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Article

Analysis of the Influence of the Angular Position of the Cleat in Kinematics and Kinetics

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Abstract: Objective: The aims that we set in this study were to analyse the kinematic and kinetic changes in the lower limbs of cyclists when using the utility model, nº of publication ES 1078023, which consists of a system of cleats with an exact omnidirectional adjustment and without oscillations between the shoe and the pedal. Methods: This is a quasi-experimental, longitudinal and prospective study with a non-randomized sampling. The sample was made up of 34 cyclists. The variables studies focused on the kinematics of each joint of the lower limb in three planes and the kinetics in the function of the angular position of the cleat. They had the Bioval® system put in place, through which the kinematic parameters were recorded at the points marked on the lower limb for 20 s. Three systems were used for the kinetics (Power Tap, Rotor and Pioneer), recording the power developed for 1 min for each of the study situations. Results: Regarding the kinematic variables, statistically significant differences were found for the three planes in all of the structures studied. As for the kinetics, statistically significant differences were also observed, both when analyzing them globally and when doing so for each of the systems. Conclusions: The variation in the rotational position of the cleat influences both the cyclist's kinematics and dynamics.

Keywords: bicycling; cleat pedal; biomechanics

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1. Introduction

Performance in professional cyclists depends on a many factors. This is why there have been many studies in the last decade that have centered on the adjustment of the bicycle from both the kinematic and kinetic point of view of pedaling [1,2]. Efficiency in cycling is influenced by biomechanical factors, and it is necessary to understand how the strength on the pedal is applied [3]. There is a consensus that bad positioning on the bicycle can reduce the sport performance in professional cyclists [4]. A bad position can also result in different types of injuries due to overload [5]. An increase in injuries has been recorded since the introduction of automatic pedals [6]. The injuries due to overload at the knee level are approximately 25% in cyclists [7,8]. Pains at the previous level to the knee, along with patellar tendinitis are the most frequent, making up 60% of the total [8,9]. This idea has been supported by research by Ruby et al. [10], which found that anatomical variations between asymptomatic individuals corresponded to variations in knee kinematics.

While cyclists show very consistent results in the sagittal plane, there is little quantitative information regarding the effect of variations in the shoe-pedal interface on lower limb kinematics with respect to the frontal and transverse planes. One of the few studies was carried out by Lafortune [11] in which the pedal-shoe-shoe interaction was studied in relation to performance. Standard shoes and standard pedals, drag pedals and clipless pedals were used, with the focus being on sliding between the two parts.

Performance in professional cyclists depends on a large number of factors, which is why, in the last decade, several studies have focused on the adjustment of the bicycle from both the kinematic and kinetic point of pedaling [1,2]. However, there is no agreement regarding the optimal configuration of the position on the bicycle. Likewise, there is much skepticism about the application of forces on the pedal (mechanical efficiency) being a useful indicator of pedaling efficiency [1].

Hannaford et al. [7] examined the movement of the frontal plane in cyclists with and without problems at the knee level. The symptoms improved by reorienting this plane in movement. In cycling, cyclical loads causing injuries can be due to a misalignment between the cyclist and the bicycle. The foot is restricted by the pedal to a circular pattern in the sagittal plane and this union allows small or zero movements between the sole of the shoe and the pedal in the frontal and transverse planes [12].

Chen et al. [13] showed that the foot-ankle complex underwent movements associated with the loads applied to the pedal. If these movements are controlled at the level of this joint, the contractile forces are more optimally developed [14]. These associated loads are the result of the anatomical contractions of the articular structures and of the tridimensional orientation of their axes, these being determined by the orientation of the foot [15]. Therefore, the forces required for propulsion will be transmitted through the articulations of the lower limb.

The pedal has been an element about which little has been researched, and it has been ignored in many investigations. It has three main adjustments, that is to say, anterior-posterior, latero-medial and rotation. For both the first adjustment and the second, research has been developed to determine to what degree they influence the cyclist's performance [16–18]. However, with respect to the rotational adjustment, few investigations have been carried out [19]. It is necessary to conduct new research for a better knowledge of the union between the orientation of the forces applied to the pedal and the changes in muscular efficiency, given that this knowledge is limited [20].

It has been demonstrated that one of the factors that influences the application of forces generated by the lower limb on the pedal is the longitudinal adjustment of the cleat, but nothing is known about that influence on the rotational adjustment. This can be due to the lack of cleat systems that enable determining this adjustment exactly [21,22]. Groot et al. [23] indicated that muscular work depended on the longitude tension and force-speed of the muscles involved. The effectiveness of the force is affected by the joint ranges, the muscular length and their lever arms, in such a way that all the factors indicated depend on the point of the beginning of the movement; that is to say, the orientation of the cleat on the shoes and, thereby, on the spatial arrangement of the lower limb.

The utility model nº ES 1078023 is a cleat system with a unidirectional adjustment that permits quantifying exactly the degrees of rotation that this has with respect to the shoes and has the characteristic of not allowing oscillations between the shoe and the cleat. By means of this system, we can quantify the influence of the rotational position of the cleat on the kinematic and kinetic parameters of the lower limb (Figure 1).

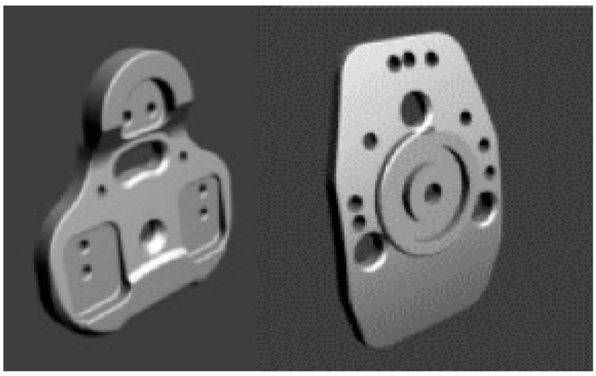

Figure 1. Cleat system nº ES 1078023.

The aims that we set were to analyze the kinematic and strength changes in the cyclist's lower limb when modifying the rotational position of the cleat, taking as a reference the cyclist's own cleat.

2. Materials and Methods

2.1. Design

This study was consecutive non-probabilistic quasi-experimental with a group [24,25]. The institutional review board approved the study (code project 0744-N-17). All participants signed an informed consent as dictated by the Declaration of Helsinki. After being informed of the aims and procedures of the study and prior to commencement of testing, all subjects gave their written consent in accordance with the Declaration of Helsinki.

2.2. Participants

The sample was made up of 34 cyclists of which 9 used the Rotor system, 13 the PowerTap system and 12 the Pioneer system to record power. The inclusion criteria were to be over 20 years old [26] and to develop a high sports intensity to not vary the sport movement [27]. The exclusion criteria established were having suffered serious injuries or operations in the lower limb and to have undergone injuries due to overload in the lower limb.

2.3. Material

To record the kinematics the Bioval Systems® (RM Ingénierie, Rodez, France) [28] device was used, a system of inertial sensors that enables visualizing and quantifying the ranges of movement of three planes of space with a frequency of 30 Hz. The points on which the inertial sensors were placed were: the posterior upper spine, the greater trochanter of the femur, the anterior tuberosity of the tibia and the dorsal area of the tip of the boot [29].

Three different systems were used to record the power: the Pioneer Power Meter SGY-PM9100® [30], PowerTap® [31] system (PowerTap G3, 8 extensiometric gauges, $\pm$1.5 of precision, transmission Ant+) and the Rotor 2INPower® system.

For the rotational position of the cleat, first this was recorded with the cleat that the cyclist brought (Look Keo® grey and Time®), which allowed movement between it and the pedal. Later, the utility model nº de publication ES 1078023 was used. This permit determines exactly the angulation according to the hole in which the pin is placed. This enables placing the cleat exactly in an external rotation of 0°, 2°, 4° and 6°. Moreover, once fixed this cleat system does not allow movement between the pedal and the shoe, avoiding any factor of confusion in the recording of the variables, such as the oscillation between these elements as takes place in the cleats on the market.

2.4. Variables

The descriptive variables were: age, sex, weight, height and BMI. The independent variable was the degrees of rotation of the cleat and the cleat of the cyclist. The dependent variable was the power recorded by the different systems (AVG_Power_Right) and the movement in the 3 planes of the ankle (ABD, Prono, Flex), knee (Flex, Rot, Yawn), hip (Flex, Rot, ABD) and pelvis (Ap, Rot, Ml). All the variables were recorded for the right limb.

The AVG_Power_Right was measured in watts, and the kinematic variables were registered in degrees.

Protocol

For the research, participants brought their own bikes and they used the same trainer with the same resistance to minimize the external factors. During the whole study, each participant had to maintain the same cadence of pedal, which had to be comfortable for him.

The protocol consisted of the recording of the kinematic variables during 20 s and the kinetic variable during 1 min. First, this was recorded with the cyclist's cleat, and then with the model ES 1078023 at 0°, 2°, 4° and 6°.

2.5. Statistical Analysis

The data were analyzed with the SPSS version 22.0 software packet for Windows. The Shapiro-Wilk normality test was applied. For the descriptive statistics, maximum, minimum, average and statistical deviation were used. For the statistical inference, one-factor ANOVA for the kinematic variables and the total power were used. The Student *t*-test for related samples studied the power by systems, comparing each situation with the value of the cleat itself. An $\alpha = 0.05$ statistic was considered significant.

3. Results

Next are the details of the descriptive statistics of the sample. The sample was made up of 34 subjects ($n = 34$), all men. The average age was 29.58 ± 4.08 years old, height 1.72 ± 0.03 m, weight 69.28 ± 6.07 kg and BMI 23.33 ± 2. Of the cyclists taking part, 20 had cleats of the type Look Keo® gris (4.5° freedom) and 14 Time® (5° freedom). Table 1 shows the average values of the variables.

Table 1. Descriptive statistics. Ankle flex: ankle flexion; Ankle_ABD: ankle abduction; Knee_flex: knee flexion; knee_rot: knee rotation; knee_yawn: knee varus/valgus; Hip_flex: hip flexion; Hip_rot: hip rotation; Hip_ABD: hip abduction; Pelvis_AP: pelvis front-back; Pelvis_Rot: pelvis rotation; Pelvis_ML: pelvis mid-lateral; AVG: average.

Variable	Minimum	Maximum	Average	St. dev.
Ankle_Flex_PP	13°	46°	29.41°	8.75°
Ankle_Flex_0	11°	42°	28.76°	8.12°
Ankle_Flex_2	12°	43°	27.38°	8.52°
Ankle_Flex_4	13°	50°	28.52°	9.01°
Ankle_Flex_6	12°	40°	27.59°	8.55°
Ankle_ABD_PP	5°	17°	9.72°	3.68°
Ankle_ABD_0	4°	21°	10.00°	4.33°
Ankle_ABD_2	3°	21°	10.69°	4.25°
Ankle_ABD_4	5°	19°	10.97°	4.07°
Ankle_ABD_6	4°	27°	11.38°	5.20°
Ankle_Prono_PP	3°	17°	7.28°	2.76°
Ankle_Prono_0	3°	12°	7.55°	2.46°
Ankle_Prono_2	4°	14°	7.07°	2.40°
Ankle_Prono_4	4°	15°	7.41°	2.93°
Ankle_Prono_6	4°	33°	8.72°	5.54°
Knee_Flex_PP	53°	89°	72.07°	8.64°
Knee_Flex_0	51°	90°	71.62°	8.72°
Knee_Flex_2	56°	85°	70.93°	7.66°
Knee_Flex_4	54°	90°	71.52°	7.88°
Knee_Flex_6	58°	89°	71.34°	8.20°
Knee_Rot_PP	5°	24°	14.07°	5.55°
Knee_Rot_0	6°	31°	14.97°	6.29°
Knee_Rot_2	7°	32°	16.07°	5.85°
Knee_Rot_4	7°	23°	14.97°	4.78°
Knee_Rot_6	6°	27°	15.45°	5.58°
Knee_Yawn_PP	4°	31°	13.34°	6.08°
Knee_Yawn_0	6°	29°	14.52°	5.41°
Knee_Yawn_2	6°	30°	14.86°	6.39°
Knee_Yawn_4	7°	27°	13.83°	4.62°
Knee_Yawn_6	8°	31°	15.31°	5.43°
Hip_Flex_PP	21°	47°	34.59°	6.24°
Hip_Flex_0	21°	48°	34.59°	6.74°
Hip_Flex_2	21°	47°	35.31°	6.80°
Hip_Flex_4	20°	45°	35.00°	6.57°
Hip_Flex_6	21°	48°	34.83°	6.66°

Table 1. *Cont.*

Variable	Minimum	Maximum	Average	St. dev.
Hip_Rot_PP	6°	25°	12.34°	4.47°
Hip_Rot_0	6°	21°	13.72°	4.08°
Hip_Rot_2	8°	20°	13.14°	3.91°
Hip_Rot_4	9°	19°	13.83°	3.13°
Hip_Rot_6	7°	21°	13.28°	3.65°
Hip_ABD_PP	6°	25°	13.93°	6.57°
Hip_ABD_0	5°	28°	15.86°	7.02°
Hip_ABD_2	5°	27°	14.34°	6.14°
Hip_ABD_4	6°	25°	15.21°	5.90°
Hip_ABD_6	5°	26°	14.34°	5.93°
Pelvis_AP_PP	2°	13°	5.59°	2.63°
Pelvis_AP_0	2°	13°	5.79°	2.55°
Pelvis_AP_2	3°	13°	5.90°	2.26°
Pelvis_AP_4	3°	12°	6.31°	2.24°
Pelvis_AP_6	3°	12°	6.24°	2.37°
Pelvis_Rot_PP	3°	10°	5.52°	2.11°
Pelvis_Rot_0	3°	10°	5.97°	2.10°
Pelvis_Rot_2	2°	10°	5.83°	2.02°
Pelvis_Rot_4	3°	11°	6.21°	2.29°
Pelvis_Rot_6	3°	11°	6.21°	2.37°
Pelvis_ML_PP	3°	10°	5.72°	2.02°
Pelvis_ML_0	3°	11°	6.31°	2.09°
Pelvis_ML_2	3°	10°	5.66°	1.59°
Pelvis_ML_4	4°	11°	6.10°	1.99°
Pelvis_ML_6	3°	11°	5.86°	1.94°
AVG_Power_Right_pp	35.4 w	233 w	122.87 w	41.76 w
AVG_Power_Right_0	66.7 w	242 w	132.87 w	47.57 w
AVG_Power_Right_2	62.8 w	234 w	137.47 w	50.01 w
AVG_Power_Right_4	62.5 w	258.3 w	141.98 w	52.70 w
AVG_Power_Right_6	63.1 w	290.51 w	143.20 w	57.20 w

Through the Shapiro-Wilk Test it was determined that the variables had a normal distribution.

An ANOVA was performed to analyze the differences between the variables studied in each situation of study (0°, 2°, 4°, 6°) and the results obtained with the cleat itself (Table 2).

Table 2. ANOVA of the kinematic variables Ankle flex: ankle flexion; Ankle_ABD: ankle abduction; Knee_flex: knee flexion; knee_rot: knee rotation; Hip_flex: hip flexion; Hip_rot: hip rotation; Hip_ABD: hip abduction; Pelvis_AP: pelvis front-back; Pelvis_Rot: pelvis rotation; Pelvis_ML: pelvis mid-lateral.

	Sig.		Sig.		Sig.		Sig.
Ankle_Flex_0	0.01	Knee_Flex_0	<0.001	Hip_Flex_0	<0.001	Pelvis_AP_0	>0.001
Ankle_Flex_2	<0.001	Knee_Flex_2	0.002	Hip_Flex_2	0.01	Pelvis_AP_2	>0.001
Ankle_Flex_4	0.01	Knee_Flex_4	0.002	Hip_Flex_4	0.06	Pelvis_AP_4	0.03
Ankle_Flex_6	0.02	Knee_Flex_6	<0.001	Hip_Flex_6	<0.001	Pelvis_AP_6	>0.001
Ankle_ABD_0	<0.001	Knee_Rot_0	0.42	Hip_Rot_0	<0.001	Pelvis_Rot_0	>0.001
Ankle_ABD_2	0.04	Knee_Rot_2	0.14	Hip_Rot_2	0.05	Pelvis_Rot_2	0.01
Ankle_ABD_4	<0.001	Knee_Rot_4	0.02	Hip_Rot_4	0.28	Pelvis_Rot_4	0.01
Ankle_ABD_6	<0.001	Knee_Rot_6	0.19	Hip_Rot_6	0.04	Pelvis_Rot_6	0.01
Ankle_Prono_0	0.04	Knee_Yawn_0	0.06	Hip_ABD_0	0.02	Pelvis_ML_0	0.06
Ankle_Prono_2	<0.001	Knee_Yawn_2	0.09	Hip_ABD_2	<0.001	Pelvis_ML_2	<0.001
Ankle_Prono_4	0.03	Knee_Yawn_4	0.01	Hip_ABD_4	0.03	Pelvis_ML_4	<0.001
Ankle_Prono_6	0.77	Knee_Yawn_6	0.01	Hip_ABD_6	<0.001	Pelvis_ML_6	<0.001

All variables, with the exception of Ankle_Prono_6, Knee_Rot_0, Knee_Rot_2, Knee_Rot_6, Knee_Yawn_0, Knee_Yawn_2, Hip_Flex_4, Hip_Rot_4 and Pelvis_ML_0, showed statistically significant differences ($p < 0.05$) compared to their values obtained with the cyclist's cleat. It is interesting because a little change in the sagittal plane of the cleat will influence the kinematics of the lower limb with repercussions for the biomechanics and the muscle action.

For the analysis of the power, first an ANOVA was performed for all of the data and later a Student *t*-test was carried out for each system (Table 3).

Table 3. Contrast tests for the power.

Variable	ANOVA	Student *t*-Test		
	Global (n = 34)	**Pioneer (n = 13)**	**PowerTap (n = 12)**	**Rotor (n = 9)**
AVG_Power_Right_0	0.04	0.44	<0.01	0.02
AVG_Power_Right_2	0.02	0.17	0.17	0.01
AVG_Power_Right_4	0.03	0.03	0.08	0.01
AVG_Power_Right_6	0.01	0.02	0.21	0.02

We can see that in all the research cases (0°, 2°, 4°, 6°), they showed differences when comparing the power values with those developed with the cyclist's cleat. Likewise, when studying the results based on the potentiometer used, we can see that at least one situation showed statistically significant differences. This confirms the data obtained when studying the kinematics. Changes in the cyclist's biomechanics will generate changes in the power developed.

4. Discussion

The objective of this study was to study the variations in the ranges of movements at the level of the pelvis, hip, knee and ankle, as well as to analyse the changes in the recording related with the power developed by the cyclist's limb when modifying the position of the cleat at the rotational level.

Although the speed can be constant during seated pedaling, the forces generated in pedaling fluctuate broadly [32]. The main reason for this is that the muscular force produced depends on the muscle length and therefore on the range of articular movement [33]. Given that the articular ranges change during the pedaling movement and that the lever of each muscle depends on the position of the rod relative to the segment of the leg, the forces applied on the pedal vary through that coupling.

Studies do not exist that have evaluated the influence of the modification of the rotational position of the cleat in cyclists. Ruby et al. [12] determined that with pedals that had freedom of movement ABD-ADD of 15° the varus-valgus moments and the rotation in the knee were reduced. Aligning all the cleats with the shoe's axis is incorrect as this depends on the parameters of the lower limb [19]. Ramos et al. [19] developed a formula through which, and taking into account certain rotational and torsional parameters of the cyclist's lower limb, they were able to adjust the angular position of the cleat. Allowing freedom in the transverse plane decreases the moments in the knee, but also the power that can be applied to the pedal. The minimum values we obtained with our utility model were also double those of the cleats that the cyclists brought. Likewise, the average values obtained with the prototype were greater in all cases. This type of pedal, which allows some degree of freedom between the shoe and the pedal, causes the muscles to spend part of their energy controlling the kinematics. When the cyclist uses a pedal with any degree of freedom between the parts, all the action of the muscles is applied to down the pedal, but if the adjustment is wrong, the joints can develop lesions.

When analyzing the results obtained with the power systems, we can see that there are differences when comparing the recordings with the cleat itself with the different situations of study, both globally and when analyzing each of the systems separately. Investigations have demonstrated that the propulsive force could be between 40–60% [34–36] of the total and that it can vary according to factors such as the position of the body, the load, rhythm and fatigue, and others [37]. The average of the power with the cleat was 122.87 and the average of the best position with the prototype was 143.2, a 16.5% increase. This is a significant value if we consider the high intensity of this sport. Some improvement is interesting if we are to obtain the objectives.

The lower limb is the main focus of attention as it is the lever in charge of generating power through muscular actions. Given the above, it is demonstrated that the modification of the rotational position of the cleat influences the power generated at the level of the pedal.

With respect to the kinematic parameters, it is also demonstrated that the modification of the position of the cleat generates significant changes in almost all the planes of the articulations studied. With respect to the pelvis, Bini et al. [29] studied the movement in the sagittal plane, finding values of 7.9–10.4° for non-professionals and of 9.1–10.3° for professionals, values higher than those recorded in our study. We obtained values that were more constant for each situation, which indicated that the pelvis is a point of stabilization in this sport. If there were differences in the values, it could be significant in that the Bioval system could be functioning in the wrong way. In the frontal plane of the pelvis, Carpes et al. [38] determined that the lateral-medial movements of the pelvis were between 1.2 and 2.6° according to the situation analyzed; values that were lower than ours in this study. Studies that quantified the transverse plane were not found. We underscore that, in our investigation, the movements of the pelvis in the three planes described similar amplitudes (5–6°). It is important because there are very few researches in this line, and it indicates that the pelvis is a point of stability that is most important to develop for the biomechanics of this sport. These values also indicated that the cyclists of this study did not have any alterations in the separation of their pelvis, a situation that is frequent in this sport.

Regarding the movement of the hips in the sagittal plane, García López et al. [1] found values of 41–46° for professionals and 40–49° for non-professionals, while those of Carpes et al. [38] were 36° and 31°, respectively. This difference in the results can be due to the reference points as, in our investigation, the point for quantifying the movement of the hip was the posterior superior iliac spine, while they took the horizontal as a reference. That is to say that, in a way, our reference was a mobile point as we have described before, with certain degrees being masked, while the reference of García López et al. [1] was static. On the other hand, our data are similar to those of Carpes et al. [38] as they marked the anterior superior iliac spine as a reference, the same as previously mentioned. At the level of the frontal plane, Carpes et al. [38] determined this movement as 4.5–5.9°, lower values than those that we found. This can be due their alluding to adduction, so we do not know if they only quantified this movement or whether they enunciated it in this way but quantified the complete range. With respect to the transverse plane, we found values similar to those described by Bini et al. [29] (10.5–12.7°).

At the level of the knee, there are more studies, given that this is the joint that suffers from more injuries in this sport. In our study, we found that the knee varied its range of movement in the sagittal plane from 71–72° based on the degrees of rotation of the cleat. These values are only similar to those recorded by the study of Carpes et al. [38] and are very far from those analysed by Ying Fang et al. [39]. The study of Yanci [40] recorded 68.68° for the left knee and Bailey et al. [41] 67.3°; that is to say, a difference of approximately 4°. We consider that the difference with that of Ying Fang et al. [39] (76.87–80.31°) is due to their using an ergometer on which the cyclist was placed quite far back, generating an increase in the extension. With respect to the frontal plane, the values were found to be higher than those of Bailey et al. [41] (3.7°) and Ruby et al. [42] (2.2 cm.). The knee represented degrees of rotation of 16–18.5°, while Carpes et al. [38] recorded 10° and Bini et al. [29] 18.6–24.1°. This demonstrates the difficulty that quantifying the degrees of this joint in this plane entails. The values we obtained for the frontal and transverse planes in this joint were similar, which led us to think that they were not unusual but that the rotational and varus–valgus movements of the knee were in sync, and most importantly, they were not affected either by the variations in the position of the cleat or by having used totally restricted cleats as was to be expected.

For the articulation of the knee in the sagittal plane, García López et al. [1] quantified the range as 15–31° for professionals and 13–36° for non-professionals. In Yanci's study [40], the values were 50.24° for the right limb and 34.14° for the left. Carpes et al. [38] found a range of 19° with oval gear ratios and 16° with normal ones. We note that the values at this level are very uneven, which could indicate that the articulation is what is most supported and adapts to the changes or situations. We have not found investigations that have quantified this in the frontal plane. Lastly, in the transverse plane, the movements were approximately 10°, which are very high movements taking into account that our pedal does not permit movements in this plane. We believe that these movements are due to the movement of the edge of the shoe and not of the shoe itself. That is to say, when placing the sensor on the tip of the shoe, these movements are due to the movement of the material more than to the movement of the foot itself. Carpes et al. [38] recorded a movement of 6° at the level of the foot in the same plane, similar to the value we obtained, in spite of their not indicating the type of pedal that they used.

If we compare all the results, we can see that the movements in the sagittal plane for the four joints studied are the ones that were most affected by varying the position of the cleat. This would make sense with the power data since the greatest range of motion in the sport occurs in this plane. The cyclists used pedals with some freedom in the transverse plane, which could be interpreted as instability and therefore control of it by the musculature. In this sense, part of the action would be used for something other than the sporting gesture itself and therefore result in loss of efficiency.

On the other hand, the kinematics in the transverse plane were altered at the ankle and pelvis level, the two points of stability that the cyclist has, and therefore we return to the idea of the previous paragraph. Some of that instability translates into muscle control through movement restriction. That restriction is achieved through muscular action. The displacements suffered by the ankle in this plane (10° approx.) would not make sense since the cleats of the prototype did not allow movement between the parts. This could be justified by the sensitivity of the system and the fact of having the sensor placed in the toe of the boot, which was flexible and susceptible to displacement of the material that did not come from the boot on the pedal.

Therefore, statistically significant differences were found both in the kinematics and in the recording of power when varying the rotation of the cleat.

5. Conclusions

The use of pedals without freedom of movement allows an increase in the power generated by the cyclist when compared with pedals that present freedom of movement between the shoe and the pedal. This idea is confirmed by the variation in the kinematics in the different joints of the lower limb, especially in the pelvis and foot. The increase in movement of these joints when compared to the initial situation suggests that the muscles do not perceive instability at the points of stability, and this allows the muscles to work better.

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