Journal of Applied Biomechanics, 2011, 27, 297-305

Influence of Bicycle Seat Tube Angle and Hand Position on Lower Extremity Kinematics and Neuromuscular Control: Implications for Triathlon Running Performance

Amy Silder, Kyle Gleason, and Darryl G. Thelen
University of Wisconsin–Madison

We investigated how varying seat tube angle (STA) and hand position affect muscle kinematics and activation patterns during cycling in order to better understand how triathlon-specific bike geometries might mitigate the biomechanical challenges associated with the bike-to-run transition. Whole body motion and lower extremity muscle activities were recorded from 14 triathletes during a series of cycling and treadmill running trials. A total of nine cycling trials were conducted in three hand positions (aero, drops, hoods) and at three STAs (73°, 76°, 79°). Participants also ran on a treadmill at 80, 90, and 100% of their 10-km triathlon race pace. Compared with cycling, running necessitated significantly longer peak musculotendon lengths from the uniarticular hip flexors, knee extensors, ankle plantar flexors and the biarticular hamstrings, rectus femoris, and gastrocnemius muscles. Running also involved significantly longer periods of active muscle lengthening from the quadriceps and ankle plantar flexors. During cycling, increasing the STA alone had no affect on muscle kinematics but did induce significantly greater rectus femoris activity during the upstroke of the crank cycle. Increasing hip extension by varying the hand position induced an increase in hamstring muscle activity, and moved the operating lengths of the uniarticular hip flexor and extensor muscles slightly closer to those seen during running. These combined changes in muscle kinematics and coordination could potentially contribute to the improved running performances that have been previously observed immediately after cycling on a triathlon-specific bicycle.

Keywords: running, triathlon, bike geometry, electromyography, cycling

A triathlon involves the consecutive events of swimming, cycling, and running. Although race performance depends upon success in all three disciplines (Hue et al., 1998), the best predictor of an athlete's overall finish is the time to complete the run segment (Dengel et al., 1989). When compared with an isolated run, a preceding bout of cycling has been shown to significantly reduce running speed (Bernard et al., 2003; Hausswirth et al., 1997, 1999; Millet, Millet et al., 2000; Millet & Vleck, 2000), particularly in the early stages of the run (Hausswirth et al., 1997).

It is believed that riding a triathlon-specific bicycle, which is characterized by aero bars and a steep seat tube angle (STA) may influence subsequent running performance. For example, a previous study found that triathletes exhibited faster running speeds after completing a 40-km ride on a stationary bike configured with an 81° STA, compared with a more traditional 73° STA (Garside et al., 2000). However, the underlying mechanisms by which STA can affect running performance are not well understood. One hypothesis is that increasing the STA positions the pelvis further forward relative to the crank, thereby increasing activation of the hamstrings and gluteus muscles (Garside et al., 2000; Heil et al., 1995; Price et al., 1997). Such a change in coordination may allow for less power fluctuation across the crank cycle (Coyle et al., 1988; Garside et al., 2000; Heil et al., 1995) and thus reduce quadriceps fatigue before running. A second explanation is that a triathlon-specific bike allows for greater hip extension (Garside et al., 2000; Hausswirth et al., 1996; Hue et al., 1998). This may enable lower extremity muscles to operate at lengths more comparable to those seen during running, thereby easing the neural and biomechanical adjustments when transitioning from cycling to running.

The purpose of this study was to first characterize differences in muscle kinematics and activation patterns between cycling and running. We then investigated how

Amy Silder is with the Department of Biomedical Engineering, University of Wisconsin–Madison, Madison, WI. Kyle Gleason is with the Department of Mechanical Engineering, University of Wisconsin–Madison, Madison, WI. Darryl G. Thelen *(Corresponding Author)* is with the Departments of Biomedical Engineering, Mechanical Engineering, and Orthopedics and Rehabilitation, University of Wisconsin–Madison, Madison, WI.

varying STA and hip flexion angle affect kinematics and muscle activation patterns during cycling to better understand how a triathlon-specific bike might mitigate the biomechanical challenges associated with the bike-to-run transition.

Methods

Fourteen competitive triathletes (9 male, 5 female; 32 ± 9 years; 1.78 ± 0.07 m; 69 ± 9 kg) provided written informed consent to participate in this study, in accordance with a protocol approved by the University of Wisconsin's Health Sciences Institutional Review Board. Cycling and running speeds were based on each participant's 40-km cycling and 10-km running performance, which are representative of the distances performed in an Olympic distance triathlon. Average cycling speed was 9.2 ± 0.7 m/s and average running speed was 3.9 ± 0.5 m/s. These speeds correspond to a 40-km bike time of 1 hr 12 min and a 10-km run time of 43 min.

The cycling portion of the protocol was conducted on a fully adjustable cycle ergometer secured to a fluid resistance unit (Travel Trac 3; Performance Bicycle; Chapel Hill, NC, USA) that was outfitted with a custom flywheel. The ergometer was equipped with a power tap on the rear hub (PowerTap; Saris Corp., Madison, WI, USA), 175-mm crank arms, drop handlebars, and adjustable clip-on aerobars (Profile Design Carbon Stryke; Long Beach, CA, USA) (Figure 1). All participants rode on the same type of saddle (Bontrager Inform RL; Waterloo, WI, USA), with saddle size selected based on each participant's ischial tuberosity width (Bontrager Saddle Sizer; Waterloo, WI, USA). The saddle was centered over the seat tube and adjusted so that it was level with respect to ground. We determined the STA on each participant's personal bicycle and found that six subjects rode a 72 or 73° frame, two rode a 76° frame, and five rode a 78° or 79° frame. We also measured the radial distance from the crank axis to the top of the saddle, and the radial distance from the crank axis to the top center of the stem. During testing, the STA was adjusted on the ergometer in such a way that maintained these radial distances, which were specific to each participant's personal bicycle.

Desired steady-state cycling power was estimated for each participant using a mathematical model that accounts for bicycle velocity, air resistance, and rolling resistance (Martin et al., 1998). Model inputs included average 40-km triathlon race speed, an assumed wind speed (v_w = 3.22 km/hr), air density (ρ = 1.223 kg·m^{-3}), incremental drag coefficient from the spokes (F_w = 0.0044), coefficient of rolling resistance (C_{rr} = 0.0032), and percent grade (0.3%). The drag coefficient (C_d) and frontal area of the rider (A) were estimated from body mass (Heil, 2001). For all trials, the participant was asked to pedal at a fixed cadence of 90 rpm, which was maintained using a metronome and visual feedback from the power tap. The desired power output was obtained by adjusting the rear gear such that the average power output reading was equal to the desired power. This gear was then maintained for the remainder of the testing. Average estimated power output across all participants was 169 ± 37 W. Post hoc evaluation of the kinematic data demonstrated that the average cadence of the riders had a variance of <1 rpm across all nine trials.

Data were collected for three hand positions at each of three STAs (73°, 76°, and 79°). Hand position was varied between the aero, drops, and hoods positions (Figure 1), which allowed us to alter hip flexion angle independently of STA. The seat tube angle was

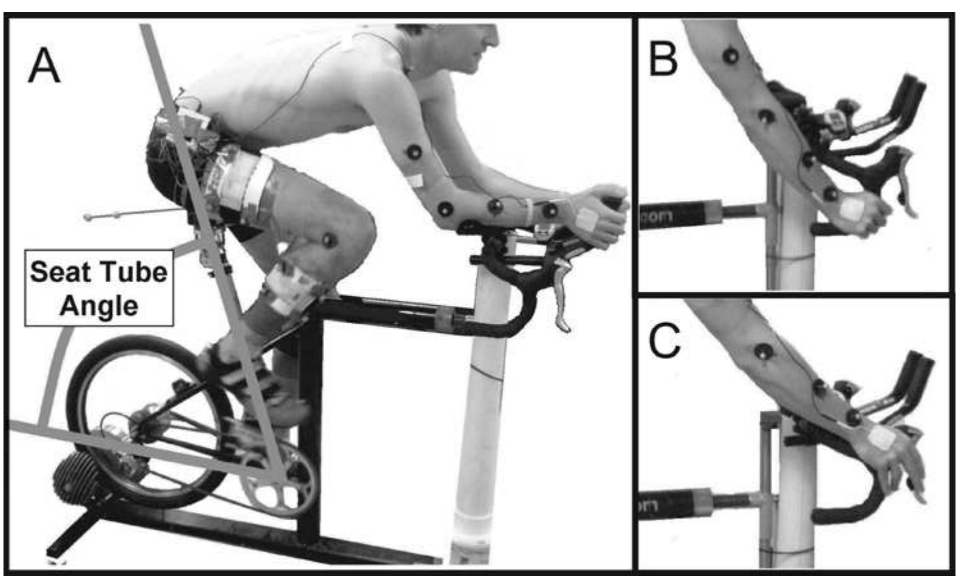

Figure 1 — Participants rode on a modified stationary ergometer at three seat tube angles (73°, 76°, 79°) in three common hand positions: (A) an aerodynamic position using aerobars, (B) the drops position, and (C) with their hands placed on the brake hoods.

manipulated in a way that maintained the radial distance between the crank axis and the saddle, and between the crank axis and the top center of the stem. This changed the orientation of the cyclist with respect to gravity, but theoretically could allow the cyclist to ride with the same hip, knee, and ankle kinematics (Rankin et al., 2010). The testing order of hand positions was randomized at each fixed STA, and each STA condition was tested before adjusting the ergometer to a new STA. Participants were given approximately 5 min of warm-up and familiarization each time STA was adjusted. Data were collected for approximately 15 s in each hand position, after the participant achieved a steady cadence.

The running portion of the test consisted of treadmill running at three speeds: 80, 90, and 100% of each participant's 10-km triathlon race pace. Following adequate warm-up and familiarization, the three speeds were collected in random order, with approximately 15 strides collected for each speed.

Whole-body kinematics were recorded during all cycling and running trials using an eight-camera passive marker system (Motion Analysis Corporation, Santa Rosa, CA, USA). Nineteen markers were placed over anatomical bony landmarks, and an additional 14 tracking markers were attached to plates that were strapped tightly to the thigh and shank segments. The markers remained in place for all cycling and running trials, with the exception of foot markers that were adhered to the participants' cycling and running shoes. The angular orientation of the pedals and crank were measured using three markers placed on each pedal. Following cycling, and before running, a set of two calibration trials were obtained with the participant standing upright. In addition, a functional hip joint center algorithm was implemented (Piazza et al., 2004) to estimate the hip joint center location in the pelvis reference frame. Marker kinematics recorded during the calibration trials were used to generate a scaled lower extremity musculoskeletal model (Delp et al., 1990). A 6-degree-of-freedom (*df*) pelvis was the base segment, and each lower limb included a 3-*df* hip, a 1-*df* knee, with translations and nonsagittal rotations defined as a function of knee flexion (Walker et al., 1988), and a 2-*df* ankle with nonintersecting talocrural and subtalar joints (Delp et al., 1990).

For each frame of a motion trial, we used a global optimization inverse kinematics routine to compute three-dimensional pelvic position and orientation, and lower extremity joint angles that minimized the discrepancy between measured marker positions and body segment markers (Lu et al., 1999). Joint angles and musculotendon lengths were computed based on a kinematic model of the lower extremity musculoskeletal system (Delp et al., 1990). Musculotendon lengths were computed as the distance from muscle origin to insertion, with wrapping about bone segments accounted for (Delp et al., 1990). Musculotendon lengths were subsequently normalized to the length from a standing posture. Joint angles and musculotendon lengths were computed separately for the right and left limbs.

For each trial, data were interpolated and then averaged across 10 successive crank or gait cycles. The start and end of each crank cycle was defined as the position at which the right crank was aligned with the vertical axis. The running gait cycle was defined by two successive foot contacts of the same limb, as determined from the treadmill vertical ground reaction forces (Bertec Corporation; Columbus, OH, USA).

Muscle activities were recorded on the right limb from the rectus femoris (RF), vastus lateralis (VL), biceps femoris (BF), medial hamstrings (MH), soleus (SOL), and medial gastrocnemius (GAS). Signals were recorded at 2000 Hz using preamplified single differential electrodes (DE-2.1, DelSys, Inc, Boston, MA, USA) interfaced with an amplifier/processor unit (CMRR > 85 dB at 60 Hz; input impedance > 100 MΩ). The EMG signals were band-pass filtered at 20–500 Hz and full wave rectified. Muscle activity data were then divided into 10 gait or crank cycles. The root mean square (RMS) activity across an entire crank or gait cycle was then normalized to the mean RMS signal from the 100% (10-km race pace) running speed (Schmitz et al., 2008; Yang et al., 1984). The onset, offset, and duration of muscle activity, relative to a gait or crank cycle, were manually determined (Li et al., 2005) and used to determine the average length of the musculotendon when it was active. The duty cycle for each muscle was computed as the duration of the muscle activity normalized to either the duration of the gait cycle (running) or pedal stroke (cycling). The active shortening period was defined as the percentage of the activation time that the muscle was shortening.

The following kinematic variables were compared between cycling conditions and across running speeds: maximum and minimum sagittal pelvis, hip, knee, and ankle angles; active lengths, peak lengths, and excursions of the gluteus maximus (GMAX), psoas, RF, VL, semitendinosus (referred to as *MH*), biceps femoris long head (referred to as *BF*), GAS, and SOL musculotendons. Differences in neuromuscular coordination between cycling conditions and running speeds were investigated by comparing RMS activity, duty cycle, and active shortening periods.

A three-factor (limb, STA, hand position) repeated-measures ANOVA was used to statistically compare cycling kinematic measures, and a two-way (STA, hand position) repeated-measures ANOVA was used to compare cycling EMG activities. For running data, a two-factor (limb, speed) ANOVA was used to compare kinematic measures and a one-way ANOVA was used to compare EMG activities. There were no significant bilateral differences or interactions, such that significant main effects for running speed, hand position, and STA were followed up with Tukey's post hoc pairwise comparisons. Since small changes in cycling cadence could influence the magnitude of muscle activity, average cadence was used as a covariate when assessing RMS muscle activities across cycling conditions. Running measures from the 10-km race pace were compared against cycling trials using paired *t* tests.

Results

Running

Stride rate and stride length averaged 174 ± 10 steps/min and 5.6 ± 0.07 m respectively at 10-km race pace. All peak lower extremity joint angles (except knee extension) increased significantly with speed (Table 1). Average RMS muscle activities also increased significantly with speed for the RF, BF, MH, SOL, and GAS, but not the VL.

Running vs. Cycling

Peak hip extension, knee extension, knee flexion, ankle dorsiflexion, and ankle plantar flexion angles were all significantly greater during running than cycling. These differences necessitated significantly larger excursions (Figure 2) and greater peak musculotendon lengths (Table 2) for the uniarticular VL and SOL and the biarticular RF, hamstrings, and GAS. Greater peak musculotendon, but not excursions, were observed for the uniarticular psoas during running (Figure 2). Only the uniarticular

Table 1 Mean (*SD*) peak joint kinematics for each running speed tested, with significant speed effects denoted

	Running Speed (% Race Pace)			
Joint Angle (°)	**80%**	**90%**	**100%**	***p*-value**
Max Anterior Pelvic Tilt	12 (5)	13 (5)	13 (5)	0.01 [c]
Max Hip Flexion	53 (8)	56 (8)	60 (8)	<0.01 [a,b,c]
Max Hip Extension	6 (5)	7 (6)	9 (5)	<0.01 [a,b,c]
Max Knee Flexion	105 (14)	109 (14)	114 (12)	<0.01 [a,b,c]
Min Knee Flexion	10 (4)	9 (3)	9 (3)	
Max Ankle Dorsiflexion	28 (4)	29 (4)	29 (4)	<0.01 [c]
Max Ankle Plantar Flexion	23 (7)	24 (7)	25 (6)	<0.01 [a,b,c]

[a]80 ≠ 90%, [b]90 ≠ 100%, [c]80 ≠ 100%.

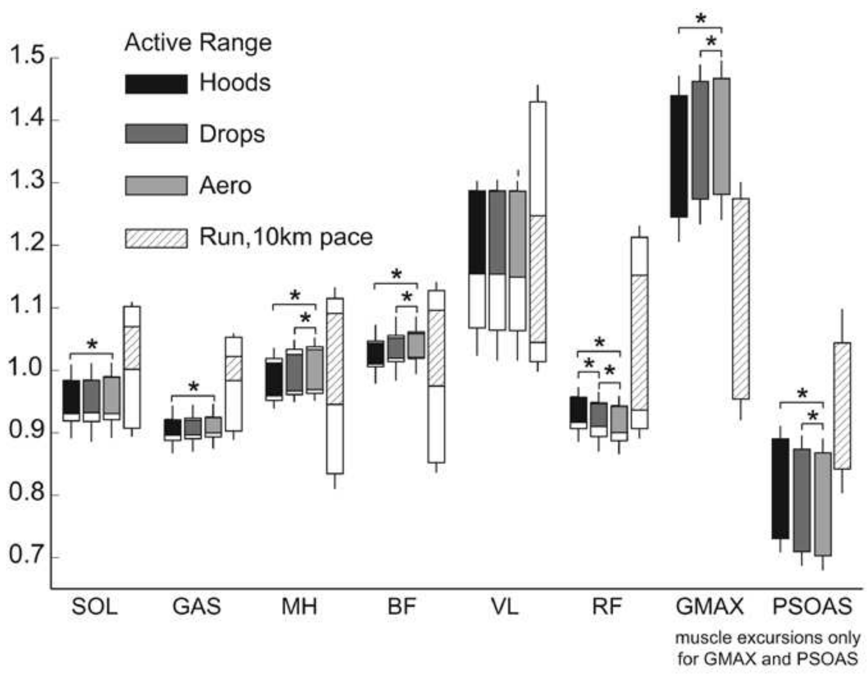

Figure 2 — Normalized musculotendon (MT) excursions and range of active lengths for running at 10-km race pace and for cycling with the hands in the hoods, drops, and aero positions (averaged across all three STAs). For all muscles, MT excursions were significantly larger during running than cycling. Peak MT lengths were also larger for running than cycling for all muscles except the psoas. The increase in hip flexion when moving the hands into the hoods hand position induced a significant (**p* < .05) shift toward longer peak MT lengths in the rectus femoris (RF) and psoas muscles, while slightly decreasing peak lengths of the gluteus maximus (GMAX), medial hamstrings (MH) and biceps femoris (BF). Relatively small changes were seen with hand position in the soleus (SOL) and gastrocnemius (GAS). Note that electromyographic recordings were not obtained from the GMAX and psoas, such that the active ranges are not denoted for these muscles.

hip extensors (i.e., the GMAX) remained at consistently shorter musculotendon lengths during running (Figure 2).

Substantial differences were observed between running and cycling in the durations of active shortening (concentric) contractions. In particular, cycling involved significantly longer periods of active shortening from the RF, VL, MH, SOL, and GAS (Table 2). The duty cycle (percent time the muscle was active) was similar for running and cycling, with running showing small reductions in the duration of activity for the VL, BF, and GAS.

Seat Tube Angle Effects

Anterior pelvic tilt increased an average of 3° for each 3° increase in STA. However, there were no other significant differences in hip, knee, or ankle angles with variations in STA (Table 3), suggesting that the main kinematic effect of STA was a reorientation of body position with respect to gravity. Despite similar kinematics across STAs, RF muscle activity increased significantly with STA (Table 4), with primary activity observed during the upstroke of the crank cycle (Figure 3).

Hand Position Effects

A significant decrease in both anterior pelvic tilt and hip flexion, along with an increase in hip extension was observed when transitioning from the aero to drops hand position. Similar affects were observed when switching from the drops to hoods (Table 3, Figure 4). These hand position changes also induced a decrease in dorsiflexion and a slight increase in plantar flexion (Table 3). At the knee, only a small difference in knee extension between the aero and hoods hand position was measured.

The joint kinematic changes with hand position produced small, but significant, shifts in some musculotendon lengths. Specifically as hip extension increased, peak musculotendon lengths significantly increased for the psoas and RF, and decreased for the GMAX, MH, and BF muscles (Figure 2). The shift toward a more plantar flexed ankle in the hoods hand position also resulted in a significant decrease in peak GAS and SOL musculotendon lengths (Figure 2). Hand position also had a significant effect on hamstring muscle activity, with cyclists exhibiting the greatest amount of RMS activity when the hip was most extended in the hoods hand position (Table 4).

Table 2 Peak musculotendon (MT) lengths were significantly longer during running than cycling for all muscles from which muscle activities recorded. While muscle duty cycles were generally similar between cycling and running, running involved significantly greater periods of active lengthening in the soleus (SOL), medial gastrocnemius (GAS), vastus lateralis (VL), and rectus femoris (RF). Cycling values represent averages across all cycling conditions. Running values were from the 10-km race pace trials.

		SOL	GAS	MH	BF	VL	RF
Peak MT Length (%)	Cycle	99 (2)	92 (2)	103 (2)	105 (3)	129 (2)	95 (2)
	Run	107 (1)**	102 (1)**	109 (1)**	110 (3)**	137 (3)**	115 (2)**
Active MT Length (%)	Cycle	97 (3)	91 (2)	101 (2)	104 (3)	124 (3)	93 (2)
	Run	104 (1)**	100 (1)**	103 (2)*	105 (3)	114 (3)**	105 (2)**
Duty Cycle (%)	Cycle	54 (13)	54 (15)	53 (14)	56 (15)	54 (15)	52 (12)
	Run	53 (11)	50 (11) **	48 (9)	51 (9) **	49 (10)*	51 (11)
Active Shortening (%)	Cycle	68 (12)	51 (14)	43 (14)	50 (17)	72 (11)	72 (15)
	Run	40 (9)**	37 (5)*	57 (7)**	56 (7)	48 (9)**	29 (12)**

Note. Significant differences between cycling and running: $*p < 0.05$, $**p < 0.01$.

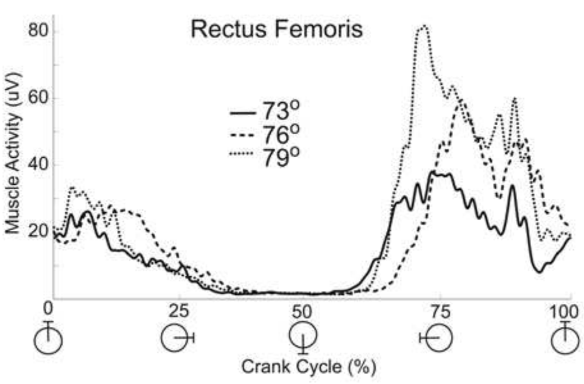

Figure 3 — Root mean square muscle activity for the rectus femoris increased significantly with seat tube angle. Shown is the ensemble-averaged rectus femoris muscle activity for each seat tube angle tested. The greatest differences were observed during the upstroke of the crank cycle (60–100% of the pedal stroke).

Discussion

Previous research suggests that riding a triathlon-specific bike frame may improve running speed in the early stages of a triathlon run (Garside et al., 2000). In this study, we quantitatively investigated the kinematic and neuromuscular differences between cycling and running. We also

Table 3 Mean (*SD*) joint kinematics for each seat tube angle (STA) and hand position (aero, drops, hoods) tested, with significant differences denoted

	Seat Tube Angle (STA)									*p*-value	
	73°			76°			79°				
Joint Angle (°)	Aero	Drops	Hoods	Aero	Drops	Hoods	Aero	Drops	Hoods	STA	Hand Position
Max Anterior Pelvic Tilt	29 (5)	27 (5)	20 (6)	32 (5)	30 (5)	23 (5)	36 (4)	33 (4)	27 (5)	<0.01 [a,b,c]	<0.01 [a,b,c]
Max Hip Flexion	102 (6)	99 (6)	93 (7)	101 (7)	99 (6)	94 (6)	102 (6)	100 (6)	95 (7)		<0.01 [a,b,c]
Min Hip Flexion	52 (8)	50 (7)	44 (8)	52 (8)	50 (7)	44 (8)	51 (7)	51 (7)	45 (7)		<0.01 [a,b,c]
Max Knee Flexion	106 (3)	105 (3)	105 (3)	106 (3)	105 (43)	105 (4)	105 (4)	105 (4)	105 (3)		
Min Knee Flexion	27 (9)	28 (9)	28 (9)	28 (9)	28 (9)	28 (9)	26 (9)	27 (9)	27 (9)		0.03 [a]
Max Ankle Dorsiflexion	18 (7)	17 (7)	16 (7)	18 (7)	17 (7)	16 (7)	19 (6)	17 (7)	16 (7)		<0.01 [a,b,c]
Max Ankle Plantar Flexion	9 (9)	9 (9)	10 (8)	8 (8)	8 (8)	9 (7)	9 (8)	10 (8)	10 (7)		<0.01 [c]

Note. STA effects: [a]73 ≠ 76°, [b]76 ≠ 79°, [c]73 ≠ 79°. Hand position effects: [a]aero ≠ drops, [b]drops ≠ hoods, [c]aero ≠ hoods.

Table 4 Mean (*SD*) muscle activities (normalized to activity at 100% running speed) across cycling conditions, with significant seat tube angle (STA) and hand position effects denoted

	Seat Tube Angle (STA)									p-value	
	73°			76°			79°				
Muscle	Aero	Drops	Hoods	Aero	Drops	Hoods	Aero	Drops	Hoods	STA	Hand Position
SOL	64.2 (35.7)	65.0 (37.6)	60.2 (30.6)	70.5 (43.0)	69.0 (40.0)	70.1 (45.0)	73.1 (43.4)	69.5 (43.6)	72.2 (50.6)		
GAS	94.7 (55.3)	95.5 (54.9)	94.7 (57.2)	86.5 (47.7)	88.8 (53.1)	90.8 (51.1)	88.4 (42.6)	87.5 (42.0)	90.4 (48.8)		
BF	40.7 (18.7)	43.5 (54.9)	42.8 (57.2)	40.0 (22.7)	44.1 (22.0)	47.6 (27.0)	38.6 (18.9)	43.5 (20.4)	43.6 (19.5)		0.01 [a,c]
MH	43.5 (20.8)	45.8 (20.2)	43.4 (20.2)	39.4 (16.7)	46.2 (16.7)	45.0 (20.9)	36.9 (14.1)	43.6 (14.7)	42.9 (15.8)		0.01 [a,c]
VL	86.6 (37.6)	88.2 (37.9)	85.9 (36.1)	89.1 (36.1)	85.8 (34.0)	89.2 (36.0)	87.5 (32.1)	83.5 (33.0)	88.0 (34.1)		
RF	101.5 (62.3)	103.9 (58.7)	90.7 (56.0)	133.3 (63.4)	119.2 (69.5)	122.3 (60.5)	143.7 (60.0)	119.5 (56.4)	124.5 (59.0)	<0.01 [a,c]	

Note. STA effects: [a]73 ≠ 76°, [b]76 ≠ 79°, [c]73 ≠ 79°. Hand position effects: [a]aero ≠ drops, [b]drops ≠ hoods, [c]aero ≠ hoods. Muscle abbreviations: soleus (SOL), gastrocnemius (GAS), biceps femoris (BF), medial hamstrings (MH), vastus lateralis (VL), rectus femoris (RF).

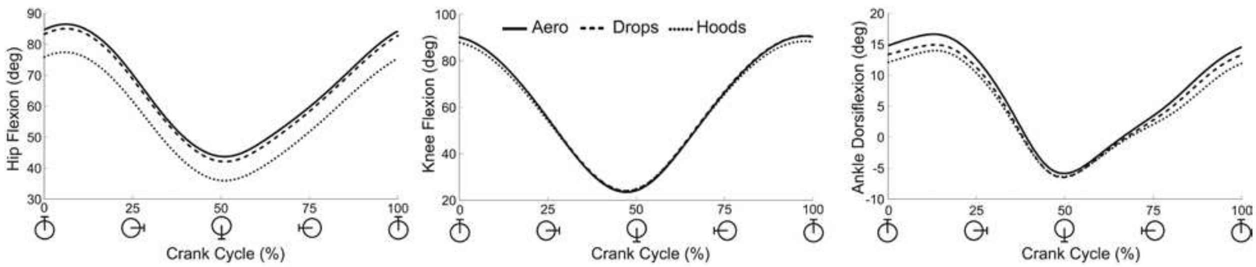

Figure 4 — Ensemble-averaged sagittal hip flexion, knee flexion, and ankle dorsiflexion angles across the aero, drops, and hoods hand positions. Hand position induced significant changes at the hip and ankle. The hip became significantly more extended when transitioning from the aero to drops and again from the drops to hoods hand positions. Ankle dorsiflexion was the greatest in the aero hand position, when the hip was most flexed.

systematically varied both STA and hand position to identify differences in cycling kinematics and neuromuscular coordination patterns, which could affect running performance after a bout of cycling. Our results show that running tends to require much greater muscle excursions, peak muscle lengths, and longer periods of active lengthening contractions than cycling. Increasing STA in isolation did not alter lower extremity kinematics, but did induce a significant increase in RF activity. Manipulating hand position from aero to hoods induced significant changes in hip and ankle kinematics, greater hamstring muscle activity, and moved the operating lengths of the uniarticular hip flexor and extensor muscles closer to that seen in running. Together, these combined effects on lower extremity kinematics and coordination may affect running performance after a fatiguing cycling bout.

It is commonly stated that the bike-to-run transition in triathlon competition is challenging because different muscles are used between the two activities. The results of this study clearly show that the same lower extremity muscles are active in both cycling and running, but substantial differences exist in the operating range and biomechanical demands placed on those muscles. Interestingly, the changes in muscle lengths that can actually be achieved by manipulating STA and hand position are relatively minor compared with changes in muscle lengths that occur when transitioning from cycling to running (Figure 2). The most salient effect was observed in the uniarticular hip flexors, with the psoas shifting toward substantially longer lengths when riding with greater hip extension (i.e., hoods) (Figure 2). During running, a nearly full range of hip extension is used at toe-off (Riley et al., 2010) such that tight hip flexors could diminish stride length. At the tissue level, viscoelastic affects necessitate a conditioning period before tendons reach steady-state mechanical behavior at new lengths (Abramowitch et al., 2010; Schatzmann et al., 1998). Thus, it is possible that similar conditioning occurs in the lower extremity muscles following the bike-to-run transition, which may contribute to the common observation of reduced stride length and more flexed torso that occur in the early stages of a triathlon run (Garside et al., 2000; Hausswirth et al., 1997). Cycling with a more extended hip may lessen these effects.

Manipulation of STA alone had little effect on kinematic patterns during cycling. This is likely due to the fact that we varied STA while maintaining the radial distance between the crank axis and saddle, and between the crank axis and the center of the stem. Previous biomechanical studies of cycling have not clearly stated how handlebar positioning was adjusted with STA (Garside et al., 2000; Heiden et al., 2003; Heil et al., 1995; Price et al., 1997; Ricard et al., 2006), thus making it difficult to compare our results to other experimental studies. A theoretical study (Rankin et al., 2010) predicted that increasing STA in the manner described in our current study would result in increased anterior pelvic tilt, while having minimal effect on lower extremity joint angles and muscle activation patterns. Interestingly, we observed that increasing STA did increase RF muscle activity during upstroke of the crank cycle. The RF is thought to aid in limb recovery and stroke transition near top dead center (Raasch et al., 1997), such that greater activity in this region could affect power fluctuations over the pedal stroke.

Altering hand position across the aero, drops, and hoods positions resulted in significant changes in both lower extremity kinematics and neuromuscular control patterns. Similar to others, we found no significant changes in muscle activation patterns in the ankle plantar flexors between an upright and aerodynamic riding posture (Chapman et al., 2008). However, moving to the hoods position reduced mean hip flexion angle and induced a significant increase in both BF and MH muscle activities (Table 4). This is in disagreement with a study by Dorel et al. (Dorel et al., 2009) who found that an aerodynamic riding position induced greater vastii activity, decreased RF activity, and caused no changes in hamstring activity. Our results are consistent with the opinion that triathlon-specific bike setups place greater emphasis on the hamstring muscles (Garside et al., 2000; Heil et al., 1995; Price et al., 1997).

The experimental approach employed in this study was not sufficient to infer how musculotendon mechanics and power development change with STA and hand position. For example, additional information on muscle force development would be needed to determine if the increased hamstring activity measured in our current study actually results in less localized quadriceps fatigue, as has been suggested previously (Garside et al., 2000; Heil et al., 1995; Price et al., 1997). The use of musculoskeletal modeling to simulate pedaling and analyze muscle contributions to crank and limb power (Raasch et al., 1997; Rankin et al., 2010) may provide greater insights into this issue. In addition, the use of full body motion analysis in this study negated the possibility of having participants perform fatiguing bouts of cycling and running, as might be experienced in a triathlon race. As a result, we cannot use our data to assess how neuromuscular control and kinematic might change with fatigue (Hausswirth et al., 2001).

In summary, we showed that running requires longer musculotendon lengths and a greater duration of active lengthening (eccentric) contractions, when compared with cycling. Increasing the STA and configuring the handlebars to allow participants to ride with a more extended hip increased the biarticular muscle activity and moved the lower extremity muscle lengths slightly closer to those seen in running. These combined changes in neuromuscular coordination and musculotendon mechanics may contribute to the improved running performances that have been observed immediately after cycling on a triathlon-specific bicycle.

Acknowledgments

We would like to acknowledge Bryan Heiderscheit, Rick Neptune, and Mark Timmerman.

References

Abramowitch, S. D., Zhang, X., Curran, M., & Kilger, R. (2010). A comparison of the quasi-static mechanical and non-linear viscoelastic properties of the human semitendinosus and gracilis tendons. *Clin Biomech (Bristol, Avon), 25*(4): 325–331.

Bernard, T., Vercruyssen, F., Grego, F., Hausswirth, C., Lepers, R., Vallier, J.M., et al. (2003). Effect of cycling cadence on subsequent 3 km running performance in well trained triathletes. *British Journal of Sports Medicine, 37*(2), 154–158, discussion 159.

Chapman, A.R., Vicenzino, B., Blanch, P., Knox, J.J., Dowlan, S., & Hodges, P.W. (2008). The influence of body position on leg kinematics and muscle recruitment during cycling. *Journal of Science and Medicine in Sport, 11*(6), 519–526.

Coyle, E.F., Coggan, A.R., Hopper, M.K., & Walters, T.J. (1988). Determinants of endurance in well-trained cyclists. *Journal of Applied Physiology (Bethesda, Md.), 64*(6), 2622–2630.

Delp, S.L., Loan, J.P., Hoy, M.G., Zajac, F.E., Topp, E.L., & Rosen, J.M. (1990). An interactive graphics-based model of the lower extremity to study orthopaedic surgical procedures. *IEEE Transactions on Bio-Medical Engineering, 37*(8), 757–767.

Dengel, D.R., Flynn, M.G., Costill, D.L., & Kirwan, J.P. (1989). Determinants of success during triathlon competition. *Research Quarterly for Exercise and Sport, 60*(3), 234–238.

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

Garside, I., & Doran, D.A. (2000). Effects of bicycle frame ergonomics on triathlon 10-km running performance. *Journal of Sports Sciences, 18*(10), 825–833.

Hausswirth, C., Bigard, A.X., Berthelot, M., Thomaidis, M., & Guezennec, C.Y. (1996). Variability in energy cost of running at the end of a triathlon and a marathon. *International Journal of Sports Medicine, 17*(8), 572–579.

Hausswirth, C., Bigard, A.X., & Guezennec, C.Y. (1997). Relationships between running mechanics and energy cost of running at the end of a triathlon and a marathon. *International Journal of Sports Medicine, 18*(5), 330–339.

Hausswirth, C., & Lehenaff, D. (2001). Physiological demands of running during long distance runs and triathlons. *Sports Medicine (Auckland, N.Z.), 31*(9), 679–689.

Hausswirth, C., Lehenaff, D., Dreano, P., & Savonen, K. (1999). Effects of cycling alone or in a sheltered position on subsequent running performance during a triathlon. *Medicine and Science in Sports and Exercise, 31*(4), 599–604.

Heiden, T., & Burnett, A. (2003). The effect of cycling on muscle activation in the running leg of an Olympic distance triathlon. *Sports Biomechanics, 2*(1), 35–49.

Heil, D.P. (2001). Body mass scaling of projected frontal area in competitive cyclists. *European Journal of Applied Physiology, 85*(3-4), 358–366.

Heil, D.P., Wilcox, A.R., & Quinn, C.M. (1995). Cardiorespiratory responses to seat-tube angle variation during steady-state cycling. *Medicine and Science in Sports and Exercise, 27*(5), 730–735.

Hue, O., Le Gallais, D., Chollet, D., Boussana, A., & Prefaut, C. (1998). The influence of prior cycling on biomechanical and cardiorespiratory response profiles during running in triathletes. *European Journal of Applied Physiology and Occupational Physiology, 77*(1-2), 98–105.

Li, X., & Aruin, A. (2005). Muscle activity onset time detection using Teager-Kaiser energy operator. *Conference Proceedings; ... Annual International Conference of the IEEE Engineering in Medicine and Biology Society. IEEE Engineering in Medicine and Biology Society. Conference, 7,* 7549–7552.

Lu, T.W., & O'Connor, J.J. (1999). Bone position estimation from skin marker co-ordinates using global optimisation with joint constraints. *Journal of Biomechanics, 32*(2), 129–134.

Martin, J.C., Milliken, D.L., Cobb, J.E., McFadden, K.L., & Coggan, A.R. (1998). Validation of a Mathematical Model for Road Cycling Power. *Journal of Applied Biomechanics, 14,* 276–291.

Millet, G.P., Millet, G.Y., Hofmann, M.D., & Candau, R.B. (2000). Alterations in running economy and mechanics after maximal cycling in triathletes: influence of performance level. *International Journal of Sports Medicine, 21*(2), 127–132.

Millet, G.P., & Vleck, V.E. (2000). Physiological and biomechanical adaptations to the cycle to run transition in Olympic triathlon: review and practical recommendations for training. *British Journal of Sports Medicine, 34*(5), 384–390.

Piazza, S.J., Erdemir, A., Okita, N., & Cavanagh, P.R. (2004). Assessment of the functional method of hip joint center location subject to reduced range of hip motion. *Journal of Biomechanics, 37*(3), 349–356.

Price, D., & Donne, B. (1997). Effect of variation in seat tube angle at different seat heights on submaximal cycling performance in man. *Journal of Sports Sciences, 15*(4), 395–402.

Raasch, C.C., Zajac, F.E., Ma, B., & Levine, W.S. (1997). Muscle coordination of maximum-speed pedaling. *Journal of Biomechanics, 30*(6), 595–602.

Rankin, J.W., & Neptune, R.R. (2010). The influence of seat configuration on maximal average crank power during pedaling: A simulation study. *Journal of Applied Biomechanics, 26,* 493–500.

Ricard, M., Hills-Meyer, P., Miller, M., & Michael, T. (2006). The effects of bicycle frame geometry on muscle activation and power during a Wingate anaerobic test. *Journal of Sports, Science, and Medicine, 5,* 25–32.

Riley, P.O., Franz, J., Dicharry, J., & Kerrigan, D.C. (2010). Changes in hip joint muscle-tendon lengths with mode of locomotion. *Gait & Posture, 31*(2), 279–283.

Schatzmann, L., Brunner, P., & Staubli, H.U. (1998). Effect of cyclic preconditioning on the tensile properties of human quadriceps tendons and patellar ligaments. *Knee Surgery, Sports Traumatology, Arthroscopy, 6*(Suppl. 1), S56–S61.

Schmitz, A., Silder, A., Heiderscheit, B., Mahoney, J., & Thelen, D.G. (2008). Differences in lower-extremity muscular activation during walking between healthy older and young adults. *Journal of Electromyography and Kinesiology, 19*(6): 1085–1091.

Walker, P.S., Rovick, J.S., & Robertson, D.D. (1988). The effects of knee brace hinge design and placement on joint mechanics. *Journal of Biomechanics, 21*(11), 965–974.

Yang, J.F., & Winter, D.A. (1984). Electromyographic amplitude normalization methods: improving their sensitivity as diagnostic tools in gait analysis. *Archives of Physical Medicine and Rehabilitation, 65*(9), 517–521.