

Gastrocnemius and soleus muscle length, velocity, and EMG responses to changes in pedalling cadence

David J. Sanderson^{a,*}, P.E. Martin^b, G. Honeyman^a, J. Keefer^a

^a School of Human Kinetics, University of British Columbia, 210-6081 University Boulevard, Vancouver BC, Canada V6T 1Z1

^b Department of Kinesiology, Penn State University, University Park, PA 16802, United States

Received 18 February 2005; received in revised form 23 September 2005; accepted 1 November 2005

Abstract

Several authors have shown different excitation patterns for soleus and gastrocnemius muscles in response to cadence manipulation during cycling. The purpose of this study was to examine gastrocnemius and soleus length and velocity change as a function of pedalling cadence to consider mechanisms underlying these excitation differences. Ten male and two female cyclists rode at five randomly assigned cadences (50, 65, 80, 95, and 110 rpm) at a nominal 200 W power output while EMG of the gastrocnemius and soleus and sagittal plane video were recorded. Joint-coordinate data for the knee and ankle were used with equations of Grieve et al. [Grieve D, Pheasant S, Cavanagh PR. Prediction of gastrocnemius length from knee and ankle joint posture, in: E. Asmussen, K. Jorgensen, editors. International Series on Biomechanics, vol. 2A, Baltimore: University Park Press; 1978. p. 405–412] to compute gastrocnemius and soleus length and velocity. Consistent with previous publications, gastrocnemius displayed a significant ($p < 0.05$) increase in integrated EMG with increased cadence, whereas cadence had no significant effect on integrated EMG of the soleus. The ankle became significantly ($p < 0.05$) more plantar flexed and reflected a reduced range of motion with increased cadence while the knee became significantly ($p < 0.05$) less extended. Soleus decreased its range of motion by 29%, whereas gastrocnemius decreased its range of motion by 9%. In contrast, soleus increased its velocity range by 32% and gastrocnemius increased by 45%. These data show that with increased cadence gastrocnemius operated over a narrower range of operating lengths but at a higher range of shortening velocity than soleus. The higher range of velocity may have resulted in the need for a relatively higher excitation, as indicated by the integrated EMG, as the muscle was working at a different range on its force–velocity curve. During the recovery portion of the pedalling cycle, the soleus was acting eccentrically while the gastrocnemius acted concentrically indicating the triceps surae complex did not always act in unison.

© 2005 Elsevier Ltd. All rights reserved.

Keywords: Force; Muscle; Length; Electromyography; Cycling; Stretch–shorten cycle

1. Introduction

A key component to understanding complex human movement is the nature of load sharing among muscles and movement control. An example of this concerns the contribution of the triceps surae to plantar flexion during cycling. Of the many muscles that contribute to plantar-flexor torque (gastrocnemius, soleus, plantaris, tibialis pos-

terior, the long flexors of the toes, peroneus longus, and peroneus brevis) it has been estimated that the triceps surae group accounts for 60–80% of that torque (Sale et al. [19]). That the soleus is a mono-articular muscle and gastrocnemius is bi-articular provides a unique control situation, where two joints affect muscle lengths and velocities, which in turn influence force-producing capacity. Without giving exact values Goto et al. [5] reported that gastrocnemius excitation increased in a linear fashion with increased cadence from 40 to 100 rpm. Ericson et al. [4] showed that both soleus and gastrocnemius were sensitive to cadence increases, from 40 to 100 rpm, although the gastrocnemius

* Corresponding author. Tel.: +1 604 822 4361; fax: +1 604 822 6842.
E-mail address: david.sanderson@ubc.ca (D.J. Sanderson).

response was much greater (190% for the gastrocnemius compared to 35% for the soleus). Marsh et al. [14] reported that peak gastrocnemius excitation increased by 87% in experienced cyclists when cadence was increased from 50 to 110 rpm, whereas peak soleus excitation was unaffected by this cadence manipulation (approximately 1% increase).

This important difference was used by Cresswell et al. [3] to explore the relative contribution of gastrocnemius to isometric plantar-flexor torque at knee angles ranging from 180° (full extension) to 60°. Peak plantar flexor torque decreased sigmoidally with increased knee flexion. Cresswell et al. also reported a decrease in the EMG root mean square with increased knee flexion for the gastrocnemius, whereas soleus EMG remained constant. The differing sensitivity of these muscles has implications for their recruitment in a cycling task, where both knee and ankle joint angle changes would be made as would the requirements for force at the pedal throughout the pedalling cycle.

A potential explanation for these differences might be found in an examination of the mechanical properties of the muscles themselves (Chapman et al. [2]). It has been well established that striated muscle exhibits a force–length–velocity relationship that dominates its response to the need for variations in power output. Grieve et al. [9] devised a method of determining the length of gastrocnemius and soleus muscles during dynamic actions. Using cadaver specimens, the knee and ankle joints were manipulated independently as length changes in the gastrocnemius and soleus were measured at the common site of attachment of the Achilles tendon. Grieve and colleagues then established a polynomial relationship between knee and ankle joint angles and gastrocnemius and soleus muscle lengths. Gregor et al. [7] used this technique with recorded surface EMG and tendon force from an implanted buckle transducer to provide useful insights into muscle function during cycling of a single subject. The gastrocnemius was active throughout the power phase (top dead center to bottom dead center), which coincided with modest changes in muscle length, first lengthening and then shortening. Most of the length changes of the gastrocnemius occurred during the recovery phase when the muscle showed little excitation. The soleus was most active in the first 90° of the pedal cycle, which also coincided with a period of limited muscle shortening. Like the gastrocnemius, most of the soleus' shortening and lengthening occurred when the muscle was not highly active. Gregor and colleagues suggested the soleus contribution to tendon force was more significant in the first 90° of the pedal cycle, whereas that of the gastrocnemius was more significant in the second 90° of the pedal cycle. Further, the gastrocnemius force production may be enhanced by a modest stretch–shorten pattern reflected during the power phase.

When the cyclist remains seated, cycling is a relatively constrained motion. The path of the feet are dictated by the geometry of the crank which when combined with the limited degrees of freedom at the joints suggests that changing pedalling rate should not have an influence on

the kinematics of the lower limb. Nonetheless, there are changes in the segment patterns and the differential excitation of the soleus and gastrocnemius muscles with changing cadence indicate some level of differential control (load sharing?) that may reflect the underlying mechanical properties of these muscles. Marsh et al. [14] have shown that the two members of triceps surae reflected different EMG responses to increased cadence at a given power output, and suggested the soleus was more sensitive to changing force demands, whereas the gastrocnemius was more sensitive to changing speed needs. However, no estimates of muscle length and velocity profiles which may be the key mechanical factors affecting recruitment were included in their analysis. Consequently, our objective was to replicate their procedures and include estimates of muscle length and velocity for the gastrocnemius and soleus. Our hypothesis was that the velocity of contraction of the soleus and gastrocnemius muscles would not vary with cadence because even though the riders were pedalling more rapidly changes in the length of the muscles, and hence velocity, would be modulated to restrict variations in length change.

2. Methods

2.1. Participants

The study group consisted of 12 cyclists (10M, 2F) with a mean (SD) age of 25.7 (3.7) years, body mass of 73.1 (6.3) kg, and height of 179.4 (5.7) cm. Participants were experienced, competitive cyclists who had either a current Canadian Cycling Association (CCA) road racing license of category three or higher or CCA mountain bike racing license of Sportsman, Expert or Elite. All persons were made fully aware of the experimental details prior to assuming their involvement in the program and signed informed consent that conformed to the ethical guidelines of the University.

2.2. Data collection

A standard bicycle mounted on a Schwinn Velodyne electronically braked cycle ergometer which simulates inertial characteristics of road riding and modulates power outputs based on cadence (Attaway et al. [1]) was used in this study. The bicycle was set up so that it would match, as closely as possible, the athlete's own bicycle with the exception that seat height was set to 100% of trochanteric height. This height was found to be optimal in terms of oxygen consumption by Nordeen-Snyder [16]. Optical encoders with 1024 steps monitored pedal and crank angle and top dead centre (TDC). A cadence monitor (Cateye) attached to the crank provided feedback to participants who were instructed to maintain cadence within ± 1 rpm of the target cadence.

Participants subsequently rode for a minimum of 5 min at each of six randomly presented cadences (50, 65, 80, 95, and 110 rpm) at a constant nominal power output of

200 W while heart rate, EMG, and sagittal-plane video were recorded. Heart rate was recorded continuously during the entire ride using a Sport Tester PE 3000 portable heart rate monitor (Polar Electro Inc., Finland). Heart-rate data were downloaded from the portable heart-rate monitor using a Polar interface system for processing. Heart rate was considered stable when it did not vary by more than two beats per minute for 1 min.

EMG data were recorded at a rate of 600 Hz through a 12-bit analogue-to-digital converter. The optical encoder indicating TDC triggered data collection. Data collection occurred in the final minute of the test protocol and lasted for three revolutions of the crank per collection period. The participant was kept unaware of the exact timing of data collection. EMG data of the soleus and gastrocnemius muscles were recorded on the left leg using surface electrodes (1.5 cm center-to-center, Therapeutics Unlimited) attached via double-sided adhesive pads. Electrodes were placed longitudinally on the distal half of the medial gastrocnemius and the distal half of the soleus. The skin sites were shaved, abraded and cleaned with isopropyl alcohol to reduce source impedance. The electrodes provided pre-amplification (gain = 35) at the recording site (CMRR = 87 dB at 60 Hz).

Kinematics of the limb segments were recorded simultaneously with EMG data at 60 Hz using a Panasonic video camera (WDV 5100), with the lens axis oriented orthogonal to the sagittal plane of the participant, positioned 6 m from the rider. A reflective marker was placed over each of the greater trochanter, the estimated centre of rotation of the knee joint, the lateral malleolus, the base of the calcaneus and the head of the fifth metatarsal prior to the test. Out-of-plane motion of the lower-limb was restricted by the riders wearing cleated cycling shoes.

Video records for the same three pedal cycles for which EMG was recorded were digitised using the Peak Performance Technologies (version 5) software package, filtered using a 2nd-order, dual-pass Butterworth filter with a cut-off frequency of 5 Hz. These data were then time-matched to the corresponding EMG data file. Time matching was achieved using a square-wave pulse from the bicycle crank, generated at TDC, and recorded as one channel of the A/D. This pulse also generated a white square on the video signal using an event synchronization unit. These data were adjusted in time via linear interpolation so that the arrays became the same length, from one TDC to the next.

Marker kinematics were used first to compute the angle of the knee and ankle joints and then the muscle lengths of the soleus and gastrocnemius muscles over the complete pedal cycle using equations developed by Grieve et al. [9]. These equations present muscle lengths changes as a percent of segment length. These data were then differentiated to generate muscle velocities, which were used to identify periods of concentric and eccentric activity. Joint angle, muscle length, and muscle velocity data were divided into three pedal cycles, normalized in

time, and the mean value determined. From these data, maximum, minimum, and ranges were computed for statistical analysis.

Raw EMG data from each muscle for each cadence condition for three crank revolutions were full-wave rectified. A linear envelope was created using a 2nd-order, dual-pass Butterworth filter with a cut-off frequency of 5 Hz in the same fashion as Marsh et al. [14]. Using the TDC pulse, the linear envelope was then divided into each of the three pedal cycles and a mean linear envelope was computed for each muscle. For amplitude scaling, the average value over the complete pedalling cycle from the 50-rpm condition was determined. All individual data were scaled to this value before the within subject and between-subject means were calculated. A number of dependent variables were computed from the average data for each muscle: minimum, maximum, and range of ankle and knee joint angle, muscle length and velocity, and peak EMG.

For each variable, a one factor ANOVA with repeated measures on cadence was performed with a 0.05 level of significance. Follow up tests were completed using the Scheffé procedure. For two subjects, difficulties with the electrodes resulted in some lost data for some trials. Thus, EMG analyses were based on a sample size of 10 rather than 12.

3. Results

With increasing cadence the ankle joint became more plantar flexed and the knee joint more flexed (Fig. 1). There was also a significant decrease in the range of motion for both the knee ($F_{4,44} = 12.2$, $p < 0.0001$) and ankle joints ($F_{4,44} = 8.8$, $p < 0.0001$). Soleus and gastrocnemius muscles functioned at shorter lengths and over a shorter range with increased cadence (Fig. 2A and D). This effect was significant for soleus ($F_{4,44} = 6.2$, $p = 0.0005$) but not gastrocnemius ($F_{4,44} = 0.461$, $p = 0.7641$). Both muscles exhibited a decreased maximum length as cadence increased (soleus $F_{4,44} = 26.7$, $p < 0.0001$; gastrocnemius $F_{4,44} = 18.8$, $p < 0.0001$). However, only the gastrocnemius demonstrated a significant decrease in minimum length ($F_{4,44} = 6.2$, $p < 0.0005$).

As expected, changes in soleus and gastrocnemius muscle velocities, presented in Fig. 2B and E, tended to increase systematically with increased cadence. Maximum velocity of shortening significantly increased for both soleus and gastrocnemius ($F_{4,44} = 6.9$, $p = 0.0002$; $F_{4,44} = 9.6$, $p < 0.0001$, respectively). The gastrocnemius also showed significantly higher maximum lengthening velocity at higher cadence ($F_{4,44} = 3.5$, $p = 0.01$). The range of velocity, from maximum lengthening to maximum shortening, was significantly affected by cadence for both soleus and gastrocnemius muscles ($F_{4,44} = 3.0$, $p = 0.03$; $F_{4,44} = 7.9$, $p < 0.0001$, respectively). The normalized peak EMG data from the soleus (Fig. 2C) was not significantly affected by cadence ($F_{4,36} = 0.246$, $p = 0.9103$; $F_{4,36} = 0.233$, $p = 0.9177$), whereas normalized and peak iEMG for the gas-

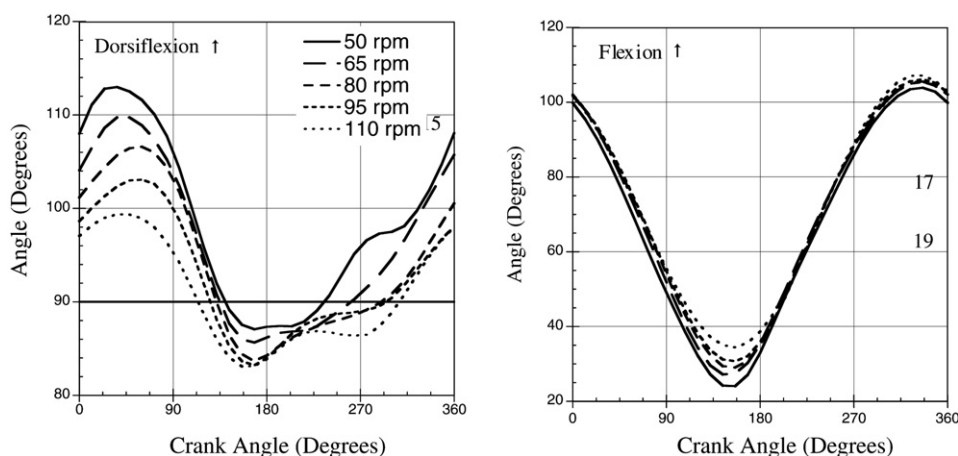


Fig. 1. Mean joint angles for ankle (left panel) and knee joints (right panel) normalized for the pedalling cycle. Neutral position for the ankle joint was 90° where values greater than 90° are dorsi-flexion. For the knee joint, full extension was 0° .

trocnemius (Fig. 2F) increased significantly with increased cadence ($F_{4,36} = 7.3$, $p = 0.0002$; $F_{4,36} = 7.2$, $p = 0.0002$).

4. Discussion

Consistent with Marsh et al. [14], gastrocnemius excitation was significantly affected by cadence manipulation, whereas soleus excitation was not. Further, these data, consistent with those presented by Gregor et al. [6–8], suggested a potential difference in function between soleus and gastrocnemius such that the soleus was primarily involved in generating initial propulsive force and the gastrocnemius overlapped that excitation and prolonged excitation later in the cycle to provide continual force at the pedal.

Similar to data reported by Gregor et al. [7], over the range of cadence manipulation, the reduction in the maximum length of each muscle was about the same magnitude, which indicated that ankle joint angle changes were the dominant cause. The reduced range of motion at the knee of about 4° over the range of cadence manipulation, while significant, was much smaller than the change in the range of motion at the ankle joint of about 10° . Even though the knee was in a more flexed position the changes were so small that the flexion was unlikely to have affected the ability of gastrocnemius to generate force or to affect his excitation as was shown by Cresswell et al. [3]. The stronger effect of the ankle joint on muscle length was explained partially by examining the moment arms of each muscle at ankle and knee joints. The moment arm for the Achilles tendon has been estimated as 52 mm when the ankle was in the neutral position, whereas the moment arm for the gastrocnemius at the knee has been estimated as approximately 12.5 mm (Spoor et al., [20]). One would expect, therefore, the movement at the ankle joint would dominate the length change of both muscles. Hoy et al. [11] have estimated that the gastrocnemius moment arm is slightly larger than the gastrocnemius moment arm. This has the effect of reducing the large difference in peak force between these

muscles so that they generate similar torques around the ankle joint. This may explain how they work together to provide a longer duration force. The velocity of shortening of soleus near TDC is low and that enables the primarily slow twitch muscle fibres to initiate plantar flexion. As the pedal proceeds through the first 180° of crank rotation and the velocity of shortening increases (Fig. 2B) the excitation from soleus falls off (Fig. 2C) and gastrocnemius is recruited to overlap (Fig. 2F).

If joint and segment kinematics were to remain unchanged increasing cadence from 50 to 110 rpm would in effect require a doubling of the rate at which these muscles changed length. However, data show that the mean velocity of shortening, between 50 and 100 rpm, increased on average by only 50% for both muscles and there was a concomitant narrowing of the ranges of length change (soleus decreased by 33% and gastrocnemius by 11%) and a shift to a smaller mean length (soleus reduced by 91% and gastrocnemius by 68%). Simultaneously, with increased cadence, ankle motion was reduced from an average of 99° to 90° , which occurred primarily in dorsiflexion during the first half of the cycle. In other words, at the higher cadence (110 rpm) the cyclists held the ankle angle at 90° relative to the shank. This had the effect of (perhaps) keeping the ankle muscles closer to a neutral length, perhaps a position of greater maximum strength. This effect was stronger for the soleus than the gastrocnemius. It is not clear why this occurs. On one hand, there was the increase in the speed of movement and that would require the muscles work in a portion of the force–velocity relationship with reduced maximum force. To reduce this effect, the muscle can reduce the speed increase by operating over a shorter length change. However, with increased cadence at constant power output the force needed is decreasing (Sanderson [18]).

Speculating, this may reflect a strategy, whereby each muscle operated in a restricted range of motion to maximize mechanical power output by operating in a more productive portion of their length–tension curves. However,

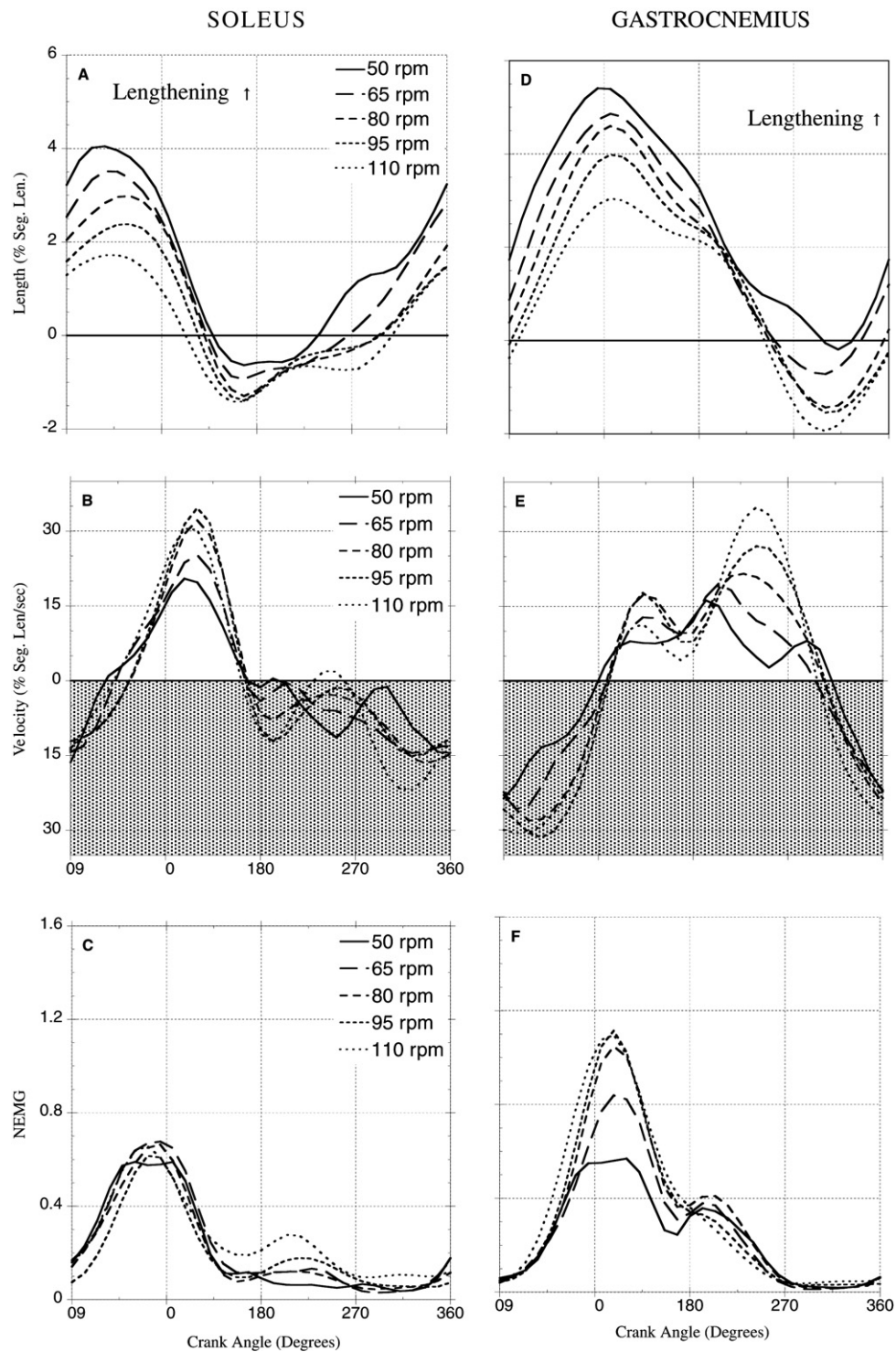


Fig. 2. Mean length, velocity, and EMG for the soleus muscle, left panel, and the gastrocnemius muscle, right panel. Panels A and D are length change, B and E are muscle velocity, and C and F are the normalized EMG data. In panels B and E the shaded portion indicates lengthening velocity.

we need more information regarding the length–tension relationship of these muscles. There did not appear to be an increase in the duration of this effect in response to cadence manipulation in that the points of change from lengthening to shortening, and vice versa, were at approximately the same point in the cycle, probably reflecting the restricted options available in a constrained motion like

cycling. There is a possibility that the structure of these muscles could influence their force production. Maas et al. [13] have shown that during isometric contractions of extensor digitorum longus, tibialis anterior, and extensor hallucis longus changes in muscle force were the combined result of muscle length changes and changes of the relative position of the muscles. They concluded that mus-

cle position relative to surrounding tissues and muscle length co-determines muscle force. It was not possible with the dataset here to assess whether this also occurs in triceps surae. While it does seem possible, the fact that soleus and gastrocnemius appear to act in opposite directions suggest that these muscles are more independent than the dorsiflexors. More work, however, is needed to verify this. Murakami et al. [15] showed that passive properties of the gastrocnemius provide little resistance to compression and that passive elastic force increases with length. However, the amount of this increase was only in the order of 1.5 N which is unlikely to impact the force production during cycling.

Considering the pedalling cycle from TDC to TDC, both muscles exhibited a period of lengthening prior to shortening as was observed by Ericson et al. [4], Gregor et al. [6–8] and Hawkins et al. [10]. This so-called stretch-shorten cycle is a common characteristic of muscle activity and these data show clearly that it occurs during cycling and over a wide range of cadence manipulation, a factor that has not been

previously shown. When the soleus was most active (between 45° and 135°), it was shortening and shortening at progressively higher rates as cadence increased. When the gastrocnemius was most active (between 60° and 150°, perhaps even up to 235°), it shifted from lengthening to shortening, i.e., it was functioning close to its transition between lengthening and shortening and at relatively lower velocity (−15% to +15% segment lengths/s). For the gastrocnemius the cadence effect on muscle lengthening and shortening velocities was more pronounced during periods when it was at relatively low excitation. It may be the case that when the gastrocnemius was making its strongest contribution, it was functioning on a more effective portion of its force–velocity relationship, which in turn would make it less sensitive to force decrements that could be associated with increases in speed due to cadence increases. The period in which cadence dramatically affected gastrocnemius EMG was in that 60–150 degree window.

It was interesting to note that the gastrocnemius showed a second, albeit small, excitation peak just after BDC

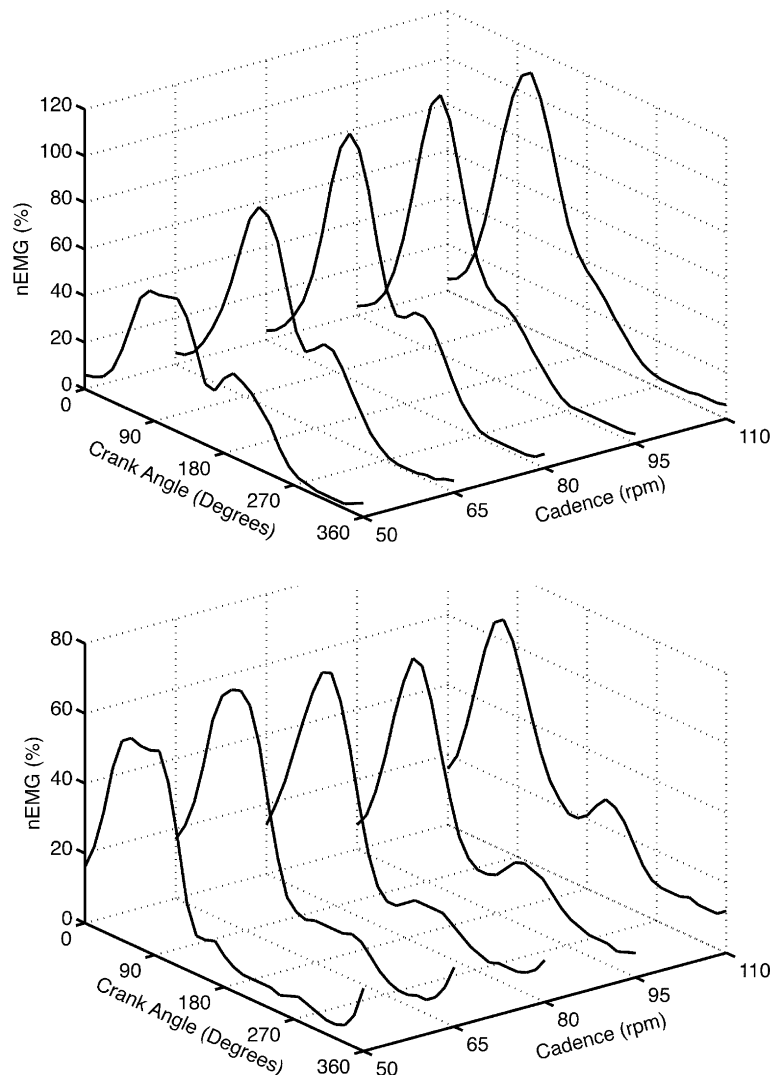


Fig. 3. Mean normalized EMG data plotted in a 3D format to illustrate the appearance of a second bump in soleus (bottom panel) and the simultaneous disappearance of the second bump in gastrocnemius (top panel).

(approximately 210–270°), which occurred as gastrocnemius shortening velocity was approaching its peak. This second excitation peak of the gastrocnemius was most prominent at 50 rpm but decreased in prominence as cadence increased such that at 110 rpm this second peak no longer existed. In contrast, the soleus developed a modest second excitation peak as cadence increased. The three-dimensional representation of the activation changes (Fig. 3 A and B) with cadence and crank angle illustrate the load sharing between these two muscles. EMG data presented by Marsh et al. [14] and Reimer et al. [17] show similar cadence-related changes in gastrocnemius and soleus excitation just after BDC, but no mention was made of the significance of these changes. Van Ingen Schenau et al. [12] reported that when knee extension and plantar flexion of the ankle were combined the gastrocnemius underwent slower shortening velocities than the other plantar flexors. In the current data, during the recovery phase the opposite occurred, that is knee flexion and ankle dorsi-flexion. This secondary change in soleus and gastrocnemius excitation appears to occur at the transition from ankle plantarflexion motion to dorsiflexion, which coincides with the moment of maximum knee extension. During the period of this secondary excitation phase the gastrocnemius continued to shorten while the soleus lengthened. We speculate that these motions put the gastrocnemius muscle at a disadvantage for effective contraction as gastrocnemius was shortening and to account for the decrease in its force contribution at the ankle joint a concomitant increase in activation of soleus was called for, however we have no data to support this.

In conclusion, while these data provided important information on how the length and velocity profiles of gastrocnemius and soleus muscles might dictate rider response to variations in cadence it remains unclear why the excitation of gastrocnemius is sensitive to cadence manipulations and soleus is not. With regard to the secondary peak in excitation just after BDC gastrocnemius continued to shorten and moved into a length range, where it was less effective at producing force soleus activation was increased thus allowing for maintenance of the ankle moment. To explore this we would consider restricting ankle motion and then observing how the excitation of these muscle muscles was modulated. These data confirm the earlier EMG data indicating that the triceps surae complex did not always act in unison, i.e. both in eccentric and/or concentric actions. Exactly how this level of control can be achieved is a question for a future study as the exact nature of this interplay remains to be explored. A limitation to this work is that we have not been able to account for different fibre orientation nor have we separated the effects of a long tendon of gastrocnemius versus a shorter tendon in soleus. In essence the Grieve model simplifies these muscles and treats them as a unipennate units. This makes it difficult to extract the exact nature of velocity and length. However, the data presented here do provide a useful analysis of the internal excitation changes with these two muscles.

References

- [1] Attaway RW, Bartoli WP, Pate RR, Davis JM. Physiologic and perceptual responses to exercise on a new cycle ergometer. *Can J Sport Sci* 1992;17:56–9.
- [2] Chapman AE, Sanderson DJ. Muscular co-ordination in sporting skills. In: Winters J, Woo S, editors. Multiple muscle systems. New York: Springer-Verlag; 1990. p. 608–20.
- [3] Cresswell AG, Löscher WN, Thorstensson A. Influence of gastrocnemius muscle length on triceps surae torque development and electromyographic activity in man. *Exp Brain Res* 1995;105:283–90.
- [4] Ericson MO, Nisell R, Arborelius UP, Ekholm J. Muscular activity during ergometer cycling. *Scand J Rehabil Med* 1985;17:53–61.
- [5] Goto S, Toyoshima S, Hoshikawa T. Study of the integrated EMG of leg muscles during pedalling at various loads, frequency and equivalent power. In: Assmussen E, Jorgensen K, editors. International series on biomechanics, vol. 2A. Baltimore: University Park Press; 1978. p. 246–52.
- [6] Gregor RJ, Brooker JP, Ryan MM. Biomechanics of Cycling. *Exercise Sport Sci Rev* 1991;19:127–69.
- [7] Gregor RJ, Komi PV, Järvinen M. Achilles tendon forces during cycling. *Int J Sport Med* 1987;8:9–14 [Supplement].
- [8] Gregor RJ, Komi PV, Browning RC, Järvinen M. A comparison of the triceps surae and residual muscle moments at the ankle during cycling. *J Biomech* 1991;24:287–97.
- [9] Grieve D, Pheasant S, Cavanagh PR. Prediction of gastrocnemius length from knee and ankle joint posture. In: Assmussen E, Jorgensen K, editors. International series on biomechanics, vol. 2A. Baltimore: University Park Press; 1978. p. 405–12.
- [10] Hawkins D, Hull M. A method for determining lower extremity muscle-tendon lengths during flexion/extension movements. *J Biomech* 1990;23(5):487–94.
- [11] Hoy MG, Zajac FE, Gordon ME. A musculoskeletal model of the human lower extremity: the effect of muscle, tendon, and moment arm on the moment arm relationship of musculotendon actuators at the hip, knee, and ankle. *J Biomech* 1990;23:157–69.
- [12] van Ingen Schenau GJ, Bobbert MF, van Soest AJ. The unique action of bi-articular muscles in leg extensions. In: Winters J, Woo S, editors. Multiple muscle systems. New York: Springer-Verlag; 1990. p. 639–52.
- [13] Maas H, Baan GC, Huijing PA. Muscle force is determined also by muscle relative position: isolated effects. *J Biomech* 2004;37(1):99–110.
- [14] Marsh AP, Martin PE. The relationship between cadence and lower extremity EMG in cyclists and non-cyclists. *Med Sci Sport Exercise* 1995;27(2):217–25.
- [15] Muraoka T, Chino K, Muramatsu T, Fukunaga T, Kanehisa H. In vivo mechanical properties of the human gastrocnemius muscle belly. *J Biomech* 2005;38:1213–9.
- [16] Nordeen-Snyder KS. The effect of bicycle seat height variation upon oxygen consumption and lower limb kinematics. *Med Sci Sport* 1977;9(2):109–12.
- [17] Reimer B, Sanderson DJ, Seburn K. Effect of cadence on the EMG of vastus lateralis and oxygen uptake during steady-state sub maximal cycling. In: Gregor R, Zernicke RF, editors. Proceedings of the 12th international congress of biomechanics, Los Angeles; 1989. p. 137–138.
- [18] Sanderson DJ. The influence of cadence and power output on the biomechanics of force application during steady-rate cycling in competitive and recreational cyclists. *J Sport Sci* 1991;9:191–203.
- [19] Sale D, Quinlan J, Marsh E, McComas AJ, Belanger AY. Influence of joint position on ankle plantarflexion in humans. *J Appl Physiol* 1982;52(6):1636–42.
- [20] Spoor CW, van Leeuwen JL, Meskers GM, Titulaer AF, Huson Q. Estimation of instantaneous moment arms of lower-leg muscles. *J Biomech* 1990;23(12):1247–59.



David J. Sanderson, Ph.D., is a Professor in the School of Human Kinetics at UBC. His specialty is in repetitive motion analysis, amputee locomotion, and wheelchair propulsion biomechanics. He received his B.Sc. and M.Sc. degrees in Kinesiology at Simon Fraser University and his doctorate from the Pennsylvania State University. He came to UBC in 1986.



Gillian Honeyman, BHK, BPT, received her BHK degree from the School of Human Kinetics in 1999 and completed her post graduate degree in Physiotherapy from UBC in 2000, and received her APTA certification as a sports certified specialist in 2005. Currently, she is Head Physical Therapist for the US Snowboard team based out of Park City, Utah.



Philip E. Martin, Ph.D., is Professor and Head of the Department of Kinesiology at Penn State University. He is a specialist in exercise and locomotion biomechanics. Dr. Martin received his B.S. and M.S. degrees from the University of Illinois and his Ph.D. from the Pennsylvania State University. He served as the faculty at Arizona State University for 19 years before moving to Penn State University in 2002.



Jennifer Keefer, BHK, BPT, received her BHK degree from the School of Human Kinetics in 1999 and completed her post graduate degree in Physiotherapy from UBC in 2000. After working as a physiotherapist with the high performance athletes in Whistler, BC, she has recently relocated to Vancouver. Jennifer is using her biomechanics background to start her own company with a focus on sport specific biomechanical assessments as well as working at Sport and Spinal Physical Therapy in Yaletown.