

2001

Investigation of the differences between inertial and cadence effects on neuromuscular coordination during cycling

Brian Svercauski Baum

Louisiana State University and Agricultural and Mechanical College

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INVESTIGATION OF THE DIFFERENCES BETWEEN INERTIAL AND
CADENCE EFFECTS ON NEUROMUSCULAR
COORDINATION DURING CYCLING

A Thesis

Submitted to the Graduate Faculty of the
Louisiana State University and
Agricultural and Mechanical College
in partial fulfillment of the
requirements for the degree of
Master of Science

in

The Department of Kinesiology

by
Brian S. Baum
B.S., Louisiana State University, 1999
December 2001

ACKNOWLEDGEMENTS

This thesis represents another accomplished goal in my life and yet another step in my pursuit of knowledge and learning. This experience has opened my eyes and my mind to what is out there, what can be done, and how much more I have to learn. As is often the case, I could not have completed this task without the help and encouragement of many people. Unfortunately, I cannot name everyone whom I would like to thank, but several people deserve special recognition. First, I would like to thank Drs. Li Li, Richard Magill, and Moshe Solomonow, for serving on my committee. It was an honor to work with such distinguished academicians and outstanding people. Each of your doors was always open to my questions, and I thank you all for your patience and willingness to take time out of your busy schedules to help me. To my advisor, Dr. Li, I wish I knew how to thank you properly for educating me both in academics and in life. Your mentorship and friendship has been invaluable throughout this learning and growing process, and I will continue to strive to reach the standard bar that you have set. Dr. Magill, I wish to thank you for your continual support and helpful comments. You have been a pivotal part of my education throughout my time at LSU. Dr. Solomonow, I thank you for your invaluable comments on EMG usage, and for entertaining my many questions. I also wish to thank Dr. Dennis Landin. I learned much through assisting you in your research, and your questions and comments regarding this paper were incredibly beneficial.

My fellow graduate students, especially Jennifer Jeansonne, Lorna Ogden, Laurie Trapp, and Janene Grodesky, I thank you for your friendship, for your support, and for putting up with me. I enjoyed going through this process with all of you and I

hope that you may have learned something from me, as I have learned so much from all of you. To Lindsey Mooney, thank you for your friendship and your many hours of help during data collection and processing. This thesis would have taken many times longer without you. I have also made many other friends here at LSU, and many of you will remain close friends for life. Thank you all for the uncountable great times we have had together and for keeping a smile on my face.

Finally, I wish to thank my family. Mom and Dad, you have taught me the importance of working hard, playing hard, and pursuing excellence. I am incredibly lucky to have you as parents, and I hope that someday I may do as fantastic a job as you have done. To my sister, Dana, it is difficult to put into words how much you mean to me. Your work ethic, your kindness, and your positive attitude towards everything are only some of the attributes that make you so amazing, and I continue to strive to become as wonderful a person as you already are. To my loving family, I thank you for being there for me. I love you all very much.

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ABSTRACT

Muscular activity and coordination may be influenced by movement speed and the inertial properties of the limbs. Some observed effects from investigations using cycling have been attributed to inertia, especially at greater pedaling speeds (cadences); however, in these investigations, movement speed and inertia were coupled. Therefore, the purpose of this experiment was to investigate and distinguish between the effects of cadence and inertial influences on lower extremity neuromuscular coordination during cycling. This was achieved by subjects cycling at different cadences and with different loads attached to the distal ends of their thighs. Electromyographic (EMG) data of seven lower extremity muscles were collected from sixteen university-aged males cycling at 250 W across three pedaling cadences (60, 80, and 100 rpm) and five loads (0, 0.5, 1.0, 1.5, and 2.0 kg). Onset, offset, duration, peak magnitude, and peak timing values from the EMG linear envelopes were calculated, as were cross-correlation coefficients and phase differences. Results showed that cadence manipulations affected the timing values of the muscles and the coordination of mono- and bi-articular antagonist pairs. Altering the inertial properties of the thigh produced changes only in peak magnitudes. These results led to the conclusion that movement speed effects have a greater influence on the lower extremity muscles during cycling than do inertial effects.

LITERATURE REVIEW

Introduction

To produce a smooth, coordinated movement, muscles must act together in a harmonized way. There must be an underlying organizational pattern and a means of controlling the muscles; otherwise, a person could not properly complete a physical task, such as pedaling a bicycle or running, in a coordinated fashion. Biomechanical studies often analyze movements in order to understand how a motor system functions. These analyses can involve kinematics, kinetics, or electromyography (EMG). Kinematics examine motion in relation to space and time by studying position, velocity, and acceleration; kinetics examine how forces act on a system and interactions among the elements of the system; and EMG analyzes the electrical signals that stimulate a muscle during contractions to study muscle activity and functioning. Researchers may manipulate a task, or aspects of it, and evaluate the changes that consequently occur. Determining why particular changes take place in response to specific task alterations will contribute to the overall understanding of how an entire system was organized, controlled, and coordinated to perform a given task.

Cycling tasks are often employed in biomechanical research because cycling uses “a constrained cyclical movement that allows for a controlled investigation of test conditions” (Neptune & Herzog, 2000). Therefore, cycling enables researchers to validly and reliably explore many questions, and then generalize their results to other activities. This study used cycling to investigate how neuromuscular coordination may be affected by external components, such as cadence (quantified by crank revolutions per minute) and the inertial properties of the lower extremity.

The analysis of inertia and cadence on neuromuscular coordination during cycling performance has relevance in athletic, prosthetic, and rehabilitation settings. Marsh and Martin (1995) analyzed cadence effects on muscular activity between cyclists and non-cyclists using EMG. Their results indicated that cycling experience does not influence muscle activity significantly, but they suggested that cyclists have pedaling techniques that are more efficient. Mattes et al. (2000) observed that matching the inertial characteristics of a prosthetic limb with the residual limb in transtibial amputees produced a greater energy cost during gait and less symmetrical step length, swing time, and stance time compared with prosthetic limbs with lesser inertial properties. Li and Caldwell (Li, 1999; Li & Caldwell, 1993, 1997) have proposed that the inertial properties of the lower extremities become more influential on muscular activity and coordination at greater cadences. These studies suggest that altering both inertia and cadences will affect the musculature. However, the literature lacks EMG data related to load manipulations in order to discern between cadence and inertia. The distinction of these effects on the lower extremities may provide insights to improve training techniques in both athletic and rehabilitation settings. For example, various cadences or inertial loads may target specific muscle groups for training. Furthermore, neuromuscular coordination changes due to alterations in inertia may provide ideas for prosthetic design, attachment, and rehabilitation. Moreover, separating inertial and cadence effects will provide insights into muscle mechanics and neuromuscular control during human movements.

Pedal Forces and Joint Moments

Inverse dynamics have been used to study the effects of movement speed, specifically cadence changes in cycling, on joint moments and pedal forces. Normal pedal forces begin increasing just before top-dead-center (TDC, 0°) until they peak near 90° of the crank cycle (Gregor et al., 1985; Kautz & Hull, 1993). Mean joint moment patterns reflect the pedal force observations. Gregor et al. (1985) reported mean joint moment patterns for the hip, knee, and ankle during cycling (see Fig. 5 in Gregor et al., 1985). They observed a net hip extensor moment for the first three quarters (from 0° to 270°) of the crank cycle with the peak magnitude occurring just after 90° . The hip joint exhibited a small net flexor moment for the last quarter of the crank cycle. The net knee joint demonstrated a net extensor moment from about 325° to just after 90° of the crank cycle and a peak extensor moment near 45° . A knee joint net flexor moment was observed from just after 90° to about 325° and peaked in magnitude at 154° . Finally, the ankle joint exhibited a net plantar flexor moment for virtually the entire crank cycle. The peak plantar flexor moment occurred at about 100° .

Investigations regarding the pedal forces and joint moments in the lower extremity provide a means to better understand the adaptation of the neuromotor system to movement speed alterations. After altering cadences during cycling, researchers have observed changes in total pedal forces (Bolourchi & Hull, 1985; Li, 1999; Neptune & Herzog, 1999; Patterson & Marenco, 1990; Takaishi et al., 1998) and joint moments (Hull & Jorge, 1985; Hull & Gonzalez, 1988; Li, 1999; Redfield & Hull, 1986). Takaishi et al. (1998) observed that, at a power output of 200 W, experienced cyclists' peak pedal forces decreased from about 500 N to approximately 325 N as cadence

increased from 45 to 105 rpm. At 150 W, cyclists' peak pedal forces decreased from about 400 N to just less than 300 N. Non-cyclists' peak pedal forces behaved in the same way, but with slightly lesser peak forces. Other researchers observed similar peak pedal force trends at constant power within this cadence range (Bolourchi & Hull, 1985; Sanderson, 1991). Average pedal forces seem to behave differently from the peak values, however. Patterson and Moreno (1990) and Neptune and Herzog (1999) observed quadratic trends for pedal forces across cadences, with minimum values occurring near 90 rpm. The joint moment profiles across cadences reflect the pedal force patterns. Hull and Jorge (1985) reported decreases in total hip moment and total knee moment as cadence increased from 63 to 97 rpm. Furthermore, the knee moment adjustments were not as considerable as those at the hip. Hull and Gonzalez (1988) extended this cadence range to include from 60 to 140 rpm. Using the standard crank length of 170 mm these authors observed that the "absolute average joint moments" decreased for the hip, knee, and ankle when cadence increased from 60 to about 95 rpm, as in Hull and Jorge (1985), but at greater cadences, the hip moment dramatically increased. Overall, both the hip and knee moments experience quadratic trends with minima at near 95 and 120 rpm, respectively. The hip moment was most sensitive to cadence changes, and only the ankle moment exhibited steady decreases with cadence across the entire range.

Interpreting Muscle Activity with EMG

Since pedal forces and joint moments partially result from muscle contractions that act on the segments, muscle activity of the lower extremity will affect the forces and moments. Muscle activity is stimulated by neural stimulation. The neural

stimulation is an electrical event that electrodes from a surface electromyography (EMG) system can detect. EMG is often used as a window to observe muscle activity as it can be used to estimate muscle activation. The information relating to muscle activity can also be used to infer how that activity may relate to joint moments and pedal forces. However, the EMG signal can be difficult to interpret due to the many factors that affect it, including the motor unit recruitment strategy, fiber-type (slow twitch versus fast twitch) composition of the muscle, the motor unit size, the number of motor units recruited, the instrumentation used to detect the signal, and electrode placement (Ascierno et al., 1999; Solomonow et al., 1986, 1987). Considering this information, EMG may be used to observe muscle responses to various activities and environmental adaptations.

One important topic is the relationship between EMG activity and muscle force. In general, as the muscle force increases, the EMG signal will also increase; however, this relationship is not always linear. The relationship strongly depends on the joint angle during flexion and extension of a joint. Solomonow et al. (1991) reported a linear EMG signal to force ratio at relative ankle joint angles in cats near full extension (120 and 135 degrees), but this relationship became progressively less linear as the muscle lengthened and the joint flexed from 105 degrees to 45 degrees. Furthermore, a time delay (electromechanical delay, or EMD) can be expected from the onset of the EMG burst to actual force production (Vint et al., 2001). Cavanagh and Komi (1979) reported this EMD as between 30 and 100 ms, depending on the muscle. At a pedaling rate of 100 rpm, these delays account for between 18° and 60° of the crank cycle, respectively. When making any kind of force timing assumptions from EMG linear

envelopes, this information must be taken into account. Keeping the limitations of the EMG signal in mind, it may be used as a powerful tool to observe muscle responses to various activities, but without heeding the limitations of the EMG signal and the factors that influence it, researchers can easily construe misconceptions.

EMG and Cadence

By increasing the pedaling rates, an EMG burst onset of muscle activity may occur at a lesser crank angle for some muscles (Li & Caldwell, 1997; Neptune et al., 1997; Suzuki et al., 1982). The cadence increase may cause a muscle to react earlier in order to produce forces at the appropriate point in the crank cycle. Increases in EMG peak magnitude, or a shift of these peaks to earlier in the crank cycle, correspond to increases in kinematic joint moments. Some researchers have reported that peak muscle activity measured by EMG increased with cadence (Neptune et al., 1997), whereas others observed only a peak shift with no apparent magnitude change (Marsh & Martin, 1995). Since the kinematic joint moments increase to produce increased limb segment accelerations, these accelerations will require either more forceful muscle contractions or a muscle burst shift to earlier in the crank cycle in order to appropriately decelerate the limb segment at the end of the range of motion. If these adjustments do not take place, then the lower extremity would no longer follow the constrained cyclical motion of cycling. Consequently, some subjects might have attempted to adjust their muscle activity to produce more forceful contractions, suggested by increased peak EMG magnitudes, in order to decelerate the limb segments (Neptune et al., 1997). Conversely, other subjects may shift their muscular activity earlier in the crank cycle to

account for the EMD in order to appropriately decelerate the limb segments (as suggested by Marsh & Martin, 1995).

Onset and Offset Threshold Subjectivity

In addition to analyzing the peak timing and magnitudes of EMG profiles, determining the onset and offset timings of a muscle's EMG burst can yield valuable information about muscle activities. Currently, methods to determine threshold values for muscle burst onset and offset are highly subjective. One method is to calculate the maximum EMG value the muscle burst reaches, and then consider the muscle active once the burst surpasses a predetermined percentage of that maximal value. For example, Hull and Hawkins (1990) used a threshold of 30% while Li and Caldwell (1998) used 25% as their threshold values. Another common method for determining muscle burst onset and offset criterion is to find a "quiet" portion of the EMG signal for a muscle and consider the muscle active once the burst exceeds several standard deviations of the baseline for a specified period of time (e.g. Karst & Hasan, 1991; Raasch et al., 1997). This method also employs subjectivity since researchers must arbitrarily decide how many standard deviations above the baseline to use and the amount of time the muscle burst must remain above that threshold. These discrepancies are quite prevalent in the literature. For example, Raasch et al. (1997) used a threshold of three standard deviations above a relaxed baseline for a period of at least 55 ms to determine muscle activity. However, using the same method, Karst and Hasan (1991) decided upon ten standard deviations above the baseline for a 7.5 ms period as appropriate. Using these arbitrary methods can affect results, and therefore conclusions, about muscle activity differently depending on the criterion chosen, but no agreement

on a standard method to determine muscle burst onset and offset thresholds has been reached (Hodges & Bui, 1996; Li & Caldwell, 1999).

Because of the subjectivity of onset and offset threshold values, Li and Caldwell (1999) developed a technique using the coefficient of cross-correlation to compare activity patterns and phase differences between curves. This method can be used to determine muscular coordination changes and coupling between conditions. The method uses objective measures, so a researcher can confidently state that significant differences exist due to variable manipulations and not because of a subjective event threshold.

Muscle Function and Coordination

Upon determining the methods to investigate muscle activities during an action, the information gained can be used to make inferences regarding muscle functioning. Knowledge of muscle functions can shed light on why muscles may coordinate in a certain way. Several studies have examined the functional roles of the muscles during the cycling action, and muscles of the lower extremity may have different roles in cycling depending on how many joints the muscles traverse, as hypothesized by many, e.g., Ingen Schenau et al. (1992, 1994). It was suggested that muscles spanning one joint (mono-articular muscles) primarily function to produce force (Bolhuis et al., 1998; Ingen Schenau et al., 1992, 1994, 1995). In contrast, muscles spanning two joints (bi-articular muscles) may control the direction of the forces by regulating the net distribution of moments over the joints (Ingen Schenau et al., 1992, 1994). Jacobs et al. (1996) also observed that bi-articular muscles contribute to the power transfer from proximal to distal joints. They explained that this transfer allows for joints with small

muscles (i.e. distal joints) to have greater power outputs. Moreover, co-contraction of antagonistic bi-articular muscles may provide joint stability by reducing bone displacement and rotation (Hirokawa et al., 1991) or by equalizing the pressure distribution in the articular surface (Solomonow et al., 1988). Consequently, co-contraction of bi-articular muscles with their mono-articular counterparts may occur for several important reasons.

In order to pedal effectively, the mono- and bi-articular muscles must work together to produce a coordinated movement. Both the timing of the muscle contractions and the force production must be coordinated for this to occur. As the task changes, such as a change in cadence or load, the mono- and bi-articular muscles may shift their coordination patterns accordingly. Mono-articular muscles may react differently from bi-articular muscles, as mono-articular muscles may experience onset, offset, and peak timing shifts (Li & Caldwell, 1997; Marsh & Martin, 1995; Neptune et al., 1997; Suzuki et al., 1982), while bi-articular muscles may experience increased bifunctionality at greater cadences (Suzuki et al., 1982). Furthermore, Neptune et al. (1997) reported that some muscles respond differently to changes in cadence, and activities of some muscles may have no association with cadence at all. Therefore, if some muscles react to cadence while other muscles have no response, then one may predict that cadence changes will affect muscular coordination.

Inertial Effects in Gait

Several studies relating to gait exist that alter the inertial properties of the lower extremities. For example, Myers and Steudel (1985) reported that adding mass to the ends of limbs produced a greater energy cost than adding to the center of mass (at the

torso). This implies that increased mass, which will increase inertia, has a definite influence on muscular function. It was further demonstrated that the position of the added mass also affected the magnitude of the inertial influence. Mattes et al. (2000) studied the effects of various prosthetic masses and moments of inertia on patients with transtibial amputation. They reported that compared to a prosthetic limb with lesser inertial properties, a limb that closely matched the mass and moment of inertia of the patient's residual limb resulted in an increased energy cost and less symmetrical step length, swing time, and stance time. In a recent study, Chang et al. (2000) reported that although increasing and decreasing masses (altering inertial properties) affected the horizontal forces in running, gravity influenced these forces more than inertia. Martin and Cavanagh (1990) added up to 0.5 kg to the thigh and the foot and observed that loads affect the kinetics during running, but the change of inertia did not significantly affect the movement pattern of the swing leg. They noticed increases in joint reaction forces and net moments at the joints proximal to the added load. They suggested that the lower extremity muscles increased their output in response to the inertial loading. Combining the results of these studies suggests that changing the inertial properties of the limbs will affect the musculature, but EMG data to support this interpretation is scarce.

Inertial Effects in Cycling

During cycling, changing the cadence has been shown to elicit changes in pedal forces, joint moments, and muscular activity (e.g. Hull & Jorge, 1985; Kautz & Hull, 1993; Li, 1999; Neptune & Herzog, 1999; Redfield & Hull, 1986). Recently, researchers have begun to decompose these measurements into components to further

understand the mechanisms controlling the various responses to cadence. For example, pedal force and crank torque have been decomposed into muscular and non-muscular components (Fregly & Zajac, 1996; Kautz & Hull, 1993; Neptune & Herzog, 1999). The muscular component consists of forces or torques due solely to muscular activity, while the non-muscular component comprises all other forces or torques acting on the pedal or crank and may include inertia and gravity (Kautz & Hull, 1993). Neptune and Herzog (1999) observed that overall pedal forces experienced a quadratic trend with a minimum value at 90 rpm. The muscular component of the pedal force may remain relatively constant (Kautz & Hull, 1993) or decrease linearly (Neptune & Herzog, 1999) as cadence increases, but the overall pedal force may exhibit a quadratic trend because of a marked increase in the non-muscular component of the pedal force with cadence. Kautz and Hull (1993) observed that the non-muscular component of pedal forces increased as cadence increased from 70 to 110 rpm, and that a substantial non-muscular component existed even at 70 rpm. Since gravitational effects should remain fairly constant across cadence conditions at the same body position (Brown et al., 1996), the increase in the non-muscular component reflects an increase of the inertial influence on pedal forces at greater cadences. Neptune and Herzog (1999) also observed that the average non-muscular pedal forces increased with cadence, influencing the pattern of the overall pedal forces. These authors also observed a similar trend for the non-muscular crank torques as cadence increased. Li (1999) reported that muscular coordination changed as cadence increased. Based on the magnitude of the changes in a proximal (greater changes) to distal (lesser changes) order, he concluded that the greater inertial properties of the proximal limb segments played an important role in the

coordination changes. These experiments support the notion that inertia accounts for significant portions of kinetic and muscular changes at greater cadences. However, the effects of inertial properties were compounded with cadence effect in previous studies; consequently, the influence of inertia was not investigated independent of cadence, so attributing observed effects solely to inertia may result in misconceptions. Therefore, a question that has arisen is, does inertia, independent of cadence, affect muscular coordination? Keeping cadence constant and manipulating the inertial component with added masses to lower extremity segments may provide a window to investigate this issue. Few data exist related to altering inertial properties during cycling, and suggestions that inertial property alterations in gait and prosthetic literature affect the musculature of the lower extremity lack EMG data confirmation. Furthermore, since walking and running are complex forms of locomotion, it is more difficult to isolate the effect of inertial properties. Cycling is a more constrained form of motion, and the effect of inertial property manipulation on neuromuscular coordination can be investigated independent of other variables.

Purpose and Hypotheses

The purpose of this study was to investigate and distinguish between the effects of cadence and inertial influences on lower extremity neuromuscular coordination during cycling. More specifically, several hypotheses were examined. First, because some muscles may respond differently to cadence manipulations than others (Neptune et al., 1997), it was hypothesized that there would be coordination changes among mono- and bi-articular antagonistic muscles due to both cadence and load manipulations. Literature has reported that cadence effects on the lower extremities are

more considerable proximally than distally because of the greater inertial properties of the limbs in this order (Li, 1999). Therefore, the second hypothesis was that both cadence and load manipulations would have greater effects proximally than distally. Furthermore, functional differences were thought to exist between mono- and bi-articular muscles (i.e. Ingen Schenau et al., 1992, 1994), so there was reason to expect that the two muscle types may respond differently to external manipulations. Consequently, it was hypothesized that mono- and bi-articular muscles would react to load and cadence changes differently: the mono-articular muscles would react by shifts in timing whereas the bi-articular muscles would experience changes in function. Finally, it has been reported that the non-muscular component increases with cadence (Kautz & Hull, 1993; Li & Caldwell, 1999; Neptune & Herzog, 1999). Therefore, a load at the distal end of the thigh, which will increase the inertia of the thigh, should show increased influences at greater cadences.

METHODS

Subjects

Sixteen male subjects (mean \pm standard deviation: age: 23 ± 5 years; height 1.8 ± 0.2 m; body mass: 85 ± 10 kg) were recruited from the Louisiana State University community. The number of subjects was chosen based on power tables for repeated measures designs (see Table 13.5 in Stevens, 1996). Each subject had at least two years of recreational cycling experience. In compliance with the university human subject policy, subjects gave their informed consent before the experiment. The experiment was conducted in the Motor Behavior Laboratory of Louisiana State University.

Experimental Protocol

The muscle activities of gluteus maximus (GM), rectus femoris (RF), biceps femoris (BF), vastus lateralis (VL), tibialis anterior (TA), medial gastrocnemius (GAS), and soleus (SOL) of the left lower extremity were monitored with surface EMG. After sites were shaved, lightly abraded, and cleaned with alcohol, pairs of silver/silver chloride pre-gelled surface electrodes of 1.5 cm diameter (Marquette Medical Systems, Jupiter, Florida) with a center to center distance of 2.5 cm were applied along the muscle fibers over the bellies of seven muscles for EMG data acquisition. A common reference electrode was placed on a bony site at the distal end of the left ulna. A bipolar differential amplifier with an input impedance of $1 \text{ M}\Omega$, a gain up to 10,000, a Common Mode Rejection Ratio (CMRR) of 90 dB at 60 Hz, and a signal to noise ratio of $< 0.8 \mu\text{V}$ at 60 Hz was used during data collection to optimize the resolution of the EMG signal.

The seven muscles were chosen based on several criteria. First, the muscles chosen cross all three of the major joints in the lower extremity (hip, knee, and ankle); second, the muscles represent mono- and bi-articular antagonist muscle groups that cross each of the joints; and finally, the seven muscles chosen include those most commonly found in literature using cycling as a means of investigation.

The subjects rode a Monarch cycling ergometer (model 868, Stockholm, Sweden) at a constant power output of 250 watts. The resistances were changed according to the desired pedaling cadences to meet this requirement. The seat height was adjusted such that the distance between the seat and the crank center was 100% of the subject's greater trochanter length (e.g. Hull & Gonzalez, 1988; Hull & Jorge, 1985; Jorge & Hull, 1986). Cadence conditions of 60, 80, and 100 revolutions per minute (rpm) represented low, medium, and high cadence conditions, respectively. The pedaling cadence was monitored with a cadence monitor attached to the ergometer. The mean (SD) cadences calculated from kinematic data of each subject were 63.6 (1.1), 83.4 (0.9), and 101.9 (0.9) rpm. Five different loads of 0, 0.5, 1.0, 1.5, and 2.0 kg were attached to the distal end of the thigh via Flexi-Wrap (Cramer, Gardener, KS). Loads were created using plastic Ziplock[®] bags reinforced with duct tape and filled with sand. Positioning the loads at the exact same relative position on each subject was not possible due to inter-subject anthropomorphic differences and the variability of sand as a load. However, since the goal of the load was simply to change the inertial properties at the distal end of the thigh, the precise load location was not considered a major drawback (Martin & Cavanagh, 1990). The testing order was randomized to minimize possible order effects, such as fatigue and learning. A random number generation

computer program produced the order for each subject. That order was then altered so that the same load was collected for three consecutive trials at the different cadences. The pseudo-random order was used to control the load location variability across different cadences. In each condition, the subjects pedaled for approximately one minute to establish a steady state before a ten-second data collection trial. Subjects were pedaling at a power output of 250 W, and since this power output is relatively small in relation to maximal power output values of 1300 W or more between 60 and 100 rpm (McCartney et al., 1983), fatigue was not considered a major factor. Nonetheless, each subject rested for at least one minute between trials with an extra five minutes of rest after every three trials to control for fatigue.

Data Collection

An internally synchronized motion analysis system was employed to collect data. The system included a 2-D 60 Hz camera (MotionAnalysis, Santa Clara, CA) to capture kinematic data with passive reflective markers and a 16-channel surface EMG system (Run Technologies, Laguna, CA) to capture EMG activity.

Reflective markers were placed on the crank center and pedal spindle of the bicycle, and on the subject's greater trochanter (representing the hip joint), posterior third of the knee line for knee joint, lateral malleolus for the ankle joint, fifth metatarsal-phalange joint, and posterior surface of the heel at approximately the same height of the fifth metatarsal phalange joint marker. The coordinates of these markers were recorded by the camera and automatically digitized by the MotionAnalysis system. The crank arm, identified by the crank center and pedal spindle markers, was used to identify crank cycles. One complete cycle (Figure 1) was defined by consecutive

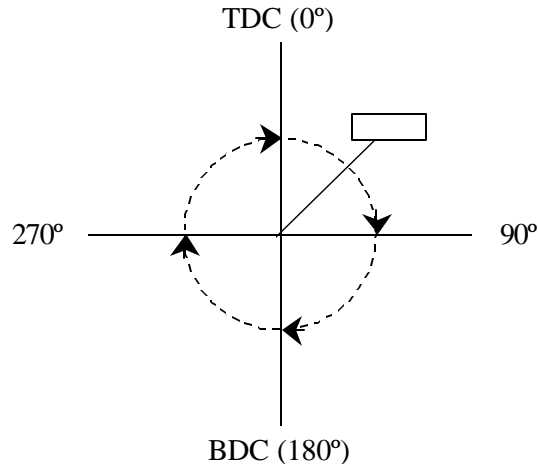


Figure 1. One complete cycle, defined by a full rotation of the pedal from top-dead-center (TDC) to TDC.

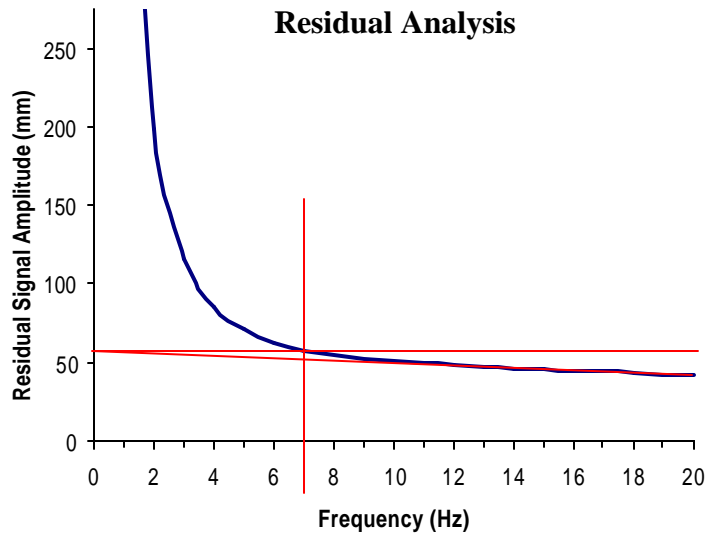


Figure 2. Residual analysis of GM EMG. The analysis suggests a cutoff frequency of 7 Hz.

occurrences of crank top-dead-center (TDC), with the crank arm vertical and the pedal at its highest position.

The EMG data were collected at a sampling rate of 960 Hz. Raw data were unbiased, full-wave rectified, and then smoothed with a low pass, fourth order, zero lag Butterworth filter at 7 Hz to create the linear envelope. The cutoff frequency of 7 Hz

was chosen upon completing a residual analysis (Figure 2; Winter, 1990), and the linear envelope at this frequency reflected the pattern of the rectified raw data (Figure 3). The lab configuration was free of high frequency noises, and the motion artifact was less than 2 Hz, so it was not deemed necessary to use a bandpass filter.

After smoothing the data, ensemble curves were created from five consecutive cycles. Finally, for each subject, the EMG magnitudes of different conditions were normalized to a percentage of the maximum value found across all conditions for each individual muscle. For example, Subject 1's GM obtained a maximum value in Condition 2. Therefore, all fifteen conditions for the GM in Subject 1 were normalized as a percentage of that maximum value in Condition 2.

In order to assess muscle coordination and pattern changes, onset and offset thresholds along with the coefficient of cross correlation were calculated. A threshold value of 10% of the maximum value across conditions was chosen as the onset and

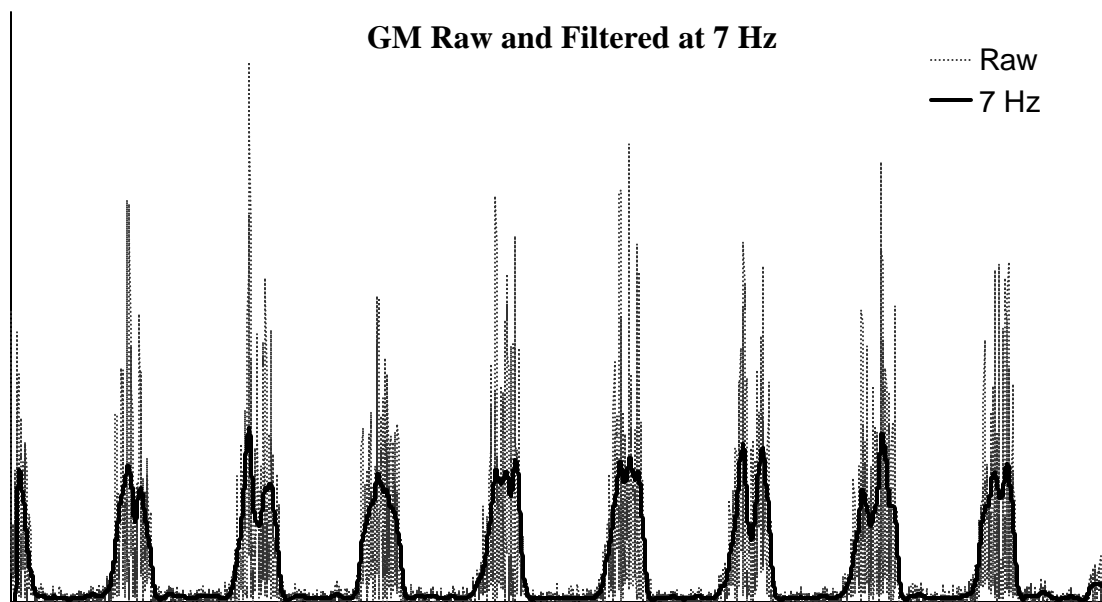


Figure 3. Filtered data at 7 Hz superimposed on the raw, rectified data. The linear envelope of the filtered data reflects the pattern of the raw data.

offset criterion. Visual inspection determined if this threshold was appropriate. Appropriate thresholds reflected easily identifiable onset and offset points and minimal discrepancies in identifying non-meaningful bursts. In the case that 10% was considered inappropriate, the threshold was raised to 20% of the maximum value across conditions (see Figure 4). Upon reaching the determined threshold, the muscle was considered active, and the muscle “burst” duration was defined as the duration, in degrees, of the crank angle between the onset and offset values (Figure 5). The coefficient of cross correlation and time domain correspondence was employed for pattern recognition and phase shifting comparisons (see Li & Caldwell, 1999 for a detailed description of using this method). Inter-task cross-correlations with no phase shift yielded the $r_{\text{task}}(0)$ value to test the change of muscular activity patterns across

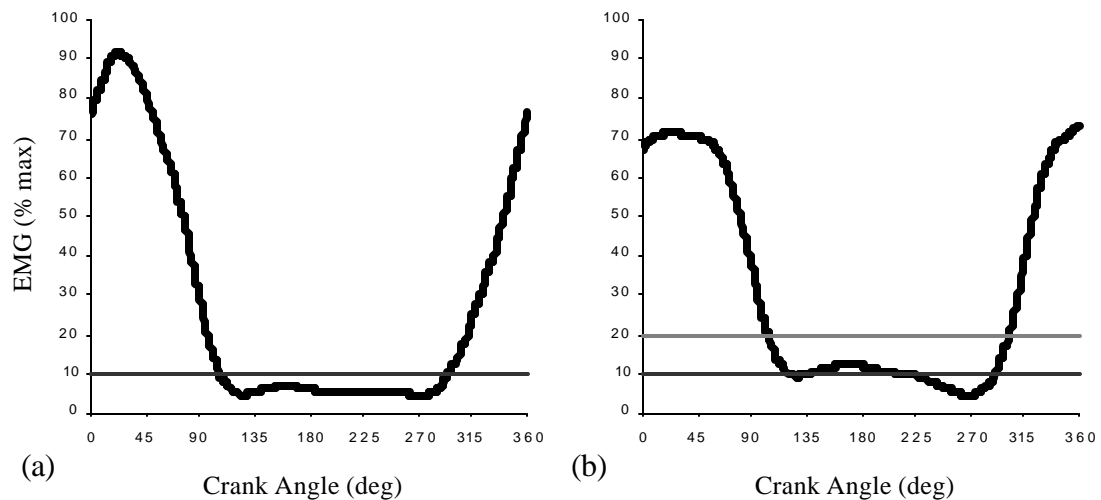


Figure 4. The graphs display examples of EMG linear envelopes with onset and offset threshold criterion. In (a) 10% of the maximum value across conditions was considered appropriate for the onset and offset threshold criterion. In (b) 20% of the maximum value was considered appropriate for the threshold criterion. The 10% threshold value was deemed inappropriate because it created a problematic determination of when the muscle was considered “on” or “off.”

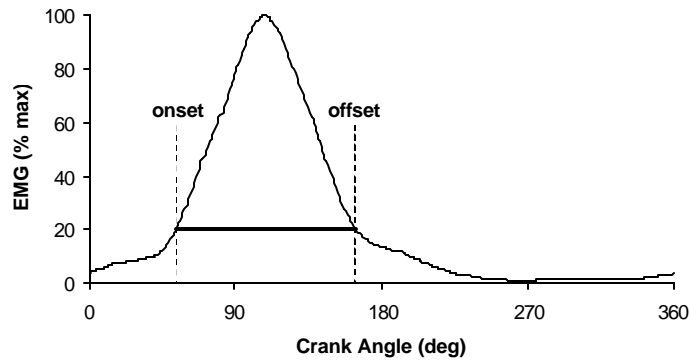


Figure 5. EMG muscle burst onset and offset values as functions of crank angle. The horizontal black line represents the duration of the active muscle. An arbitrary threshold of 20% of maximum EMG is demonstrated.

conditions. The 95% confidence interval (CI) of $r_{\text{task}}(0)$ was used to compare the muscle responses to the cadence and load changes. The $r_{\text{task}}(k)$ [parenthetical values (k) represented the degrees shifted] value indicated the highest correlation value between the two conditions tested, where k represented the degrees of the crank angle shifted to reach this r -value. The between muscle cross-correlation with a zero phase shift was represented by the $r_{\text{pair}}(0)$ value, which was calculated for the mono- and bi-articular antagonist muscle pairs GM/RF, BF/VL, VL/GAS, and TA/GAS across the three cadence and five load conditions. High positive $r_{\text{pair}}(0)$ values indicated tight coupling (i.e. co-contraction) between the muscles, whereas, on the opposite end of the continuum, high negative $r_{\text{pair}}(0)$ values indicated little to no coupling. The 95% CI of $r_{\text{pair}}(0)$ was used to compare the muscle pair responses to the changing conditions. The angle of the highest $r_{\text{pair}}(k)$ value was used to indicate a coordination change due to a change in condition. If a significant difference was seen between conditions for the $r_{\text{task}}(0)$ or $r_{\text{pair}}(0)$, then a coordination change was declared. Low $r_{\text{task}}(0)$ or $r_{\text{pair}}(0)$ values indicated large amounts of variance originated either from differences of firing pattern

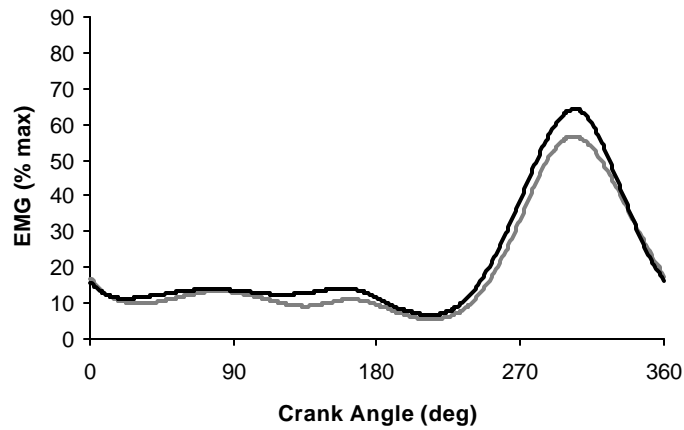


Figure 6. Demonstrates a high r-value (0.997) indicating similar patterns and phase timing between the two curves.

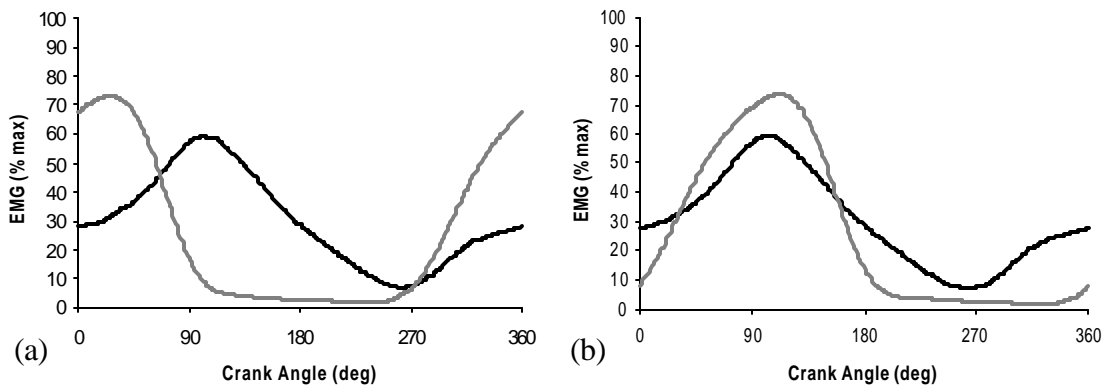
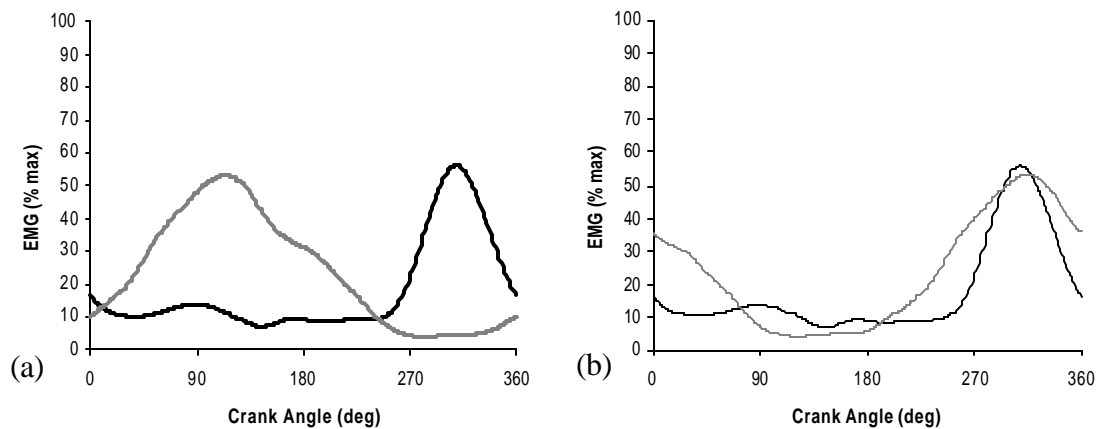


Figure 7. (a) demonstrates a low r-value (0.071) indicating dissimilar patterns, phase difference, or both between the two curves. (b) demonstrates a phase shift of $k = 86^\circ$ between the curves in (a). The phase shift improved the correlation values from $r(0) = 0.071$ to $r(86) = 0.945$. This suggests that most of the variability between the two curves is due to phase difference rather than pattern difference, although some pattern differences are apparent.



Figures 8a-b. Demonstrate r -value variability due to both phase difference and pattern differences. The difference in r -value magnitude between (a) $r(0) = -0.590$ and (b) $r(k) = 0.824$ indicates that phase difference accounts for some variability, but the low $r(k)$ value in (b) indicates variability due to activity pattern differences.

or from patterns shifted in the time domain, or possibly both. A low $r_{\text{pair}}(k)$ indicated a dramatic pattern difference, where a significant phase shift was identified if the k value was significantly different from 0. If the k value was not different from 0 then much of the variance between muscles was attributed to a pattern difference (Figures 6-8).

Statistical Analyses

A 3 (cadence) X 5 (load) factorial analysis of variance (ANOVA) with repeated measures on both factors was employed to test for statistical significance. Tukey's HSD post-hoc analysis was employed when necessary. Differences between the muscular responses to cadence and load manipulations were assessed by the cross-correlation method. Furthermore, trend analyses were performed to identify significant trends for cadence, load, and interaction. The significance level for all statistical tests was set at $\alpha = 0.05$.

RESULTS

Patterns of muscle activity are displayed in Figures 9a-b as EMG ensemble curves of all seven muscles in the three cadence conditions and the five load conditions, respectively. Figures 10a-b show the EMG burst onsets and offsets and their changes with respect to varying cadences and loads.

Onset

Table 1 displays the EMG crank angle values for EMG onset across cadence conditions for the seven muscles investigated. As pedaling cadence increased, GM, RF, BF, VL, TA, and SOL exhibited significant ($p < 0.05$) changes in crank angles of muscle burst onset. Out of these muscles, all but SOL exhibited a significant linear trend with the onset timing shifting to an earlier crank angle with increased cadence. Although SOL exhibited significant changes in onset angle, no significant trend was observed, though visual inspection of the data showed a slight timing shift to a later crank angle as cadence increased. GAS did not exhibit significant differences in onset timing or a significant trend for onset due to cadence alterations. Table 2 shows the crank angle values for EMG onset across load conditions. Only RF and SOL exhibited significant differences in onset crank angle due to changing the inertial loads. RF and BF displayed significant linear trends having a general pattern of the muscle burst onset shifting to an earlier crank angle as the loads increased. SOL exhibited a significant quadratic trend with its minimum onset crank angle at a load of 1.0 kg. Changing loads had no significant effect on the EMG burst onsets of GM, VL, TA, or GAS. No significant interactions between frequency and load conditions were apparent to the onset angles of any muscle; however, several significant interaction trends existed. The

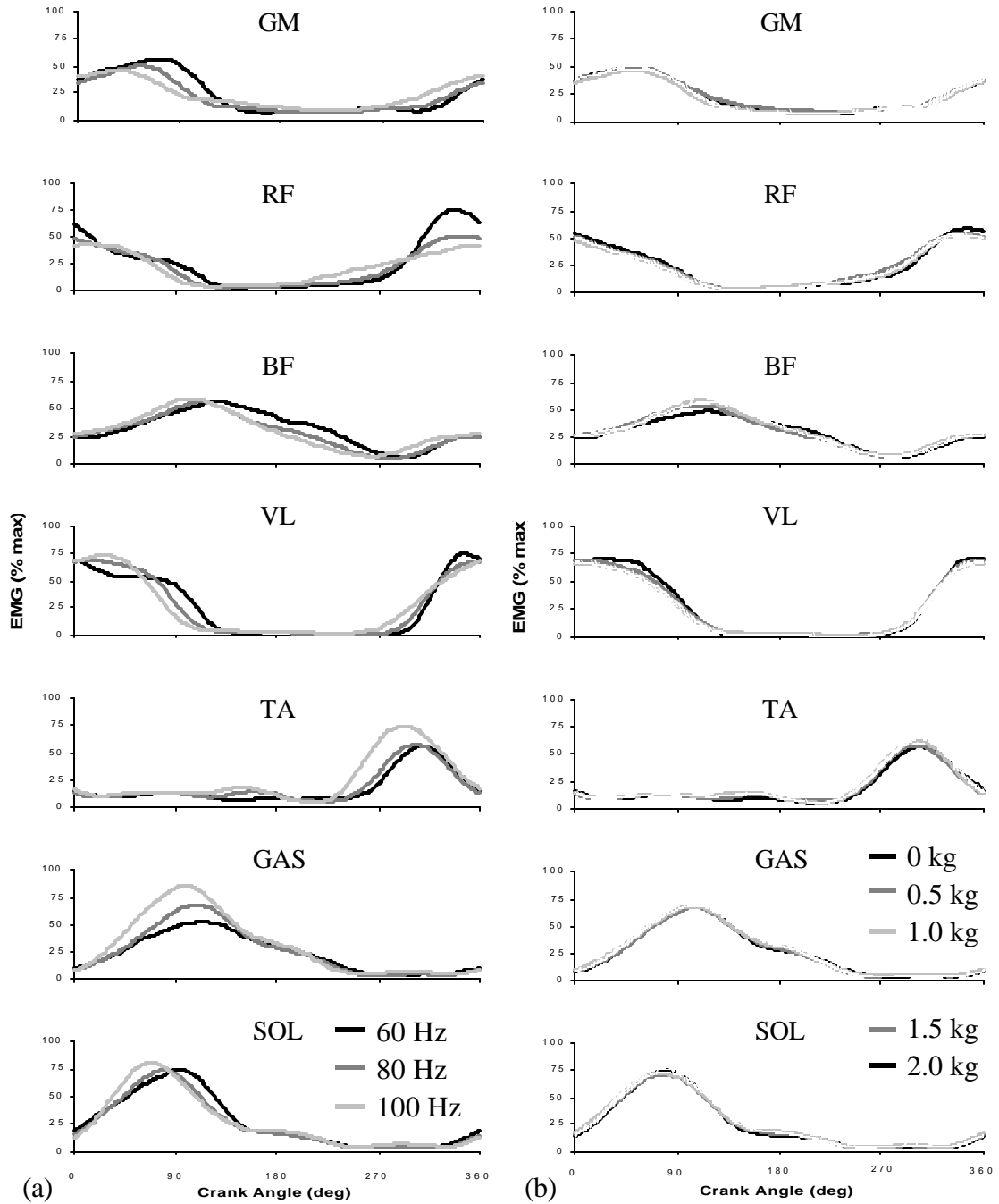
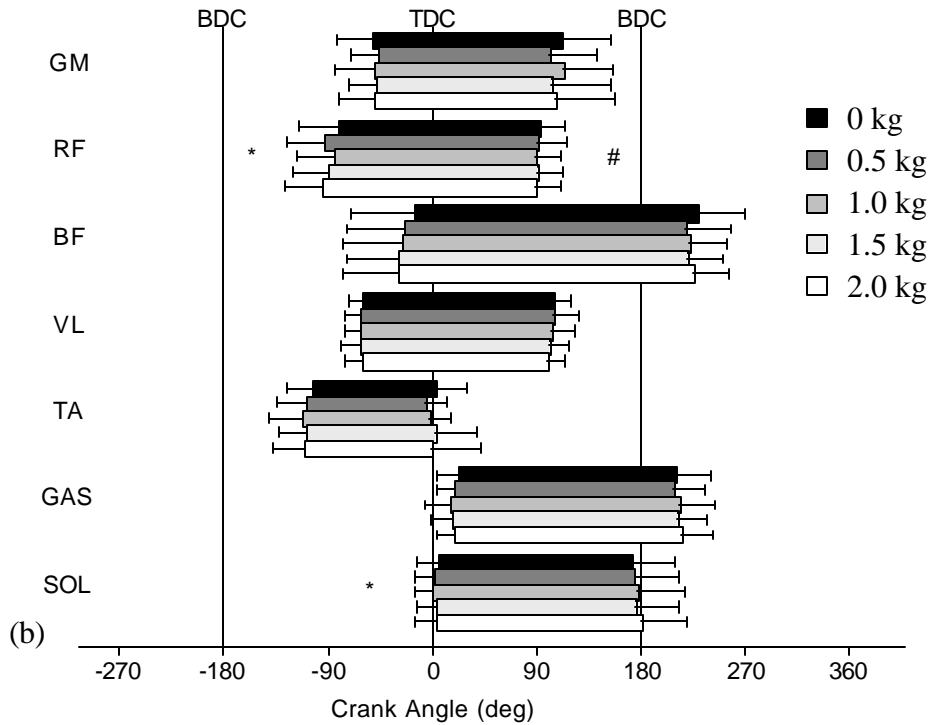
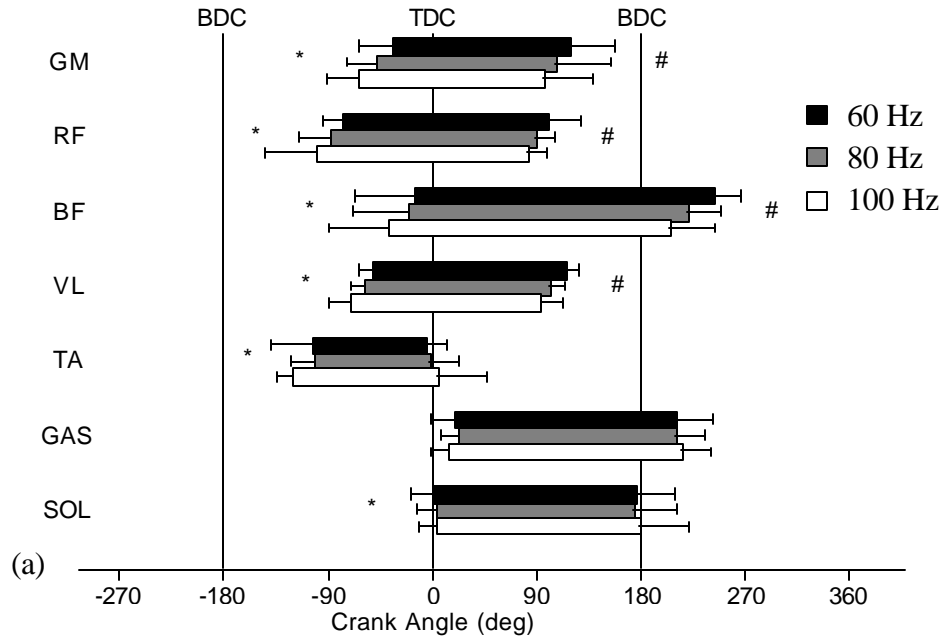


Figure 9. Mean ensemble curves of EMG activity for each muscle across (a) cadence and (b) load conditions. The crank angle range represents TDC to next TDC, 0 to 360°. EMG curves for each subject were normalized to the maximum value observed across all 15 conditions. The mean curves here were calculated from the normalized individual curves of gluteus maximus (GM), rectus femoris (RF), biceps femoris (BF), vastus lateralis (VL), tibialis anterior (TA), gastrocnemius (GAS) and soleus (SOL).



Figures 10a-b. Mean onset, offset, and duration of EMG linear envelopes of gluteus maximus (GM), rectus femoris (RF), biceps femoris (BF), vastus lateralis (VL), tibialis anterior (TA), gastrocnemius (GAS), and soleus (SOL) across (a) cadence and (b) load conditions, respectively. The left and right edges of each rectangle represent mean onset and offset values, respectively. Error bars represent one standard deviation of the mean onset and offset. * and # indicate a statistically significant difference ($p < 0.05$) between cadence or load conditions for onset and offset, respectively.

onset crank angles of GM and SOL showed a quadratic by linear (cadence by load, respectively) interaction trend; a linear by linear interaction trend was observed for BF and VL; and TA displayed a linear by quadratic interaction trend.

Offset

Table 3 shows the crank angle values for EMG offsets for the muscles investigated across cadence conditions. GM, RF, BF, and VL experienced both significant differences and significant linear trends where the muscle bursts offset at earlier crank angles as cadence increased. TA, GAS, and SOL had no apparent

TABLE 1. Onset of EMG activity across cadences expressed as a function of the crank angle, in degrees.

Cadence (rpm)	60	80	100
GM*	326 (28) ^a	312 (27) ^a	297 (28) ^b
RF*	283 (18) ^a	272 (26) ^{ab}	260 (44) ^b
BF*	344 (49) ^a	340 (47) ^a	323 (53) ^b
VL*	309 (11) ^a	302 (13) ^b	290 (19) ^c
TA*	257 (35) ^a	259 (19) ^a	241 (15) ^b
GAS	21 (22)	24 (17)	16 (18)
SOL	0 (18) ^a	3 (16) ^{ab}	4 (15) ^b

Values are mean (\pm SD)

* Significant linear trend

^{a,b,c} Indicate significantly different heterogeneous groups

TABLE 2. Onset of EMG activity across loads expressed as a function of the crank angle, in degrees.

Load (kg)	0	0.5	1.0	1.5	2.0
GM	310 (32)	314 (25)	311 (36)	313 (26)	311 (32)
RF*	279 (33) ^a	268 (33) ^b	275 (32) ^{ab}	271 (31) ^{ab}	267 (33) ^b
BF*	345 (55)	337 (51)	335 (52)	331 (45)	331 (49)
VL	301 (14)	301 (16)	300 (16)	299 (18)	301 (17)
TA	257 (22)	252 (26)	249 (29)	252 (24)	251 (28)
GAS	23 (19)	21 (18)	17 (23)	19 (20)	21 (17)
SOL [#]	5 (18) ^a	1 (16) ^{ab}	0 (15) ^b	3 (17) ^{ab}	3 (18) ^{ab}

Values are mean (\pm SD)

* Significant linear trend

[#] Significant quadratic trend

^{a,b} Indicate significantly different heterogeneous groups

TABLE 3. Offset of EMG activity across cadences expressed as a function of the crank angle, in degrees.

Cadence (rpm)	60	80	100
GM*	117 (39) ^a	107 (47) ^{ab}	97 (42) ^b
RF*	99 (29) ^a	89 (19) ^b	81 (18) ^b
BF*	242 (24) ^a	220 (28) ^b	200 (39) ^c
VL*	115 (11) ^a	101 (14) ^b	92 (20) ^c
TA	355 (19)	356 (27)	364 (43)
GAS	209 (32)	209 (26)	214 (25)
SOL	175 (35)	172 (39)	180 (42)

Values are mean (\pm SD)

* Significant linear trend

^{a,b,c} Indicate significantly different heterogeneous groups

TABLE 4. Offset of EMG activity across loads expressed as a function of the crank angle, in degrees.

Load (kg)	0	0.5	1.0	1.5	2.0
GM	111 (41)	102 (40)	112 (42)	104 (50)	106 (42)
RF*	93 (21) ^a	90 (27) ^{ab}	88 (23) ^{ab}	89 (24) ^{ab}	87 (24) ^b
BF	221 (42)	217 (40)	221 (32)	220 (30)	224 (31)
VL*	105 (14)	105 (23)	103 (21)	102 (16)	99 (15)
TA	361 (30)	354 (20)	356 (21)	362 (36)	359 (43)
GAS	209 (31)	207 (27)	213 (31)	211 (25)	213 (27)
SOL*	170 (40)	173 (40)	177 (40)	175 (37)	181 (38)

Values are mean (\pm SD)

* Significant linear trend

^{a,b} Indicate significantly different heterogeneous groups

differences in offset due to cadence. Table 4 shows the crank angle values for EMG offsets as related to inertial loads. Significant differences in offset crank angle due to loads were only observed in RF. RF and VL exhibited significant linear trends with their offsets occurring at earlier crank angles with increasing loads. The offset crank angle of BF experienced a significant interaction between the two conditions. BF also displayed a significant linear by linear interaction trend.

TABLE 5. Duration of EMG activity across cadences expressed as a function of degrees, from onset to offset, through the crank angle.

Cadence (rpm)	60	80	100
GM	152 (41)	155 (50)	160 (54)
RF	176 (35)	177 (31)	181 (52)
BF*	258 (65) ^a	240 (61) ^b	241 (72) ^b
VL	166 (15)	159 (16)	163 (27)
TA*	97 (44) ^a	98 (36) ^a	123 (45) ^b
GAS	188 (39)	185 (35)	198 (33)
SOL	175 (38)	169 (44)	175 (48)

Values are mean (\pm SD)

* Significant linear trend

^{a,b} Indicate significantly different heterogeneous groups

TABLE 6. Duration of EMG activity across loads expressed as a function of degrees, from onset to offset, through the crank angle.

Load (kg)	0	0.5	1.0	1.5	2.0
GM	162 (54)	148 (37)	162 (55)	150 (44)	155 (50)
RF	174 (39)	182 (40)	173 (38)	178 (41)	181 (42)
BF	243 (69)	240 (75)	246 (65)	249 (57)	253 (65)
VL*	164 (15)	164 (26)	163 (22)	163 (20)	158 (17)
TA	104 (39)	102 (37)	107 (37)	110 (49)	108 (52)
GAS	186 (40)	186 (33)	195 (43)	192 (32)	193 (65)
SOL*	165 (48) ^a	172 (44) ^{ab}	177 (42) ^{ab}	173 (40) ^{ab}	178 (44) ^b

Values are mean (\pm SD)

* Significant linear trend

^{a,b} Indicate significantly different heterogeneous groups

Duration

Tables 5 and 6 illustrate the duration of EMG activity, in degrees of the crank cycle, for each muscle across cadence and load conditions, respectively. Only BF and TA showed significant differences in duration due to cadence, and only SOL exhibited a significant difference in duration due to load. BF and TA demonstrated significant linear trends due to cadence. The duration of BF decreased while the duration of TA increased with increasing cadence. In relation to the changing loads, VL and SOL displayed significant trends. Here, the duration of VL decreased whereas SOL's

duration increased with increasing loads. No significant interactions were apparent; however, both RF and BF displayed significant linear by linear interaction trends.

Peak EMG Magnitude

Peak EMG magnitude values were calculated by normalizing all the values within each muscle to the maximum value found across the fifteen conditions within each subject. Tables 7 and 8 present data for the peak EMG magnitude values for cadence and load conditions, respectively. RF, VL, TA, and GAS demonstrated significant differences in peak EMG magnitude values due to cadence changes. RF presented both significant linear and quadratic trends. In addition, TA, GAS, and SOL displayed significant linear trends with increasing peak values as cadence increased. Only BF showed significant differences in peak magnitudes due to load. BF and TA showed significant linear trends due to load where the peak magnitude generally increased with increasing loads. Both VL and TA exhibited significant interactions. Four out of the seven muscles displayed a significant interaction trend with BF, TA, and SOL demonstrating a linear by linear trend and VL exhibiting a quadratic by linear interaction trend.

TABLE 7. Mean peak of EMG activity per cycle across cadences expressed as a percentage of the maximum value of each muscle.

Cadence (rpm)	60	80	100
GM	66 (25)	57 (23)	58 (21)
RF ^{*#}	83 (17) ^a	58 (20) ^b	52 (21) ^b
BF	73 (20)	69 (18)	71 (21)
VL	82 (14) ^a	76 (17) ^b	78 (14) ^{ab}
TA [*]	63 (19) ^a	65 (18) ^a	79 (16) ^b
GAS [*]	58 (16) ^a	72 (14) ^b	90 (11) ^c
SOL [*]	77 (14)	77 (13)	84 (12)

Values are mean (\pm SD)

* Significant linear trend

Significant quadratic trend

^{a,b,c} Indicate significantly different heterogeneous groups

TABLE 8. Mean peak of EMG activity per cycle across loads expressed as a percentage of the maximum value of each muscle.

Load (kg)	0	0.5	1.0	1.5	2.0
GM	60 (21)	63 (21)	58 (26)	60 (25)	61 (24)
RF	68 (24)	65 (23)	62 (24)	65 (24)	61 (23)
BF*	67 (24) ^a	68 (19) ^a	71 (22) ^{ab}	77 (19) ^b	72 (18) ^{ab}
VL	82 (12)	80 (13)	76 (17)	79 (17)	76 (15)
TA*	66 (17)	65 (20)	70 (19)	73 (19)	72 (20)
GAS	73 (16)	71 (21)	74 (19)	74 (20)	74 (17)
SOL	80 (12)	77 (14)	79 (12)	82 (13)	80 (14)

Values are mean (\pm SD)

* Significant linear trend

^{a,b} Indicate significantly different heterogeneitic groups

Peak EMG Timing

This value is equivalent to the point in the crank cycle, in degrees, where the peak EMG magnitude was observed. Tables 9 and 10 present data for the peak EMG timing related to changes in cadence and load, respectively. Figures 9a-b display the EMG linear envelopes of mean muscle activity for the subjects. BF, VL, TA, GAS, and SOL exhibited significant differences in peak EMG timing due to cadence. Six out of the seven muscles (all but RF) showed significantly linear trends. GM, BF, TA, GAS, and SOL displayed significantly earlier peak EMG timing with increased cadences, whereas VL exhibited later peak timing as cadence increased. No significant effect of load on peak timing was observed, and only GM had a significant linear trend because of load changes. In general, the peak GM value occurred earlier in the crank cycle as the load increased. No significant interactions or interaction trends were evident.

TABLE 9. Mean crank angle, in degrees, at which the peak EMG activity per cycle occurred across cadences.

Cadence (rpm)	60	80	100
GM*	49 (52)	37 (46)	32 (49)
RF	341 (18)	340 (30)	341 (49)
BF*	147 (64) ^a	121 (49) ^b	107 (51) ^b
VL*	355 (40) ^a	14 (28) ^b	22 (25) ^b
TA*	311 (23) ^a	303 (20) ^b	294 (18) ^c
GAS*	120 (29) ^a	111 (15) ^a	98 (14) ^b
SOL*	92 (13) ^a	80 (13) ^b	71 (14) ^c

Values are mean (\pm SD)

* Significant linear trend

^{a,b,c} Indicate significantly different heterogeneous groups

TABLE 10. Mean crank angle, in degrees, at which the peak EMG activity per cycle occurred across loads.

Load (kg)	0	0.5	1.0	1.5	2.0
GM*	45 (50)	49 (50)	30 (47)	36 (49)	36 (50)
RF	347 (35)	339 (37)	337 (35)	340 (36)	340 (31)
BF	133 (55)	129 (63)	128 (61)	111 (44)	124 (62)
VL	10 (39)	9 (32)	16 (34)	9 (37)	7 (24)
TA	302 (18)	303 (24)	301 (15)	303 (20)	305 (28)
GAS	108 (24)	111 (24)	113 (27)	108 (20)	108 (17)
SOL	81 (14)	82 (16)	80 (16)	81 (17)	81 (16)

Values are mean (\pm SD)

* Significant linear trend

Coordination of Muscles

Individual Muscle Activity Patterns Across Cadences

Figures 11a-c compare the muscular coordination of the seven lower extremity muscles across the cadence conditions using the cross correlation technique. Low $r_{\text{task}}(0)$ values indicate variance between the compared curves (i.e. the curve for GM 60 versus the curve for GM 100). This variance may be caused by activity pattern differences, phase timing differences, or a combination of the two. Dramatic increases in the r-value from $r_{\text{task}}(0)$ to $r_{\text{task}}(k)$ indicate variance due to phase differences of k

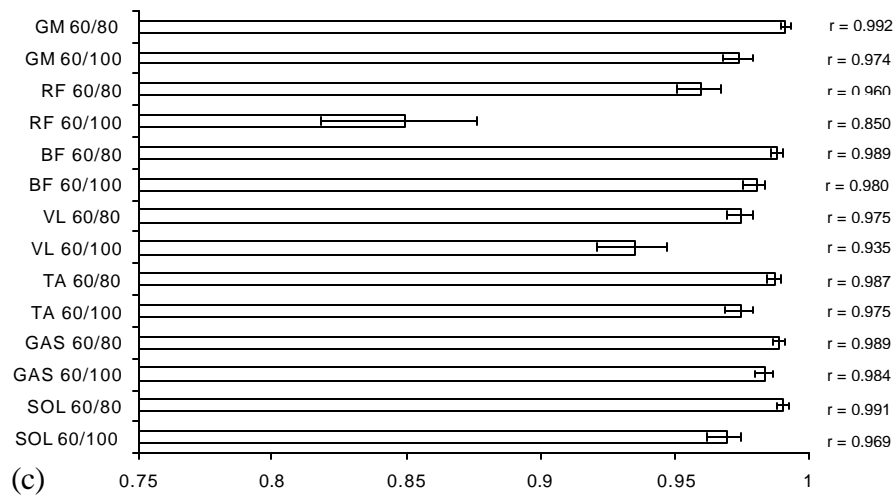
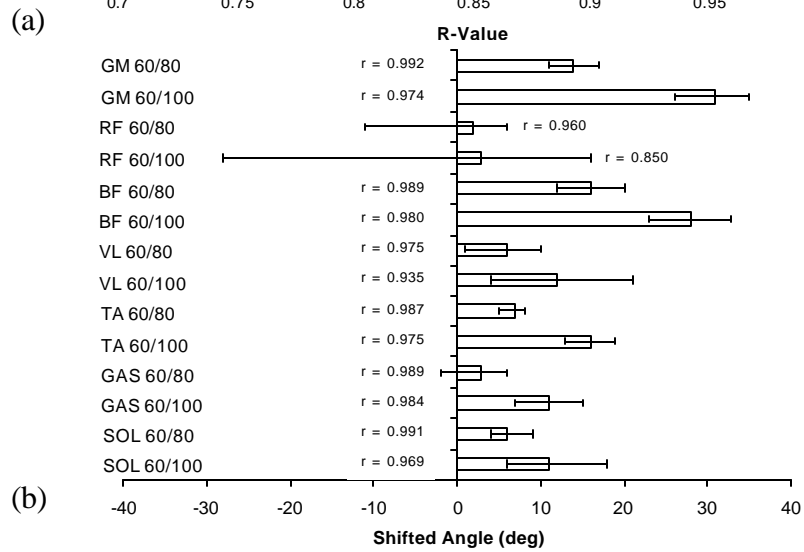
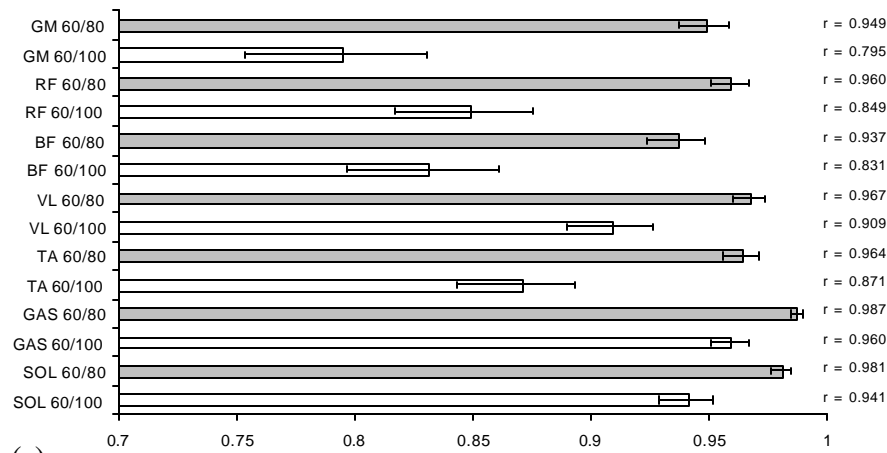


Figure 11. Cross-correlation and shifting values of gluteus maximus (GM), rectus femoris (RF), biceps femoris (BF), vastus lateralis (VL), tibialis anterior (TA), gastrocnemius (GAS), and soleus (SOL) comparing cadence conditions. (a) shows the $r_{\text{task}}(0)$ value (correlation with no shifting), (b) displays the shifted k value to reach the highest correlation, or $r_{\text{task}}(k)$ value, seen in (c).

degrees. A low $r_{\text{task}}(k)$ value indicates variance due to activity pattern changes. Cadence or load effects can be observed when comparing several r - or k -values. Significant differences between these values (i.e. no overlap in CI error bars) will indicate a cadence or load effect.

For GM the low r -value (0.795, Figure 11a) between the 60 and 100 rpm conditions improved dramatically, to 0.974, after a 31-degree phase shift (Figure 11b). This indicates that a difference in the timing of the two curves accounts for most of the variance. However, significant differences (the 95% confidence interval [CI] error bars exhibit no overlapping) exist when comparing the shift between 60 and 80 rpm and the phase shift between 60 and 100 rpm. This indicates a cadence effect on the phase shifting of GM across cadences. Comparing to 60 rpm, GM activity patterns at 80 and 100 rpm shifted 14 and 31 degrees, respectively. Also, significant differences between the 60/80 rpm maximal r -value (0.992) and 60/100 rpm maximal r -value (0.974) indicate the activity pattern change was also influenced by cadence (Figure 11c). Similar results were observed for BF (although with a lesser emphasis on a cadence effect on pattern change) and TA. The variance in the RF 60/100 rpm comparison (Figure 11a) could not be attributed to timing differences between the curves (Figure 11b), since the shifts were not significant (the error bars included 0° within their range). The maximal r -values of RF 60/80 rpm and RF 60/100 rpm were 0.960 and 0.850, respectively (Figure 11c), indicating a pattern change due to cadence. Variation in VL improved with a 6- and 12-degree phase shift for the 60/80 and 60/100 rpm comparisons, respectively, but the cadence effect was not significant indicated by overlapping error bars. Cadence did affect the EMG activity pattern of VL, however.

The VL 60/80 rpm comparison yielded an $r(k)$ value of 0.975 compared to $r(k) = 0.935$ for VL 60/100. The GAS 60/100 rpm comparison exhibited a significant phase shift ($k = 11$ degrees) which accounted for most of the variability within the $r_{\text{task}}(0)$ value, even though the variability was small. Cadence did have an effect on the shifting ($k = 3$ and 11 degrees for GAS 60/80 and 60/100, respectively; Figure 11b), but not on the pattern (Figure 11c). SOL variability across cadences was small, and phase differences of 6 and 11 degrees accounted for most of the variability between the 60/80 and 60/100 rpm comparisons, respectively. Cadence did not affect the phase difference, but the remaining variability across conditions can be accounted for by the cadence effect on the muscle activity pattern.

In summary, cadence effects were apparent on the phase shifting of GM, BF, TA, and GAS, and cadence affected the EMG activity patterns of GM, RF, BF, VL, TA, and SOL.

Coordination of Antagonist Pairs Across Cadences

This study investigated the coordination of four mono- and bi-articular muscle antagonist pairs in the lower extremity, including GM/RF, BF/VL, VL/GAS, and TA/GAS. The EMG linear envelope patterns of the selected muscle pairs across cadence conditions are displayed in Figures 12a-d. Visual inspection yields noticeable differences in some of the activity patterns and couplings across cadence conditions, indicating a coordination change. Figures 13a-c display the $r_{\text{pair}}(0)$ values, phase shifting, and maximal r -values after shifting, respectively, for each muscle pair. High positive $r_{\text{pair}}(0)$ values indicate coupling between the two muscles (i.e. high occurrences of co-contraction), whereas, on the other end of the continuum, high negative $r_{\text{pair}}(0)$

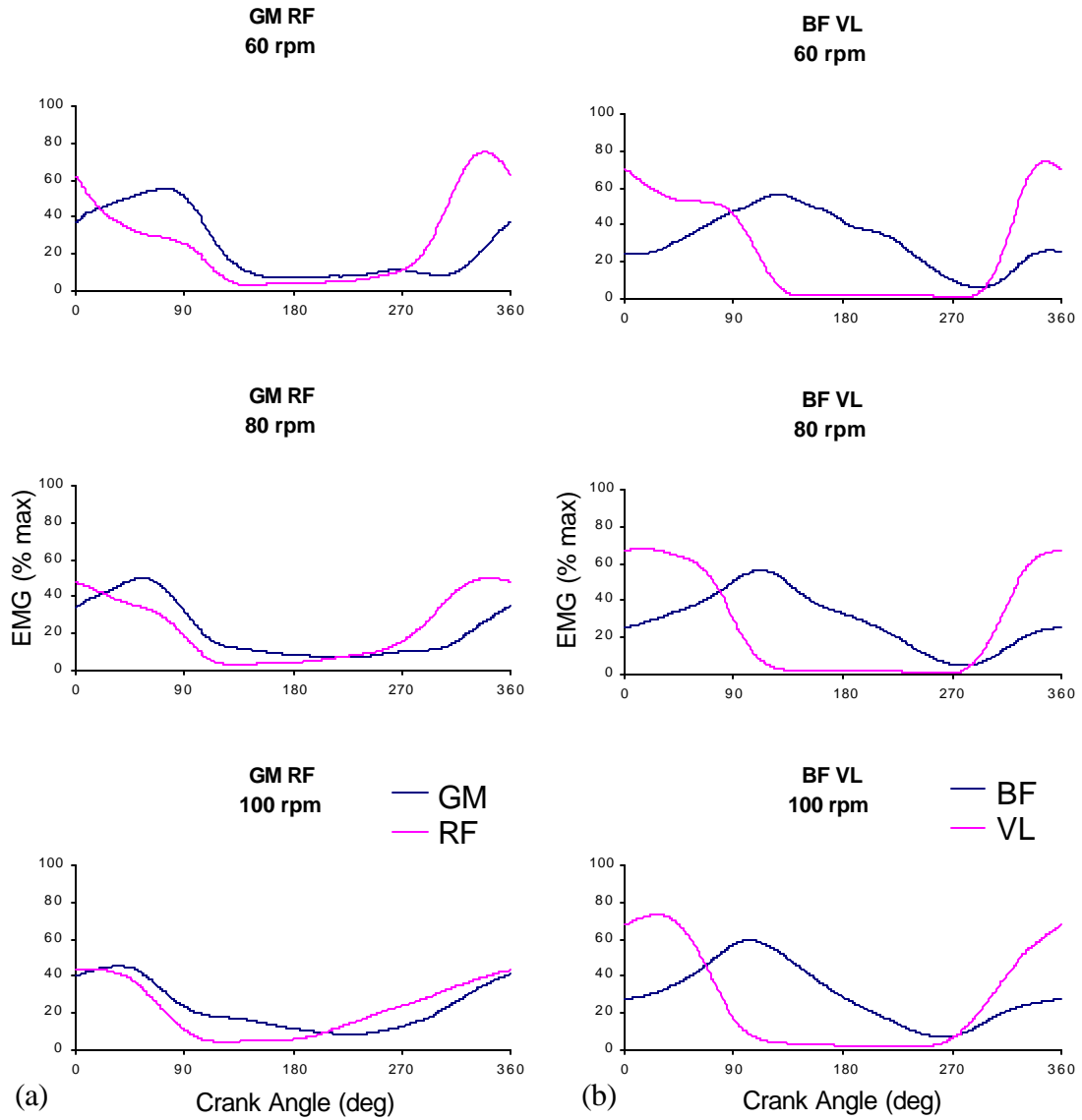


Figure 12. Mean EMG linear envelopes of selected mono- and bi-articular pairs across cadence conditions. The pairs include (a) gluteus maximus/rectus femoris (GM RF), (b) biceps femoris/vastus lateralis (BF VL), (c) vastus lateralis/gastrocnemus (VL GAS), and (d) tibialis anterior/gastrocnemus (TA GAS).

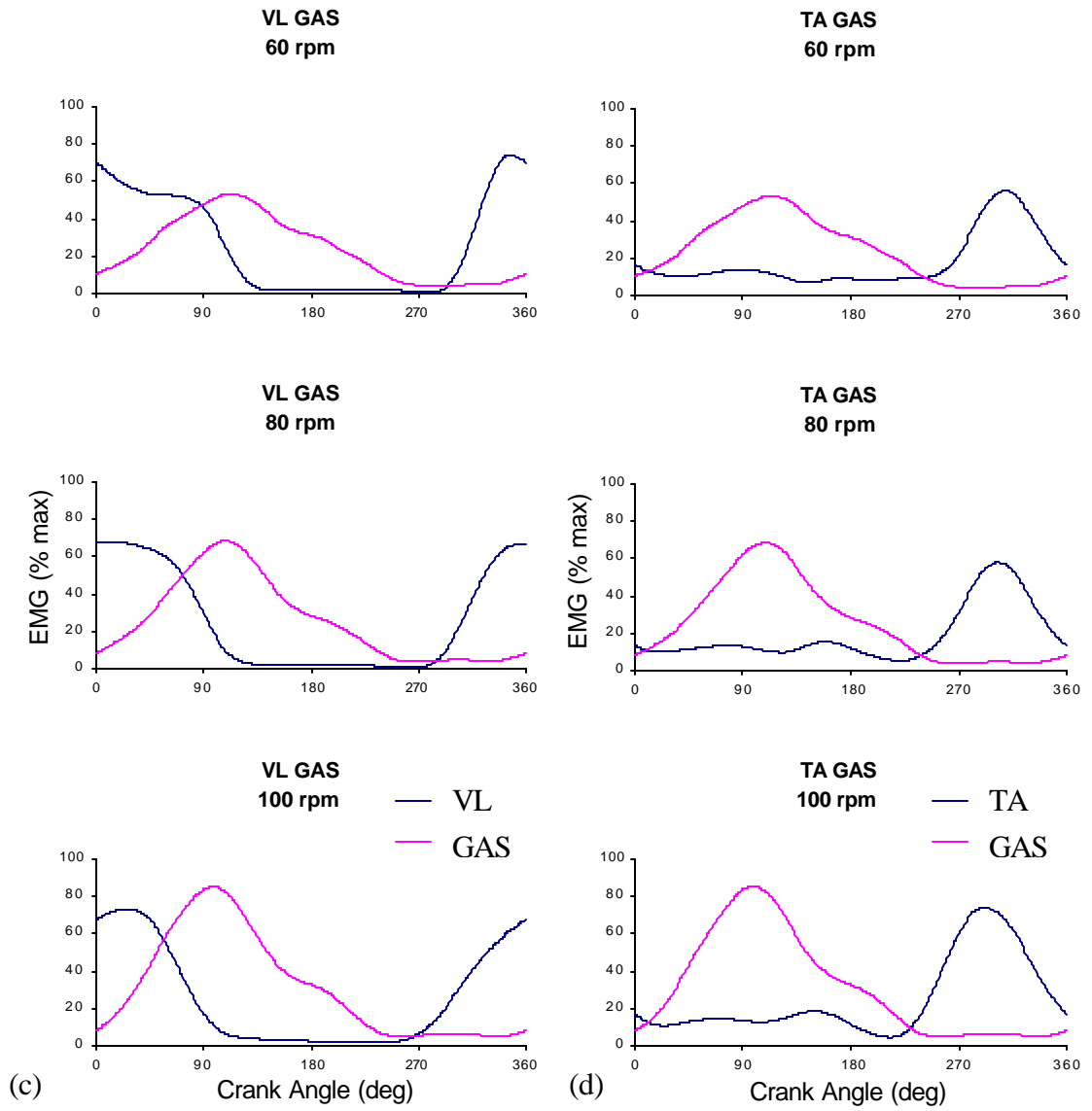


Figure 12 (cont'd).

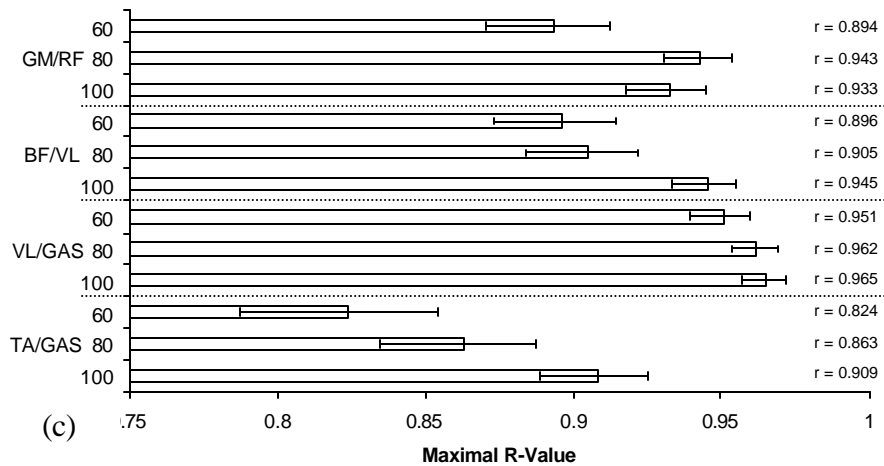
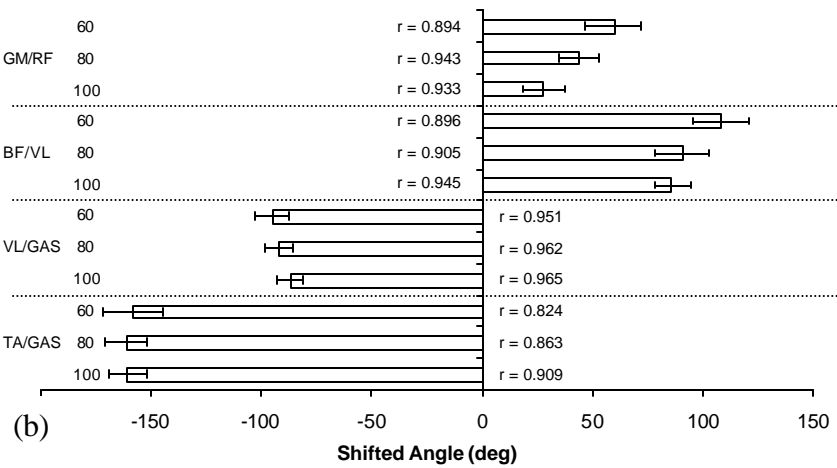
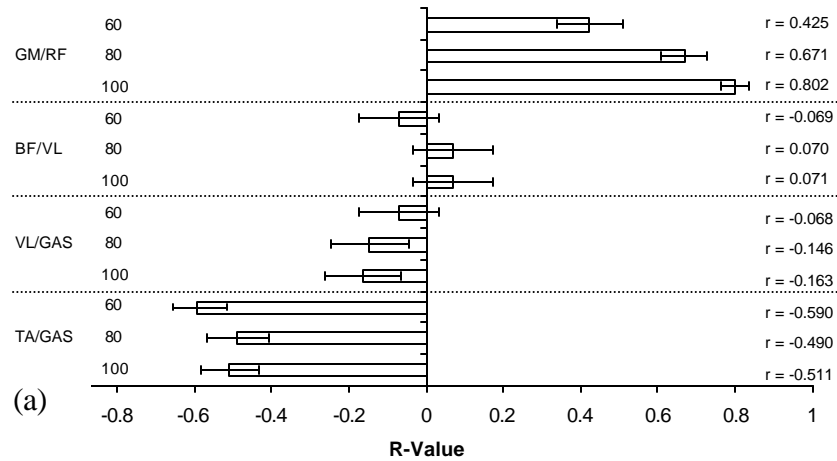


Figure 13. Cross-correlation and shifting values of the gluteus maximus/rectus femoris (GM RF), biceps femoris/vastus lateralis (BF VL), vastus lateralis/gastrocnemus (VL GAS), and tibialis anterior/gastrocnemus (TA GAS) muscle pairs across cadence conditions. (a) shows the $r_{\text{pair}}(0)$ value (correlation with no shifting), (b) displays the shifted k value to reach the highest correlation, or $r_{\text{pair}}(k)$ value, seen in (c).

values indicate a lack of coupling (i.e. little to no co-contraction). The $r_{\text{pair}}(0)$ values for GM/RF, 0.425, 0.671, 0.802 across cadence conditions, show different degrees of coupling of the muscle activity. The co-contraction increased with cadence with the highest amount of co-contraction occurring in the 100 rpm condition. GM/RF shifting of 60, 44, and 28 degrees of the crank angle for the 60, 80, and 100 rpm conditions, respectively, partially accounted for the variance between the two muscles; however, cadence affected both the phase shift (Figure 13b) and the EMG activity pattern (Figure 13c). BF/VL variance was high with no shifting ($r[0] = -0.069, 0.070, \text{ and } 0.071$; Figure 13a), and the phase differences accounted for most variation. Cadence had a very small, but significant effect on the phase difference between BF/VL 60 ($k = 109$) and 100 ($k = 86$). Furthermore, cadence affected the activity patterns of the muscles ($r[k] = 0.896, 0.905$ for BF/VL 60 and 80, respectively) with the most similar activity patterns (after shifting) observed at 100 rpm ($r[k] = 0.945$). Low $r_{\text{pair}}(0)$ values for VL/GAS 60, 80, and 100, (-0.068, -0.146, and -0.163, respectively) were accounted for with phase difference ($k = -95, -92, -87$ degrees, respectively; Figure 13a-b). Since the pair responded to cadence conditions similarly, no cadence effect is seen on either shifting or on the activity patterns of the muscles. Similarly, cadence did not affect TA/GAS phase difference; however, the increase in maximal r-values from 0.824 to 0.863 to 0.909 with cadence indicates a cadence effect on the activity patterns of the TA/GAS muscle pair.

In summary, cadence affected the GM/RF and BF/VL phase relation, although the BF/VL cadence effect was to a lesser extent. Cadence had an effect on the patterns of each muscle pair except for VL/GAS.

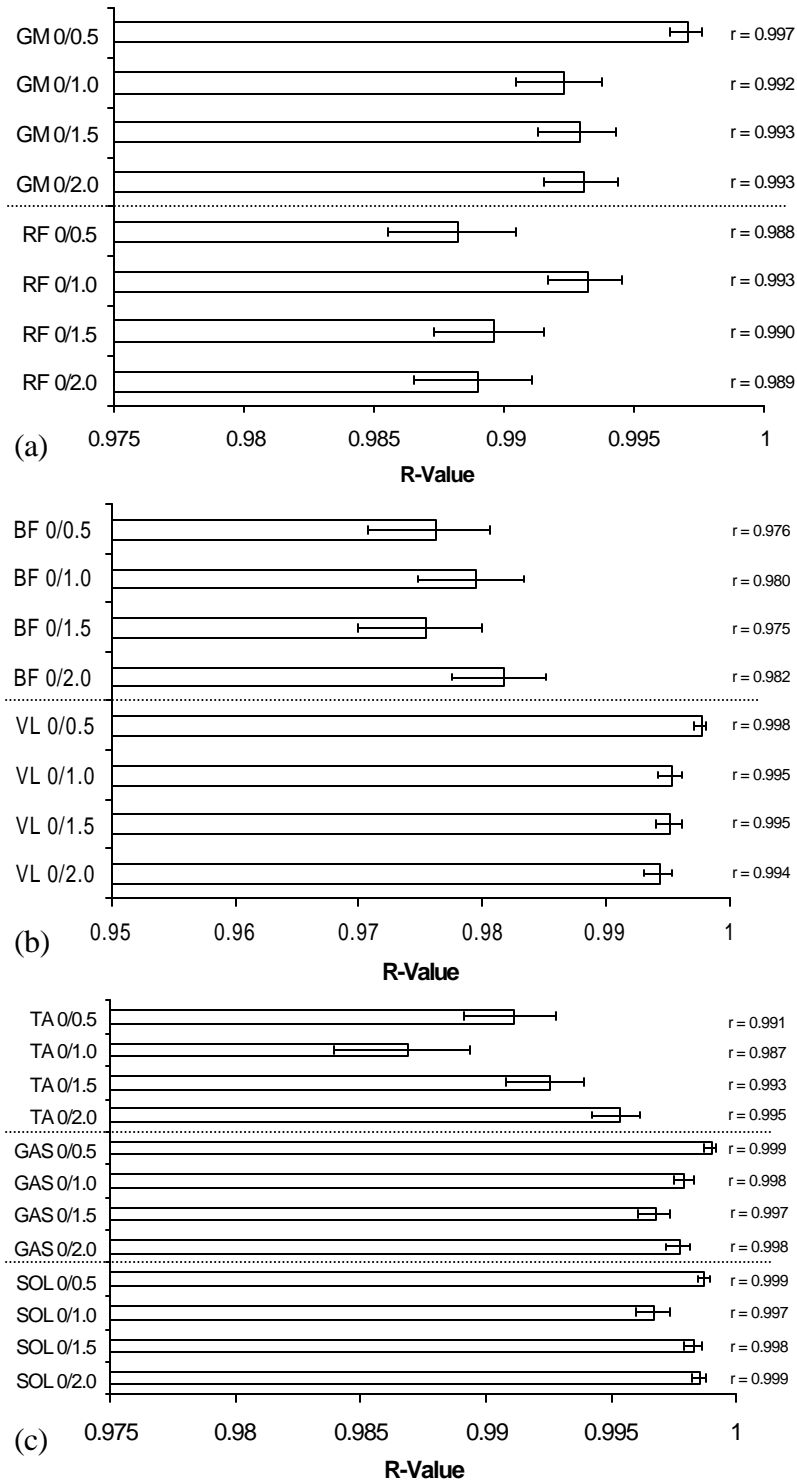


Figure 14. Cross-correlation and shifting values of gluteus maximus (GM), rectus femoris (RF), biceps femoris (BF), vastus lateralis (VL), tibialis anterior (TA), gastrocnemius (GAS), and soleus (SOL) comparing load conditions. Graphs (a)-(c) show the $r_{\text{task}}(0)$ value (correlation with no shifting) for the muscles, (d)-(f) display the shifted k value to reach the highest correlation, or $r_{\text{task}}(k)$ value, seen in (g)-(i).

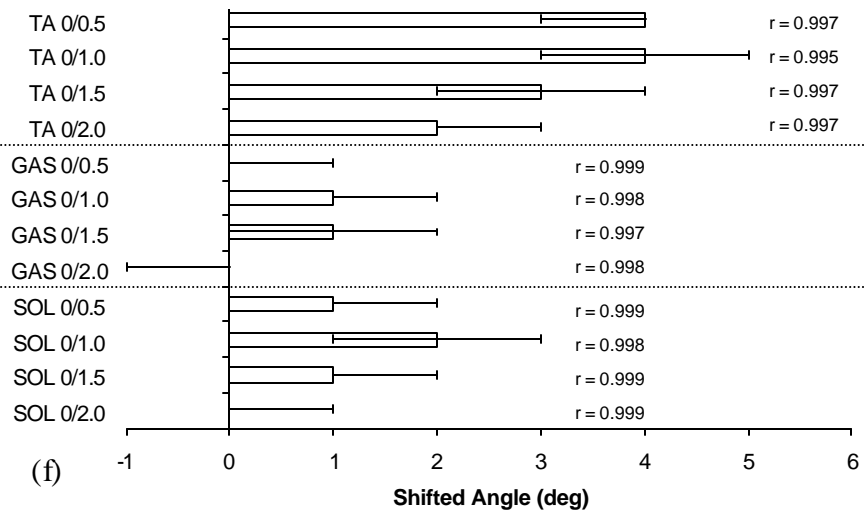
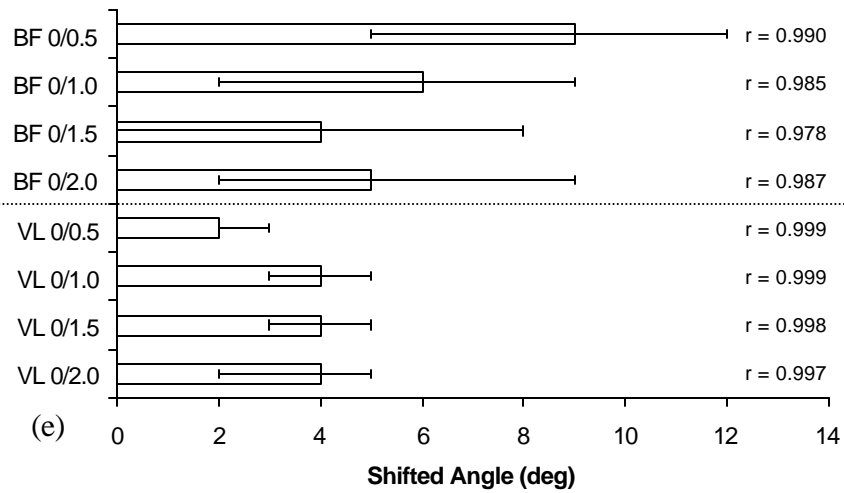
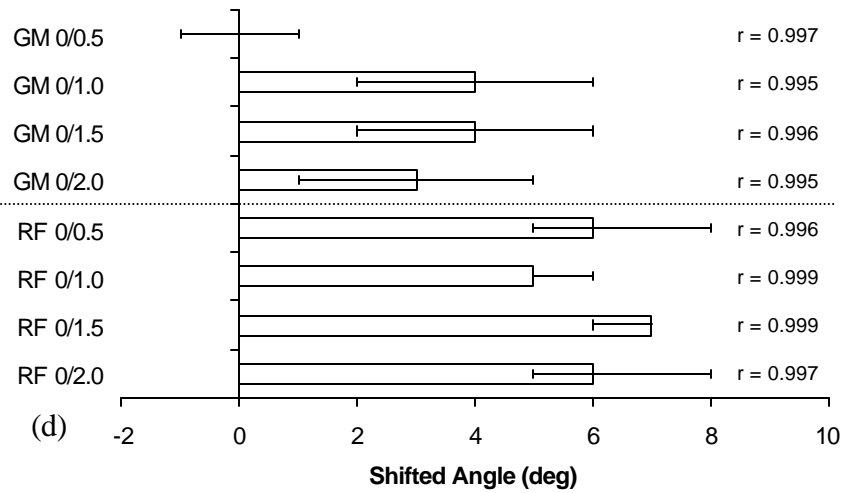


Figure 14 (cont'd).

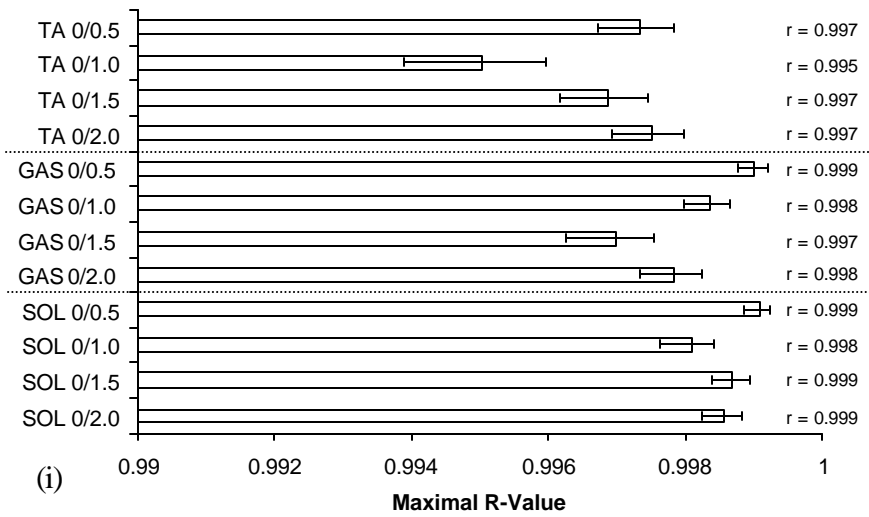
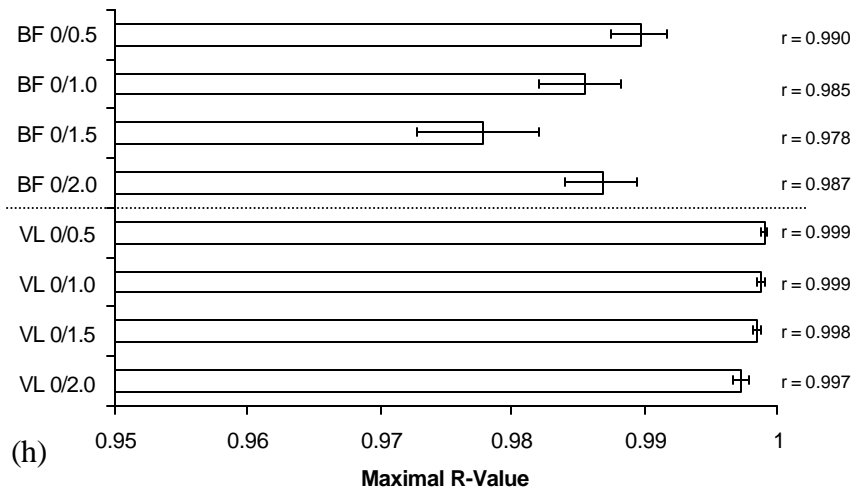
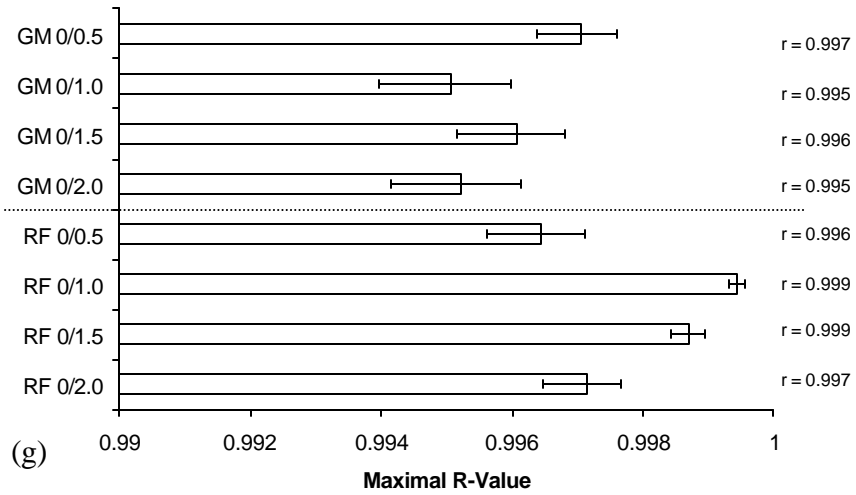


Figure 14 (cont'd).

Individual Muscle Activity Patterns Across Loads

Figures 14a-i display the comparisons of each lower extremity muscle tested across load conditions. Extremely high $r_{\text{task}}(0)$ values, between 0.975 and 0.999 (Figures 14a-c), indicate low variance between the load conditions for each muscle. Phase difference (Figures 14d-f) accounted for most of what little variance existed, and overlapping of the 95% CI error bars among most conditions for each muscle indicates that loads did not affect the phase relations. This result was also observed in Figures 14g-i, which display the maximal r-values achieved across load conditions.

Antagonist Pair Coordination Across Loads

Figures 15a-d display the EMG linear envelope patterns of the selected antagonistic muscle pairs across the load conditions. Visual inspection reveals little to no alterations in the activity patterns and coupling of the muscles across loads. Figures 16a-f present the coordination comparisons for the mono- and bi-articular muscle pairs tested. The lower $r_{\text{pair}}(0)$ values increased considerably with phase shifting (Figures 16c-d), and similar shifting patterns across loads signify no load effects on the shifts for any muscle pair. Additionally, the overlapping 95% CI error bars in Figures 16e-f indicate that adding loads to the distal end of the thigh did not affect the coordination patterns of the mono- and bi-articular antagonist muscle pairs in the lower extremity.

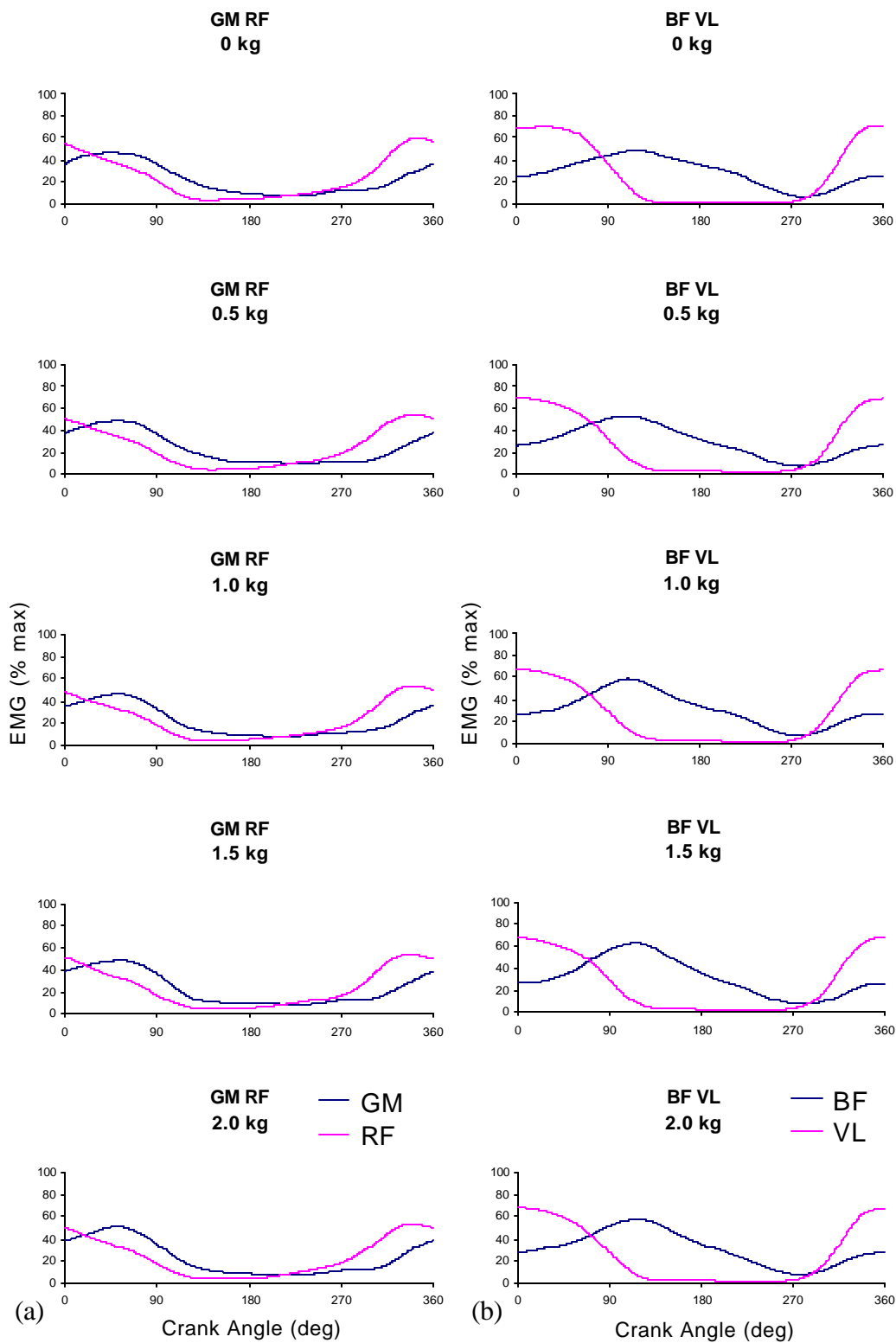


Figure 15. EMG linear envelopes of selected mono- and bi-articular muscle pairs across load conditions. The pairs include (a) gluteus maximus/rectus femoris (GM RF), (b) biceps femoris/vastus lateralis (BF VL), (c) vastus lateralis/gastrocnemus (VL GAS), and (d) tibialis anterior/gastrocnemus (TA GAS).

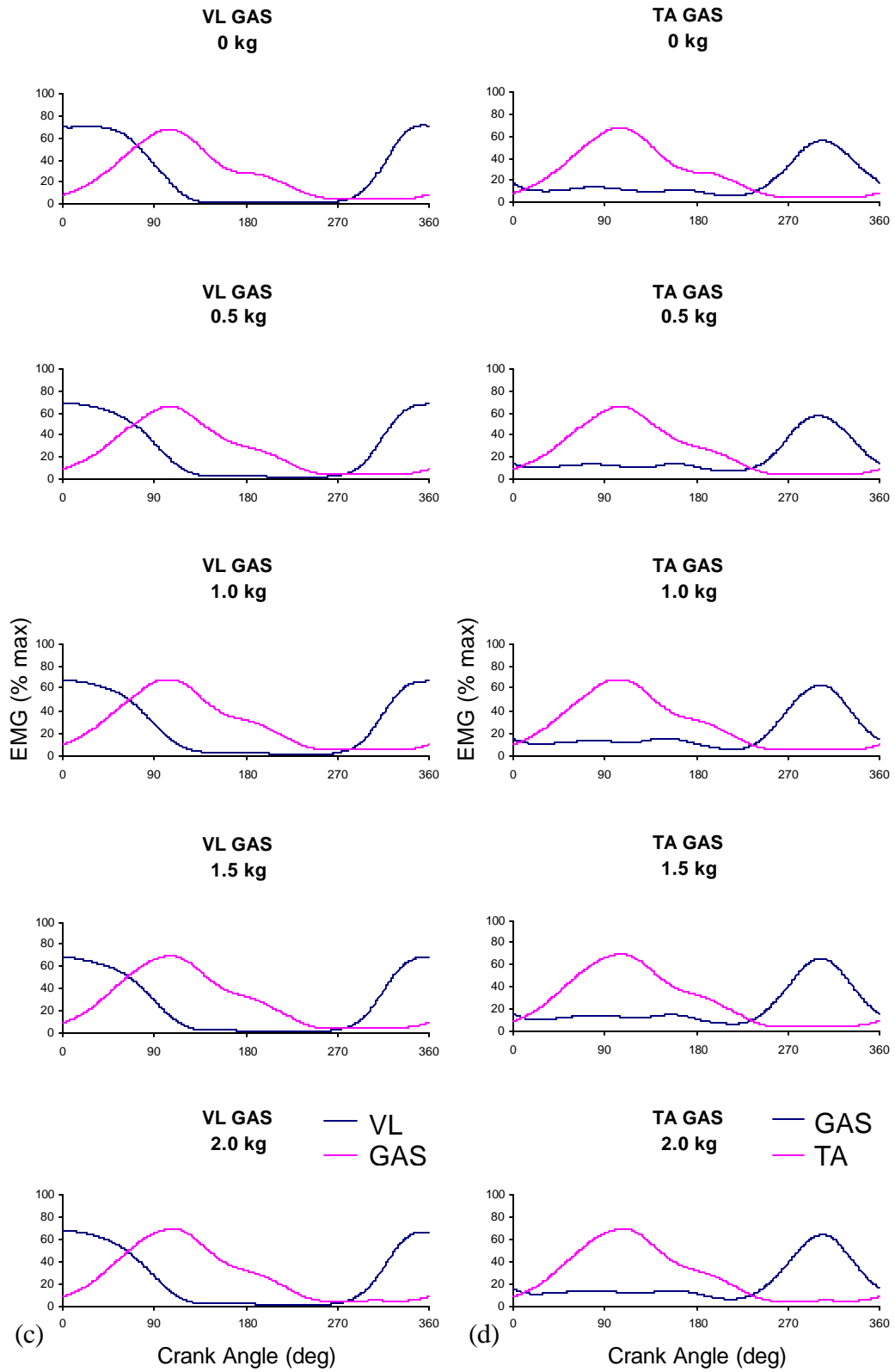


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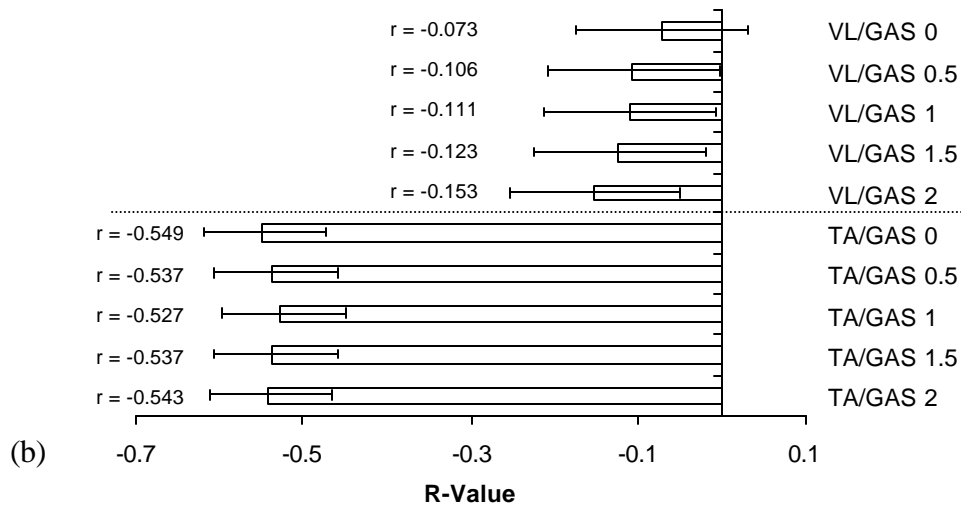
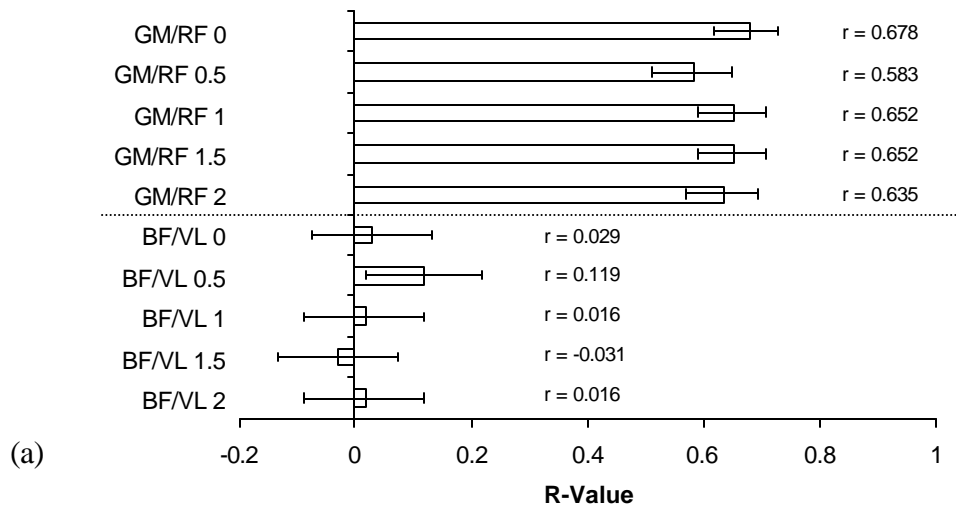


Figure 16. Cross-correlation and shifting values of the gluteus maximus/rectus femoris (GM RF), biceps femoris/vastus lateralis (BF VL), vastus lateralis/gastrocnemeus (VL GAS), and tibialis anterior/gastrocnemeus (TA GAS) muscle pairs across load conditions. Graphs (a)-(b) show the $r_{\text{task}}(0)$ value (correlation with no shifting) for the muscles, (c)-(d) display the shifted k value to reach the highest correlation, or $r_{\text{task}}(k)$ value, seen in (e)-(f).

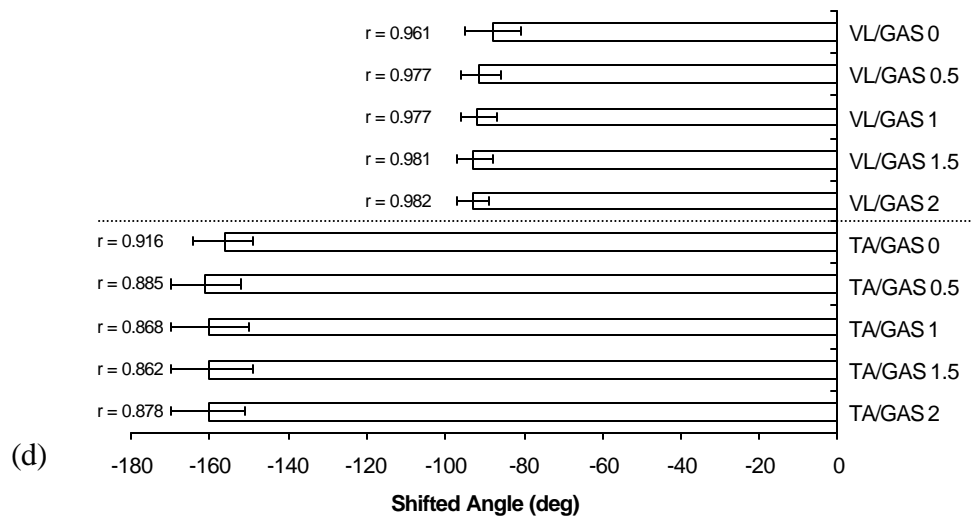
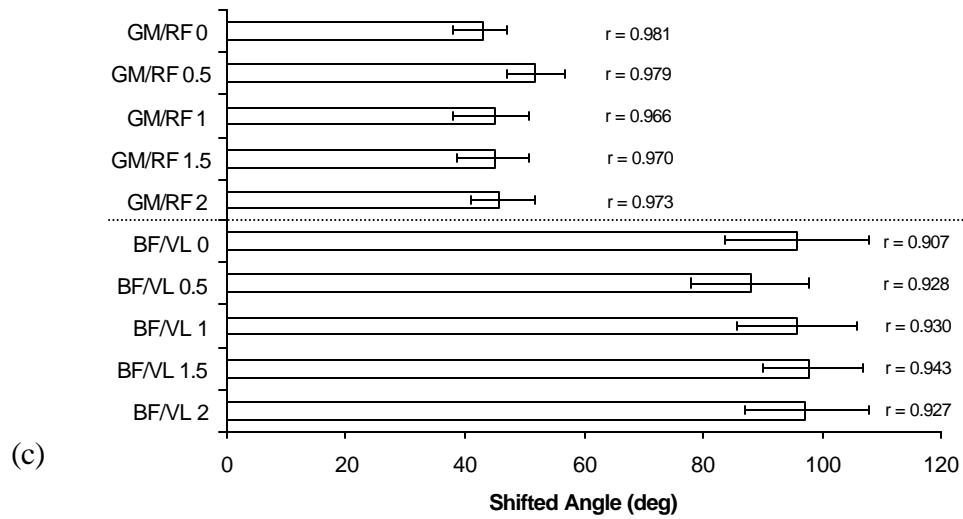


Figure 16 (cont'd).

Effects Across Joints

When comparing the differences in values between the 60 and 100 rpm cadence conditions and the 0 and 2 kg load conditions, only onset and offset values across cadences appeared to have more dramatic effects proximally than distally, and only the peak magnitude timing exhibited this trend for the loads, although the changes were very small. Using mono-articular muscles to illustrate this point, the mean onset timing

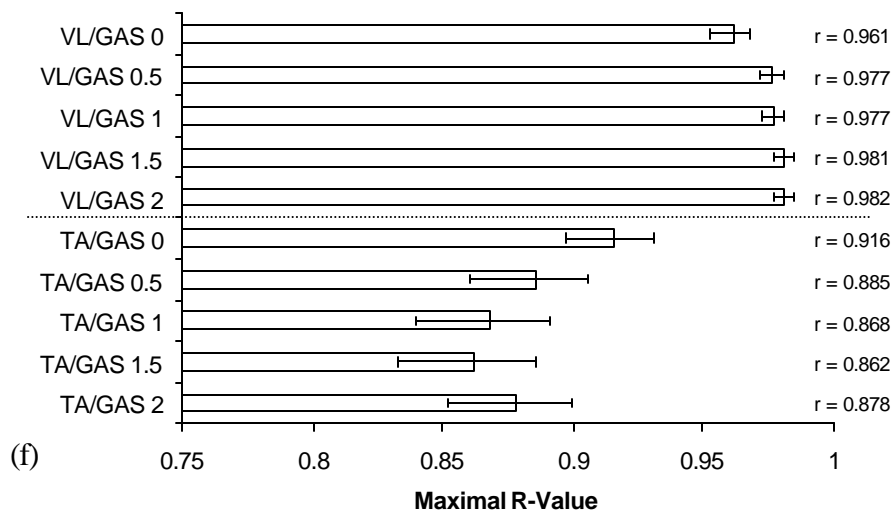
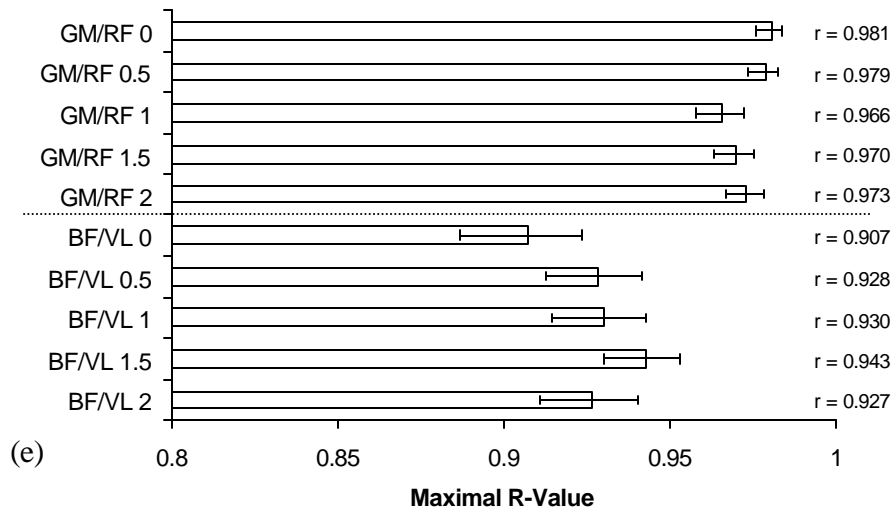


Figure 16 (cont'd).

from 60 to 100 rpm changed 29° for GM, 19° for VL, 16° for TA, and 4° for SOL.

Between 60 and 100 rpm, offset values shifted 20° for GM, 23° for VL, 9° for TA, and

5° for SOL. For peak EMG magnitude timing across load conditions, GM shifted 9°,

VL and TA shifted 3°, and SOL exhibited no change in timing between 0 and 2 kg.

These data illustrate that the muscles around the proximal joints had greater changes

than those muscles located distally. All other variables examined either produced no

pattern shifts or very similar shifts across the musculature. For the peak magnitudes

across cadences, the ankle musculature values increased with cadence, whereas the hip and knee musculature peaks decreased with increasing cadence (Table 7). Mean TA peak magnitude increased 16 percentage points of the maximum value across conditions from 60 to 100 rpm, and GAS peak EMG magnitude increased 32 percentage points with the cadence increases. At the knee and hip joints, RF peak EMG magnitudes significantly decreased with an increase in cadence, from 83 percent at 60 rpm to 52 percent at 100 rpm, a net change of 31 percentage points. Phase shifts due to cadence tended to be greater at the hip than the knee, and the ankle musculature displayed the smallest shifts. Inspecting Figures 12a-d reveals noticeable magnitudes of differences in the patterns and coupling changes in a proximal to distal hierarchy.

DISCUSSION

The goal of this study was to examine and distinguish between the neuromuscular responses to cadence and inertial manipulations during cycling, specifically as they related to the coordination of mono- and bi-articular antagonists. In this light, the magnitude, timing, and correlation values of EMG linear envelope data calculated from the activities of seven lower extremity muscles were explored.

Li (1999) proposed that muscular coordination changes at greater cadences might result from the muscles adapting to an increased inertial demand, but cadence and inertia were not separated in that study. We separated the two components by both altering cadences and adding loads to the distal ends of the thighs so that distinct conclusions about cadence and inertial influences on muscular coordination could be made. Using a similar loading method, Martin and Cavanagh (1990) reported that kinematic patterns of gait did not change after manipulating inertial properties of the lower extremity, but increases in net joint moments and reaction forces did occur. They supported these findings by suggesting that the inertial loading would increase muscular activity as seen by Lestienne (1979), who employed a simple elbow flexion-extension task to examine how loads may affect muscular activity. Little empirical evidence exists related to physically altering the inertial properties of the lower extremity during cycling, but several researchers have speculated about inertial influences on both the kinetics and muscular activities of this action. Kautz and Hull (1993) reported an increased magnitude of the non-muscular component (specifically the inertial subcomponent) of the pedal force with cadence and Li (1999) and Li and Caldwell (1993, 1997) proposed that the cadence effects observed on the neuromuscular

coordination stemmed from a systematic increase of the inertial influence with increasing pedal speeds. The results from the current experiment, however, support these ideas only partially. The inertial influence on the EMG patterns was observed in the onset and offset timing as well as the peak magnitudes but not with the cross-correlation related analysis with the increasing loads. Neuromuscular coordination and muscular activity patterns displayed differential reactions to cadence and load. Timing and magnitude changed with both greater loads and cadences, but more profound pattern and coordination changes were observed with cadence changes. The results indicate that muscles may respond differently to both cadence and load manipulations across the discrete events, so it was speculated that muscular coordination might change due to these task alterations (see Tables 11 and 12). Furthermore, examining Figure 10a reveals distinct differences in the way the ankle musculature reacted to cadence compared with the reactions of the hip and knee musculature. Results from the cross-correlation technique suggested that coordination changes occurred between mono- and bi-articular antagonists due to cadence, but not due to load manipulations. Because of the discrepancy in the results relating to coordination changes due to loads, it is proposed that inertial influences may be more relevant to event magnitudes rather than timing. The cross-correlation technique used does not consider curve magnitudes whereas discrete event comparisons depend on the onset/offset threshold magnitude.

Analyzing coordination differences due to cadence between mono- and bi-articular antagonists with the cross-correlation technique revealed greater effects at the hip than at the knee or ankle, suggesting a proximal to distal order of effects. In the GM/RF relationship, coordination changed due to cadence. The coordination changes

TABLE 11. General discrete event responses to increased cadence for mono- and bi-articular muscle pairs. Only cases when the muscles responded differently are displayed.

	GM/RF	BF/VL	VL/GAS	TA/GAS
Onset timing			VL earlier GAS \emptyset	TA earlier GAS \emptyset
Offset timing			VL earlier GAS \emptyset	
Duration		BF \downarrow VL \emptyset		TA \uparrow GAS \emptyset
Peak EMG magnitude	GM \emptyset RF \downarrow		VL \emptyset GAS \uparrow	
Peak EMG timing	GM earlier RF \emptyset	BF earlier VL later	VL later GAS earlier	

\emptyset represents no significant change

\uparrow and \downarrow represent increases or decreases in the discrete event value, respectively

Earlier and later represent timing shifts to earlier or later in the crank cycle, respectively

TABLE 12. General discrete event responses to increased loads for mono- and bi-articular muscle pairs. Only cases when the muscles responded differently are displayed.

	GM/RF	BF/VL	VL/GAS	TA/GAS
Onset timing	GM \emptyset RF earlier	BF earlier VL \emptyset		
Offset timing	GM \emptyset RF earlier	BF \emptyset VL earlier	VL earlier GAS \emptyset	
Duration		BF \emptyset VL \downarrow	VL \downarrow GAS \emptyset	
Peak EMG magnitude		BF \uparrow VL \emptyset		TA \uparrow GAS \emptyset
Peak EMG timing	GM earlier RF \emptyset			

\emptyset represents no significant change

\uparrow and \downarrow represent increases or decreases in the discrete event value, respectively

increased from 60 to 100 rpm. At the knee, the BF/VL pair exhibited a coordination change only due to phase difference, and the VL/GAS pair showed no coordination changes. At the ankle, no coordination changes were noted in the mono- and bi-articular pair TA/GAS. This illustrates a proximal to distal prominence of effects due to cadence in mono- and bi-articular antagonists. However, using the cross-correlation method, these distinct effects were not observed across the load conditions. The small

magnitude of coordination changes due to load manipulations and the lack of consistent patterns across these changes suggest that meaningful changes in muscular activity timing did not take place. However, changes in discrete event timings and magnitudes between mono- and bi-articular antagonists (Table 12) do indicate a possible proximal to distal order of load effects. Previous researchers have reported the proximal to distal significance of effects, especially in mono-articular muscles (Li, 1999; Neptune et al., 1997), and these effects were explained by the greater inertial properties proximally in the lower extremities. Consequently, the muscles acting on the knee and ankle joints reacted to a lesser and lesser extent due to the lesser inertial properties of the shank and foot, respectively. The results only partially support these suggestions because of the discrepancy between the cross-correlation analysis and the discrete event responses. Consequently, movement speed seems to have a greater influence on the lower extremity muscles than does inertia, which may only affect event magnitudes rather than timing.

The third hypothesis, that mono-articular muscle timing and bi-articular muscle functioning would change due to cadence and load manipulations, was also only partially supported by the results. Peak timing shifts due to cadence in this experiment are similar to those found in Marsh and Martin (1995). Those authors reported significant timing differences due to cadence in VL, RF, BF, GAS, and SOL, and linear trends for the peak magnitude to occur earlier in the crank cycle as cadence increased from 50 to 110 rpm for each of the muscles except RF. The only discrepancy between these results and those of Marsh and Martin (1995) is that a difference in RF peak timing was not observed. This may be attributed, however, to the lesser range of

cadences used in the current study. Still, these data support the idea that changing movement speeds will affect the muscular activity timing values. We also observed onset and offset timing changes due to cadence alterations. Neptune et al. (1997) hypothesized that EMG muscle burst onset values must shift earlier in the crank cycle as cadences increase in order to develop pedal forces in the same relative area of the crank cycle. Although their results indicated muscle burst onset timing changes, different muscles responded differently to the cadence changes thus suggesting a coordination change between muscles. They observed linear trends shifting the EMG burst onset to earlier in the crank cycle for GM, RF, BF, and vastus medialis (VAS). Their SOL shifted later in the crank cycle, GAS exhibited a quadratic trend, and TA had no trend for EMG onset due to cadence. In this study similar trends were observed for GM, RF, BF, and VL; no trend for GAS or SOL; and a linear trend for TA onset timing due to cadence. Differences between TA, GAS, and SOL trends between the studies may be attributed to the use of different onset and offset threshold criterion (Hodges & Bui, 1996; Li & Caldwell, 1999). Furthermore, Neptune et al. (1997) used a greater range of cadences, from 45 to 120 rpm, than the range used in this study. Timing changes also occurred due to load manipulations, although not to the extent of the changes due to cadence manipulations. These timing differences were observed for both mono- and bi-articular muscles. However, the hypothesis that cadence and load manipulations would alter bi-articular muscle functioning was not supported by our results. Suzuki et al. (1982) reported that as maximum pedaling velocity was approached, the bi-articular muscles BF and RF began to exhibit double bursting patterns of activation, corresponding to contributions of muscle activation at both the

hip and the knee. This double-bursting activity was not observed in the current study. Instead, the bi-articular muscles exhibited one burst, but they remained active so as to act at both joints that the muscles crossed.

The fourth hypothesis of this study was not supported. The results did not reveal a greater load influence at greater cadences. However, at even greater cadences, perhaps a more pronounced load effect would be seen. Neptune and Herzog (1999) noticed negligible amounts of negative muscular crank torque created at 90 rpm, but greater pedaling rates (105 and 120 rpm) produced substantial negative crank torque and that torque increased with the increasing pedaling rates. These authors also noticed a similar trend in average pedal forces due to the non-muscular component. Consequently, if some of the main effects in Neptune and Herzog (1999) only became prevalent at cadences over 100 rpm, especially relating to non-muscular components, perhaps similar effects would occur with inertial effects on the muscular activity.

Finally, our results suggest that joint power transfers may occur primarily in a proximal to distal order. If GM and VL are concurrently active along with RF, then power will be transferred from the hip to the knee, and conversely, BF coactivation with GM and VL indicates a power transfer from the knee to the hip (Ingen Schenau et al., 1990; Jacobs et al., 1996). Greater coupling with increased cadence for the GM/RF muscle pair indicated by greater $r(0)$ values and lesser phase differences (Figures 13a and 13b) suggest firing patterns indicative of a hip to knee joint power transfer. Conversely, the BF/VL pair coupling and firing pattern coordination was not enhanced as much with cadence indicating that a power transfer from the knee to the hip joint was less evident. This observation differs from that of Ingen Schenau et al. (1990) who

described net power transfers from the knee to the hip joint during the second half of the downstroke.

In conclusion, our results indicate that changing motion speed and altering the inertia of the thigh will affect the muscular activity and coordination in the lower extremity during cycling. The results supported some of our hypotheses and did not support others. Coordination changes were observed via both discrete event timing and magnitude changes and cross-correlation revealed pattern and phase differences due to cadence. However, load manipulations only affected the discrete events. This suggests that coordination changes due to inertia may only relate to changes in magnitude, and the effects of movement speed seem to have a greater influence on the lower extremity muscles' activity and coordination during cycling than do inertial effects. Cadence had greater effects proximally than distally for onset timing, offset timing, and antagonist pair coordination changes, while greater load effects were observed proximally for peak EMG timing and antagonist pair discrete event differences. Mono-articular discrete event timing shifts due to loads for VL and SOL were observed, but bi-articular muscles did not alter their functioning as predicted; instead the bi-articular muscles appeared to act bifunctional at all cadences and loads. The data did not support the hypothesis that loads would have an increased influence at greater cadences.

The discrete event parameters were based on an onset/offset threshold criterion value, and inherent limitations exist when using a threshold criterion for determining the onset/offset values. Using a threshold value is subjective, and results may vary depending on the threshold chosen (Hodges & Bui, 1996; Li & Caldwell, 1999). Also, it is possible that the frequency content of the EMG signal might change with increasing

cadences, and this frequency content alteration may influence the linear envelope (M. Solomonow, personal communication, October 30, 2001). Therefore, the objective cross-correlation method was also employed to detect and compare the shapes of the curves. Another limitation of this study was the small range of cadences used. The method of load creation and attachment prevented the use of cadences greater than 100 rpm.

Based on the results and limitations presented in this study, several opportunities for conducting future research have developed. The idea that inertia affects event magnitude changes but not timing changes merits further consideration. These results provide additional information to current views and experimental results in the literature that suggest significant inertial influences on the muscular activity and kinetics in running and cycling (i.e. Kautz & Hull, 1993; Li, 1999; Martin & Cavanagh, 1990). Since we did not include kinetic investigations in our experiment, we could not directly compare our results with those implying net moment and pedal force changes due to inertia. Consequently, direct comparisons between kinetics, kinematics, and EMG distinguishing movement speed and inertial effects warrant future investigation. Studies of this construct would clarify the relationship between cadence and inertia and their respective effects on the kinetics, kinematics, and muscular activities during movement. Furthermore, since this study revealed several limitations due to methodology constraints, it is recommended that the inertial influences on cycling be investigated at a wider range of added loads and cadences in order to confirm the findings herein over a range that will encompass a larger portion of practical situations.

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APPENDIX

CONSENT FORM

Motor Behavior Laboratory, Department of Kinesiology
Louisiana State University, Baton Rouge

- Title: ***Neuromuscular Coordination Affected by Inertial Loading and Cadences during Cycling***
- Performance Site: Data collection will be performed in the Motor Behavior Laboratory located in room 3 of the Gym-Armory, LSU, Baton Rouge.
- Contacts: Dr. Li Li and Brian Baum can be reached at the Department of Kinesiology (Phone: 578-2036) between 7:30 and 4:30, Monday through Friday.
- Purpose of the Study: The purpose of this study is to differentiate between the effects of cadence and inertial influences on muscle coordination in the legs.
- Subjects: Fifteen male subjects between the ages of 18 and 35 years who have at least two years of recreational cycling experience and no pathology will be recruited from the population of students at Louisiana State University, Baton Rouge.
- Exclusion: If you answer “yes” to any of the following questions, you will be excluded from the experiment. Signing this consent form implies that you have answered “no” to all questions.
1. Has your doctor ever said you have heart trouble?
 2. Do you frequently have pains in your heart and chest?
 3. Do you often feel faint or have spells of severe dizziness?
 4. Has a doctor ever said your blood pressure was too high?
 5. Has your doctor ever told you that you have a bone or joint problem, such as arthritis, that has been aggravated by exercise, or might be made worse with exercise?
 6. Is there a good physical reason not mentioned here why you should not follow an activity program even if you wanted to?

Study Procedure:	<p>Upon entering the laboratory, you will be requested to read the consent form and then discuss the experiment with the investigator. Any questions concerning the research will be answered.</p> <p>You will cycle on a stationary bicycle for a two-minute warm-up period to allow for stabilization. You will cycle at three different cadences (60, 80, and 100 rpm) and five different weight conditions (0, 0.5, 1.0, 1.5, and 2.0 kg) for a total of fifteen trials. A weight condition consists of attaching a specified weight near the knee during the cycling trial. After the warm-up period, EMG data will be collected for ten seconds per trial via electrodes placed on seven separate muscles.</p> <p>The experiment will take approximately two hours to complete all trials.</p>
Benefits:	<p>The study does not provide direct health, monetary, or mental benefits to the subjects involved. The results of the study will benefit the society, as they will lead to a better understanding of muscle coordination in the lower extremity.</p>
Risks/Discomfort:	<p>If you exert more effort than is necessary or act differently than instructed during the experiment, minor skin and/or joint irritation is possible in the lower extremity. If you act differently than instructed, you will be told to stop the activity. As with all physical activities, there is a potential for adverse effects, including a heart attack or similar problem. There is no predictable psychological and/or sociological risk to you for participating in this research.</p> <p>CPR-trained personnel will be present or on call nearby during all cycling trials.</p>
Right to Refusal:	<p>Participation in this study is voluntary and you may change your mind and withdraw from the study at any time without penalty or loss of any benefit to which you may otherwise be entitled.</p>
Privacy:	<p>The study will be confidential. The data collected will be linked to your identity via a code and this data will be kept confidential unless release is legally compelled. A</p>

hard copy with your name and code number will be stored in a locked office. Only the code numbers will be used as data on the computer.

Financial Information: There will be neither monetary compensation nor cost to you associated with participating in this research.

“The study has been discussed with me and all my questions have been answered. I may direct additional questions regarding study specifics to the investigators. If I have questions about subjects' rights or other concerns, I can contact Robert C. Mathews, Chairman, LSU Institutional Review Board, (225) 578-8692. I agree to participate in the study described above and acknowledge the researchers' obligation to provide me with a copy of this consent form if signed by me.”

Subject Signature

Date

VITA

Brian Baum was born on August 19, 1977 and resided in Glastonbury, Connecticut until age 12, when his family moved to Albuquerque, New Mexico. After a year and a half, the family moved once more to Carson City, Nevada, where Brian's parents still live. After completing high school, Brian was recruited by the Louisiana State University swimming team, where he earned All-American honors two times. He received his bachelor of science degree in kinesiology in May 1999 from Louisiana State University and earned a graduate assistantship from the same institution to pursue his master of science degree in biomechanics, which was awarded in December 2001.