THE EFFECT OF BODY CONFIGURATION ON CYCLING PERFORMANCE

Danny Too
Department of Kinesiology
University of Illinois at Urbana-Champaign

Introduction

In the 1930's, Francois Faure, a relatively unknown racing cyclist, defeated the world champion Lemoire, in a 4 km pursuit race. What was unique aboubt this feat, is that Faure used a supine recumbent bicycle and broke track records that had been established on conventional bicycles. In 1980, the single rider Vector tricycle established a new human powered speed record at 56.66 mph (25.33 m/s) with the cyclist seated in a supine recumbent position.

It is well documented that recumbent human power vehicles are more effective aerodynamically than the standard cycling position (Kyle, 1974, 1982; Kyle & Caiozzo, 1986; Kyle, Crawford & Nadeau, 1973; 1974; Whitt & Wilson, 1982). With speeds of some human powered vehicles exceeding 60 mph (96.6 km/hr) (Gross, Kyle & Malewicki, 1983), it is obvious as to the importance of minimizing aerodynamic drag. However, when the drag coefficient and effective frontal area have been reduced in some human powered vehicles to 0.11 and 0.5 square feet, respectively (compared to 1.1 and 6.0 square feet, respectively for a standard upright bicycle) (Gross et al., 1983), it is questionable as to: 1) how much lower the aerodynamic drag can be reduced; and 2) how significant such changes would be. The design of human powered vehicles has focused exclusively on the aerodynamic properties of the vehicle with the cyclist. To further improve performance, it becomes necessary to focus on some aspect other than the aerodynamic properties. The most logical area to explore would be the human engine which powers the vehicles.

Review of Literature

The functional effectiveness of force production by a muscle is dependent upon the interaction of the muscle length at a particular joint angle and the muscle moment force arm length at that angle. Changes in joint angles which alter this interaction will affect the production of force, resulting in changes in performance. It has been documented with isometric contractions that there are joint angles which optimize: 1) force production (Kulig, Andrews, & Hay, 1984; Lunnen, Yack and LeVeau, 1981); 2) muscle moment arm length (Nemeth & Ohlsen, 1985; Pohtilla, 1969); and 3) muscle length (Elftman, 1966). However, the optimal joint angles which maximize performance in a dynamic task such as cycling have never ben clearly established.

Investigations with a standard racing bicycle have often manipulated only the height of the seat (Hamley and Thomas, 1967; Nordeen-Snyder, 1977; Shennum & DeVries, 1976; Thomas, 1967); or the crankarm length (Inbar, Dotan, Trousil, & Dvir, 1983) which alters both the hip and knee angles. It is then unknown as to whether improved cycling performance is attributed to changes in hip angles, knee angles or both; and what the most effective ranges of hip and knee angles are for one pedal revolution. Therefore, the purpose of this investigation was to determine the effect of changes in hip angels on cycling performance as measured by cycling duration and total work output.

Methods

Sixteen male subjects (21-35 years of age) were tested in five different body positions (0, 25, 50, 75 and 100 degrees), as defined by the angle formed between the seat tube and a vertical line. By rotating the seat to maintain a backrest perpendicular to the ground, a systematic decrease in hip angle (body configuration) from the 0 to 100 degree positions was induced. The mean hip angles corresponding to the 0, 25, 50, 75 and 100 degree seat tube angles were 130.9, 113.4, 100, 76.8, and 59.9 degrees, respectively. For each seat tube angle, the seat to pedal distance was adjusted to remain 100% (to within 3/4 of an inch or 1.905 cm) of each subject's total leg length, as measured from the greater trochanter of the femur of the right leg to the ground. All subjects were tested in each of the five positions on a Monark bicycle ergometer according to a pre-selected sequence of workloads and pedaling frequencies, with increments occurring every 3 minutes until exhaustion (Table 1).

Table 1

Bicycle Ergometer Test Protocol

Brake Load (kp)	Pedal Rate (rpm)	Time (min)	Work Rate (kpm/min)	(watts)	(hp)
1	60	3	360	58.9	.089
2	60	6	720	117.7	.158
3	60	9	1080	176.6	.237
3	70	12	1260	206.0	.276
3.5	70	15	1470	240.4	.322
4	70	18	1680	274.7	.368
4.5	70	21	1890	309.0	.414
4.5	75	24	2025	331.0	.444
5	75	27	2250	367.9	.493
5	80	30	2400	392.4	.526
5.5	80	33	2640	431.0	.579

Note: 1. Work Rate (power) = Brake Load x Pedal Rate
2. 1 HP = 746 watts = 4562.4 kpm/min

The testing sequences for the 5 positions were randomly selected for each subject with a minimum of 48 hours between test sessions. All subjects were strapped to the seat-backrest at the waist and hips, and toe-clips were used during all test sessions.

Results and Discussion

For each seat tube angle, the mean, minimum, and maximum angles, and range of motion at the hip, knee, and ankle were obtained for one complete pedal revolution (Table 2).

Table 2

Hip, Knee, and Ankle Angles at Five Seat Tube Angles

		Seat Tube Angle (deg)				
		0	25	50	75	100
Hip Angle (deg)						
Mean	Mean	130.9	113.4	100.0	76.8	59.9
	(SD)	(5.25)	(3.72)	(5.49)	(4.38)	(4.85)
Minimum	Mean	112.2	94.0	81.0	56.5	37.6
	(SD)	(5.64)	(4.24)	(5.94)	(4.46)	(5.38)
Maximum	Mean	149.6	132.8	119.1	97.1	82.2
	(SD)	(5.64)	(4.49)	(7.80)	(6.82)	(5.85)
Range	Mean	37.4	38.8	38.1	40.6	44.6
	(SD)	(6.78)	(4.58)	(8.47)	(7.46)	(5.68)
Knee Angle (deg)						
Mean	Mean	95.5	97.9	103.3	103.6	103.8
	(SD)	(6.42)	(5.36)	(4.00)	(5.58)	(8.04)
Minimum	Mean	62.7	62.2	65.1	65.7	67.5
	(SD)	(5.85)	(5.91)	(1.78)	(5.74)	(6.22)
Maximum	Mean	128.3	133.7	141.6	141.6	140.1
	(SD)	(8.83)	(6.57)	(6.93)	(6.52)	(10.83)
Range	Mean	65.6	73.9	77.0	75.2	72.6
	(SD)	(7.73)	(12.0)	(5.89)	(4.96)	(7.69)
Ankle Angle (deg)						
Mean	Mean	113.5	95.3	93.6	96.0	91.8
	(SD)	(6.47)	(6.30)	(7.90)	(5.48)	(9.23)
Minimum	Mean	91.8	87.4	87.1	88.8	83.4
	(SD)	(9.23)	(6.04)	(8.86)	(7.23)	(11.49)
Maximum	Mean	135.3	103.1	100.2	103.2	100.1
	(SD)	(12.02)	(8.00)	(8.00)	(7.19)	(9.24)
Range	Mean	43.6	15.8	13.2	14.5	16.1
	(SD)	(17.09)	(6.55)	(6.00)	(9.37)	(9.53)

As can be observed from Table 2, there is a systematic decrease in hip angle with changes in seat tube angles, whereas the knee and ankle angles are fairly similar across seat tube angles. It was found with repeated measures MANOVAs that there were significant differences (p , .01) in cycling duration and total work output with changes in body position (seat tube angle) and body configuration (mean hip angle). It is apparent, from observations of Table 3, Figure 1 and 2 that: 1) peak cycling performance, as measured by total work output and cycling duration occur in the 75 degree position with a mean hip angle of 76.8 degrees; and 2) a quadratic trend in cycling performance exists with changes in hip angles.

Table 3

Cycling Duration and Total Work Output at Five Hip Angles

		Mean Hip Angles (deg)				
		130.9	113.4	100.0	76.8	59.9
Cycling Duration (min)						
	Mean	9.50	12.05	15.01	16.03	13.43
	(SD)	(2.926)	(3.589)	(4.317)	(4.403)	(4.736)
Total Work Output (kpm)						
	Mean	7368	10737	15327	16968	13201
	(SD)	(3751)	(4852)	(7290)	(7656)	(8120)

Trend analysis was used to identify the function which best described the characteristics of the two performance variables with changes in cycling position. Dunnett's Multiple Comparison Test, was used as a post-hoc test to compare the hip angle in the 75 degree position with each of the other cycling positions. It was concluded from post-hoc tests that: 1) hip angles in the 75 degree position resulted in a significantly greater cycling duration ($p < .05$) than in all the other positions; 2) except for the 50 degree position, hip angles in the 75 degree position resulted in a significantly greater total work output than in all the other positions ($p < .01$); and 3) a second order function best describes the change in total work output and cycling duration with changes in hip angles ($p < .01$).

Based upon the results of this investigation, it is concluded that the optimal mean hip angle which maximizes cycling duration and total work output with incrementing workloads is approximately 77 degrees, with a minimum of 57 degrees, a maximum of 97 degrees, and a hip angle range of motion of 41 degrees. Therefore, it is suggested, with all other things being equal, that coaches and cyclists explore the possible use of these hip angles to enhance cycling performance.

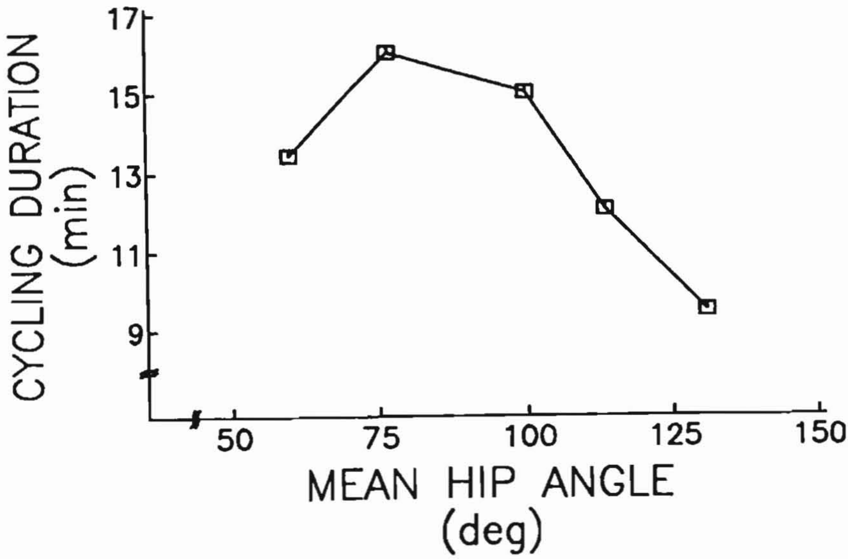

FIGURE 1: Cycling Duration with Changes in Body Configuration

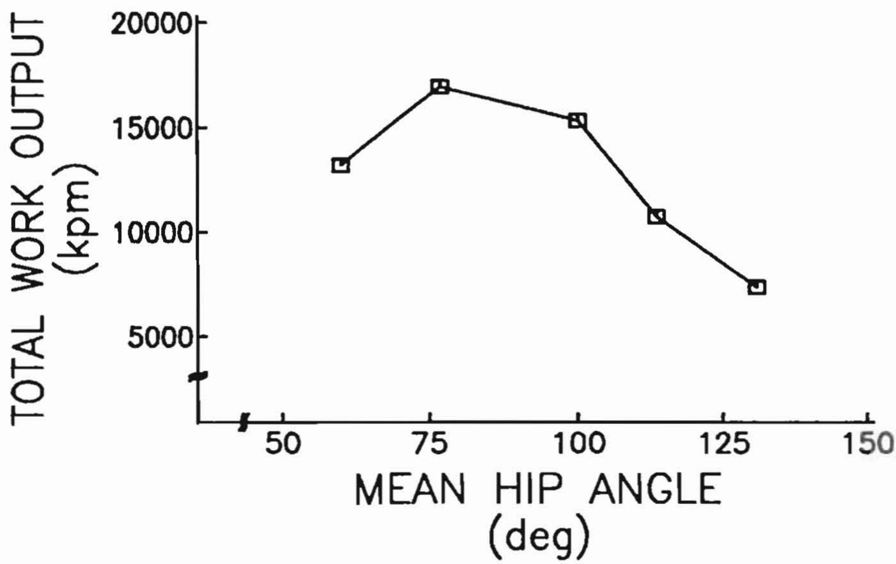

FIGURE 2: Total Work Output with Changes in Body Configuration

References

Elftman, H. (1966). Biomechanics of muscle: with particular application to studies of gait. *Journal of Bone and Joint Surgery*, 48-A, 363-377.

Gross, A.C., Kyle, C.R., & Malewicki, D.J. (1983). The aerodynamics of human-powered land vehicles. *Scientific American*, 249(6), 142-152.

Hamley, E.J., & Thomas, V. (1967). Physiological and postural factors in the calibration of the bicycle ergometer. *Journal of Physiology*, 191, 55-57P.

Inbar, O., Dotan, R., Trousil, T., & Dvir, Z. (1983). The effect of bicycle crank-length variation upon power performance. *Ergonomics*, 26(12), 1139-1146.

Kulig, K., Andrews, J.G., & Hay, J.G. (1984). Human strength curves. *Exercise and Sport Science Review*, 12, 417-466.

Kyle, C.R. (1974). The aerodynamics of man powered land vehicles. *Proceeding of the Seminar of Planning, Design, and Implementation of Bicycle Pedestrian Facilities*, pp. 312-326. San Diego, CA

Kyle, C.R. (1982). Go with the flow: aerodynamics and cycling. *Bicycling*, 23(4), 59-60, 62, 64-66.

Kyle, C.R., & Caiozzo, V.J. (1986). Experiments in human ergometry as applied to the design of human powered vehicles. *International Journal of Sport Biomechanics*, 2, 6-19.

Kyle, C.R., Crawford, C., & Nadeau, D. (1973). Factors affecting the speed of a bicycle. *CSULB Engineering Report No 73-1.*

Kyle, C.R., Crawford, C., & Nadeau, D. (1974). What affects bicycle speed? Part I. *Bicycling*, 5 (7), 22-24.

Lunnen, J.D., Yack, J., & LeVeau, B.F. (1981). Relationship between muscle length, muscle activity, and torque of the hamstring muscles. *Physical Therapy*, 61, 190-195.

Nemeth, G., & Ohlsen, H. (1985). In vivo moment arm lengths for hip extensor muscles at different angles of hip flexion. *Journal of Biomechanics*, 18, 129-140.

Nordeen-Snyder, K.S. (1977). The effect of bicycle seat height variation upon oxygen consumption and lower limb kinematics. *Medicine and Science in Sports*, 9, 113-117.

Pohtilla, J.F. (1969). Kinesiology of hip extension at selected angles of pelvifemoral extension. *Archive of Physical Medicine and Rehabilitation*, 50, 241-250.

Shennum, P.L., & DeVries, H.A. (1976). The effect of saddle height on oxygen consumption during bicycle ergometer work. *Medicine and Science in Sports*, 8, 119-121.
Thomas, V. (1967). Scientific setting of saddle position. *American Cycling*, 6(4), 12-13.
Whitt, F.R., & Wilson, D.G. (1982). *Bicycling Science*, (2nd ed.). Cambridge: M.I.T. Press.

This investigation was partially funded by a grant-in-aid of research from Sigma XI, The Scientific Research Society.

SELECTED METABOLIC AND HEMODYNAMIC RESPONSES TO REPEATED STEADY-STATE BOUTS OF INDOOR CYCLING, UTILISING MARGINAL INCREASES IN MECHANICAL POWER OUTPUT: CONSIDERATIONS FOR THE EVALUATION OF INDIVIDUAL COMPETITIVE ROAD CYCLISTS USING A PORTABLE ON-BICYCLE COMPUTER

Douglas W. Briggs
Frank E. Fedel
James A. Foulke, Charles B. Woolley
Paul D. Stein, Hani N. Sabbah
Lawrence D. Rink, Edward Getts, David Burnworth

Introduction

It has been demonstrated by Sanderson, Cavanaugh et al. (1985), and the authors, (1987), that impulse and average net power distributions (W) generated about the pedal spindle and crank arms, vary with individual cyclists, either creating a mechanically desirable circular cycling pattern where the impulse is 'smoothed', or a 'butterfly' distribution indicating unequal force distribution(s) throughout each pedaling cycle.

Based on research performed indoors by Cavanaugh (1985), and Anderson (1986), and this group outdoors at the United States Cycling Federation Camp in Colorado in 1987 and 1988, it appears that techniques employed to reduce the counter-propulsive tangential crank arm forces could possible improve average net power magnitudes produced by individual elite cyclists outdoors during competition, and thus improve their overall time(s) recorded for selected events.

Telemetric feedback of both the magnitude(s) and location of the negative force component(s) within each leg's pedaling revolution, and display of this information on a handlebar mounted devise is possible with current technology. Appropriate technology could be designed to improve cycling techniques of individuals should continued research confirm the importance of these parameters.

Our group felt it appropriate to design a portable sprocket computer to measure the instantaneous force pattern for every second or $2n^{th}$ crank revolution for use both indoors and outdoors, in an attempt to modify existing cycling patterns of elite cyclists if warranted.

The main purpose of the research was to investigate the relationship(s) between mechanical power output (Watts), cardiac output (L/min), heart rate (bpm), and selected non-invasive estimates of ventilatory threshold (V.T.) in an attempt to prescribe repeated 20-30 minute tempo workouts for elite cyclists and professional triathletes.

A secondary purpose was to document and investigate anticipated heart rates, metabolic and bloodflow response pattern differences for individual cyclists betwen 30 minute indoor bouts of steady-state cycling versus 30 minutes of variable leg speed cycling. The latter were performed to simulate outdoor road racing where bursts of speed are required to remain with the leading pack of riders. These bouts of exercise were equivalent to ventilatory threshold interval workouts.

Initially, 35 road cyclists, including two National champions, cycled to exhaustion using a 4 gear size by 3 leg speed combination of power outputs using a continuous incremental workload $V0_2$ with V.T. protocol designed to exhaust subjects in 6 minutes following a 5 minute steady-state warm up at leg speeds betwen 90 and 120 RPM. Variable gear selection was possible during the test, with the rear portion of each subject's bicycle frame attached to a momentum trainer, the Road Machine.

Blood gases were sampled every 15 seconds using a Sensormedics Horizon TM Metabolic Cart programmed in the average mode using measured $F10_2$.

Pedal force patterns were measured with a custom-built self-contained data acquisition package. Low power microprocessor technology implemented a minimum component system. In addition to the processor circuit memory a four channel analog to digital converter and a strain gauge amplifier and sampling control circuit was used. The complete force pattern data acquisition system is contained in the crank sprocket space.

Pedal forces were measured by transducing bending moments produced in the crank arms with pairs of strain gauges installed on the cranks. The strain gauges provide signals measuring the tangential and lateral pedal force components in each crank arm. Amplified strain gauge signals were sampled by an analog to digital converter and coded to an eight bit measurement of pedal forces. Data sampling was indexed to eight positions by the sampling index systems. Minimum power consumption was achieved by activating the processor only on an event to be samples as keyed by the pedal position index system. Memory capcity was managed by sampling and storing every $2n^{th}$ crank rotation where 'n' was selected according to the anticipated length or duration of ride(s) indoors or outdoors.

For this part of the research, the magnitude of maximum power output (W) and duration of ride(s) to exhaustion attained by the top four performers in the group tested at preferred pedal frequencies of 110-120 RPM and power outputs of 315-460 W, and designated T4C, exceeded the power outputs earlier reported by U.S.O.C. when testing top male and female individual pursuiters.

VO_2 max. values for T4C were greater than 80 $ml.kg.min^{-1}$; some of the highest reported laboratory values for race cyclists. The mean ventilation frequency for T4C of 67.6 breaths/min. was the determining factor in attaining high VE max. values of 194-240 L/min. The ventilatory response pattern for T4C was such that during the initial 5 minute warm up phase, VE was dependent on tidal volume rather than on breath frequency. Subsequently, when both leg speed and gear size were increased simultaneously, this situation reversed, increased in VE becoming frequency dependent. The highest positive correlations between VE and f occurring at power outputs of 315-460 Watts ($r = .86/.91/.88$).

At power outputs of 315-460 W, T4C subjects obtained consistently higher percentage increments in V.E. and H.R. which were matched by similar $ml.kg.min^{-1}$ increments in VO_2. T4C values for non-invasive estimates of V.T. were consistently high. Less consistency was observed for gas exchange estimates of VO_2 max. The greatest number of significantly high positive zero-order correlations greater than .94 occurred at 460 Watts, with the greatest number of significantly high positive correlations for T4C occurring at power outputs of 315-460 Watts. At 460 Watts the relationships between VE, f, O_2VE, CO_2 VE, VO_2 ($ml.kg.min^{-1}$), and heart rate, appeared optimal for T4C.

Subsequently, one member of the United States Junior World's Road Cycling Team, performed two 30 minute indoor rides using the same equipment earlier mentioned plus an exercise doppler.

Continuous non-invasive measurement of bloodflow during rest (pre-warm up, seated on the bicycle), and exercise was obtained.

The continuous wave doppler (Quinton Exerdop) was used to obtain peak accelerations (m/s^2 and velocities (m/s) of blood ejected into a varying diameter aorta, and cardiac output (L/min), stroke volume (% increase above resting value(s); end-diastolic volume, peak flow (L/min), ejection franction(s) (EDV - ESV), heart rate (bpm), and systolic time ratios (peak ejection time/ventricular ejection time all measurements of ventribular function) were recorded using a BoMed NCCOM3-R7 thoracic electrical impedance (TEP) monitor. Changes in the electrical conductivity of the thorax due to the pulsatile flow of blood through the segment, provided a basis for the TEB technology. Ejection fraction is calculated from the measurement of systolic time intervals (STIs) derived from ECG and TEP waveforms.

Results

An eight inch increase in gear size (89"/81" steady state ride - 'SSR') combined with an increase in average leg speed of 14=18 RPM (110-114 vs 96 RPM attained during 2 minute 'bursts', produced noticeable differences in the following parameters.

A. Mechanical power output - 311W for bursting ride vs 220 W for SSR. For the 89 inch gear, the mean power output (W) generated during the 2 minute burst activity exceeded that value produced during the 4 minute steady state bouts by 50 percent/101 W for an average increase in leg speed of 10% (95-105 RPM). Maximum power generated during the final burst activity was 420 Watts, equivalent to a 114 RPM/ 3.7 kp. friction-braked ergometer force-velocity combination.

B. Metabolic Cost - Significant increases occurred in favor of the 89 inch gear in:

1. $V0_2$ as a percentage of maximum = 67% max (53 mls/kg-min) during 4 minute periods, and 73% max. (58 mls/kg-min during 2 minute bursts, versus 40-48 mls/kg-min (50-50% max) for the 81 inch gear. VE was considerably higher for the burst activity run, reaching an average of 97 L/min for the 4 minute periods and 115 L/min during the 2 minute bursts, compared with an average value of 55-77 L/min for the steady state 81 inch gear ride.

2. Heart Rate (bpm) — Average rate for the 81 inch gear was 166, and 179 for the 89 inch bursting run. Maximum heart rates differed considerably; 203 for the 89 inch gear and 173 for the 81

inch gear. Significantly, the average rate attained for the 2 minute bursting activity was 193 RPM, equivalent to 93% of maximum.

3. Bloodflow response — Significant increases in cardiac output occurred above resting on-bicycle levels for both gear sizes (43 L/min vs 4-6 L/min for burst activity, and 23 vs 2.8 L/min for 81 inch gear). During the burst activity, stroke volume increased by 140% above resting level; end-diastolic volume increased by 200%, and the ejection fraction incrased by an average of 14-16% above resting level. For the 89 inch gear, during 4 minute exercise bouts the average cardiac output was 6-11 L/min, but this increased to an average value of 32-43 L/min during the 2 minute bursts, an average of 4-5 times resting value.

Conclusions

Within the limitations of the study an 8 inch increase in gear size, and a 14+18 RPM burst-induced average increase in average leg speed resulted in 50% increase in average net power production (W), and induced significant increases in heart rate, ventilation rate (breaths/min) respiratory minute volume (L/min), cardiac output (L/min above resting value), percent utilization of $V0_2$ max, and blood lactate (6.6 mMols 5 min post exercise 89 inch gear).

In terms of force production, the portable indoor-outdoor computer designed by members of the group, adequately highlighted individual left and right leg contributions to the total power output and the variations in velocity-dependant force for each leg separately produced at various power outputs.

The data gathered supports the premise that subtle changes in leg speed matched with variable gear selection has considerable effect on the heart rate, cardiac output, ventilatory, and lactate response patterns. If we are to improve individual time(s) for given differences requiring cardiovascular conditioning, to changes in mechanical power output/leg speed/gear size optimal pairings.

References

Briggs, D.W., Fedel, F., Foulke, J., Woolley, C. (1987). Measurement of competitive cycling performance using an on-board computer. *Proceedings of ISB*, Amsterdam, 1987.

Cavanaugh, P.R., Sanderson, D.J. (1985). The biomechanics of cycling; studies of the pedaling mechanics of elite pursuit riders. In E. R. Burke (eds) *Science of Cycling*, 5. 91-122.

Coast, J.R., and Welch, H.G. (1985). Linear increases in optimal pedal rate with increased pwer output in cycle ergometry. *European Journal of Applied Physiology and Occupational Physiology.* 53(4), 339-342.

Daly, D.J., & Cavanaugh, P.R. (1976). Assymmetry in bicycle pedaling. *Medicine and Science in Sports*, 8, 204-208.

Davis, R.R., and Hull, M.L. (1981). Measurement of pedal loading in bicycling: II analysis and results. *Journal of Biomechanics*, 14(12). 857-872.

Goldberg, S.J., Allen, Hugh D., Marx, Gerals, R., and Donnerstein, Richard L. (1988). *Doppler Echocardiography* 2nd Ed. Lea & Febiger.