

# Biomechanics of Cycling

## (Literature review)

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### Abstract

*The aim of this review paper is to outline the effects of several biomechanical factors on cycling efficiency and safety. The paper begins with a short introduction and listing of basic concepts important for understanding the biomechanics of cycling, followed by an explanation of mechanical forces and torques that are created during pedalling. Workloads and joint movement are detailed in chapter three, which is augmented by chapter four on muscle activation patterns. Throughout the text we have paid careful attention in interpreting the results of research studies into changes in bicycle geometry, feet position, terrain incline and other cycling-related factors. The paper closes with an overview of all issues and solutions as well as presenting proposals for additional research.*

### 1. Introduction

Cycling has become one of the world's most popular sports in recent years. But its increasing popularity has also resulted in a growing number of injuries and a subsequent need to better understand the workloads on the body during this complex movement and sports activity. In order to discover the best ways to exercise, scientists embarked on studying cycling with various methods. Laboratory studies of cycling were launched in early 20<sup>th</sup> century, when the first cycling ergometer was built (Krogh & Lindhard, 1913). The device allows for easy study of movement patterns under laboratory conditions due to its standardised workloads. It enables precise workload controlling, a feature that makes it highly suitable for other purposes besides testing and diagnostics, such as exercise.

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Development has continued since then with continual improvements to measurement technology to this very day. Modern ergometers allow for very accurate workload settings and provide data that can then be directly uploaded to a computer for analysis. The development of ergometers went hand in hand with improvements to software for capturing data and performing the subsequent detailed analysis. Technological advances do, however, not solely happen due to research. They have also been spurred on by the need to prescribe the most suitable exercise to prevent or cure numerous illness-related and post-surgical conditions and to improve the fitness of professional and amateur cyclists. Bertucci, Frederic Grappe, & Gros Lambert (2007) believed that laboratory tests differ from outdoor cycling, which is why they carried out a study to highlight the differences between laboratory and real-life cycling conditions. The result was the complete opposite of what they thought as the crank torque profiles in the two environments did not differ. They have subsequently proven that laboratory conditions mimic outdoors cycling well.

Cycling research was at first taken on by physiologists, who were mainly interested in cardiographic, vascular and respiratory aspects, which has resulted in numerous questions being raised regarding metabolism under workload. Physiologists developed numerous tests that can gauge a person's stamina and identify his or her weak spots in the laboratory, thus allowing cyclists to improve and increase their abilities and efficiency. A good knowledge of the body's functioning allowed them to pinpoint the areas of intensive activity for target motor abilities training. Moreover, their studies of cyclists' physiology and the introduction of incremental tests to exhaustion set a milestone in researching physiological responses during exertions. Tests in physiologic laboratories have to this very day remained the mainstay of testing an individual's fitness.

Cycling is one of the safest non-contact sports. As cyclists sit on a saddle, the workloads on their joints are almost completely unrelated to their weight, a fact that can be put to good use in slimming regimes. Due to light joint workloads, cycling is advisable in early phases of post-surgical and post-traumatic rehabilitation of the motor system as the injured joint can be exposed to a relatively low workload combined with a relatively large muscle effort. These efforts can be supplemented through changes to bicycle geometry, which can greatly influence the intensity of joint workloads, a topic that will be discussed in more detail below. These adjustments should be known by rehabilitation specialists, especially if they want to preserve or improve an injured person's strength or stamina while taking care not to expose the injured joint to too much mechanical work-

load. Ericson & Nisell (1986) proposed that rehabilitation of injuries to the anterior cruciate ligament (ACL) should be carried out in the following order: walking with crutches, cycling, normal walking, slow running, fast running. They came to these conclusions following a study in which they investigated biomechanical workloads on the knee joint, which they then compared to the results by researchers who studied joint workloads during other everyday activities, such as walking, walking up or down the stairs, getting up from the chair or lifting loads from the ground. Similar conclusions were also reported by other studies that examined joint workloads (Ericson, 1986; Ericson, Bratt, Nisell, Németh, & Ekholm, 1986; Ericson, Ekholm, Svensson, & Nisell, 1985; Ericson & Nisell, 1986, 1987). One of the sport's positive characteristics is that changes to bicycle geometry can influence the amplitude of individual joint movements, for example, in preventing excessive amplitudes of injured joints if that is what the physician wants and vice-versa.

Understanding movement patterns in cycling and looking for an optimum solution necessitates an accomplished diagnostics method spanning various technological approaches as well as entailing considerations of the importance of an individual's subjective feelings. These requirements invariably cause scientists to come across many questions that they want to explain in the best and most accurate way possible. Professional cyclists want to increase their mechanical efficiency, their amateur counterparts wish to enjoy the activity more and injured people seek the safest and most efficient methods for treating various conditions. The biomechanics of cycling is a wide field and we should be aware of that. We have hereby limited ourselves to presenting some basic concepts, forces exerted on pedals, workloads on the joints and their movement and the characteristics of inter-muscle coordination, even though the field spans many more topics.

In looking through available literature, we have focused on expert and scientific papers from the following databases: Pubmed, ScienceDirect, Springerlink, SportDiscus and Ebscohost. We combed through them by using keywords such as biomechanics, forces, joints, EMG and geometry and added the word cycling to them. We netted over 100 expert and scientific papers.

## **2. Explanation of Some Basic Concepts**

As well as most other sports, cycling uses several terms that are unknown to the average person. In order to better understand the rest of the paper, we will provide a short definition of terms that are used in

describing the biomechanics of cycling. When we speak about bicycle geometry settings, we usually refer to saddle position, which can be adjusted according to height, tilt and placement forward/back. The saddle's height is defined in scientific literature as the distance between the top part of the saddle and the pedal axle, with the pedal in its lowest position (Burke, 1994). Recently, however, a different definition which is mainly used by professional cyclists and bicycle retailers uses the distance between the top part of the saddle and the middle of the crank axle. For the purposes of this paper, we will use the former definition when discussing adjustments to saddle height. When talking about handlebar position, we will usually refer to its width and distance from the ground and the saddle (figure 1). Foot position on the pedals was roughly defined by scientists when studying biomechanical effects of various placements. We distinguish between the anterior position, usually used by cyclists, and the posterior position, which is not used in everyday cycling. The anterior position has the pedal's centre placed at the level of the metatarsophalangeal thumb joint, while posterior position has it in the middle of the foot. We would like to reiterate that the posterior position is not used in everyday cycling, but can be beneficial in rehabilitation of injuries to the talocrural joint and the Achilles tendon (Ericson et al., 1985). In interpreting those biomechanical parameters that change in relation to various pedal positions, we must present them in conjunction with the position of the pedal in which they occur. Identifying the top dead centre (TDC) and the bottom dead centre (BDC) allows us to present these values as a function of the pedal's angle as it changes between the topmost ( $0^\circ$ , TDC) and the bottommost position ( $180^\circ$ , BDC). The phases of a pedal's revolution are: (i) the first or the downstroke phase (from  $0^\circ$  to  $180^\circ$ ), (ii) the second or upstroke phase (from  $180^\circ$  to  $360^\circ$ ), and (iii) two transitional phases ( $\pm 5^\circ$  from the TDC and BDC). The pedalling frequency (cadence) has been the subject of many studies and is still researched today. It is defined as the number of revolutions per minute (rpm). The crank's torque is defined as the product of the force rectangular to the crank and the crank's length. This dynamic propulsive torque is the key factor in the mechanical efficiency of cycling (Coyle et al., 1991). The product of torque (Nm) and angular speed (rad/s) is power (W) which cyclists use to overcome the workloads and represents their end mechanical effect (Bertucci, Grappe, Girard, Betik, and Rouillon, 2005). We will focus on the following in this paper: (i) mechanical forces, (ii) joint workloads and movements, (iii) muscle activation, and (iv) mechanical efficiency. However, we will also mention the changes to these factors that occur through various biomechanical bicycle settings (saddle position, cadence, workload, etc).

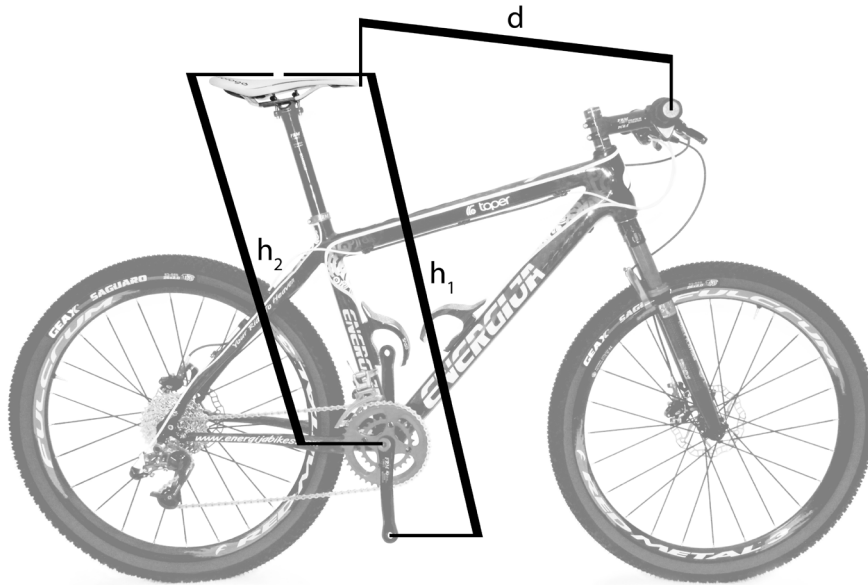


Fig. 1. Overview of the main bicycle geometry settings ( $h_1$ : Seat height definition used in sport science,  $h_2$ : seat height definition used by cyclists and bicycle retailers,  $d$ : distance between the seat and the handlebar)

### 3. Mechanical Forces and Torque

The first study into forces exerted on pedals was carried out by Hoes, Binkhorst, Smeekes-Kuyl and Vissers (1968), who discovered that they are the highest when the pedal is in the horizontal position ( $90^\circ$ ) and that their magnitude on a pedal in a single revolution was twice as high as the preset workload on the ergometer. This led them to conclude that the passive leg (the one in the upstroke phase, – i.e. between  $180^\circ$  in  $360^\circ$ ) is raised by the active leg (the one in the downstroke phase – i.e. between  $0^\circ$  and  $180^\circ$ ), which they believed reduced mechanical efficiency. The measurements were carried out with standard pedals that did not allow “pulling” in the second phase of the revolution, however they still clearly showed the active forces in the first phase. Davis and Hull (1981) meanwhile searched for ways to increase the efficiency of forces on the pedals and discovered that it increases during: (i) use of clipless pedals and (ii) higher workloads.

Ericson and Nisell (1988) investigated mechanical forces under various conditions and discovered that the tangential force is the only mechanically efficient force while the centrifugal force (perpendicular to the tangential force and running alongside the crank) does not contribute to the end mechanical efficiency (figure 2). Their data allowed them to calculate the ratio of mechanical efficiency, which is always between -1 and 1 and represents the relation of the tangential force and the sum of all forces on the pedal. Optimum mechanical efficiency is reached if the centrifugal force equals zero (efficiency ratio = 1). This means that the sum of forces is directed tangentially and acts in its entirety in the direction of the pedal's movement (Ericson & Nisell, 1988). Ericson and Nisell (1987) also discovered that the resultant of the forces throughout the revolution faces downwards as well as slightly forwards between  $0^\circ$  and  $160^\circ$  and slightly backwards between  $160^\circ$  and  $360^\circ$ . The tangential force was negative (directed opposite to the pedal's movement) between  $195^\circ$  and  $360^\circ$ , which led them to conclude that all the work in this case was done by the contralateral leg. The highest centrifugal force (as the inefficient force) was measured when the pedal was between  $120^\circ$  and  $195^\circ$  and the lowest when the pedal was in a horizontal position (between  $90^\circ$  and  $285^\circ$ ). When analysing the forces, the authors found the vertical force to be much stronger than the horizontal. They measured the ratio of mechanical efficiency while changing the workload (0, 120 and 250 W), cadence (40, 60, 80 and 100 rpm), the height of the saddle (102% (low), 113% (optimum) and 120% (high) of the distance between the medial malleolus and the ischial tuberosity) and the position of the foot on the pedal (anterior and posterior). They reported that the ratio of mechanical efficiency increases by a slight, yet statistically significant amount on higher workloads with the foot in the anterior position in comparison with the posterior one. Other factors do not significantly influence mechanical efficiency.

Coyle et al. (1991) measured the difference in the forces exerted on the pedals by professional and amateur cyclists by using clipless pedals, which allowed them to reach maximum mechanical efficiency. They discovered that professionals generated 11% greater mechanical efficiency and used only 9% more work to do so. The increase in power use was mainly due to more torque in the first phase of the revolution, which can also be seen from the higher maximum torque levels in this phase. It is worth noting that professionals increased the torque by using greater vertical force in the downstroke phase and not by pulling in the second phase. The recreational cyclists meanwhile generated less torque in the first and more in the second phase. Another thing to note is that the groups did not differ significantly in their pedalling technique during the same workloads and cadence.



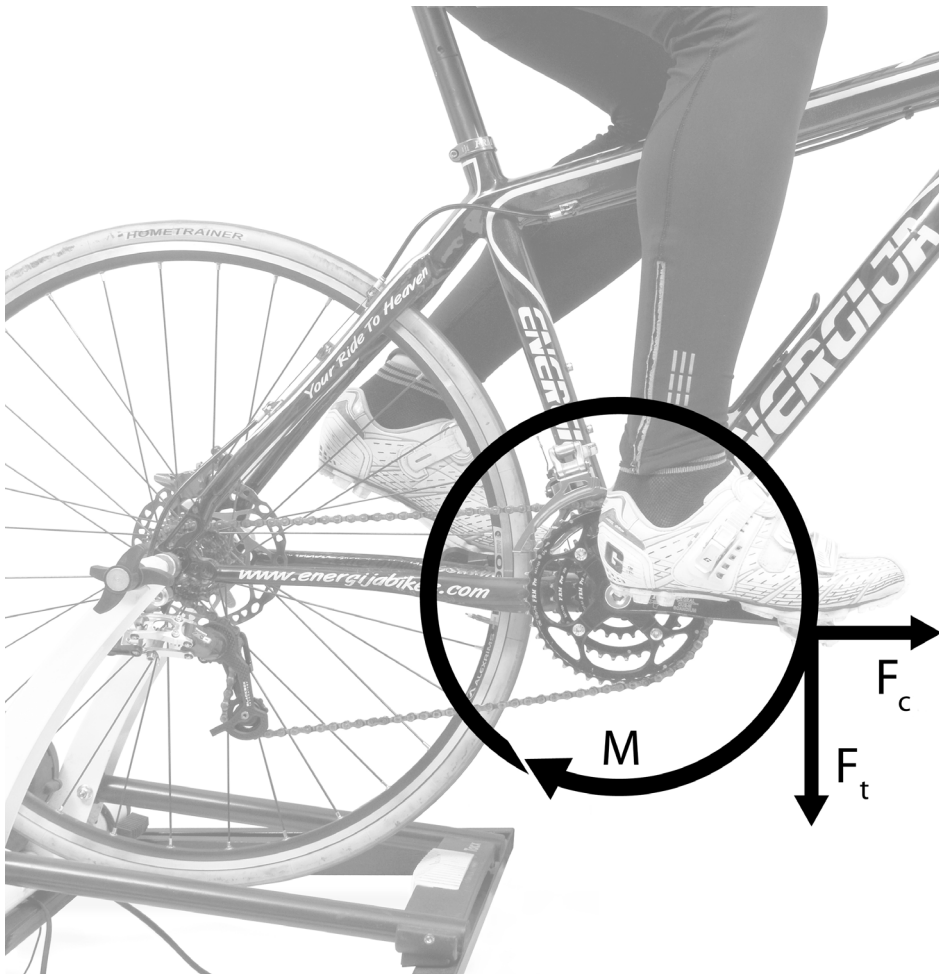


Fig. 2. Overview of all major forces directed at the pedal in the downstroke phase at  $90^\circ$ . ( $M$ =propulsive torque,  $F_c$ =centrifugal force,  $F_t$ =tangential force)

Bertucci et al. (2005) monitored torque changes when cycling under different conditions. They discovered that torque increases by 26% when managing an 8% uphill slope at the same cadence (80 rpm). Torque was statistically significantly higher at 60 rpm cadence in comparison to 80 rpm cadence on flat terrain. Torque was also statistically significantly greater than observed at 100 rpm cadence on flat terrain. The biggest difference in the torque profile was detected when comparing flat terrain cycling at 100

rpm cadence and uphill cycling at 60 rpm cadence. In general, the highest torque levels were recorded at lower cadences, as also dictated by the laws of physics. In order to produce the same power, we have to generate higher torque at lower cadences. Torque profile appears later in the revolution at higher cadences. The study was carried out under real-life conditions, meaning that measuring speed and power was more challenging than laboratory testing. The primary goal of the study was to find the optimum cadence for uphill cycling, as experience from major competitions led them to believe that it should be lower than when riding on flat terrain.

Sanderson & Black (2003) hypothesised that fatigue results in uneven patterns and sizes of effective forces on pedals. In their study they monitored the sum of all forces and the tangential force. They carried out two tests, where experienced professionals used 80% and 30% of their maximum power, respectively, while cycling at the same cadence. The cyclists' maximum power was determined during a multi-level workload test, carried out during previous testing. The authors did not detect any statistically significant differences in the magnitude of the forces at 30% of maximum power. During testing at 80% of the maximum power, they discovered that the sum of forces was statistically significantly larger during the first and the last minute, while the tangential force remained the same. They also observed significantly higher torque in the last minute. The efficiency index that they calculated dropped significantly in the last minute, but primarily in the second phase of the revolution. This led them to conclude that forces were higher in the first phase of the revolution as the tiring cyclists began losing their efficiency in overcoming the workloads in the second phase. This research also unveiled that cyclists generated a relatively high tangential force in the second phase, meaning that they were pulling up in that phase.

#### **4. Joint Movement and Workloads**

Numerous studies on workload and movement of joints in lower extremities during cycling at various workloads (0, 120, 240 W), cadences (40, 60, 80 and 100 rpm), foot position on the pedal and saddle height (low (102%), optimum (113%) and high (120%)) have been carried out by Swedish scientists (Ericson, Nisell, & Nemeth, 1988; Ericson, 1986; Ericson et al., 1986; Ericson et al., 1985; Ericson & Nisell, 1986, 1987, 1988). They reported that compression forces on the joints during standardised cycling on an ergometer (120 W workload, 60 rpm cadence and optimum saddle height) on average amount to a full body weight.



Ericson, Nisell and Nemeth (1988) observed changes in the range of motion (ROM) in the joints of the lower extremities under the same variable settings as mentioned above. Their results have shown that hip ROM did not change statistically when adjusting the saddle height from low to high, that extension increased by 19° and flexion decreased by 16°. Changing the workloads from 0 to 240 W, caused hip flexion to decrease by 2° and extension to increase by 3°. Hip ROM did not statistically significantly change when altering the workloads. Anterior foot position increased the extension and hip ROM by a marginal but statistically significant 7%.

Raising the saddle from low to high position increased knee ROM by 15° and knee extension by 41°, while knee flexion decreased by 22°. The workload change from 0 to 240 W did not show any statistically significant changes for ROM and flexion, while knee extension decreased by 7°. Anterior foot position on the pedal caused a statistically significant knee ROM decrease (by 3°). Knee flexion meanwhile increased by 7°, while extension decreased by 10°.

Raising the saddle from low to high position caused ankle ROM to increase by 18° and plantar flexion to rise by 20°, while dorsal flexion remained statistically unchanged.

The workload increase from 0 to 240 W increased the ankle ROM by 8° and dorsal flexion by 9°, while plantar flexion was not significantly altered. Anterior foot position increased ankle ROM and ankle dorsal flexion by 5°, respectively, compared to posterior position. Plantar flexion remained statistically unchanged with various foot positions. See Table 1 for detailed view of changes in ROM for all joints under all mentioned conditions (seat height, intensity, foot position).

Table 1. Table depicting the change of joint movement under various biomechanical settings (summarized data by Ericson et al (1998))

	<b>Seat height</b> <b>From low to high</b>	<b>Intensity</b> <b>From 0 to 240 W</b>	<b>Foot position</b> <b>Instep to ball of the foot</b>
<b>Hip ROM</b>	Not influenced	Not influenced	- 7°
<b>Hip extension</b>	+ 19°	+ 3°	+ 2°
<b>Hip flexion</b>	- 16°	- 2°	Not influenced
<b>Knee ROM</b>	+ 15°	Not influenced	+ 7°
<b>Knee extension</b>	+ 41°	- 7°	+ 10°
<b>Knee flexion</b>	- 22°	Not influenced	- 7°
<b>Ankle ROM</b>	+ 18°	+ 8°	- 5°
<b>Ankle plantar flexion</b>	+ 20°	Not influenced	Not influenced
<b>Ankle dorsal flexion</b>	Not influenced	+9°	- 5°

The authors (Ericson et al., 1988) compared ROM data for individual joints that they collected in their studies with the average maximum range of motion as reported by the American Academy of Orthopaedic Surgeons (AAOS). The comparison indicated normal ergometer cycling only generated 28% of hip ROM, 45% of knee ROM and 40% of ankle ROM, compared to the maximum range of motion as reported by the AAOS. A similar study was carried out by Sanderson & Amoroso (2009), who monitored ROM at various saddle heights. They discovered that knee ROM increased by 17° and tarocrural joint ROM by 14° at high saddle setting. It is worth noting that the tarocrural joint is in dorsal flexion for the entire revolution at low saddle position, while at high position it reaches 66° of plantar flexion at the lowest point of the revolution. The study involving kinematic analysis, carried out by Chapman et al. (2008) has shown that ROM of joints during bent or more aerodynamic posture does not change in comparison to normal or upright posture. Cadence in no way changes the joint movement (Ericson et al., 1988).

The workload on the knee joint rises significantly if: (i) the intensity of cycling is increased, (ii) the height of the saddle is lowered (Ericson & Nisell, 1986, 1987). The authors also discovered that the largest compression force on the tibiofemoral joint appears at between 60° and 100°. The shear force directed forward to the anterior cruciate ligament reaches the highest magnitude between 80° and 140°, while the shear force directed backwards to the posterior cruciate ligament peaks at between 330° and 80°. Compression forces on the knee ligament are statistically significantly reduced if saddle height is increased. The forward shear force on the anterior cruciate ligament is statistically significantly increased during higher workloads or the use of posterior foot position. The cadence and foot position on the pedal does not impact the compression forces on the knee. All the forces that are exerted on the knee joint are almost completely independent from a cyclist's body weight.

Ericson et al. (1985) observed tarocrural joint forces during ergometer cycling and monitored their changes under varying conditions (different cadence, workload, saddle height and foot position). The highest moment in the tarocrural joint appears at 11° of dorsal flexion. They discovered that the tarocrural joint experiences significantly greater workloads in case of posterior foot position (compared to the anterior position), however, the Achilles tendon experiences significantly smaller ones. Changing the saddle height while using the anterior foot position does not cause statistically significant changes in the forces on the tarocrural joint, however, these forces significantly increase during higher workloads.

Many cyclists complain about pains in the lumbar region during cycling. It has been proven that some people are more and some less susceptible to such pain, which can be caused by unoptimised bicycle settings. This is why Salai, Brosh, Blankstein, Oran, & Chechik (1999) carried out a biomechanical study of forces that operate on the lower back during cycling. They discovered that these forces in the sagittal plane are solely traccial. They used the fluoroscopic method for imaging the lumbo-sacral part and monitored the angle between the pelvis bone and the ground in three different saddle positions. By adjusting saddle tilt, they discovered that a greater angle (lowered front part of the saddle) reduces the angle between the pelvis bone and the ground, consequently reducing traction forces and workloads. In their opinion, optimised saddle setting could eliminate lower back pain, to which purpose they carried out a clinical test, during which they changed the tilt between 10° and 15° for 80 cyclists on various types of bicycles (road, mountain and city bikes). The cyclists were called back six months later and quizzed about lumbar pains. According to the results, 72% of cyclists reported that the pain was gone, 20% reported significantly lesser pain and 7% recorded no change in its intensity. It is recommended that the saddle not be tilted by more than 20° as that could cause the cyclist to inadvertently slide forward.

## **5. Muscle Activity During Cycling**

In order to improve rehabilitation protocols and the cyclists' stamina, it is important to be well acquainted with the functioning of the muscles in the lower extremities during cycling. Accurate determination of muscle group/chain activation patterns is of key importance in order to understand a cyclist's motions during riding. The simplest and most often used method for measuring muscle activation is electromyography (EMG). This diagnostic method measures the electric activity of the muscles with the aid of electrodes that can be placed above (surface method) or in the target muscle (needle method).

The first to use EMG for studying cycling were Houtz and Fischer (1959), who monitored the activity of nearly all the main muscles in the lower extremities, with the exception of m. Soleus. This study was later characterised as very unreliable, due to methodological deficiencies (Hug in Dorel, 2009). Subsequent studies used more representative patterns and accurately described the patterns of muscle coordination during cycling in normal conditions (Ericson, Nisell, Arborelius, & Ekholm, 1985; RJ Gregor, Broker, & Ryan, 1991; Ryan & R. Gregor, 1992; Hug, Decherchi, Marqueste, & Jammes, 2004; Dorel, Couturier, & Hug, 2008). Many re-

searchers later studied changes in muscle coordination by varying the conditions under which the tested subjects cycled. Ericson et al. (1985) used EMG at different saddle heights, cadences, workloads and leg position on the pedal. Quite a number of studies monitored EMG readings while changing the posture and the slope (Chapman et al., 2008; Duc, Bertucci, Pernin, & Grappe, 2008; Li & Caldwell, 1998; Sanderson & Amoroso, 2009). As described below, muscle activity patterns can change according to the position of the body, cadence, saddle height, workload, etc.

As mentioned above, EMG is mainly used in two ways. The first method is invasive and performed by inserting needle electrodes into a muscle. This method has a number of deficiencies. It only records a relatively small number of activated fibres (several cubic millimetres) and is invasive, meaning that it can cause pain and hinder normal movement. The other (and the most commonly used) method in cycling is surface EMG (sEMG) that records the changes in a muscle's electric activity during contraction with the help of electrodes that are placed on the skin above the target muscle. Its advantage is that it records data from a larger muscle volume and is thus more directly linked to the end mechanical muscle efficiency. It is mainly dependent on nerve factors and muscle fibre properties (speed of transmitting action potentials in the muscle's membrane). The downside of this method is that signal interpretation requires consideration of some factors that greatly impact signal amplitude. It is important to correctly place the electrodes, prepare the skin and electrodes and later correctly interpret (process) the signals. . The first issue to be encountered is interference caused by neighbouring muscles (crosstalk). These can be largely eliminated by correctly placing the electrodes and by proper subsequent signal processing. The recommendations for correct electrode placement are dealt with by the SENIAM organisation, which promotes the use of unified electrode placement in research to facilitate the comparison of normalised values among various studies.

Muscle activation and coincidence with regard to a pedalling cycle is studied with the use of averaged and superimposed signals over several consequent cycles. By defining the TDC and BDC, we can display the EMG profile as a temporal function rendered as a percentage of the entire cycle duration, which is why it is possible to compare it to other cycles of differing duration (different cadence). By setting the threshold on non-activity or activity of muscles it is possible to pinpoint the times within the cycle when the muscles are activated. The level of muscle activity in cycling is best quantified through the use of the root mean square (RMS) (Laplaud, Hug and Grélot, 2006; Dorel et al., 2008; Duc, Bertucci, Pernin and Grappe,

2008) or with the use of integrated EMG (EMGi) (Ericson et al., 1985; Jorge and Hull, 1986; Takaishi, Yamamoto, Ono, Ito and Moritani, 1998).

To compare muscle activity between individual muscles, the values need to be normalised, which is mainly achieved through the use of isometric maximum voluntary contraction (IMVC). However, such an approach drew criticism from several authors, which managed to exceed IMVC values during short-time intensive cycling (Hautier et al., 2000). The issue of normalisation was tackled by Rouffet & Hautier (2008), who concluded that dynamic normalisation method would be more suitable than isometric for dynamic activities. Their results show this method to be less strenuous, easier to implement and have the same repeatability as the IMVC during maximum-strength pedalling. Some studies simply used maximum or average values obtained during maximum cycling as normalisation values (Hug & Dorel, 2009). Fernández-Peña, Lucertini, & Ditroilo (2009) proved in their study that maximum cycling on an isokinetic ergometer presents a good approach to normalisation. Normalisation is not as important in studies that compare the same muscle in different conditions (e.g. different posture, workloads, etc.).

A threshold, usually set at 15-25% of maximum EMG normalisation amplitude, is used to determine the time interval of muscle activity (use of IMVC for normalisation). 1, 2 or 3 standard deviations from the average amplitude value during no motion can also be used. This method allows for determining the intervals of muscle activity and non-activity and the subsequent interpretation of this data and the data on crank angle, allowing us to pinpoint the areas of muscle activation during specific parts of the cycle. Such identification of the activation threshold can be contentious in some cases, so some researchers set it according to their instincts. This, however, brought numerous criticisms by others, who claimed this manner of setting would be completely subjective (Li & Caldwell, 1998). As far as we know, a general and precise agreement regarding the optimum threshold setting to determine muscle activity (or non-activity) has not yet been reached.

Active muscles in cycling are roughly divided into single-joint and two-joint muscles. Single-joint muscles are tasked with generating the force that is delivered via two-joint muscles in the correct direction to the pedals. As stated in the chapter on mechanical forces, cyclists with better pedalling technique generate higher effective forces than those with more rudimentary pedalling. A hypothetical explanation, using the results that will be presented in this chapter, allows us to tentatively conclude that



this difference comes from varying degrees of activation of two-jointed muscles. As far as we are aware, studies that would investigate the link between various activation levels and pedal forces in more detail have not yet been carried out.

Single-joint muscles that are active and most frequently measured during cycling are: m. Gluteus Maximus m. (GMax), m. Gluteus medius (GMed), m. Vastus Lateralis (VL), m. Vastus Medialis (VM), m. Tibialis Anterior (TA), m. Soleus (SOL) and m. Iliopsoas (IP). The two-joint ones include: m. Rectus Femoris (RF), m. Semimembranosus (SM), m. Semitendinosus (ST), m. Biceps Femoris (BF), m. Gastrocnemius Lateralis (GL) and m. Gastrocnemius Medialis (GM). A good description and identification of muscle activity timing was published by Ryan & Gregor (1992) (figure 3 - see next page). GMax extends the hip and is active between  $340^{\circ}$  and  $130^{\circ}$ , peaking at  $80^{\circ}$ . VL and VM extend the knee and are active in the same part of the cycle – between  $300^{\circ}$  and  $130^{\circ}$ , peaking at  $30^{\circ}$ . RF acts as knee extensor and hip flexor and is active between  $200^{\circ}$  and  $110^{\circ}$ , peaking at  $20^{\circ}$ . SOL stabilises the tarocrural joint between  $340^{\circ}$  and  $270^{\circ}$ , peaking at  $90^{\circ}$ , when the forces exerted on the pedal are the highest. GM and GL have the same function – tarocrural joint stabilisation and knee flexion. They are active between  $350^{\circ}$  and  $270^{\circ}$  and peak at  $110^{\circ}$ . TA also serves to stabilise the tarocrural joint and at the same time flexes it. It is active throughout the cycle, peaking at  $280^{\circ}$ . SM and ST flex the knee and are active between  $10^{\circ}$  and  $230^{\circ}$ , both peaking at  $100^{\circ}$ . BF flexes the knee and extends the hip. It is active between  $350^{\circ}$  and  $230^{\circ}$ , peaking at  $110^{\circ}$ .

The same researchers also studied the variability coefficient of individual muscles by comparing surface and needle EMG. Single-joint hip and knee extensors (m. Vastus Medialis, m. Vastus Lateralis, m. Gluteus maximus) had the lowest variability coefficient. However, with the exception of m. Tibialis Anterior, m. Semitendinosus and m. Gastrocnemius, all muscles showed very low variability. They also discovered that results obtained by the use of surface and needle EMG are very similar.

Raasch & Zajac (1999) divided muscle activities into three types, according to their tasks. The first group comprises single-joint hip (GMax) and knee (VM and VL) extensors, alongside single-joint hip (IP) and knee (short head BF) flexors. The authors labelled this group the E/F group (extensor/flexor). The second group includes the two-joint RF and TA muscles (the RF/TA group) and the third the hamstring (HAM) muscles, namely the ST, SM and the long head of the BF, alongside the SOL, GL and GM (the HAM/SG group). The main purpose of the E/F group is



to generate energy for pedalling, while the RF/TA and HAM/SG groups mainly act as rigid transmitters to improve the efficiency of energy transfer between the segments. The RF/TA group provides energy at the end of the second phase of the revolution and helps in the transition to the new revolution cycle. The HAM/SG groups are mainly active at the end of the first phase and help in the transition to the second phase. Raasch & Zajac's muscle activation pattern requires the tarocrural stabilisation muscles to maintain activity at all times to successfully transfer the energy to the pedals.

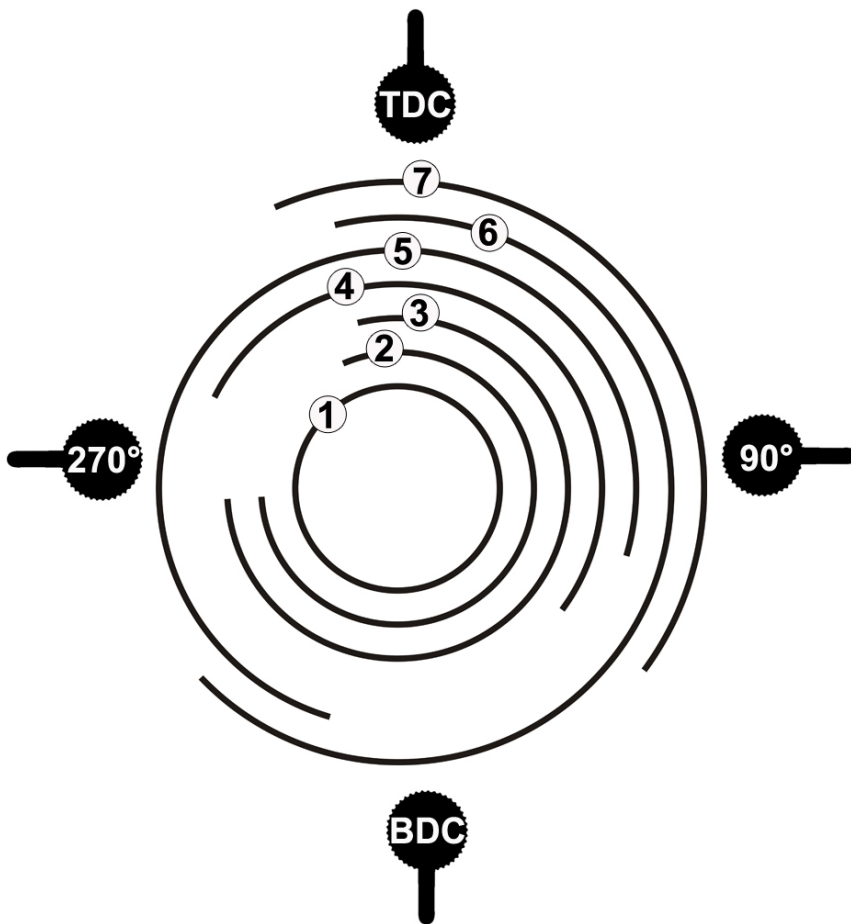


Fig. 3. Overview of muscle activity timing in lower extremities during cycling in relation to the crank angle (1=TA, 2=SOL, 3=GM, 4=VL&VM, 5=RF, 6=BF and 7=GMax). Based on the results of Ryan & Gregor, (1992)

Differences in activation between single- and two-joint muscles were first noticed by Ericson et al. (1985), who reported that single-joint muscles were more active than two-joint ones at 120 W (around 54% of maximum aerobic power) – VM at 45%, VL at 44% and SOL at 32% of IMVC (standard normalisation). Two-joint muscles reached lower activation levels and were at 22% (RF) and 18% (BF). Two-joint muscles serve various purposes in different revolution phases. RF, for example, takes part in hip flexion during the second phase and is active in knee extension in the first phase. A similar pattern was recorded for the GS, which flexes the knee in the second phase and stabilises the tarocrural joint in the first phase (Raymond, Joseph, & Gabriel, 2005). According to Gregor et al. (1991), the BF and RF reach two peaks at high cadences. Two-joint muscles are also active in transitional phases during TDC and BDC (transitions from the first to second and second to first phase). Relative power of individual muscles also plays an important role as strong single-joint hip extensors can cause a spike in the RF's activity. The opposite happens with weak extensors, as the BF needs to jump in to finalize hip extension (Li & Caldwell, 1998). This makes it necessary to take into account power ratios between individual muscles or muscle groups in designing training sessions and rehabilitation protocols.

Numerous co-activation patterns can be detected in the first phase of the revolution. For example, plantar flexors (GM, GL and SOL) and dorsal flexors (TA) act together to provide good stabilisation of the tarocrural joint in the first phase of the revolution, where the forces reach their peak. If we look at the hip muscles, we see that the VL, VM and RF are aided by hamstring muscles during hip extension (Gregor et al., 1991; Jorge & Hull, 1986). Detailed studies of muscle coordination have showed that co-activation is to a large extent present between single- and two-joint antagonists, which results in a coordinated transfer of mechanical energy between joints. Muscle co-activation does not only improve energy transfer between segments, it also protects the joints.

As already noted, muscle coordination changes with varying cycling conditions. Sanderson & Amoroso (2009) monitored the activity of distal leg muscles (GM and SOL) using various saddle heights. The EMGi value was statistically significantly lower for both muscles when using a lower saddle position. GM meanwhile was only 32% active at a low saddle setting, while soaring 20% more at a high setting. The effects of saddle height adjustments on other muscles were studied by Ericson et al. (1985), who reported increased levels of activity for GMax, ST, SM, GM and GL at high saddle settings. Chapman et al. (2008) used the needle EMG method

to compare tarocrural joint muscle activation during normal and aerodynamic posture for recreational cyclists, triathlon athletes and professional cyclists. Aerodynamic posture decreased the main EMG amplitude and increased co-activation for amateur cyclists and triathlon athletes, but not for professional cyclists. Based on various EMG values for different cyclists, the researchers concluded that differences in muscle activation are caused by neurological and not biomechanical factors.

Muscle activation does not statistically significantly change when cycling up a 4%, 7% and 10% slope with 80% of maximum aerobic power, which was determined beforehand during a multi-level workload test. Such cycling only resulted in increased activity of the GMax and m. Erector spinae (ES) muscles. The results were likely influenced by cadence as well, which was not set in advance and dropped for most of the cyclists during uphill cycling. This could have caused lower EMG values than otherwise (Duc et al., 2008). However, Clarys, Alewaeters, & Zinzen (2001) recorded generally increased leg muscle activation when changing the incline from 0% to 12%. Their results refer to general leg muscle EMG patterns and not to individual muscles. Muscle activation during uphill cycling (especially for steeper inclines) has not yet been definitely researched and requires future studies.

Since diverse researchers have arrived at different conclusions, the impact of cadence on EMG values has also not been well explained yet. MacIntosh, Neptune, & Horton (2000) reported that inter-muscle activity relations change in accordance with the workload and cadence. The lowest values are found at higher cadences during larger workloads and vice versa. Some researchers reported increased muscle activation at higher cadences (Ericson et al., 1985). Cadence change impacts the position and duration of an individual muscle's activity within a cycle. The higher the cadence, the sooner in the cycle activation increases for nearly all the muscles of the lower extremities, with the exception of the SOL, which shows activity later in the cycle (Neptune, Kautz, & Hull, 1997). Electromechanical delay is not impacted by the cadence and usually lasts between 10 and 100 ms (Cavanagh & Komi, 1979).

Ericson et al. (1985) found increased RF, BF and TA activity and reduced VM, VL and SOL activity when using clipless pedals compared to ordinary ones. They recorded a decrease in GMed and RF activation and an increase in SOL activation in posterior foot position in comparison to anterior position.

Cannon, Kolkhorst, & Cipriani (2007) carried out a study to investigate the effect of muscle activity and mechanical efficiency for two different pedalling techniques. The first one was the dorsal technique, where the foot was at all time kept in dorsal flexion and the second a plantar technique, where it remained in plantar flexion. The subjects were given a couple of days to acquaint themselves with the two and were then asked to do ergometer tests with 80% of maximum aerobic power at a 90 rpm cadence. The researchers monitored the activation of the VL, TA, the lateral head of the GS and the BF. They reported increased GS activity in dorsal technique and increased BF activity in plantar technique. VL and TA activation was not affected by the use of the two techniques, none of which is used in cycling under normal conditions.

Nerve muscle fatigue during submaximum workloads is shown as an increase in the EMG amplitude and changes in the EMG signal frequency spectrum towards lower frequencies, caused by the activation of fresh motor units (De Luca, 1984). In cycling and in many other sports the abovementioned electro-physiological changes of the nerve-muscle system occur before the drop in ability of a cyclists to overcome specified workloads occurs. Fatigue is defined as the inability to continue overcoming the workload and is seen as a drop in performance (Asmussen, 1979). Psek & Cafarelli (1993) meanwhile came across an interesting case of fatigue when researching activation in antagonistic muscles during cycling. They discovered that fatiguing the VL caused greater activation of the BF. Contrary conclusions were presented by Hautier et al. (2000), who reported that co-activation during maximum pedalling decreases in line with the drop in the force, exerted by the subject. Hettinga, De Koning, Broersen, Van Geffen, & Foster (2006) recorded a decrease in speed and increase in EMGi in the VL and BF during their study into fatigue in 4000m cycling. They concluded that fatigue was peripheral. However, as they only carried out EMG measurements at every 200 metres, they were unable to specifically determine the trend of its increase.

## **6. Discussion**

Even though biomechanics of cycling was and remains researched throughout the world, there are still several blind spots in certain areas. This paper used a wide range of references to shed light on the subject from three viewpoints which have been scrutinised the most so far. In order to facilitate rehabilitation, the researchers mainly studied different bicycle settings and cycling techniques while keeping an eye on joint workloads. Ericson et al. (1985, 1986, 1988) and Ericson & Nisell (1987,

1986) published extensive studies on joint workloads and motions. While they were carried out on professional cyclists, they did not mimic racing conditions – e.g. no use of clipless pedals, racing posture and workloads that would suit each individual cyclist. All their studies used a standardised (as opposed to customised) workload, which can prove misleading, as certain workloads present no problem whatsoever for well-trained cyclists while being extremely difficult for others. The authors have acknowledged that in their conclusions. Workloads play an important role in monitoring biomechanical parameters, which is why they should be tailored to an individual's abilities. The studies changed the height of the saddle, cadence, workload and feet position on the pedals. When observing forces exerted on the pedals they discovered that only the first (downstroke) phase is mechanically efficient. The passive leg (the foot of which is in the upstroke phase) was said to be raised by the effort of the active leg. These conclusions were confirmed by some and rejected by others. Many questions remain unanswered regarding the forces on pedals. It should again be noted that not all researchers used clipless pedals, thereby rendering numerous studies unrepresentative. Interesting findings have been posted by Coyle et al. (1991) in their force analysis. They discovered that while less experienced cyclists used more force in the second phase, the more experienced ones still had a higher mechanical efficiency, mainly due to exerting more force in the first phase. Bertucci et al. (2005) monitored torque to find the optimum cadence for flat terrain and uphill cycling. The study carried out in natural surroundings showed differences in amplitude and timing of the torque in a revolution. Greater torque appears at lower cadences and a slight, while statistically significant higher one in cycling uphill. Despite interesting findings, Bertucci was unable to discover the optimum cadence model in uphill cycling as this also depends on other, probably more important, factors. It is a shame that he only monitored the torque in the downstroke phase, as the use of racing equipment would have allowed him to accurately show torque patterns in the upstroke phase as well.

Diagnostic equipment has come a long way since the time that the majority of research was carried out. Forces and torque on pedals will likely be described well only in subsequent studies. The biggest unknowns are still the forces in the second phase of a revolution, which have not been explained to the same degree by researchers. While some have said there are no effective forces in the second phase, Sanderson & Black (2003) proved the opposite by recording a tangential force in the upstroke phase. The majority of other major studies were carried out over 15 years ago, when not only diagnostics equipment, but also bicycles and pedalling techniques lagged far behind their modern counterparts. Efficiency in the second

phase of a revolution can also be observed by monitoring torque throughout a revolution, however newer studies only focused on its first phase.

## **7. Conclusion**

Even though cycling is generally leg movement in a predefined circular trajectory (Hug & Dorel, 2009), it is far from a simple motion as it can be influenced by the smallest geometrical, geographical and changes to various other factors that can have larger or smaller impact on biomechanical parameters. Knowledge about some of these factors can be used to assist in setting and achieving goals, be it for rehabilitation or racing. The majority of studies that included modified cycling conditions were carried out to shed light on behavioural patterns of a cyclist and only a very limited number of research resulted in practically usable results. Thus, additional research is needed to find ways for safer and more efficient cycling. We believe that advances in diagnostic equipment will lead to more detailed insights into other open questions in the field of biomechanics of cycling and thereby increase scientific knowledge about this sports and recreational activity.

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