

Functional Roles of the Leg Muscles When Pedaling in the Recumbent Versus the Upright Position

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An understanding of the coordination of the leg muscles in recumbent pedaling would be useful to the design of rehabilitative pedaling exercises. The objectives of this work were to (i) determine whether patterns of muscle activity while pedaling in the recumbent and upright positions are similar when the different orientation in the gravity field is considered, (ii) compare the functional roles of the leg muscles while pedaling in the recumbent position to the upright position and (iii) determine whether leg muscle onset and offset timing for recumbent and upright pedaling respond similarly to changes in pedaling rate. To fulfill these objectives, surface electromyograms were recorded from 10 muscles of 15 subjects who pedaled in both the recumbent and upright positions at 75, 90, and 105 rpm and at a constant workrate of 250 W. Patterns of muscle activation were compared over the crank cycle. Functional roles of muscles in recumbent and upright pedaling were compared using the percent of integrated activation in crank cycle regions determined previously for upright pedaling. Muscle onset and offset timing were also compared. When the crank cycle was adjusted for orientation in the gravity field, the activation patterns for the two positions were similar. Functional roles of the muscles in the two positions were similar as well. In recumbent pedaling, the uniarticular hip and knee extensors functioned primarily to produce power during the extension region of the crank cycle, whereas the biarticular muscles crossing the hip and knee functioned to propel the leg through the transition regions of the crank cycle. The adaptations of the muscles to changes in pedaling rate were also similar for the two body positions with the uniarticular power producing muscles of the hip and knee advancing their activity to earlier in the crank cycle as the pedaling rate increased. This information on the functional roles of the leg muscles provides a basis by which to form functional groups, such as power-producing muscles and transition muscles, to aid in the development of rehabilitative pedaling exercises and recumbent pedaling simulations to further our understanding of task-dependent muscle coordination. [DOI: 10.1115/1.1865192]

Introduction

Recumbent pedaling is an exercise well suited for the diseased and disabled population. Recumbent ergometers, unlike upright bicycles, have large seats with backrests to provide support for the upper body and are low to the ground, permitting easier access for wheelchair riders and individuals with mobility impairments [1]. Furthermore, recumbent pedaling has been demonstrated to be a therapeutic modality for exercise and rehabilitation for the diseased and disabled [2–6]. Better understanding the coordination of the leg muscles in recumbent pedaling could lead to innovations that will improve the efficacy of recumbent pedaling as a therapeutic modality. Coordination involves the relative timing and activity levels and the functional roles of the muscles participating in the pedaling motion.

Prior experimental and theoretical research of muscle coordination while pedaling has resulted in a theoretical framework by which to classify the functional roles of the leg muscles in upright pedaling [7,8]. This framework is based on observations that the muscles can be partitioned into phase-controlled functional groups (PCFG) based on the extent of muscle activity as indicated by the percentage of whole cycle integrated muscle activation (iACT) in

a particular functional region of the crank cycle [7,8]. This framework has been used in the development of computer models to simulate pedaling under different conditions, such as forward and backward pedaling [9,10] and pedaling while reclined at different body orientations [11].

Coordination of muscles in recumbent pedaling has received limited attention. Although the position of recumbent pedaling is different from that of normal upright pedaling, similarities in muscle coordination should exist. Lower extremity kinetics and kinematics (i.e., musculoskeletal system outputs) are similar for recumbent and upright pedaling when adjusted for the different orientation of the rider with respect to gravity [1,4]. If concomitant similarities in muscle activity patterns between recumbent and upright pedaling could be established, then this would motivate the application of PCFG analysis for upright pedaling to analyze muscular coordination in the recumbent position. Therefore, the first objective was to determine whether muscle activity patterns are similar for the two pedaling positions when the crank cycle is rotated to account for the different orientation of the rider with respect to gravity.

Although previous research has demonstrated that muscle activity levels change as a result of different body orientations while pedaling in the reclined position [12], we are aware of no prior studies that have examined the functional roles of the leg muscles in recumbent pedaling. An understanding of the functional roles of the leg muscles would be useful to the development of simulations of recumbent pedaling that could be used to design rehabilitative

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Table 1 Demographic and United States Cycling Federation (USCF) category level for the study subjects. Two of the subjects were classified as competitive endurance cyclists (CEC). The mean age, height, and weight (± 1 standard deviation) were 29.9 (± 12.0), 1.81 (± 0.06) m, and 71.8 (± 5.2) kg, respectively.

Subject	Gender	Age (yrs)	Height (m)	Weight (kg)	Level
1	M	24	1.73	68.9	Cat 5
2	M	18	1.91	68.0	Cat 3
3	M	27	1.79	76.2	Cat 4
4	M	28	1.88	79.4	Cat 3
5	M	60	1.83	73.5	CEC
6	M	23	1.91	79.4	Cat 5
7	M	27	1.88	70.3	Cat 4
8	M	54	1.77	72.6	CEC
9	M	24	1.78	67.1	Cat 4
10	M	22	1.75	70.3	Cat 4
11	M	23	1.83	66.7	Cat 4
12	M	21	1.73	70.3	Cat 4
13	M	37	1.79	64.4	Cat 4
14	M	30	1.78	68.0	Cat 5
15	M	30	1.83	81.6	Cat 4

pedaling exercises. Assuming that the muscle activity patterns are similar for the two pedaling positions when adjusted for the different orientation of the rider with respect to gravity, it could be concluded that the musculoskeletal system inputs to perform the pedaling task are the same for the recumbent and upright positions. If the musculoskeletal system inputs and outputs are similar for the two pedaling positions, then the internal activity of the musculoskeletal system (i.e., muscle functional roles and mechanical energetics) should be similar as well, thereby making it possible to draw upon the results of a PCFG analysis and the literature on upright pedaling to determine the functional roles of the leg muscles in recumbent pedaling. Thus the second objective was to perform a PCFG analysis to compare the functional roles of the leg muscles while pedaling at a set pedaling rate and work rate in the recumbent and upright positions.

Previous research has shown that leg muscle onset and offset timing in upright pedaling are affected by pedaling rate [8,13,14]. Additionally, the functional roles of the leg muscles in upright pedaling did not change over a large range of pedaling rates [8]. If the onset and offset timing of the leg muscles in both the recumbent and upright positions respond similarly to changes in pedaling rate, then the functional roles of the leg muscles in recumbent pedaling would be applicable to more pedaling rates than the one investigated. The third objective was to determine whether leg muscle onset and offset timing for recumbent and upright pedaling respond similarly to changes in pedaling rate.

Methods

Experiments. Written informed consent was obtained from 15 cyclists who volunteered for the study. The age of the subjects ranged from 18 to 60 years (mean 30 years), the heights ranged from 1.73 to 1.91 m (mean 1.81 m), and the weights ranged from 64 to 82 kg (mean 72 kg) (Table 1). The experimental protocol was approved by the Institutional Review Board of the University of California at Davis.

Kinematic, kinetic, and electromyographic (EMG) data were collected from the subjects as they pedaled a recumbent ergometer (Scifit ISO1000R, Tulsa, OK) and a conventional racing bicycle mounted on an electronically braked Velodyne ergometer (Frontline Technology, Inc., Irvine, CA). Both ergometers allowed a constant work rate to be set independent of pedaling rate. The subjects adjusted the bicycle to match their own bicycle's geometry. The subjects all used zero-float clipless pedals and chose their own cleat angle.

The subjects pedaled at 90 rpm with a work rate of 120 W for 15 min to warm up, and thereby account for temperature dependencies of muscle function [15]. The subjects then pedaled at 75,

90, and 105 rpm with a work rate of 250 W on both the recumbent and upright ergometers. The pedaling rate was regulated by a metronome. The subjects pedaled at each pedaling rate for 5 min. Data were collected ten times for 3 s randomly selected intervals during the last 2.5 min of the 5 min test period. The ergometer order and pedaling rates were assigned randomly for each subject to control for possible interactions and fatigue.

Ergometer crank angle data were determined using high-resolution video-based motion analysis (Motion Analysis Corp., Santa Rosa, CA). Spherical reflective markers were placed at either end of a 30 cm long bar attached in line with the top surface of the pedal and at three fixed points on the ergometers. A virtual marker indicating the point connecting the pedal spindle to the crank was developed at the midpoint of the two markers in the pedal-fixed frame. The three markers attached to the ergometers were necessary to establish an ergometer-fixed coordinate system to track a virtual marker located at the point where the axis of the crank spindle intersected a plane coincident with the outer surface of the crank arm. The position of this virtual marker was calculated using a static calibration trial where the center of a spherical reflective marker was attached at the point. Crank angles were determined from the two virtual markers. The right crankarm in the upward vertical position (i.e., top-dead-center) defined the beginning of the crank cycle (0 deg of a 360 deg cycle) for both the recumbent and upright positions. Four high-speed video cameras recorded the three-dimensional marker positions. Video data were sampled at 120 Hz and filtered using a zero phase shift Butterworth low-pass filter with a cutoff frequency at 10 Hz, the maximum frequency of normal human movement [16]. The crank data were interpolated by cubic spline (MATLAB, The Math Works, Natick, MA) to make it synchronous with the EMG data (see below).

To examine muscle activity, surface EMG electrodes were placed over the belly of the soleus (SOL), medial gastrocnemius (MGAS), lateral gastrocnemius (LGAS), tibialis anterior (TA), vastus medialis (VASM), vastus lateralis (VASL), rectus femoris (RF), biceps femoris (BF), medial hamstring (SM), and the gluteus maximus (GMAX) of the right leg. The preamplified surface electrodes (Model MA-300-10, Motion Lab Systems, Baton Rouge, LA) were fit with custom-made silver-silver chloride electrode cups (In Vivo Metric, Healdsburg, CA) and placed according to the recommendations of Delagi et al. [17]. The electrode cups were filled with electrode cream, and the electrode was attached to the shaved, abraded skin surface with adhesive washers. Following placement, an adhesive elastic wrap was wrapped around the leg to secure the electrode attachment.

The EMG outputs were collected and synchronized with the video data by the motion capture system. EMG signals were sampled at 1200 Hz to ensure the ratio of sampling frequency to signal frequency was greater than five for a mean amplitude error less than 5% [18]. The EMG gains were set to yield the maximum resolution of the digitized signal without saturation. To reduce low-frequency motion artifacts and high-frequency noise, the EMG data were passed through a bandpass analog filter with low-pass cutoff of 500 Hz and high-pass cutoff of 40 Hz (manufacturer's recommendation). The 12-bit A/D board contained in the motion analysis system digitized the analog inputs. At the end of the pedaling trials, resting baseline EMG values were collected for 10 s while the subject rested in a supine position [8]. The mean values of the rectified baseline data were used to subtract baseline offset in the EMG records.

EMG Processing. The raw sampled EMG data $e(t)$ were full-wave rectified, filtered using a zero phase lag fourth-order Butterworth digital filter with 12 Hz cutoff frequency [8,16,19], demeaned, and normalized to the highest value measured for the respective muscle while pedaling the respective ergometer. The normalized EMG data $u(t)$ were then subjected to two separate processes to obtain (i) the burst onset and offset crank angles, and (ii) the muscle activation, $a^m(t)$.

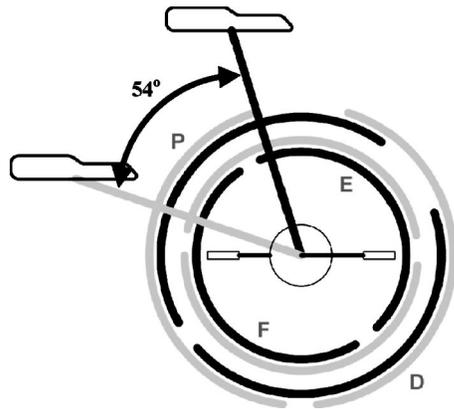


Fig. 1 The phase-controlled functional regions for the recumbent position (gray) and the upright position (black). The four primary regions for the upright position are based on the work of Raasch et al. [7] and include the extension (E), distal transition (D), flexion (F), and proximal transition (P) regions. The corresponding regions for the recumbent position have been rotated by 54 deg in the counterclockwise direction to account for the different seat angle with respect to the upright position.

The EMG burst onset and offset crank angles were determined with reference to the resting baseline data. A custom-automated wave-form-processing program written in the MATLAB language was used to identify the burst onset and offset angles. The criteria for the burst onset and offset angles were a minimum threshold of three standard deviations of the resting baseline data within a 50 ms moving rectangular window and a minimum 50 ms burst duration [7,8,19]. The results for each cycle were examined graphically, and the threshold was adjusted when necessary to identify the burst onset and offset angles [7,8].

To determine muscle activation $a^m(t)$, the normalized EMG data, $u(t)$, were input into a first-order differential equation modeling the activation dynamics [20]. The muscle activation dynamics were represented by the following first-order equation:

$$\dot{a}^m = \begin{cases} [u - a^m] \cdot [c_1 u + [c_2 \dots c_2]^T] & u \geq a^m \\ [u - a^m] \cdot c_2 & u < a^m \end{cases} \quad (1)$$

where $c_1 = \tau_{act}^{-1} - \tau_{deact}^{-1}$ and $c_2 = \tau_{deact}^{-1}$ [7,21]. The muscle activation and deactivation time constants were 20 and 60 ms, respectively [7,8,22].

Because the second objective of this study was to perform a PCFG analysis to compare the functional roles of the leg muscles, a common definition of the functional regions was used. The functional regions for upright pedaling defined by Raasch et al. [7] and subsequently modified by Neptune et al. [8] were used. However, the functional regions for recumbent pedaling were rotated by 54 deg (counterclockwise from the right side of the ergometer) from those of upright pedaling to account for the different orientation of the rider with respect to gravity in recumbent pedaling (Fig. 1). The change in the angle formed by the line from the hip joint to the crank spindle with the vertical was used to determine the extent by which the regions defined for upright pedaling were rotated. The four primary regions associated with the recumbent position were labeled as the extension (E) region (283–80 deg), distal transition (D) region (18–174 deg), flexion (F) region (95–270 deg), and proximal transition (P) region (187–337 deg). The unions between adjacent regions in the crank cycle were defined as E-D, D-F, F-P, and P-E [8].

The PCFG analyses commenced by computing the integrated muscle activation (iACT) for the whole crank cycle based on the following equation:

$$iACT = \int_0^T a^m dt \quad (2)$$

where T is the time of the crank cycle. Next, the percentage of iACT within each of the four primary regions (i.e., extension (E), distal (D), flexion (F), and proximal (P)) was calculated for each muscle. Subsequently, the percentage of iACT for each of the four union regions (i.e., E-D, D-F, F-P, and P-E) was computed. All of the dependent variables were computed on a cycle-by-cycle basis and averaged across cycles for each subject. Data were analyzed for one whole crank cycle during the 3 s intervals. Thus the number of cycles included in the subject averages was 10.

Statistical Analysis. To determine whether muscle activation patterns in the recumbent position were similar to those of upright pedaling at 90 rpm, the averaged whole cycle activation data for each muscle were plotted as a function of crank angle with the angle adjusted to account for the difference in the gravity field between the two body positions. Also the percentage of iACT in each of the functional regions of the crank cycle was determined and compared statistically. A preliminary two-factor repeated-measures analysis of variance (ANOVA) test was performed for each of the ten muscles using SAS (Release 8.02, SAS Institute Inc. Cary, NC). The two factors were crank cycle region at eight levels (four primary and four union) and body position at two levels (recumbent and upright). The dependent variable was the percent of whole cycle iACT. Because the results of these preliminary analyses indicated that significant ($p < 0.05$) and important interactions were present, eight one-factor repeated-measures ANOVA tests were performed separately for each of the regions for each muscle, yielding a total of 80 analyses (8 regions \times 10 muscles). The single factor was body position at two levels (recumbent and upright). To detect significant body position effects, the level of significance was set at $p < 0.05/n$ where $n = 8$ for the Bonferroni correction.

To compare the functional roles of the muscles in the recumbent and upright positions, the percentage of iACT within each of the four primary regions (i.e., extension (E), distal (D), flexion (F), and proximal (P)) was used to determine the principal function of the leg muscles while pedaling in both positions. The muscles were determined to have a single function (extensor, distal transition, flexor, or proximal transition) based on the region with the greatest percentage iACT. The muscle function was redefined as bifunctional if the percentage of iACT in the union regions exceeded that of the primary regions by 20% [8].

To examine whether leg muscle onset and offset timing for recumbent and upright positions responded similarly to changes in pedaling rate, preliminary statistical analyses consisting of two-factor repeated-measures ANOVA tests were performed. The two factors were pedaling rate at three levels (75, 90, and 105 rpm) and body position at two levels (recumbent and upright). The dependent variables were onset angle and offset angle. For each statistical analysis, the dependent variable for the recumbent position was referenced to a crank cycle that was rotated by 54 deg (counterclockwise from the right side of the ergometer) to account for the different orientation of the rider in the recumbent position relative to the upright position. The analysis was performed separately for each muscle and dependent variable yielding a total of 20 analyses (10 muscles \times 2 dependent variables). Because the results of these analyses indicated that significant ($p < 0.05$) and important interactions were present, three separate one-factor repeated-measures ANOVA tests were performed for each of the three pedaling rates for each of the muscles yielding a total of 60 analyses (3 pedaling rates \times 10 muscles \times 2 dependent variables). In each analysis the single factor was body position at two levels (recumbent and upright). To detect significant body position effects, the level of significance was set at $p < 0.05/n$ where $n = 3$ for the Bonferroni correction.

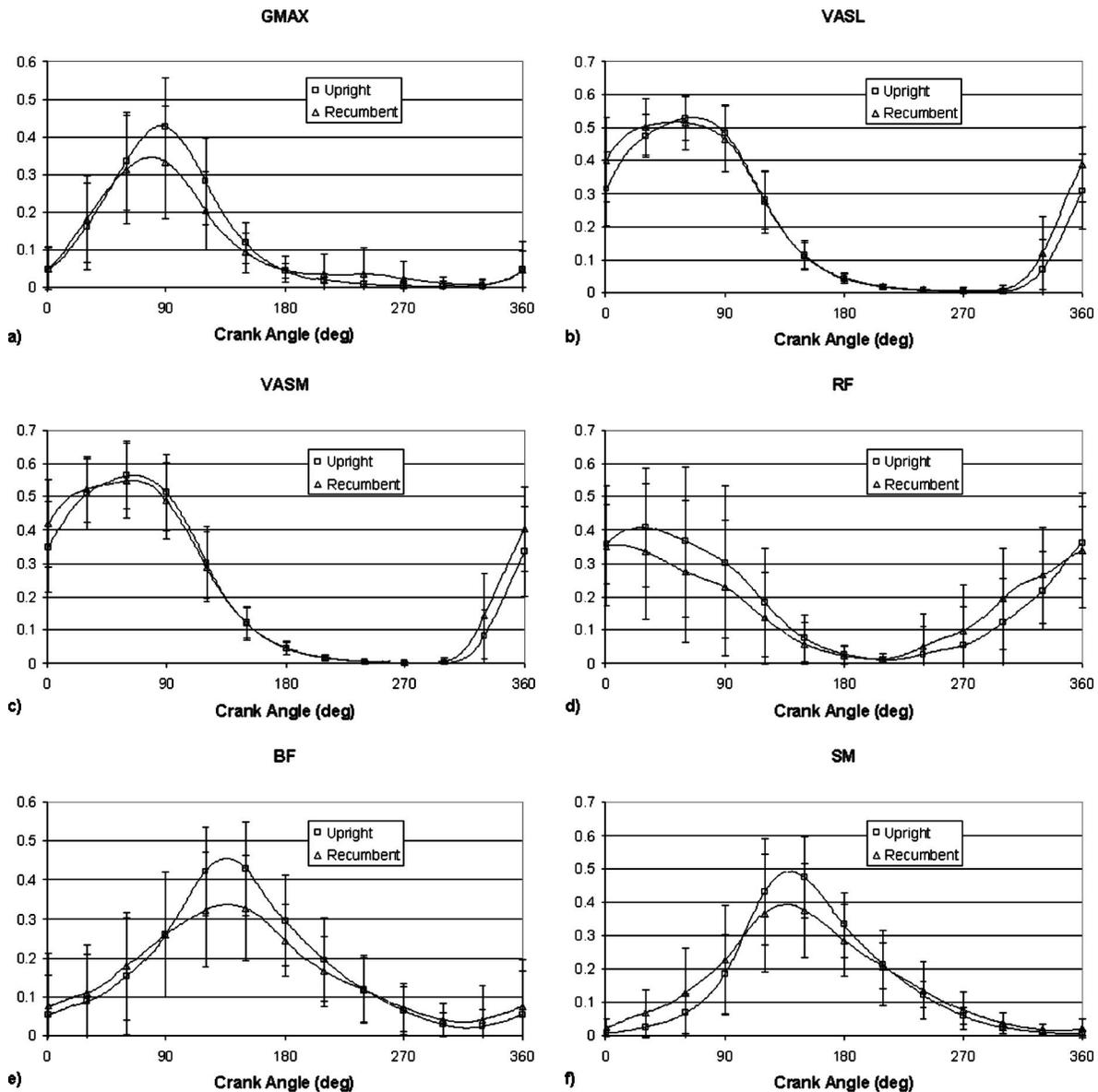


Fig. 2 (a)–(j). Muscle activation patterns for the ten muscles of the right leg while pedaling in both the recumbent and upright positions at a constant pedaling rate (90 rpm) and work rate (250 W). All patterns are the average of the 15 subjects. Error bars indicate \pm one standard deviation.

Results

The muscle activation patterns while pedaling at 90 rpm were qualitatively similar for the two body positions (Fig. 2). Furthermore, the analyses of iACT indicated that body position did not significantly affect the magnitude of muscle activity within the adjusted functional regions of the crank cycle for most of the muscle activity-functional region combinations studied. For six of the ten muscles, the body position did not significantly affect muscle activity in any of the functional regions (Table 2). For the remaining four muscles, body position effects were significant ($p < 0.05/8$ or $p < 0.0063$ for the Bonferroni correction) in only two of the eight functional regions for both VASM and RF (extension-distal (E-D) and flexion-proximal (F-P) regions), one region for SM (D region), and five regions for LGAS (E-D, E, F, F-P, and P regions). Thus in 70 of the 80 (87.5%) of the muscle activity-functional region combinations examined, the body position did not significantly affect the muscle activity.

During recumbent pedaling, six of the ten muscles were classified as unfunctional (i.e., having a single function) and four as

bifunctional. The uniaxial hip and knee extensors, GMAX, VASL, and VASM, were activated primarily during the extension region and were classified as unfunctional E muscles (Fig. 3). The RF, a biarticular hip flexor and knee extensor, was bifunctional in the proximal-extension (P-E) region. BF and SM, biarticular hip extensors and knee flexors, were activated primarily in the distal (D) region. SM was classified as an unfunctional D muscle, but BF was classified as bifunctional in the E-D region due to the higher activity level in the E region (Fig. 3).

The muscles of the triceps surae group, LGAS, MGAS, and SOL, were also activated primarily in the distal (D) region and, with the exception of LGAS, were classified as unfunctional D muscles. LGAS was bifunctional in the distal-flexion (D-F) region with over 90% of the activation occurring in these respective regions. The TA, an uniaxial ankle dorsiflexor, was classified as bifunctional in the F-P region, with 89% of its activation occurring in this region. None of the muscles were classified as unfunctional in either the P or F regions.

The functional roles of the leg muscles for the upright position

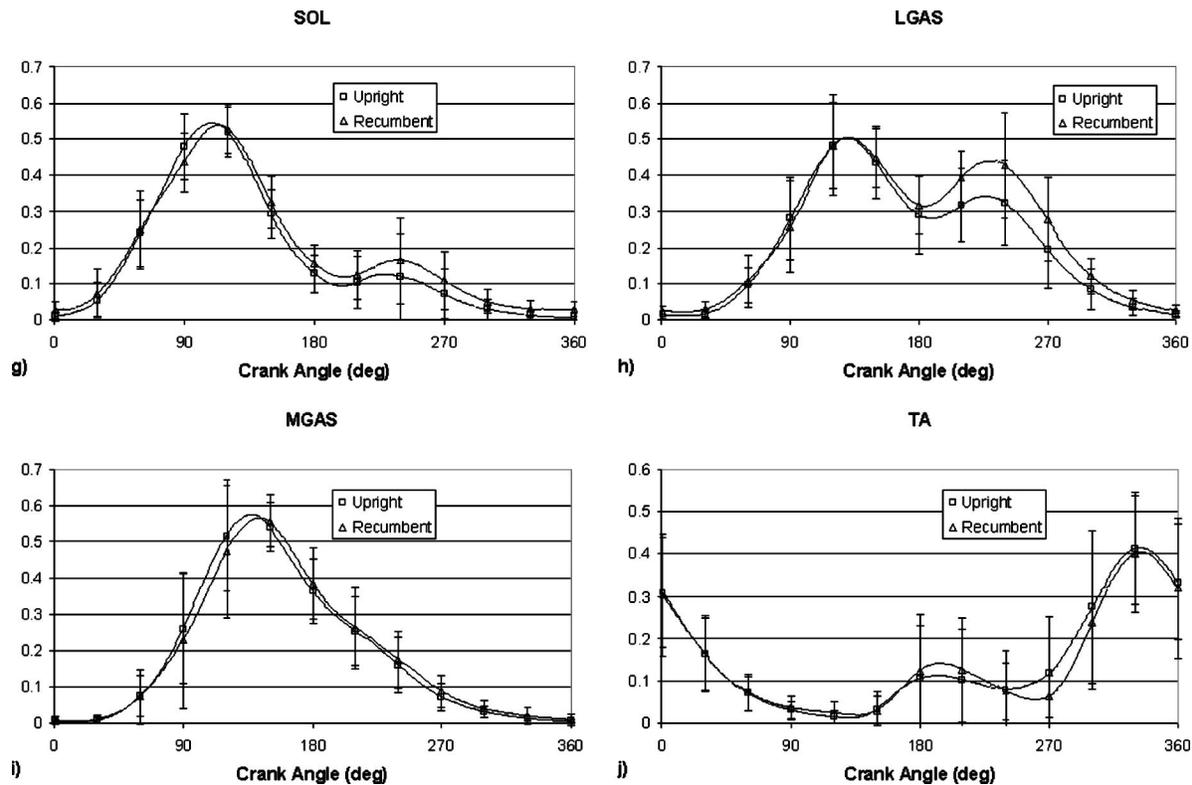


Fig. 2 (Continued).

as defined by the PCFG analysis were the same as those for the recumbent position for eight of the ten muscles examined. BF and TA were classified as unifunctional in the distal (D) and proximal (P) regions, respectively, in the upright position and bifunctional in the extension-distal (E-D) and flexion-proximal (F-P) regions, respectively, in the recumbent position. Notwithstanding the differences in classification for these two muscles, there were no significant differences ($p < 0.05/8$ or $p < 0.0063$ for the Bonferroni correction) between the percentage of integrated muscle activation (iACT) within each of the four primary and four union regions of the crank cycle (Table 2) in the recumbent versus upright position.

The muscle burst onset and offset analyses indicated that leg muscle timing for recumbent and upright pedaling responded similarly to changes in body position at all pedaling rates for all muscles with a few exceptions. For the muscle onset, body position did not significantly affect muscle onset for seven of the ten muscles examined (Table 3). For the remaining three muscles, the body position effect was significant ($p < 0.05/3$ or $p < 0.0167$) for only one pedaling rate for each of these muscles (i.e., MGAS at 105 rpm, $p = 0.0024$; TA at 75 rpm, $p = 0.0116$; and VASM at 90 rpm, $p = 0.0006$). Thus in 27 of the 30 (90%) muscle-pedaling rate combinations analyzed, the body position did not significantly

affect the onset timing. For the muscle offset, body position also did not significantly affect muscle offset for seven of the ten muscles examined (Table 4). For the remaining muscles, the body position effect was significant for only one pedaling rate for each muscle (BF at 75 rpm, $p = 0.0026$; GMAX at 105 rpm, $p = 0.0021$; and LGAS at 90 rpm, $p = 0.0134$). Thus in 27 of the 30 (90%) muscle-pedaling rate combinations analyzed, the body position did not significantly affect the offset timing.

Discussion

Because recumbent pedaling is an exercise modality well suited for the diseased and disabled population and because muscle coordination in recumbent pedaling has not been investigated previously, the objectives of this study were (i) to determine whether muscle activation patterns for recumbent pedaling are similar to those in upright pedaling, (ii) to compare the functional roles of the leg muscles while pedaling in the recumbent position to the upright position, and (iii) to determine whether muscle burst onset and offset timing in recumbent and upright pedaling respond similarly to changes in pedaling rate. One key finding of this study was that the activation patterns of the leg muscles were similar for both the recumbent and upright pedaling positions. A second key

Table 2 Summary of the p -values from the statistical analyses of the effect of body position on the functional roles of the muscles in the different regions of the crank cycle. Significance for the analyses was determined as $\alpha = 0.05/8$ or $\alpha = 0.0063$.

Region	BF	GMAX	LGAS	MGAS	RF	SM	SOL	TA	VASL	VASM
E	0.1633	0.2523	0.0150	0.0522	0.0072	0.5210	0.1077	0.9149	0.3609	0.2663
E-D	0.1444	0.1874	0.0004	0.1485	0.0043	0.1793	0.1034	0.3161	0.0144	0.0053
D	0.0093	0.0427	0.0004	0.1733	0.0125	0.0007	0.1130	0.1147	0.3054	0.0542
D-F	0.0351	0.0679	0.2960	0.9898	0.1250	0.0098	0.6039	0.7748	0.4023	0.1174
F	0.3537	0.1626	0.0049	0.0475	0.0073	0.8157	0.1055	0.8895	0.5026	0.4862
F-P	0.7140	0.1210	0.0028	0.0449	0.0041	0.8808	0.0497	0.9856	0.0460	0.0052
P	0.0077	0.1178	0.0009	0.1184	0.0066	0.0178	0.0580	0.1382	0.0735	0.0087
P-E	0.0395	0.6140	0.9012	0.1140	0.3581	0.2541	0.4617	0.3577	0.5723	0.2365

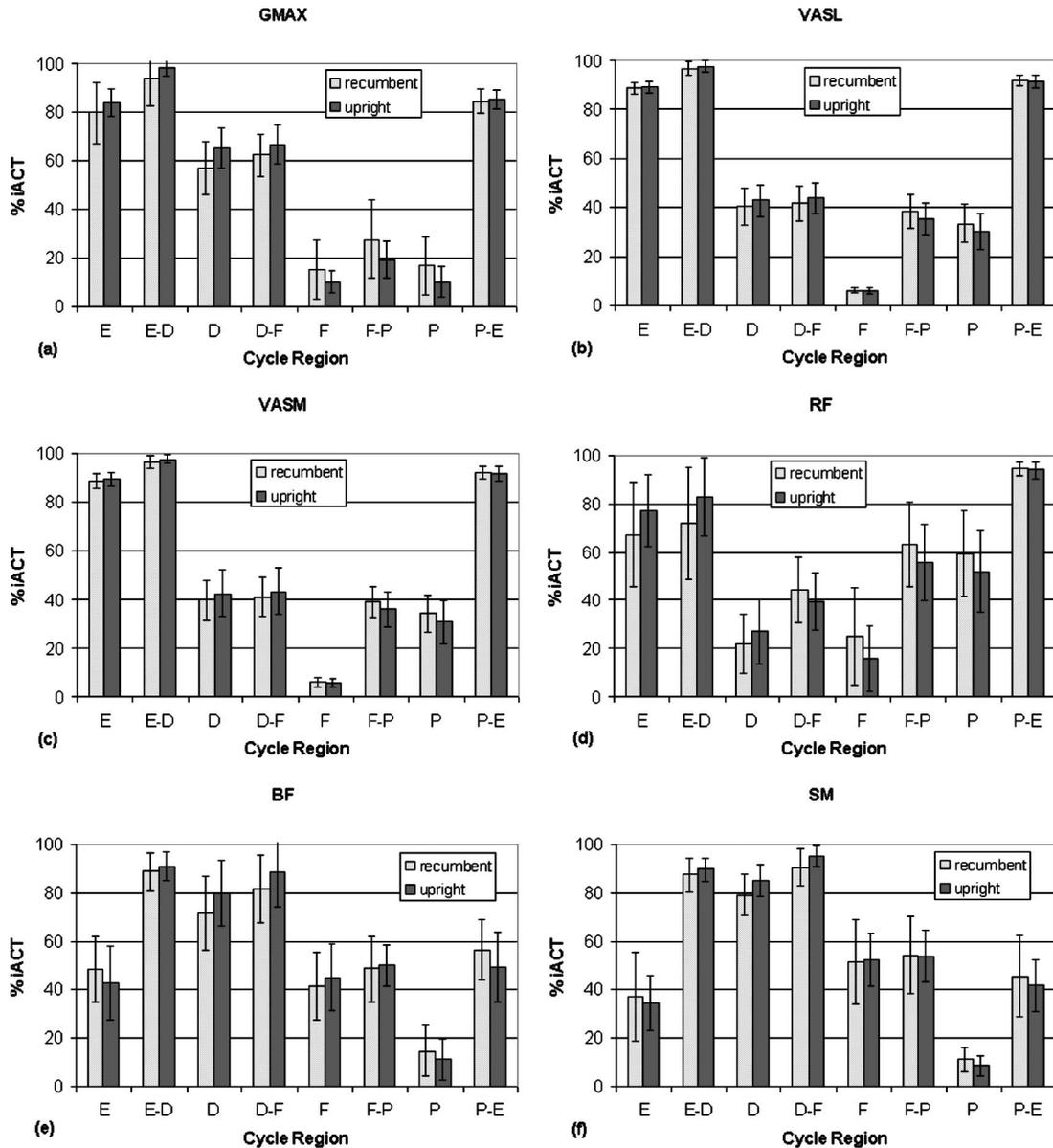


Fig. 3 (a)–(j). The percentages of whole cycle integrated muscle activation (iACT) within each of the four primary phase-controlled functional regions and the union regions between adjacent primary regions for the ten muscles of the right leg while pedaling in both the recumbent and upright positions at a constant pedaling rate (90 rpm) and work rate (250 W). All values are the average of the 15 subjects. Error bars indicate \pm one standard deviation.

finding was that the leg muscles function similarly in the recumbent and upright positions. A final key finding was that the burst onset and offset timing responded similarly to changes in pedaling rate independent of body position for all but a few muscles. Before discussing the importance of these findings, several methodological issues should be reviewed to assess their potential to influence the results of the study.

Methodological Issues. To compare muscle activity patterns between recumbent and upright pedaling, the crank cycle regions defined for upright pedaling were adjusted by 54 deg (counterclockwise from the right side of the ergometer) for the recumbent position. This adjustment served to redefine the beginning of the crank cycle to account for the different orientation of the rider with respect to gravity in recumbent pedaling. A pilot study conducted to assess this adjustment indicated that the shift in peak crank torque angles between the recumbent and upright positions were similar to the difference in the two seat angles. In addition,

both Gregor et al. [1] and Perell et al. [4] demonstrated that lower extremity intersegmental joint moment patterns are similar for recumbent and upright pedaling when differences in the seat position with respect to gravity are taken into consideration. Thus the adjustment to the crank cycle and functional regions was based on experimental evidence, indicating that the musculoskeletal system outputs were similar for the two body positions when the different orientation in the gravity field was considered.

Different ergometers were used to test subjects in the recumbent and upright positions. Although it would have been desirable to use the same ergometer for both body positions, it was not possible to find a single ergometer that could accommodate both the recumbent and upright positions required by this study. Because differences in work rate and/or the seat distance from the crank spindle between the two ergometers could have affected the EMG records, the raw EMG data were normalized with respect to

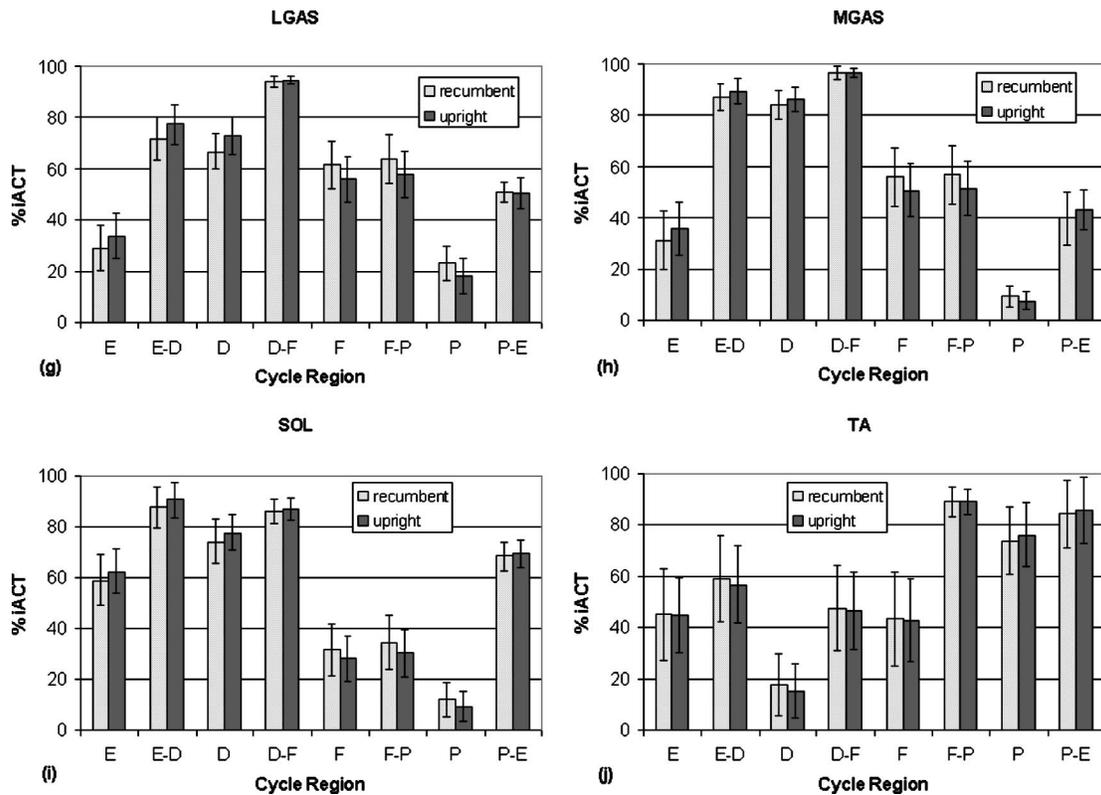


Fig. 3 (Continued).

each ergometer. However, this normalization did not affect the results of any of the statistical analyses because the percentage of iACT was the dependent variable of interest.

Instead of standardizing rider position, the seat position and cleat angle were set to match those of each subject's own bicycle. The subjects in this study were all competitive cyclists who were highly accustomed to their personal riding position; therefore, standardizing rider position would likely have introduced more variability into results. Additionally, it has been argued that anatomy and bicycle fit are rider specific and, hence, should not be standardized [23,24].

Fundamental to the accuracy of identifying the burst onset/offset were the quality of the EMG records and the criteria used in the automated wave-form processing program. Two sources of variability associated with the use of surface EMG are electrode

placement and signal degradation. To minimize the variability associated with electrode placement, the same individual placed all electrodes for all subjects. However, differences in the electrode placement between subjects could be a source of some of the variability in the EMG records because different regions of the muscle are responsible for different functional activity [25,26], are comprised of different fiber types [27], and are susceptible to different degrees of cross talk from other muscles [28].

Because the limitations of a systematic method to reliably identify burst onset/offset timing due to the variability of EMG records across muscles and subjects have been addressed previously [8], the specifics as they pertain to this study will only be summarized here. In this study, the parameters of the automated wave-form processing program were set to provide an accurate determination of the EMG onset and offset timing. Subsequent

Table 3 The average crank angle in degrees (± 1 standard deviation) of muscle burst onset for the two body positions and the three different pedaling rates. Significance for the analyses was determined as $\alpha=0.05/3$ or $\alpha=0.0167$.

	75 rpm		90 rpm		105 rpm	
	recumbent	upright	recumbent	upright	recumbent	upright
BF	316 \pm 49	14 \pm 37	310 \pm 47	11 \pm 37	308 \pm 47	10 \pm 41
GMAX	300 \pm 18	354 \pm 14	299 \pm 23	353 \pm 18	293 \pm 18	346 \pm 19
LGAS	331 \pm 15	21 \pm 12	337 \pm 16	28 \pm 9	341 \pm 13	28 \pm 8
MGAS	346 \pm 32	32 \pm 20	350 \pm 26	37 \pm 16	350\pm17^a	35\pm12^a
RF	213 \pm 27	278 \pm 25	207 \pm 33	271 \pm 33	194 \pm 31	257 \pm 39
SM	344 \pm 42	41 \pm 28	340 \pm 41	39 \pm 30	340 \pm 40	36 \pm 28
SOL	313 \pm 16	5 \pm 13	321 \pm 19	16 \pm 13	329 \pm 13	19 \pm 10
TA	207\pm37^a	225\pm59^a	208 \pm 39	243 \pm 52	200 \pm 34	243 \pm 41
VASL	260 \pm 11	314 \pm 14	256 \pm 12	314 \pm 13	248 \pm 14	308 \pm 14
VASM	259 \pm 12	315 \pm 9	255\pm12^a	313\pm11^a	249 \pm 14	307 \pm 15

^aSignificant body position effects ($p < 0.0167$)

Table 4 The average crank angle in degrees (± 1 standard deviation) of muscle burst offset for the two body positions and the three different pedaling rates. Significance for the analyses was determined as $\alpha=0.05/3$ or $\alpha=0.0167$.

	75 rpm		90 rpm		105 rpm	
	recumbent	upright	recumbent	upright	recumbent	upright
BF	151\pm50^a	228\pm48^a	156 \pm 51	215 \pm 45	150 \pm 48	211 \pm 31
GMAX	73 \pm 33	129 \pm 10	60 \pm 12	122 \pm 13	54\pm11^a	120\pm9^a
LGAS	218 \pm 15	267 \pm 18	216\pm13^a	262\pm13^a	211 \pm 15	259 \pm 15
MGAS	175 \pm 31	240 \pm 21	181 \pm 31	232 \pm 27	182 \pm 26	234 \pm 22
RF	39 \pm 47	112 \pm 29	32 \pm 50	103 \pm 30	19 \pm 56	100 \pm 29
SM	174 \pm 39	244 \pm 29	169 \pm 39	222 \pm 26	163 \pm 28	215 \pm 24
SOL	124 \pm 51	168 \pm 43	128 \pm 55	163 \pm 41	131 \pm 55	187 \pm 52
TA	313 \pm 16	5 \pm 18	313 \pm 20	14 \pm 18	312 \pm 29	23 \pm 15
VASL	68 \pm 13	128 \pm 10	63 \pm 11	120 \pm 10	56 \pm 12	120 \pm 25
VASM	71 \pm 17	132 \pm 19	64 \pm 14	118 \pm 13	57 \pm 14	113 \pm 13

^aSignificant body position effects ($p < 0.0167$)

interactive visual inspection on a trial-by-trial basis was performed to verify the onset and offset determination and, when the burst duration did not fall within three standard deviations of the resting baseline, to systematically adjust the threshold level parameter to identify the primary temporal burst characteristics of the EMG record.

Importance/Interpretation of Results. The results of this study demonstrated that the musculoskeletal system inputs (i.e., muscle activation patterns) were similar between recumbent and upright pedaling when adjusted for the different orientation in the gravity field. Moreover previous research has demonstrated that limb kinematics and kinetics in pedaling are similar for different body orientations [1,4]. Because both the musculoskeletal system inputs and outputs are similar for the two pedaling positions, the internal activity of the musculoskeletal system (i.e., muscle functional roles and mechanical energetics) should be similar as well. The classification of muscle function based on the PCFG analysis, which used crank cycle regions previously defined for upright pedaling [7,8] but adjusted for the different orientation in the gravity field, confirmed the similarity in muscle functional roles. Accordingly, a forward dynamic simulation of recumbent pedaling to determine appropriate regions and define muscle functions based on those regions was unnecessary.

The unifunctional role in the extension (E) region of the uniaxial muscles that extend the hip and knee (i.e., GMAX, VASL, VASM) indicates that the principal function of these muscles was to develop power during the E region of the crank cycle [7,8]. The role of the GMAX and vastii muscles as power producers is fitting because their large physiological cross-sectional areas and fiber-type distributions make them capable of generating large extensor moments at the hip and knee, respectively. Raasch et al. [7] reported that together GMAX and vastii provided 55% of the net mechanical energy produced by the muscles over a crank cycle in upright pedaling.

The bifunctional role of the biarticular hip flexor and knee extensor muscle, rectus femoris (RF), in the proximal-extension (P-E) region indicates that this muscle generates both a flexor moment at the hip and an extensor moment at the knee, enabling RF to assist in propelling the leg through the proximal transition region [7]. In the E region, where GMAX activity generates a dominant hip extensor moment, RF activity contributes to the knee joint extensor moment also generated by the activity of the vastii muscles.

The biarticular hip extensor and knee flexor muscles, medial hamstring (SM) and biceps femoris (BF), were classified as uniaxial in the distal (D) region and as bifunctional in the extension-distal (E-D) region, respectively. Notwithstanding this difference, the principal function of both the SM and BF is to drive the crank through the distal region of the crank cycle. Both

the SM and BF serve to transfer energy from the leg to generate a pedal reaction force to propel the crank through the distal transition [7].

Two (MGAS and SOL) of the three muscles comprising the triceps surae group were classified as uniaxial in the distal (D) region and thereby functioned to transfer power from the leg to the crank through the distal transition region. Only LGAS was classified as bifunctional in the distal-flexion (D-F) region, which indicates that LGAS may contribute to knee flexion in addition to transferring power from the leg to the crank through the distal transition region.

The uniaxial ankle dorsiflexor, tibialis anterior (TA), was activated primarily in the proximal (P) region and was bifunctional in the flexion-proximal (F-P) region, indicating that the leg flexors actively flex the leg in the F and P regions [7]. However, because power is absorbed by the leg in the F region and a portion of the P region [29,30], the flexor activity is not enough to overcome gravity.

The PCFG analysis results indicated that the functional roles of the leg muscles for both recumbent and upright pedaling were the same for all muscles except BF and TA. In upright pedaling BF was uniaxial in the distal (D) region [7,8] but was bifunctional in the extension-distal (E-D) region in the recumbent position. This change occurred because of a slight shift in muscle activity from the D region to regions earlier in the crank cycle (i.e., in the proximal (P) and E regions) compared to upright pedaling (Fig. 3(e)). The onset and offset angles at 90 rpm, which were 7 and 5 deg earlier in the crank cycle, respectively, for the recumbent position, reflect the advance of BF activity in the recumbent position. Nevertheless, the differences in the percentage of BF activity between the two body positions in the D and P regions were not statistically significant, thereby weakening the meaningfulness of the different functional classification.

Although the TA also was classified differently for the two body positions, similar to the BF the difference was not meaningful. TA was classified as bifunctional in the flexion-proximal (F-P) region for the recumbent position whereas for the upright position the TA was classified as uniaxial in the P region. However, there was only a 2% difference in iACT between the recumbent and upright positions in the P region, and no difference in the iACT between the two body positions in the F and F-P regions (Fig. 3(j)). While the 2% difference between recumbent and upright pedaling resulted in a different classification of the muscle function, the close comparison of iACT in the various regions indicates that this does not translate into a meaningful difference in the functional role of this muscle.

The similarities observed in muscle function and activity as elucidated by the PCFG analysis were also evident in the onset and offset analysis conducted over the range of pedaling rates

studied. The statistical analyses indicated that there were significant body-position effects on burst onset timing (but at only one pedaling rate each for MGAS, TA, and VASM) and offset timing (but at only one pedaling rate each for BF, GMAX, and LGAS) (Tables 3 and 4). Though the analyses indicated that the muscle burst onset and offset angles for the two body positions were different in absolute terms for the above six cases, the phase-shift trends for the muscles, with the exception of TA, were similar. The phase-shift trends for BF, RF, and VASM advanced earlier in the crank cycle as pedaling rate increased similar to the observations of Neptune et al. [8] and Marsh and Martin [13]. LGAS and MGAS burst onset and offset shifted later in the crank cycle with increased pedaling rate for both body positions. Similar to Neptune et al. [8], TA onset and offset timing did not exhibit a discernable trend in either the recumbent or upright position. Thus similar adaptations between the two body positions to altered pedaling rates indicate that the functional roles of the muscles are similar in these positions.

The results of the PCFG analysis for recumbent pedaling present new information on the coordination of the leg muscles in recumbent pedaling. This information on the functional roles of the leg muscles provides a foundation by which to form functional groups, such as power-producing and transition muscles. The results indicate that the GMAX and the vastii (VASL and VASM) muscles comprise the extensor group of the extensor/flexor pair (Fig. 1) that serve to generate power for the crank and limb. None of the muscles examined in this study were classified as flexors. The RF and TA muscles comprised the proximal group, and the hamstrings (BF and SM) and triceps surae (LGAS, MGAS, and SOL) muscles comprised the distal group of the proximal/distal pair (Fig. 1) that transfer energy from the limb to the crank and thereby effect smooth transitions between the extensor/flexor pair [7].

The formation of functional groups in recumbent pedaling will aid in the development of simulations to further our understanding of task-oriented muscle coordination. For example, alternate phasing of similar pairs of functional groups has resulted in simulations of different tasks in upright pedaling (e.g., maximum start-up pedaling in the forward direction and steady-state pedaling in the forward and backward directions) that agree well with measured kinematics, kinetics, and EMG [7,9,21,31].

The formation of the functional groups in recumbent pedaling also provides a basis to develop recumbent pedaling exercise routines that address the needs of specific populations. One example application would be to develop forward dynamic simulations designed to determine electrical stimulation patterns that enable individuals with spinal cord injury to pedal a recumbent ergometer to prevent the incidence of secondary complications associated with disuse of the legs. Based on their functional roles, the muscles included in the model could be partitioned into functional groups with all muscles included in a functional group receiving the same excitation signal. The forward dynamic simulation could then compute stimulation patterns to satisfy a task objective (e.g., stimulation patterns that either provide the best fit to the measured kinetic and kinematic data obtained from able-bodied individuals as they pedaled a recumbent ergometer or evenly distribute the force generated by all activated muscles over the crank cycle).

The results of this study could also aid in the development of training routines and equipment for microgravity environments. As the time periods that individuals spend in microgravity environments increase in both duration and frequency, a form of exercise that inhibits cardiovascular deconditioning, muscle atrophy, and bone demineralization is important. Pedaling is an activity that holds great promise in microgravity environments [32,33]. The recumbent pedaling position is well suited for this environment because the seatback would provide a reaction force necessary to enable an individual to pedal at high work rates and not "float" away. Through the use of forward dynamic simulations of

recumbent pedaling, exercise protocols and equipment could be designed to meet specific physiologic training objectives to counter the adverse reactions of microgravity exposure.

In summary, this study showed that the musculoskeletal system inputs (i.e., muscle activation patterns) are similar for both recumbent and upright pedaling when adjusted for the different orientation in the gravity field. Because the musculoskeletal system outputs (i.e., segment kinematics and kinetics) are also similar, the PCFG analysis used previously for upright pedaling provided an effective means by which to determine the functional roles of the leg muscles in recumbent pedaling. In both recumbent and upright pedaling, the uniarticular hip and knee extensors function primarily to produce power during the extension region of the crank cycle, whereas the biarticular muscles crossing the hip and knee function to propel the leg through the transition regions of the crank cycle. The muscles comprising the triceps surae group functioned to transfer power from the leg to the crank through the distal transition region. The ankle dorsiflexor functioned to aid the transition from leg flexion to extension during the proximal region. Muscle onset and offset timing responded similarly to changes in pedaling rate for the two body positions, indicating that the adaptations of the muscles to changes in pedaling rate are also similar for the two body positions.

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