L3 vertebra (Fig. 6.1)

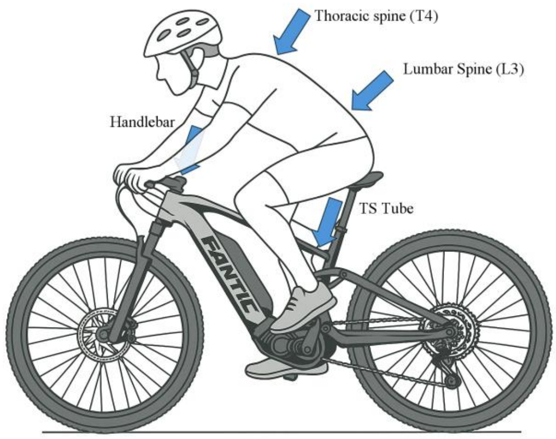


Fig. 6.1: Diagram of IMU placement on the rider's body and the bike frame; the bicycle shown in the example is the e-bike.

6.4 SUBJECTS

Eight recreational bikers (3 Female and 5 Male) (mean ± SD: age 26±3.8 years, height 176.6 ± 6.5 cm, body mass 72.4 ± 4.8 kg) with experience of riding on off-road terrain voluntary participated in the measurement’s session. All participants provided written informed consent in accordance with the Declaration of Helsinki.

6.5 FIELD TEST TRIALS

For the experimental measurements, three types of 100-meter-long test tracks were identified, each with distinct characteristics: (a) off-road terrain with high technical complexity (OFFR_H), featuring significant irregularities and an average slope of 16%; (b) off-road terrain with low technical complexity (OFFR_L), consisting of mild bumps and minor surface unevenness, with a 5% slope; and (c) paved asphalt surface (ASP), with an average slope 4% *(Fig. 5.8 a, b, c)*. The segment length was determined using a GPS device (Garmin Forerunner 255), identifying 100-meter tracts that were marked by cones. The measurements were cross-checked several times.

Both bicycles were set up with identical tire pressure, activated front and rear shock absorbers on the off-road segments, while for the ASP segment they were deactivated, and the saddle height were adjusted according to each participant's preference.

Participants took part in a single day testing session during which they rode both bicycles across all three types of terrain, in both uphill and downhill conditions. For e-bike, uphill conditions were repeated by selecting HIGH and ECO mode while downhill sections were completed with the motor turned OFF.

The conditions were administered in a randomized order to control for potential order effects on the dependent variables. The subjects were instructed to start at locations marked by cones and to ascend or descend the path in the shortest time possible. They were asked to remain seated on the saddle on the OFFR_L and ASP, and allowed a free guide position on the OFFR_H track.

For each tested combination of terrain and bicycle, subjective evaluation data were collected, specifically targeting aspects of riding performance with a main focus on the differences between e-bikes and traditional bicycles in downhill trials and between the lowest and the highest assistance in up-hill sections, Participants were asked to complete a Visual Analogue Scale immediately after the trials (VAS) (Fig. 6.3), which included the following items: (VAS_1) overall level of enjoyment, (VAS_2) how was the bike's handling? (VAS_3) how much did the bicycle allow you to use your riding skills?

a b c

Fig. 6.2 a, b, c: Images of the three ground surfaces on which the tests were performed. From the left: (a) off-road terrain with high technical complexity (OFFR_H), (b) off-road terrain with low technical complexity (OFFR_L), (c) paved asphalt surface (ASP).

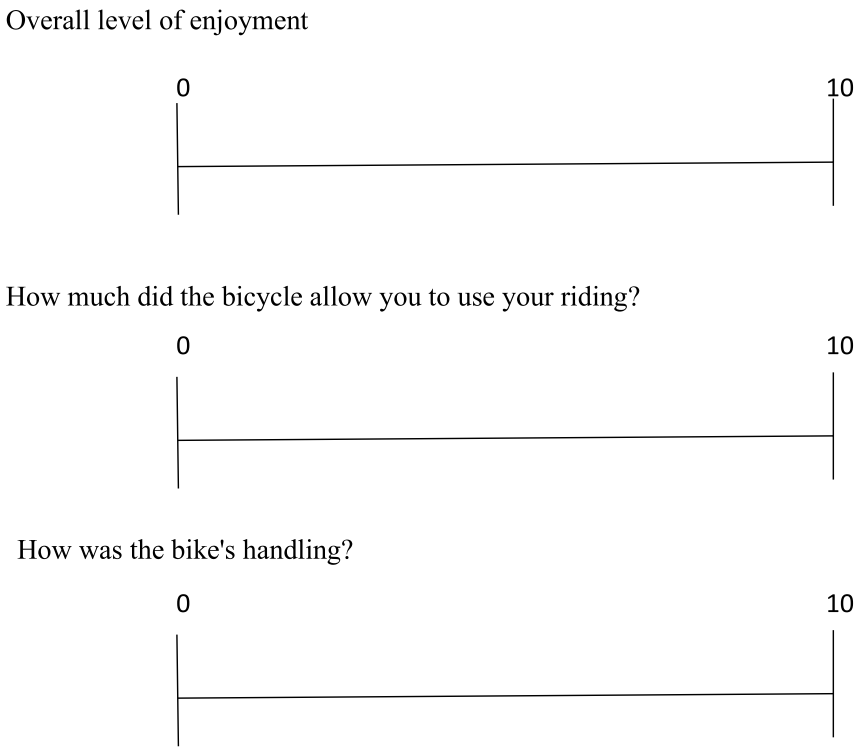


Fig.6.3: illustration of the Visual Analogue Scales (VAS) presented at the end of the trials.

6.6 DATA PROCESSING ANALYSIS

Amplitude of vibrations

All data were analyzed for total acceleration by computing the resultant of the accelerations measured along the x, y, and z axes of the devices. Prior research [9] has indicated that assessing total acceleration is effective for evaluating human vibration exposure, as soft tissue oscillations typically occur across all three orthogonal axes in studies of this nature.

To eliminate very low-frequency motion, a high-pass Butterworth filter (6th order, cutoff frequency: 0.5 Hz, zero-lag) was applied to the resultant acceleration. This filtering step also removed the gravitational acceleration component. To distinguish effect due to pedaling movement and low frequency movements of the rider during riding from high frequency vibration-related components, the resultant acceleration signal was decomposed into two separate signals: a low-frequency component (LF, <5 Hz) representing voluntary motion, and a high-frequency component (HF, >5 Hz) associated with vibrations. This approach is based on methodologies previously adopted in related research *(Macdermid et al., 2014)*,. The extraction of LF and HF components was done by applying respectively a low- and high-pass Butterworth filter (6th order, cutoff frequency: 5 Hz, zero-lag). The magnitude of vibration was then computed for both LF and HF and total vibrations (TOT) as RMS over the entire duration of each trial.

Data processing was performed using a custom-written MATLAB routine (MATLAB R2020b, The Math Works, Natick, MA, USA).

Speed over the tracts was calculated using the distance and time manually taken.

VAS scale score

For the VAS scales, the mean score was calculated for each type of trial and for each type of bicycle

Statistical analysis

The effects of bike type and terrain were tested using a two-way repeated measures ANOVA. For uphill riding, two terrain conditions (ASP and OFFR_L) and two bike types (e-bike_HIGH, e-bike_ECO) were included in the model. For downhill riding, three terrain conditions (paved, OFFR_L, and Tech) and two bike types (MTB and e-bike) were analyzed. Pair comparisons were tested in post-hoc analysis by applying Bonferroni correction for multiple comparison. Data for each condition are presented as mean and standard deviation across subjects. The significance level was set at p = 0.05.

6.7 EXPERIMENTAL RESULTS

6.7.1 Downhill riding

Acceleration

The type of the bike significantly affected speed (p= 0.015, F= 10.5) (Fig. 6.4), with lower speeds observed for e-bike. Terrain had a significant effect on speed (p = 0.001, F=15.5) where OFFR_H terrain was associated with lower speeds when compared to ASP and OFFR_L. No significant bike*terrain interaction was detected.

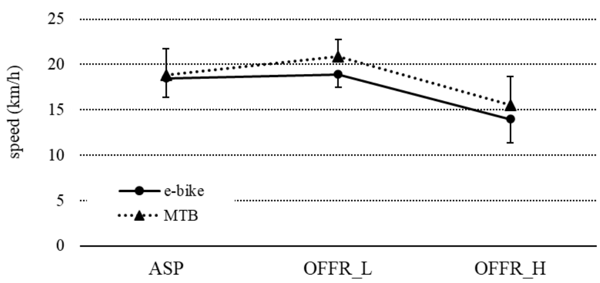


Fig. 6.4: Speed data during downhill riding. Values are reported as mean and standard deviation.

While riding downhill, the total acceleration was affected by the terrain for both body (T4 p<0.01, F= 24, L3, p<0.05, F= 7.7) and bike frame (handlebar p<0.01, F= 52.7, TS-Tube, p<0.05, F= 10.9), with increase in acceleration as the difficulties on the terrain increase. Post hoc analysis indicated significant difference between ASP and OFFR_L for T4 and handlebar, and between OFFR_L and OFFR_H for L3, handlebar and TS-Tube.

A significant effect of the bike was seen for bike frame acceleration (handlebar: p=0.007, F= 16; TS-Tube: p=0.03, F= 8;) with larger acceleration for MTB bike. A bike x terrain interaction was seen for TS-Tube, with differences in acceleration between the two bikes that become larger for the most technical tract.

Similar results were found when analyzing low frequency acceleration, (Fig. 4) with significant effect of terrain for all the measured locations (T4: p=0.005, F=24.01; L3: p=0.019, F= 24.5; handlebar: p=0.007, F=52.68; TS-Tube: p=0.030, F=10.92), with higher values for OFFR_H. An effect of the bike was found only for acceleration measured on bike frame (handlebar: p= 0.027, F=16.11; TS-Tube: p = 0.023, F=8.02), with higher values for MTB than for e-bike *(Figure 6.5)*.

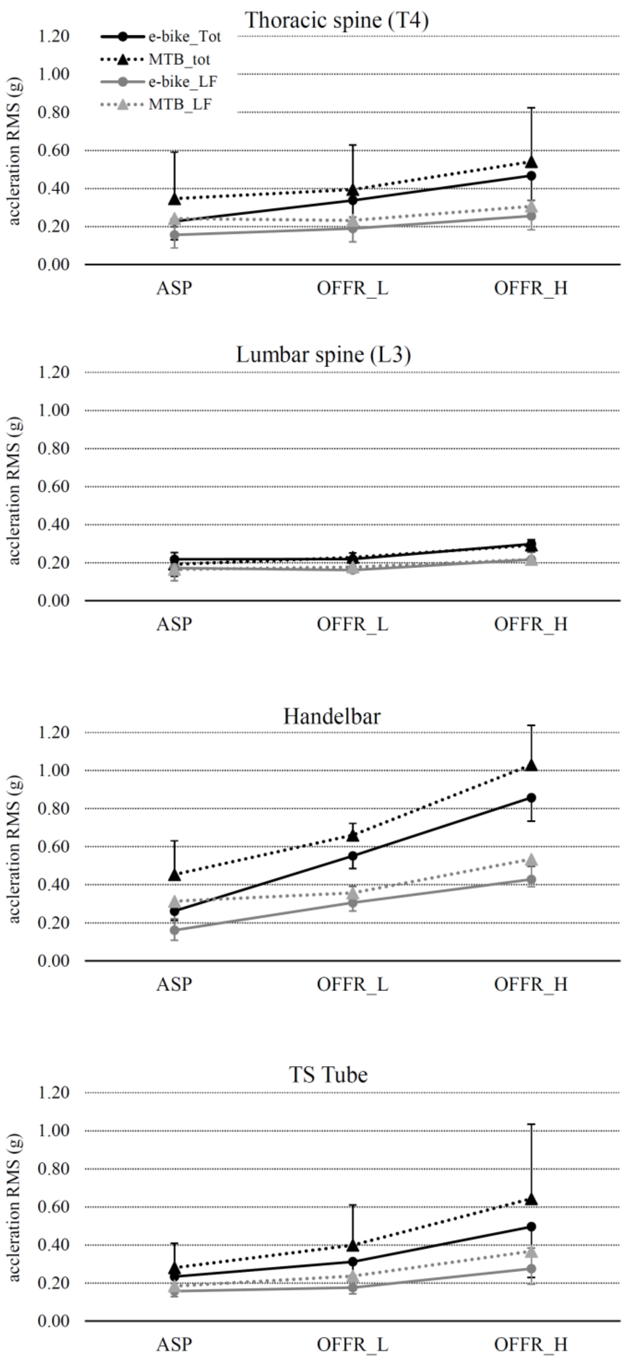


Fig. 6.5: Acceleration during downhill riding. Values are reported as mean and standerd daviation.

VAS scale

In the downhill section, for VAS_1, an effect was found both regarding the type of bicycle (P = 0.032, F=8.7) and the different terrain conditions (P = 0.004, F= 17.83). Specifically, the conventional (muscle-powered) bicycle showed higher values compared to the electric bicycle (P = 0.032); the technical terrain (DT in the graph) showed higher values than the asphalt trial (P = 0.004) and was very close to a significant difference when compared with the less technical off-road terrain (D_NT in the graph) (P = 0.051). However, no interaction effect between the two factors was found.

For VAS_2 and VAS_3, a significant effect of bicycle type was observed, with P = 0.003, F= 11.75 and P = 0.001, F=27.27, respectively, with higher values again for the conventional bicycle. No significant differences were found for terrain type or for the interaction between the two factors.

These results lead us to conclude that participants prefer using the conventional (muscle-powered) bicycle over the electric one.

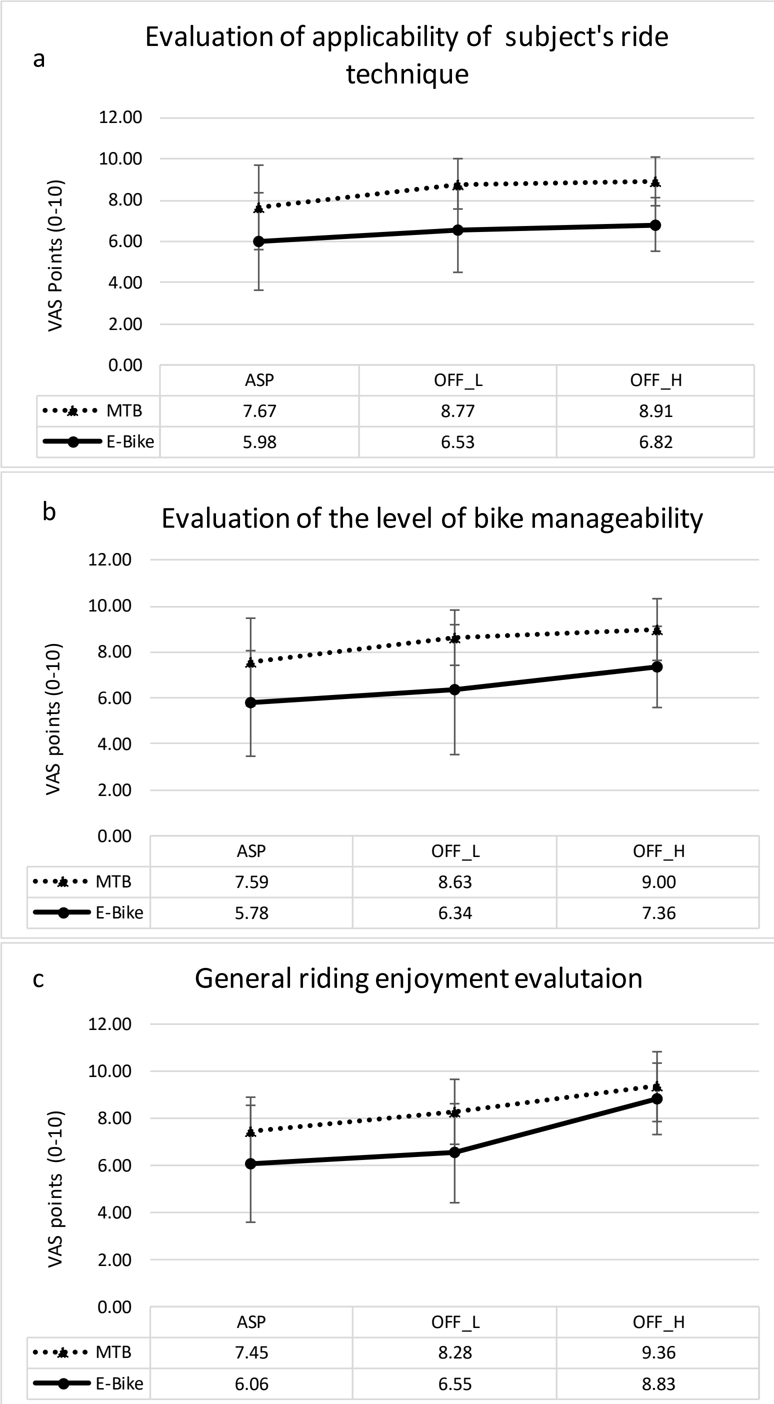


Fig.:6.6: VAS scale results: a: VAS scale n° 1; b: VAS scale n° 2; c: VAS scale n° 3.

6.7.2 Uphill riding

Acceleration

The speed (Fig. 6.7) was significantly higher for e-bike_HIGH than for ECO mode, (P=0.007, F= 6.4) with a significant bike*terrain interaction. Post hoc analysis found that only on ASP terrain, riding in HIGH mode led to greater speed.

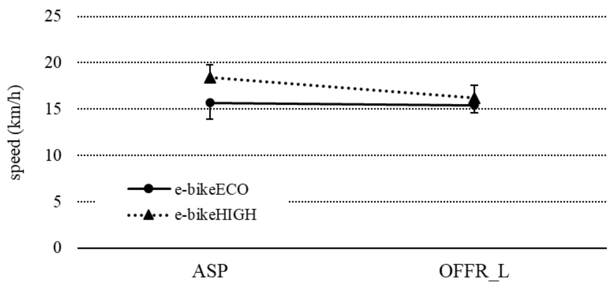


Fig. 6.7: Speed data during uphill riding. Values are reported as mean and standard deviation.

The total acceleration when riding uphill showed a significant effect of terrain on the thoracic vertebra (T4; P=0.021, 5.2) and on the bike frame (handlebar: P<0.001, F=4.9; TS-Tube: p=0.012, F= 4.6). When analyzing low frequency acceleration, a significant effect of terrain was observed only on the handlebar. No significant effect of the bike was found in either total acceleration or low frequency acceleration when comparing ECO and HIGH assistance modes *(Figure 6.8)*.

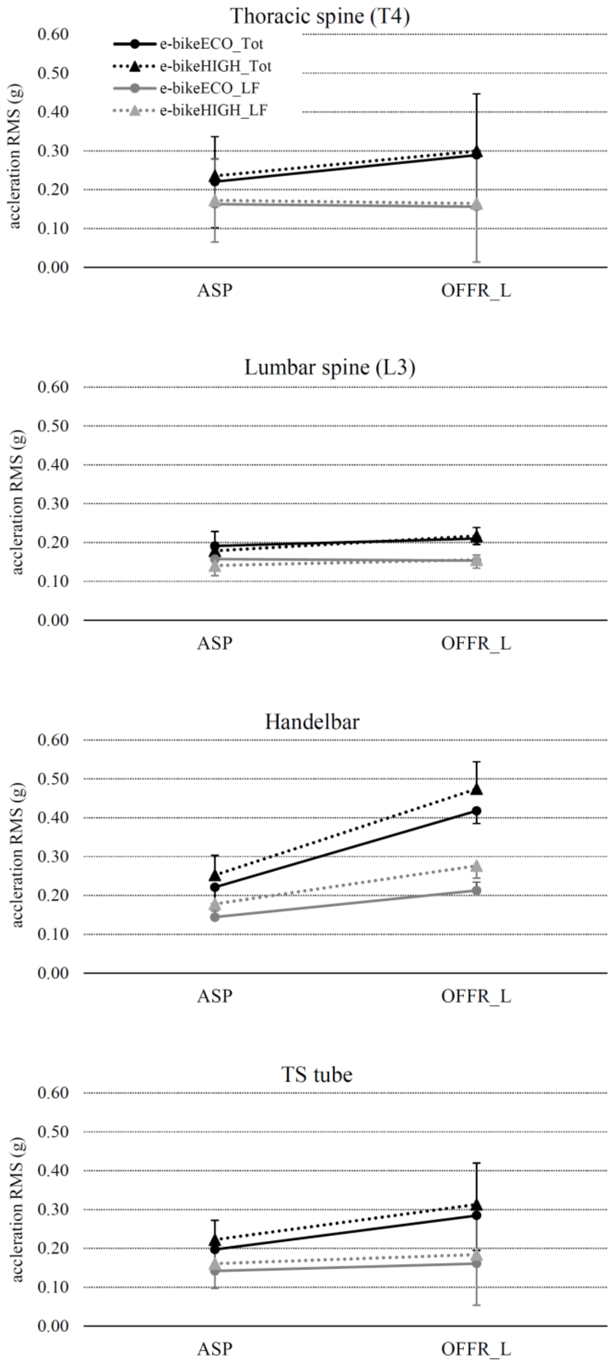


Fig. 6.8: Acceleration during uphill riding. Values are rported as mean and standard deviation.

VAS scale

Concerning the VAS scale results, in this case, neither of the two variables analysed showed a significant effect, and consequently, neither did their interaction (Fig. 6.9).

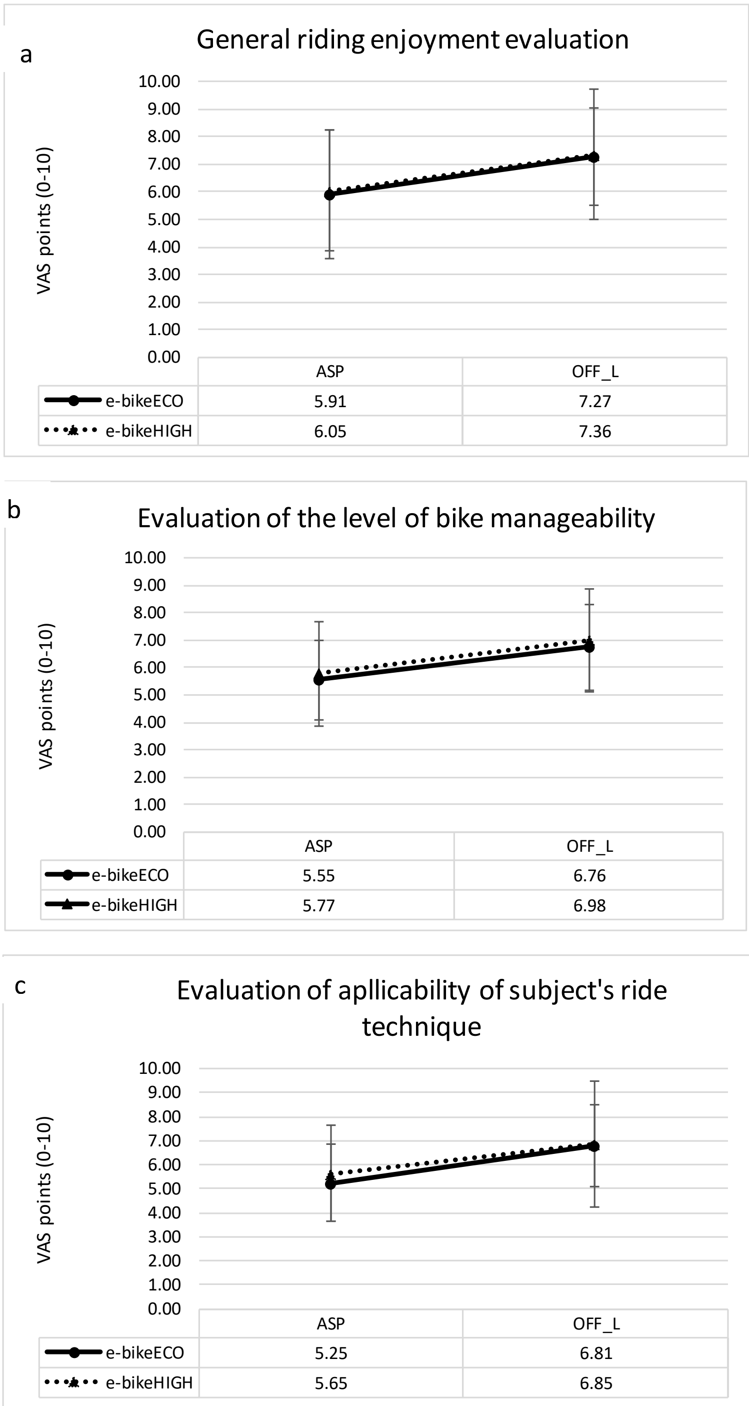


Fig.:6.6: VAS scale results: a: VAS scale n° 1; b: VAS scale n° 2; c: VAS scale n° 3.

6.8 DISCUSSION AND CONCLUSION

Measurements conducted using triaxial accelerometers allowed for the measurement and description of motion and vibration of the body and bike frame when riding a bike and a e-bike with varying assistance levels across different terrains.

The measured vibrations were generally higher at the bike frame than at the body locations. Within the frame, vibrations were higher at the handlebar than at the part of top tube located under the seat, while for the body, they were higher at the thoracic spine compared to the lumbar region. A possible explanation for the higher vibration recorded at the thoracic spine compared to the lumbar spine is that acceleration at the lumbar spine is primarily transmitted through the saddle, whereas thoracic spine vibration results from transmission through both the saddle and the handlebar, the latter exhibiting the highest vibration levels.

Furthermore, the damping capacity of soft body tissues may be greater in the region between the saddle and the spine, potentially influenced by the cushioning of the saddle itself. This may have contributed more to vibration attenuation than the damping provided by the hands on the handlebar. Additionally, the stiffness of the arms and legs, resulting from muscle tension, which was not evaluated in this study, could also play a role in dampening or enhancing vibrations transmitted from the bike frame to the body.

Off road terrain appears to induce higher accelerations compared to ASP terrain, mainly as far as the parameters accounting for total acceleration during both uphill and downhill riding. This trend was observed across all measured locations except for the lumbar site, where no significant terrain-related differences were detected during either uphill or downhill riding. These results are consistent with those of a previous investigation *(Macdermid PW., 2015)* where terrain was found to affect vibrations at the handlebar and TS-Tube, but not at the lower back.

Terrains more technical than easy off-road track (OFFR_L), tested exclusively during downhill riding (OFFR_H), elicited even greater total acceleration values, particularly at the lumbar spine, handlebar, and TS-Tube.

When focusing specifically on the slow component of acceleration, only the handlebar exhibited significantly higher values on low technical off-road terrain compared to ASP terrain during uphill riding, and further increases were noted on highly technical terrain relative to off-road during downhill. The observed enhancement of the low-frequency component may be attributed not only to the terrain characteristics, but also to the rider's decision to ride out of the saddle during the OFFR_H tract.

These findings suggest that off-road terrain predominantly influences the vibration component of acceleration, with minimal impact on the motion associated with active riding or driving. The only location where terrain type appeared to affect the slow-motion component of acceleration was the handlebar.

Different assistance levels, tested only during uphill riding, did not appear to be a factor influencing vibration, either for total acceleration or low-frequency acceleration. Slightly higher values were found for the bike frame when using HIGH mode with respect to ECO, but these differences were not statistically significant. Based on the current data, it therefore seems reasonable to say that the power delivered by the electric motor, which reduced the power output required from the rider, did not lead to changes in riding style or in the vibrations experienced.

In the comparison between the conventional mountain bike and the mountain bike provided with an electric motor, which was conducted on downhill sections, higher vibration levels were observed for the conventional MTB at the measurement sites on the bike frame. These effects were evident and consistent for both total acceleration and the low-frequency component. During the downhill assessment, the motor was turned off. To explain this finding is important to take into consideration. To explain these results, it is necessary to consider the main factors that differentiate the two bicycles. First, the electric bicycle has a higher overall weight and a lower centre of mass, which contributes to increased stability during riding. Second, the bicycles differ in terms of the front suspension: the electric bicycle is equipped with a fork offering an additional 10 mm of travel, which enhances the absorption of larger impacts but does not substantially influence low-magnitude vibrations. Moreover, the fork used on the electric bicycle is generally effective at damping medium-size impacts, whereas it is less capable of attenuating fine, high-frequency vibrations. In any case, although there are differences between the two bicycles, weight is the most macroscopically evident characteristic; it is therefore hypothesized that this factor is the primary driver of the observed differences.

Regarding the VAS scales in the comparison between the electric and conventional bicycles, all three perceptions showed higher values for the conventional bicycle, indicating that the average user still prefers the manoeuvrability offered by this type of bicycle. In contrast, with regard to the comparison among the electric bicycle’s assistance levels, no differences were observed in the perception of ride quality.

7 CHAPTER 7: CONCLUSIONS

7.1 HOLISTIC UNDERSTANDING OF RIDER-BIKE INTERACTION

This doctoral project is structured around two studies that may appear distinct, yet share the common objective of investigating the key aspects related to e-bike use, specifically focusing on models designed for both on-road and off-road applications. Adopting this broad and integrated perspective makes it possible to construct a comprehensive understanding of the various dimensions that characterize the interaction between the rider and the vehicle. The multi-parameter approach employed in this work enables the generation of robust and application-oriented scientific evidence, providing the partner company with technical insights and objective data to support the design and development of e-bikes that are increasingly aligned with the needs and expectations of end users *(Fig. 7.1)*.

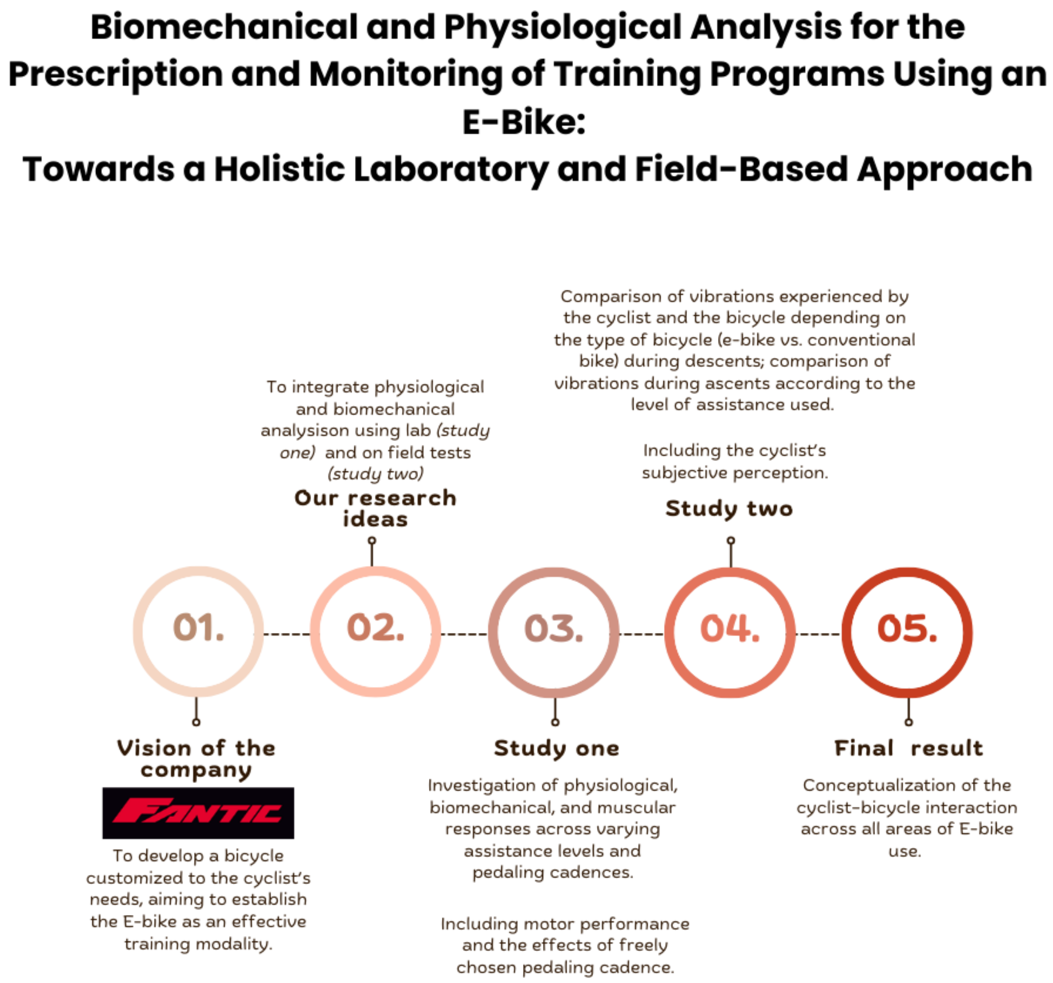


Fig. 7.1: summary of applied research project

7.2 PRACTICAL APPLICATION OF DOCTORAL PROJECT

Regarding the indoor study, an important factor that emerged concerns effect of the behaviour of the motor on the metabolic and muscular responses. Specifically, it was found that at all three levels of assistance tested, the motor works by delivering more power at 60 RPM compared to higher cadences. Consequently, this factor influences the rider's physiological workload. Another relevant aspect examined in this study concerns the freely chosen pedalling frequency adopted by the participants. As widely reported in the literature, this parameter represents the cadence that optimally balances muscular fatigue and energy expenditure, and thus serves as a key indicator of pedalling efficiency. In light of the results obtained, which indicate a stable preference around 80 rpm, it may be particularly valuable to suggest that the partner company consider developing a motor system optimized to deliver its highest performance at this specific pedalling frequency. Such a design integration would align motor operation with the cyclist's natural biomechanical preferences, thereby enhancing comfort, efficiency, and the overall quality of the riding experience.

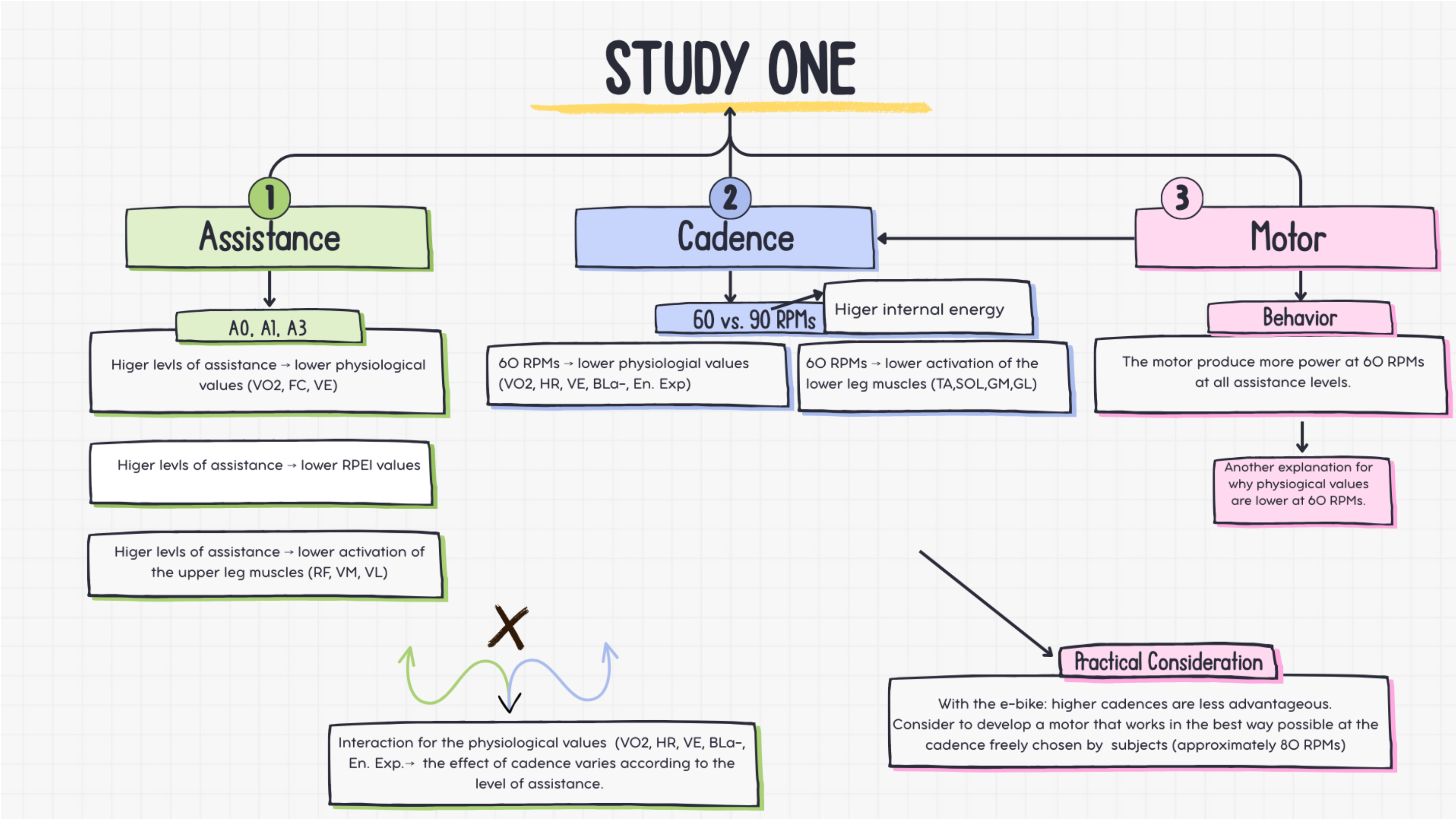


Fig: 7.2: summary outline of study one

With regard to the field study, the outcomes obtained provide meaningful and actionable insights for the partner company, particularly concerning the accelerations recorded on both the bicycle and the rider across a range of real-world riding conditions. The analyses revealed that, during downhill riding and across different terrain types, the electric bicycle consistently exhibited lower acceleration values compared to the conventional bicycle.

Furthermore, the investigation of uphill riding conditions showed that varying levels of pedal assistance did not produce detectable differences in acceleration patterns. This indicates that the motor assistance primarily affects the rider's required mechanical input rather than altering the dynamic behaviour of the bicycle in terms of accelerations. Such a result provides valuable information for understanding how assistance modes interact with the mechanical characteristics of the bike and the rider's effort.

Finally, subjective ride perception was evaluated using visual analogue scales. Participants reported a clear preference for the conventional bicycle during downhill sections, suggesting that traditional bikes are still perceived as more intuitive or predictable in terms of manoeuvrability and handling when descending.

Recommendations for the company

Study one:

- it is recommended to develop an appropriate motor mapping, or to modify the software regulating the motor's power output characteristics, so that the motor becomes more efficient at cadences close to 80 rpm (which we observed to be the cadence spontaneously selected by the participants), in order to optimize motor performance at the preferred comfort cadence.

Study two:

- the observed reduction in vibrations when using the electric bicycle suggests, in line with the literature, a decrease in non-propulsive work performed by the rider. This represents a positive implication for the company, which may consider promoting the use of e-bikes among less trained populations on riding down-hill off-rod segments, as well as for rehabilitation purposes following injury.

- The finding that, according to the participants' perceptions, the electric bicycle is associated with lower overall enjoyment, reduced handling, and a diminished ability to apply one's riding technique suggests that the company should optimize handling characteristics or frame geometry in order to improve ride quality.

7.3 OVERALL LIMITATION

This study is subject to certain limitations, both with respect to the laboratory-based study and the field-based study.

In the first study, the findings of the study cannot be extrapolated to electric bicycles in general; it would therefore be inappropriate to recommend that users adopt low pedalling cadences on the basis of their advantage over higher cadences. The study was, in fact, conducted exclusively with reference to the performance of the motor installed on the bicycle supplied by the collaborating company.

In the second study, the bicycles used for the comparison between electric and conventional bicycles were similar, but not identical. Moreover, although the path lengths were identical and, according to GPS accuracy, the driving tracks appeared the same, it is possible that, in practice, participants followed slightly different trajectories along the route. For future studies, a visual observation could be proposed to better identify differences in track traversal, and subsequently compare the same trajectories.

8 REFERENCE

Alessio HM, Reiman T, Kemper B, von Carlowitz W, Bailer AJ,Timmerman KL. Metabolic and Cardiovascular Responses to a Simulated Commute on an E-Bike. Translational Journal of the American College of Sports Medicine. 2021;6:(2):e000155. doi:10.1249/tjx.00000 00000 000155

Ansley L., Cangley Patrik. Determinants of "optimal" cadence during cycling. Europen Journal of Sport Science, 2009; 61-85. https://doi.org/10.1080/17461390802684325

Artioli A., 2019–2020, Bachelor's thesis: Pedaling cadence in cycling: strategies to improve performance.

Avina-Bravo E. G., Cassirame J., Escriba C., Acco P., Fourniols J.Y., Soto-Romeo G. Smart electrically assisted bicycles as health monitoring systems: a review. Sensors, 2022; 8;22(2):468.

Avina-Bravo E. G., Cassirame J., Escriba C., Acco P., Fourniols J.Y., Soto-Romeo G. Smart electrically assisted bicycles as health monitoring systems: a review. Sensors, 2022; 8;22(2):468.

Biomechanics of cycling – book p.13-70

Bersten S., Malnes L., Langåker A., Bere E. Physical activity when riding an electric assisted bicycle. International Journal of Behavioral Nutrition and Physical Activity, 2017; 14:55.

Bini RR, Dagnese F (2012). Noncircular chainrings and pedal to crank interface in cycling: a literature review. Brazilian Journal of Kinantropometry and Human Performance 14 (4):470-482. http://dx.doi.org/10.5007/1980-0037.2012v14n4p470

Brand C., Goodman A., Rutter H., Song Y., Ogilvie D. Association of individual, household and environmental characteristics with carbon dioxide emissions from motorised passenger travel. Applied Energy, 2013; 104(100):158-169. https://doi.org/10.1016/j.apenergy.2012.11.001

Brisswalter, J., Hausswirth, C., Smith, D., Vercruyssen, F., & Vallier, J. M. (2000). Energetically optimal cadence vs. freely-chosen cadence during cycling: Effect of exercise duration. International Journal of Sports Medicine, 21, 60-64

Broune J.E., Sauchelli S., Perry R., Page A., Leary S., England C., Cooper A. R. Health benefits of electrically assisted cycling: a systematic review. International Journal of Behavioral Nutrition and Physical Activity, 2018; 21;15(1):116.

Broune J.E., Cooper A. R., Kelly P., Kinnear F.J., England C., Leary S., Page A. The impact of e-cycling on travel behaviour: a scoping review. Journal of Transport & Health,2020; 19:100910. https://doi.org/10.1016/j.jth.2020.100910

Candotti CT, Ribeiro J, Soares DP, De Oliveira AR, Loss JF, Guimarães ACS (2007) Effective force and economy of triathletes and cyclists. Sports Biomechanics 6 (1):31-43

Coast, J. R., & Welch, H. G. (1985). Linear increase in optimal pedal rate with increased power output in cycle ergometry. European Journal of Applied Physiology and Occupational Physiology, *53*, 339-342

Contò, C., & Bianchi, N. (2023). E-Bike Motor Drive: A Review of Configurations and Capabilities. In *Energies* (Vol. 16, Issue 1). MDPI. https://doi.org/10.3390/en16010160.

De Groot, G., Sargeant, A., & Geysel, J. (1995). Air friction and rolling resistance during cycling. Medicine and Science in Sports and Exercise, 27, 1090-1095.

Diefenthaeler F, Bini RR, Carpes FP, Vaz MA (2007) Analysis of pedalling technique during a maximal cycling exercise. Paper presented at the 25 International Symposium on Biomechanics in Sports, Ouro Preto, Brazil

Diefenthaeler F, Bini RR, Laitano O, Guimarães ACS, Nabinger E, Carpes FP,Bolli C (2006) Assessment of the effects of saddle position on cyclists pedalingtechnique. Medicine & Science in Sports & Exercise 38 (5):S181

Di Prampero P.E. The energy cost of human locomotion on land and in water. International Journal of Sports Medicine, 1986; 7:55-72

Dorel S, Drouet JM, Couturier A, Champoux Y, Hug F (2009) Changes of pedaling technique and muscle coordination during an exhaustive exercise. Medicine and Science in Sports and Exercise 41 (6):1277-1286. doi:10.1249/MSS.0b013e31819825f8

Dorel S, Couturier A, Hug F (2009) Influence of different racing positions on mechanical and electromyographic patterns during pedalling. Scandinavian Journal of Medicine and Science in Sports 19 (1):44-54

Ericson M. On the biomechanics of cycling: a study of joint and muscle load during exercise on the bicycle ergometer. Scandinavian Journal of Rehabilitation. Medicine, 1986: 16(Suppl.): 1-43

Ettema, G., Lorås, H., & Leirdal, S. (2009). The effects of cycling cadence on the phases of joint power, crank power, force and force effectiveness. *Journal of Electromyography and Kinesiology*, *19*(2).

Fonda, B., & Sarabon, N. (2012). Biomechanics of Cycling. *Sport Science Review*, *19*(1–2). https://doi.org/10.2478/v10237-011-0012-0

Francescato, M., Girardis, M., di Prampero, P., Francescato, M. I., & -M Girardis di Prampero, N. P. (1995). Oxygen cost of internal work during cycling for a. In Eur J Appl Physiol (Vol. 72).

Goodman A., Sahlqvist S., Ogilvie D. Who uses new walking and cycling infrastructure and how? Longitudinal results from the UK iconnect study. Preventive Medicine Journal,2013; 57(5):518-24. https://doi.org/10.1016/j.ypmed.2013.07.007

Gojanovic B, Welker J, Iglesias K, Daucourt C, Gremion G. Electric bicycles as a new active transportation modality to promote health. Med Sci Sports Exerc. 2011;43(11):2204-2210.

Hall C., Hot T. H., Julian C., Wright G., Chaney A. R., Crookston B., West J. Pedal-assist mountain bikes: a pilot study comparison of the exercise response, perceptions and beliefs of experienced mountain bikes. JMRI Formative Research, 2019; 13;3(3): e 13643.

Hang R. D., Weis S. E., Raven P. B. Effect of pedal rate on cardiorespiratory responses during continuous exercise. Medicine and Science in Sports and Exercise, 1992; 0195-91.

Hansen D, Soors A, Deluyker V, Frederix I, Dendale P. Electrical support during outdoor cycling in patients with coronary artery disease: impact on exercise intensity, volume and perception of effort. Acta Cardiol. 2018;73(4):343-350.

Hinder J., Jäger M., Current trends and injuries in cycling: faster, further, e-bike? Der Orthopäde, 2019; 1019-1029. https://doi.org/10.1007/s00132-019-03824-4

Hoare, E., Stavreski, B., Jennings, G. L., & Kingwell, B. A. (2017). Exploring motivation and barriers to physical activity among active and inactive australian adults. *Sports*, *5*(3). https://doi.org/10.3390/sports5030047

Houtz, S.J., Fischer F. J. *An analysis of muscle action and joint excursion during exercise on a stationary bicycle.* The Journal of Bone and Joint Surgery, 1959. American Volume, 41-A (1), 123-131.

Hug, F., & Dorel, S. (2009). Electromyographic analysis of pedaling: A review. In *Journal of Electromyography and Kinesiology* (Vol. 19, Issue 2, pp. 182–198). https://doi.org/10.1016/j.jelekin.2007.10.010

Hug, F., Bendahan, D., le Fur, Y., Cozzone, P. J., & Grélot, L. (2004). Heterogeneity of muscle recruitment pattern during pedaling in professional road cyclists: A magnetic resonance imaging and electromyography study. *European Journal of Applied Physiology*, *92*(3), 334–342. https://doi.org/10.1007/s00421-004-1096-3

Johnson, N., Fitch-Polse, D. T., & Handy, S. L. (2023). Impacts of e-bike ownership on travel behavior: Evidence from three northern California rebate programs. *Transport Policy*, *140*, 163–174. https://doi.org/10.1016/j.tranpol.2023.06.014

Jorge M, Hull ML. Analysis of EMG measurement during bicycle pedalling. J Biomech 1986; 19:683-94

Kautz S, Feltner ME, Coyle EF, Bailey MP (1991) The pedaling technique of elite endurance cyclists: changes with increasing workload at constant cadence. International Journal of Sports Biomechanics 7:29 doi:http://journals.humankinetics.com/jab

Kautz, S. A., & Hull, M. L. (1993). A theoretical basis for interpreting the force applied to the pedal in cycling. Journal of Biomechanics, *26*, 155-165.

Kirkwood, L. A., Ingram, L. A., Cunningham, J., Malone, E., & Florida-James, G. D. (2017). Physiological characteristics and performance in elite enduro mountain biking. *Journal of Science and Cycling*, *6*(2), 13–21. https://doi.org/10.28985/171231.jsc.09

Langford, B. C., Cherry, C. R., Bassett, D. R., Fitzhugh, E. C., & Dhakal, N. (2017). Comparing physical activity of pedal-assist electric bikes with walking and conventional bicycles. *Journal of Transport and Health*, *6*, 463–473. https://doi.org/10.1016/j.jth.2017.06.002

Laursen PB, Shing CM, Tennant SC, Prentice CM, Jenkins DG (2003) A comparison of the cycling performance of cyclists and triathletes. Journal of Sports Sciences 21 (5):411-418

Lewis, S. F., Snell P. G., Taylor W.F., Hamara M., Graham R. M., Pettinger W. A., Blomqvist C. G., The role of muscle mass and mode of contraction in circulatory responses to exercise. Journal of Applied Physiology, 1985; 58(1):146-51. https://doi.org/10.1152/jappl.1985.58.1.146

Louis J., Brisswalter J., Morio C., Barla C., Temprado J-J. The electrically assisted bicycle: an alternative to promote physical activity. Am J Phys Med Rehabil.2012;91(11):931-940.

Lucia A., Hoyos J., Chicharro J. L. Preferred pedalling cadence in professional cycling. Medicine and Science in Sports and Exercise, 2001; 33, 1361-1366.

Lucia, A., San Juan, A. F., Montilla, M., Cañete, S., Santalla, A., Earnest, C., & Pérez, M. (2004). In professional road cyclists, low pedaling cadences are less efficient. Medicine and Science in Sports and Exercise, *36*(6), 1048–1054. https://doi.org/10.1249/01.MSS.0000128249.10305.8A

MacDonald J. H., Farina D., Marcora S.M. Response of electromyographic variables during incremental and fatiguing cycling. Medicine and Science in sports and exercise,2008.

Macdermid, P. W., Fink, P. W., & Stannard, S. R. (2015a). The Effects of Vibrations Experienced during Road vsOff-road Cycling. International Journal of Sports Medicine, 36(10), 783–788. https://doi.org/10.1055/s-0034-1398534

Macdermid PW., Fink P W., Miller M c., Stannard S.The impact of uphill cycling and bicycle suspension on downhill performance during cross-country mountain biking. Journal of Sport Science; 2017. 1355-1363.

Macdermid, Paul & Miller, Matthew & Macdermid, Fiona & Fink, Philip. (2015b). Tyre Volume and Pressure Effects on Impact Attenuation during Mountain Bike Riding. Shock and Vibration, 2015. 10. 10.1155/2015/191075.

Marsh A. P. & Martin P.E. Effect of cycling experience, aerobic power, and power output on preferred and most economical cycling cadences. Medicine and Science in Sports and Exercise, 1998; 30, 942-948. https://doi.org/10.1097/00005768-199709000-00016

Marsh, A. P., Martin, P. E., & Sanderson, D. J. (2000). Is a joint moment-based cost function associated with preferred cycling cadence? In *Journal of Biomechanics* (Vol. 33).

Mayer D., Senner V. Evaluating a heart rate regulation system for human-electric hybrid vehicles. Journal of Sports Engineering and Technology, 2018; pages 102-101. https://doi.org/10.1177/1754337117710069

McVicar, J., Keske, M. A., Daryabeygi-Khotbehsara, R., Betik, A. C., Parker, L., & Maddison, R. (2022). Systematic review and meta-analysis evaluating the effects electric bikes have on physiological parameters. In Scandinavian Journal of Medicine and Science in Sports (Vol. 32, Issue 7, pp. 1076–1088). John Wiley and Sons Inc. https://doi.org/10.1111/sms.14155

Minetti, A. E., Pinkerton, J., & Zamparo, P. (2001). From bipedalism to bicyclism: Evolution in energetics and biomechanics of historic bicycles. Proceedings of the Royal Society B: Biological Sciences, *268*(1474), 1351–1360. https://doi.org/10.1098/rspb.2001.1662

Minetti, A. E. (2011). Bioenergetics and biomechanics of cycling: The role of "internal work." European Journal of Applied Physiology, 111(3), 323–329. https://doi.org/10.1007/s00421-010-1434-6

Mitchell, R. A., Boyle, K. G., Ramsook, A. H., Puyat, J. H., Henderson, W. R., Koehle, M. S., & Guenette, J. A. (2019). The Impact of Cycling Cadence on Respiratory and Hemodynamic Responses to Exercise. Medicine and Science in Sports and Exercise, 51(8), 1727–1735. https://doi.org/10.1249/MSS.0000000000001960

Neptune, R. R., Kautzt, S. A., & Hull, M. L. (1997). The effect of pedaling rate on coordination in cycling. In *l Bmnechanics* (Vol. 30, Issue 97). Elsevex Science Ltd.

Neptune, R. R., & Herzog, W. (1999). The association between negative muscle work and pedaling rate. In Journal of Biomechanics (Vol. 32).

Ola Aasvold, L., Ettema, G., & Skovereng, K. (2019). Joint specific power production in cycling: The effect of cadence and intensity. *PLoS ONE*, *14*(2). https://doi.org/10.1371/journal.pone.0212781

Patterson RP, Moreno MI (1990) Bicycle pedalling forces as a function of pedalling rate and power output. Medicine and Science in Sports and Exercise 22 (4):512-516

Peterman, J. E., Morris, K. L., Kram, R., & Byrnes, W. C. (2016). Pedelecs as a physically active transportation mode. *European Journal of Applied Physiology*, *116*(8), 1565–1573. https://doi.org/10.1007/s00421-016-3408-9

Rauter, S., Supej, M., & Vodičar, J. (2023). Examining the Efficiency of Electric-Assisted Mountain Biking across Different Types of Terrain. *Applied Sciences (Switzerland)*, *13*(21). https://doi.org/10.3390/app132111677

Sanderson DJ (1991) The influence of cadence and power output on the biomechanics of force application during steady-rate cycling in competitive and recreational cyclists. Journal of Sports Sciences 9 (2):191-203

Sarre G., Lepers R., Maffiuletti N., Millet G & Martin A. Influence of cycling cadence on neuromuscular activity of the knee extensors in humans. European Journal of Applied Physiology, 2003; 88(4-5), 476-479.

Simons M, Vanes E, Hendrilsen I. Electrically Assisted Cycling.Medicine & Science in Sports & Exercise. 2009;41(11):2097--2102.doi: 10.1249/mss.0b013 e3181 a6aaa4

Shah, J., Shah, S., Mehta, G. et al. Vertical stiffness of bicycle frame and its influence on rider comfort. SN Appl. Sci. 2, 1629 (2020). https://doi.org/10.1007/s42452-020-03410-w

Sperlich B., Zinner C., Hobert-Losier K., Bron D-P, Holmberg H-C. Biomechanical, Cardiorespiratory, metabolic and perceived responses to electrically assisted cycling. European Journal of Applied Physiology,2012; 112(12):4015-25.

Swart, J., & Holliday, W. (2019). Cycling Biomechanics Optimization-the (R) Evolution of Bicycle Fitting. www.acsm-csmr.org

Titlesatd et al. Effect of suspension systems on the physiological and psychological responses to sub-maximal biking on simulated smoothand bumpy tracks. J Sports Sci. 2006, Feb 24 (2):125-35.

Tokui M., Hirakoba K. Effect of internal power on muscular efficiency during cycling exercise. European Journal of Applied Physiology, 2007; 101(5):565-70.

Vercruyssen F., Brisswalter J. Which factors determine the freely chosen cadence during submaximal cycling? Journal of Science and Medicine in Sport, 2008; 225-231. https://doi.org/10.1016/j.jsams.2008.12.631

Visentini PJ., McDowell AH, Pizzari T.,Factors associated with overuse injury in cyclists: a systematic review. J Sci Med Sport. 2022, May; 25 (5);391 – 398.

Wakeling JM, Nigg BM, Rozitis AI. Muscle activity damps the soft tissue resonance that occurs in response to pulsed and continuous vibrations. J Appl Physiol 2002; 93: 1093-1103

Wozniak Timmer, C. A. *Cycling Biomechanics: A Literature Review*. Journal of Orthopedic & Sports Physical Therapy (1991). www.jospt.org

Zameziati K, Mornieux G, Rouffet D, Belli A (2006) Relationship between the increase of effectiveness indexes and the increase of muscular efficiency with cycling power. European Journal of Applied Physiology 96 (3):274-281

Statista: https://www.statista.com/statistics/255653/worldwidesales-of-electric-bicycles

Statement on Originality and Use of Artificial Intelligence Tools

All content, ideas, and analyses presented in this thesis are original. ChatGPT (Free Web version) was used for grammar checking and stylistic refinement. All AI-assisted revisions were subsequently reviewed to ensure that the original meaning was preserved.