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PEDAL AND KNEE LOADS USING A MULTI-DEGREE-OF-FREEDOM PEDAL PLATFORM IN CYCLING

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Abstract—To provide a scientific basis for the design of bicycle pedals which possibly alleviate over-use knee injuries, two hypotheses were tested in the present study. The two hypotheses were: (1) that the three-dimensional pedal constraint loads; and (2) that the three-dimensional intersegmental knee loads would be reduced more significantly by a foot/pedal platform allowing both adduction/abduction and inversion/eversion rotations simultaneously than by a platform which allowed either rotation individually. To test these hypotheses, pedal load and lower limb kinematic data were collected from 10 subjects who pedaled with four pedal platforms which allowed zero, one, and two degrees of freedom. A number of quantities describing both pedal loads and intersegmental knee loads was computed for each of the four pedal platforms using a previously reported biomechanical model. The quantities included the positive and negative extremes, averages, and areas, as well as the total absolute area and RMS. Quantities were compared using analysis of variance techniques. The key results were that there were significant reductions in the coupled nondriving moments at the pedal for the dual-rotation platform compared to each of the single-rotation cases. The significant reductions in the coupled moments at the pedal were not manifest at the knee. However, a general nonsignificant reduction in both coupled knee moments was evident. Also, the valgus knee moment was significantly reduced by the dual-rotation platform compared to the inversion/eversion only design. Although the axial knee moment was not significantly reduced by the dual-rotation platform over the adduction/abduction design, there was a general nonsignificant reduction. The lack of significance in knee load results occurred because of high intersubject variability. Accordingly, load reduction benefits made by introducing the second degree of freedom need to be considered individually.

Keywords: Cycling; Loads; Pedal; Knee.

INTRODUCTION

In a previous article (Ruby and Hull, 1993), the hypothesis that allowing relative motion between the foot and pedal would generally reduce the intersegmental knee loads was tested. The motivation behind this hypothesis was that reducing intersegmental knee loads would possibly alleviate over-use knee injuries in cycling. These injuries plague cyclists at all levels of participation, from the casual recreational rider to the elite competitor (e.g. Holmes *et al.*, 1991).

The rationale behind the hypothesis was that permitting the relative motion would eliminate constraint loads at the pedal, and this would manifest as decreased intersegmental loads at the joints. In Ruby and Hull (1993), this was found to be valid at the knee when either adduction/abduction or inversion/eversion rotation was permitted separately. However, since allowing only one degree of freedom still constrains the other, it might reasonably be hypothesized that allowing both degrees of freedom simultaneously would result in greater reductions in both the pedal and knee loads over either degree of freedom allowed separately. Thus the objective of the work reported in this article was to test the two hypotheses that (1) three-dimensional pedal loads and (2) three-dimensional intersegmental knee loads both would be reduced more by a pedal platform which allowed adduction/abduction and inversion/eversion rotations simultaneously than by a platform which allowed either rotation individually.

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METHODS

Ten male cyclists from a population of competitive cyclists volunteered for participation in this study (height = 1.81 m, S.D. = 0.04 m; weight = 76.49 kg, S.D. = 3.35 kg; age = 29.6 yr, S.D. = 4.1 yr). Informed consent was obtained before the experiment. The subjects rode a conventional racing bicycle mounted on an electronically braked Schwinn Velodyne ergometer which provides a constant workrate independent of pedaling rate. The bicycle seat and handlebar heights were adjusted to match each cyclist's preferred geometry. The cleats mounted on the bottom of the shoes were aligned along the center axis of the shoe to standardize the cleat position for all subjects (Ruby and Hull, 1993).

The three-dimensional intersegmental knee loads were computed using two models. These models have been described in detail in Ruby *et al.* (1992) and will be reviewed briefly here. The bicycle–rider system was modeled as a five-bar linkage with motion in both the frontal and sagittal planes. The model was analyzed independently in each plane to obtain the intersegmental knee loads. In the sagittal plane the equations of motion for each link were solved using inverse dynamics, starting with the foot and proceeding through each link to the hip. In the frontal and transverse planes, a quasi-static model was used which neglected inertial effects. The anthropometric estimates of the mass, center of gravity, and moments of inertia of the foot, shank, and thigh were defined based on the work of Plagenhoef *et al.* (1983). Using these models, the loads exerted by the tibia on the femur were computed in a tibia-fixed coordinate system (Fig. 1). In this coordinate system the intersegmental knee forces include the anterior (+)/posterior (−) force F''_x, the medial (+)/lateral (−) force F''_y and the compressive (+)/distractive (−) force F''_z. The intersegmental knee moments are the varus (+)/valgus(−) moment M''_x, the extension (+)/flexion (−) moment M''_y, and the internal (+)/external (−) moment M''_z.

The necessary kinematic data were recorded at 60 Hz from reflective spherical markers located in the sagittal plane over the right anterior–superior iliac spine (ASIS), greater trochanter, lateral epicondyle, lateral malleolus, and crank spindle. The hip joint center was assumed to be positioned at the end of the average vector from the ASIS marker to the greater trochanter marker (Neptune and Hull, 1995). The markers used to record frontal plane motion were located over the tibial tuberosity and at a point on the lower shank proximal to the ankle joint. Data were recorded using a motion analysis system (Motion Analysis Corp., Santa Rosa, CA). Angular orientation data of the crank arm and pedal were collected simultaneously with two optical encoders sampled at 100 Hz. The crank angle was measured from the top dead center position and a positive angle corresponded to a clockwise rotation. The video data were filtered using a fourth-order zero-phase shift Butterworth low-pass filter with a cutoff frequency of 6 Hz. The filtered video data were linearly interpolated to correspond in time with the pedal force and encoder data. All derivatives to determine coordinate accelerations were calculated using a second-order central difference technique.

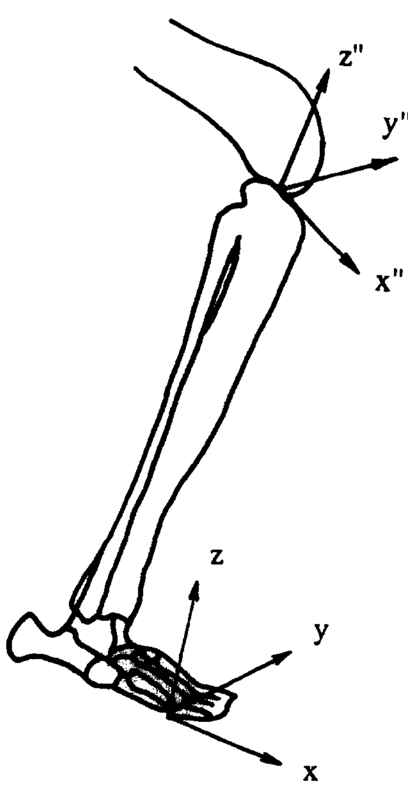

Fig. 1. Bicycle and knee coordinate systems. Inversion/eversion rotations occur about the x-axis and adduction/abduction rotations occur about the z-axis.

The pedal load data were collected simultaneously with the video and encoder data using the six load component pedal dynamometer described by Boyd *et al.* (1996) and a multi-degree-of-freedom pedal interface described by Wooten and Hull (1992). The coordinate system used to describe the loads was fixed relative to the pedal (Fig. 1) and the origin was located at the center of the bottom surface of the cleat. Using shear panels as the elastic elements and electrical resistance strain gages as the transducers, the dynamometer measured the six pedal load components with root mean squared errors (RMSEs) bounded by 3.21 N and 0.21 Nm for force and moment components, respectively. The multi-degree-of-freedom pedal interface allowed both inversion/eversion and adduction/abduction rotations either separately or in combination and measured the corresponding rotations. To minimize friction in the mechanisms allowing the rotations, all surfaces were coated with Teflon. Both rotation limits were approximately $\pm 10°$. Weight was added to the opposite pedal to counter-balance the dynamometer and interface. The pedal load data and encoder data were filtered using a fourth-order zero-phase shift Butterworth low-pass filter with a cutoff frequency of 20 Hz.

The protocol consisted of a 15 min warm-up period at a workrate of 120 W at 90 rpm. Then each subject cycled at a steady-state level of 90 rpm and 250 W using four different pedal platform setups: fixed position (FIX), inversion/eversion rotation (IN/EV), adduction/abduction rotation (AD/AB), and both rotations in combination (BOTH) yielding four different protocols. These protocols were randomly assigned to control for possible carry-over effects as a result of fatigue. After a 3-min adaptation period, data collection was randomly initiated twice during the following 2 min for 10 s each. The accuracy of control was maintained at ± 1 rpm and ± 5 W.

Significant differences for calculated quantities describing both pedal loading and knee loading for the different pedal platforms were computed using a one factor repeated measures ANOVA design (Neter *et al.*, 1990), with four levels corresponding to each of the pedal platforms. The chosen significance level was $\alpha = 0.05$. A Tukey test ($\alpha = 0.05$) was then used to examine the differences in the means of the loading quantities. Following Ruby and Hull (1993), eight descriptive quantities were computed and compared for each load for each of the four pedal platforms. Computed from the load profiles averaged over 14 cycles, the eight quantities were maximum value, minimum value, RMS value, average positive value, average negative value, absolute area, positive area, and negative area under the load vs crank angle curve.

RESULTS

Foot motion

Six of the ten subjects remained abducted throughout the crank cycle when using AD/AB (Fig. 2(A)). Three subjects reached the maximum abduction angle of approximately -10° from center position. When using IN/EV, four of the subjects remained everted throughout the entire crank cycle (Fig. 2(B)) while the other six rotated in both the inversion and eversion directions. One subject consistently rotated to the maximum allowable eversion angle of -10°.

When BOTH was used, the general patterns of both movements remained similar to the single rotations. However, the ranges increased and the motions were offset in either a positive or negative direction depending on the subject. Only two of the subjects remained abducted through the entire crank cycle, and one subject remained entirely adducted. All but two subjects remained within the $\pm 10^{\circ}$ of adduction/abduction motion allowed by the pedal platform. Three subjects remained everted throughout the entire crank cycle, while the other seven rotated in both the inversion and eversion directions.

Pedal loads

Sample results for one subject are presented to illustrate qualitatively the effect of the various foot/pedal platform rotations on the loads imparted on the pedal by the foot (Fig. 3). The pedal loads are of interest because of their relationship to the loads transmitted by the knee. Because the tests were conducted at constant workrate, the driving pedal loads F_x and F_z were relatively unaffected by the different platforms. Accordingly sample results are presented only for the nondriving loads. Average results for a single subject rather than for the subject sample are presented because within subject effects were of interest.

The nondriving pedal force F_y varied little among subjects with 9 out of 10 subjects showing a pattern similar to that in Fig. 3. However, as the pedal interface was changed, the extreme values for F_y changed within a range of ± 20 N, depending on the subject.

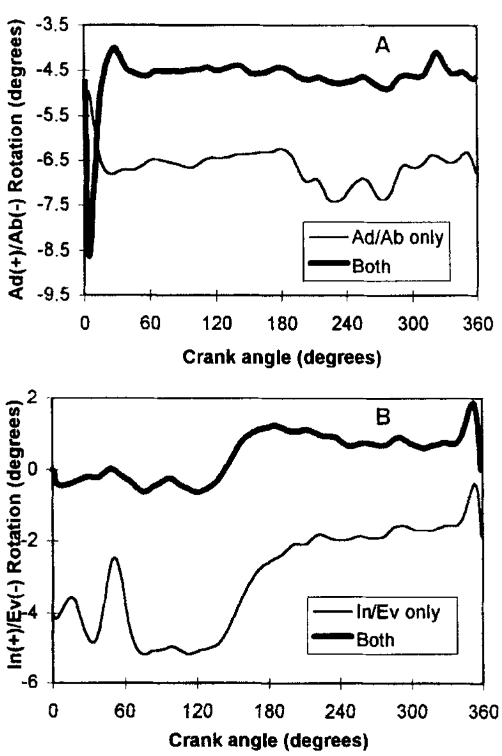

Fig. 2. Sample plot of rotation angle for single and combined rotations; (A) adduction/abduction angle, (B) inversion/eversion angle. Sample data are presented for subject 6 and averaged over 14 cycles.

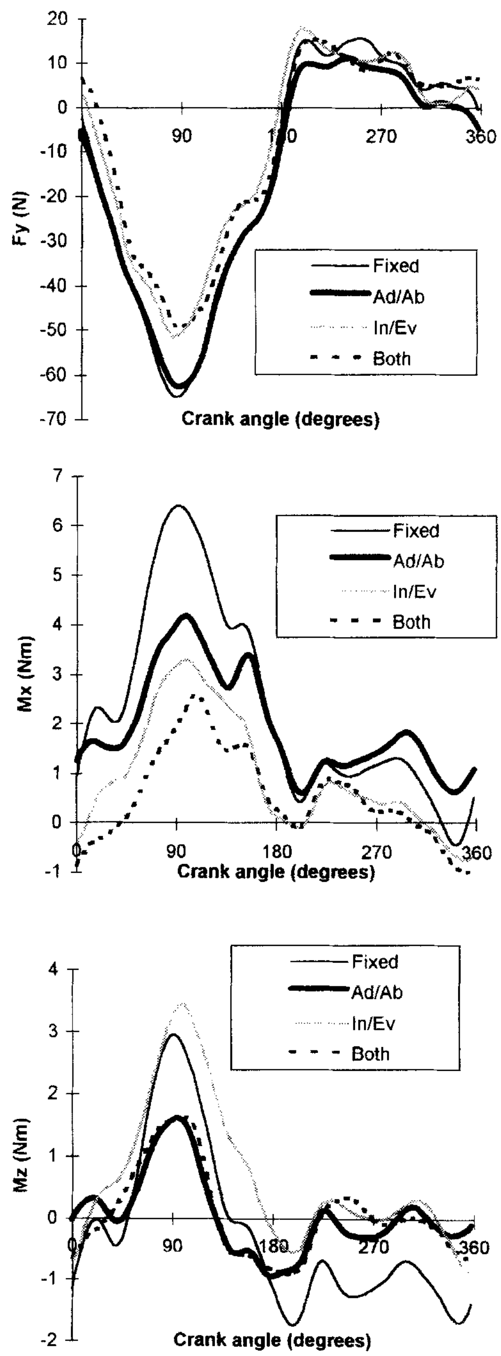

Fig. 3. Sample nondriving pedal loads for subject 6 for four peda platforms. Data are averaged over 14 cycles.

Of the two nondriving moments, the pattern of M_x was more variable among subjects, but could be generally described as having a maximum in the latter half of the downstroke (i.e. 90–180°) and a minimum near 0°. Extreme values ranged between -4.5 and 7 Nm. In contrast to M_x, M_z had the same phase for all 10 subjects reaching maximum values between 1 and 4 Nm at a crank angle near 90° and minimum values between -1 and -6 Nm during the upstroke. Although patterns remained consistent for each platform, both M_x and M_z were reduced substantially by IN/EV and AD/AB, respectively (Fig. 3). Neither moment was reduced to zero, particularly during the downstroke, because of friction in the mechanisms.

When compared to FIX, either AD/AB or IN/EV resulted in significant reductions not only in descriptive quantities associated with corresponding direct loads (M_z for AD/AB and M_x for IN/EV), but also in quantities describing some of the coupled loads (Table 1). Except for the area, the positive quantities associated with the medial pedal force, F_y, were reduced with AD/AB as were several quantities describing M_x. Two quantities describing M_z were reduced with IN/EV.

BOTH produced not only significant differences when compared to FIX (more than either AD/AB or IN/EV), but also many significant differences when compared to the individual rotation platforms. Specifically, two quantities describing F_y were lower for BOTH than for IN/EV. Four of the eight quantities describing the coupled inversion/eversion moment, M_x, were lower for BOTH than for AD/AB, and five quantities describing the coupled adduction/abduction moment, M_z, were reduced more by BOTH than IN/EV.

Knee loads

As with the pedal loads sample intersegmental knee loads for one subject are presented to illustrate qualitatively the effect of the various foot/pedal platforms on the loads imparted on the femur by the tibia (Fig. 4). Again only the nondriving loads are illustrated. The nondriving knee force F''_y is not included, however, because this load essentially duplicated F_y at the pedal.

M''_x had the same phase for nine subjects, with a maximum varus moment developed near 90° and a minimum valgus moment developed between 200 and 250°. The pattern of M''_z varied greatly among subjects. Half of the subjects had a pattern similar to that in Fig. 4 where an internal axial moment was developed in the downstroke and a smaller magnitude external axial moment was developed in the upstroke. However, there was no consistent pattern for the other subjects. Regardless of the moment, the patterns were relatively unaffected by the different platforms. However, the extreme values were affected with M''_z experiencing larger relative changes because of its smaller magnitude.

Unlike the pedal loads, there were no significant differences between descriptive quantities for either of the direct nondriving knee moments when the corresponding

Table 1. Comparisons of descriptive pedal loading quantities for four pedal platforms

	p	mu_F	mu_Z	mu_X	mu_B	Z vs F	X vs F	B vs F	B vs Z	B vs X	Z vs X
F_Y											
− Area	0.173	17.14	16.98	16.37	14.71						
Total area	0.077	21.07	19.80	20.20	17.86						
Avg. | − value|	0.038	29.67	26.27	28.83	24.82			<		<	
Avg. + value	0.042	8.96	6.91	8.46	7.09	<		<			
Maximum	0.024	16.81	13.19	16.29	13.40	<		<		<	<
Minimum	0.245	− 56.35	− 55.20	− 53.75	− 49.48						
+ Area	0.058	3.93	2.82	3.83	3.14						
RMS	0.184	27.79	27.26	26.81	24.27						
M_X											
− Area	0.018	0.37	0.35	0.15	0.13		<	<	<		>
Total area	< 0.001	1.90	1.53	1.08	0.92	<	<	<	<		>
Avg. | − value|	0.006	0.85	0.71	0.46	0.42		<	<	<		
Avg. + value	0.006	1.90	1.53	1.23	1.13		<	<			
Maximum	0.019	4.36	3.28	3.00	2.68	<	<	<			
Minimum	0.055	− 1.73	− 1.51	− 1.03	− 0.90						
+ Area	0.016	1.53	1.18	0.93	0.79		<	<			
RMS	< 0.001	2.37	1.86	1.44	1.21	<	<	<	<		>
M_Z											
− Area	< 0.001	0.84	0.26	0.65	0.24	<		<		<	<
Total area	< 0.001	1.21	0.50	0.97	0.49	<	<	<		<	<
Avg. | − value|	< 0.001	1.16	0.45	0.87	0.40	<		<		<	<
Avg. + value	0.002	1.14	0.50	0.84	0.62	<		<			<
Maximum	0.008	2.23	1.20	1.66	1.30	<		<			
Minimum	< 0.001	− 2.49	− 1.00	− 1.81	− 0.83	<		<		<	<
+ Area	0.189	0.37	0.24	0.32	0.25						
RMS	< 0.001	1.44	0.64	1.16	0.62	<	<	<		<	<

Note. Z represents adduction/abduction rotation. X represents inversion/eversion rotation. B represents both rotations. F represents no rotations. mu represents average of listed quantity for a specified rotation. p is the probability that there is no effect from any of the rotations. Units are N for the force quantities and Nm for the moment quantities. Inequalities are used to indicate which means are different according to the ANOVA and then the Tukey test at $\alpha = 0.05$.

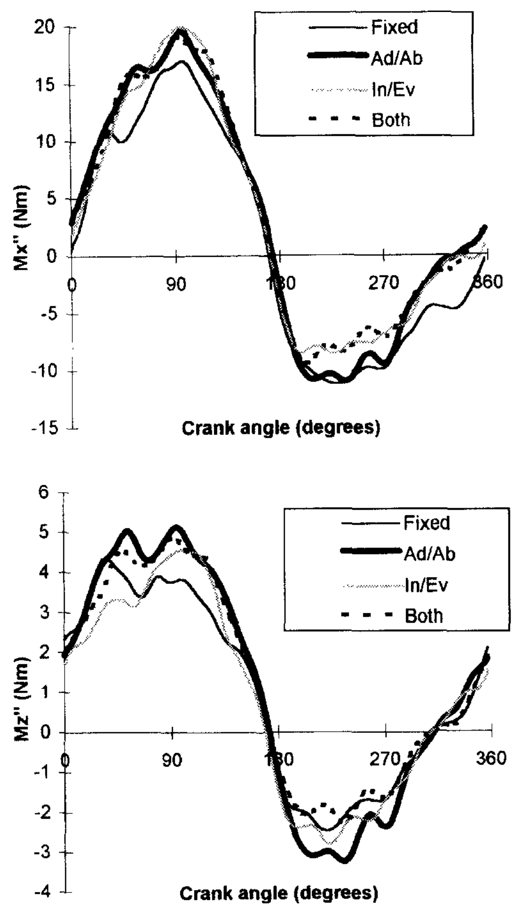

Fig. 4. Sample nondriving knee moments for subject 1 for four pedal platforms. Data are averaged over 14 cycles.

single rotation platform was compared to FIX (Table 2). However, the absolute average values of six of the eight quantities computed were lower for IN/EV than FIX. In contrast, the absolute average values of six of the eight computed quantities were greater for AD/AB than FIX.

Significant differences existed between BOTH and FIX only for M_x''. All descriptive quantities for which a difference existed were reduced for BOTH. Although there were no significant differences in any of the quantities describing M_z'', a reduction in six of the eight quantities was evident for BOTH.

Significant reductions occurred for the three quantities describing the direct valgus knee moment, $-M_x''$, when BOTH was compared to IN/EV. Although not significant, the remaining five quantities had lower values for BOTH. For the coupled moment, M_z'', all quantities were reduced for BOTH but none of the reductions was significant.

When BOTH was compared to AD/AB, seven of the eight quantities describing the direct moment, M_z'', were reduced for BOTH but none of the reductions was significant. Also all eight mean values for the coupled M_x'' moment were lower for BOTH but none was significantly lower.

DISCUSSION

It was hypothesized that allowing two rotational degrees-of-freedom simultaneously at the pedal would: (1) reduce constraint loads at the pedal over either degree-of-freedom allowed separately; and (2) achieve corresponding reductions in intersegmental knee loads. To test these two hypotheses, it was necessary to both measure pedal loads and compute intersegmental knee loads by means of a biomechanical model using data collected for all four possible combinations of the two degrees-of-freedom studied. The limitations inherent in these computations were discussed previously (Ruby *et al.*, 1992) and will not be repeated here. Also discussed previously and not repeated here is the rationale behind the eight descriptive quantities computed for purposes of quantitative comparisons (Ruby and Hull, 1993).

The pedal load results support the hypothesis that constraint loads at the pedal would be reduced more significantly by BOTH than by either AD/AB or IN/EV. BOTH caused significant reductions in quantities describing both coupled moments compared to either single rotation case (Table 1).

The benefits of BOTH did not manifest fully at the knee. Whereas at least half of the eight quantities describing each of the coupled moments was significantly reduced at the pedal (Table 1), none of the quantities associated with the coupled loads was reduced significantly at the knee (Table 2). However, all quantities associated with both coupled loads were lower on the average thus indicating a general nonsignificant reduction.

Although quantities associated with coupled loads were not reduced significantly at the knee, three quantities for the direct load M_x'' were reduced significantly by BOTH over IN/EV (Table 2). Further, all remaining quantities for this direct load and seven of the eight quantities for M_z'' as a direct load exhibited a nonsignificant reduction. Thus, BOTH was more effective than either single-rotation platform in reducing direct and coupled moments transmitted by the knee, but only the valgus moment as a direct load was significantly reduced.

Because nonsignificant reduction of descriptive quantities of intersegmental knee loads for the population of cyclists was indicated with the exception above, care must be taken in applying the results to each individual. The differences in the means of the intersegmental knee loads were often small compared to the magnitude of the load and the intersubject variation. Additionally, all subjects had increases in some quantities used to describe intersegmental knee moments for both the single and dual degree of freedom platforms compared to FIX. For example, the lowest maximum varus moment occurred for FIX for four subjects. Accordingly, the benefits of platforms with either single or dual degrees of freedom should be considered on a subject-by-subject basis.

Knee load results for both the fixed and the single rotation platforms disagree in some respects with those reported at the same workrate by Ruby and Hull (1993). One difference was the nonsignificant increase in the M_z'' quantities reported herein rather than the significant decreases reported by Ruby and Hull (1993). Because Ruby and Hull (1993) found significant reductions in four quantities associated with M_x'', another difference was

Table 2. Comparisons of descriptive knee loading quantities for four pedal platforms

	p	mu_F	mu_Z	mu_X	mu_B	Z vs F	X vs F	B vs F	B vs Z	B vs X	Z vs X
M''_X											
– Area	0.017	2.98	2.31	2.90	2.31	<		<		<	<
Total area	0.155	8.34	8.26	8.23	7.30						
Avg.| – value|	0.022	6.15	5.15	5.85	4.70	<		<		<	
Avg. + value	0.395	10.08	9.89	9.78	8.95						
Maximum	0.596	18.19	19.79	18.13	17.64						
Minimum	0.022	– 11.54	– 9.67	– 11.35	– 9.34	<		<		<	<
+ Area	0.315	5.37	5.95	5.33	4.99						
RMS	0.301	10.14	10.63	10.22	9.38						
M''_Z											
– Area	0.868	0.80	0.85	0.86	0.74						
Total area	0.515	2.36	2.59	2.57	2.26						
Avg.| – value|	0.374	1.75	1.69	1.73	1.56						
Avg. + value	0.679	2.57	2.89	2.75	2.30						
Maximum	0.640	5.22	6.64	6.01	5.79						
Minimum	0.464	– 3.57	– 3.36	– 3.63	– 3.22						
+ Area	0.747	1.56	1.73	1.72	1.52						
RMS	0.646	3.03	3.57	3.38	3.12						

Note. Z represents adduction/abduction rotation. X represents inversion/eversion rotation. B represents both rotations. F represents no rotations. mu represents average of listed quantity for a specified rotation. p is the probability that there is no effect from any of the rotations. Units are N for the force quantities and Nm for the moment quantities. Inequalities are used to indicate which means are different according to the ANOVA and then the Tukey test at $\alpha = 0.05$.

that no descriptive quantities were significantly reduced for IN/EV compared to FIX. A final difference was that the range of motion for IN/EV, 6.3–17.3°, was substantially greater than their range, 1.5–6.8°.

A factor that has the potential to explain the different results between the two studies was the bicycle setup. In the study by Ruby and Hull (1993), the seat height was standardized at 100% of trochanteric leg length for all subjects. In the study here, however, the seat and handlebar position were adjusted to match the subjects' own equipment. Inasmuch as bicycle seat height affects pedaling mechanics (Gregor *et al.*, 1991) it is possible that this difference in bicycle setup contributed to the different results. However, the average of preferred heights for competitive cyclists such as those in the sample here matches almost identically the standardized heights (Browning *et al.*, 1988). Accordingly, seat height was probably not an important factor.

Another factor that may explain the differences observed in the results for IN/EV between the two studies is the platform design. The pedal platform used herein allowed inversion/eversion rotation about an axis designed to coincide with the subtalar axis of the foot. This was presumably a more natural axis than the inversion/eversion axis used by Ruby *et al.* (1992), which allowed rotation about an axis below the foot. Possibly, the difference in elevation of the axis accounts for the differences between the two studies.

Because the position of the axis relative to the foot was the same for both designs, the foot/pedal platform design does not account for the differences in the effect of the AD/AB however. Furthermore, the procedures to establish cleat alignment on the sole of the shoe were identical.

A final factor that may explain differences in the effect of AD/AB is inherent differences in pedaling mechanics between the subjects in the two samples. As mentioned previously, the pattern of M''_z varied among the subjects used herein. Five of the ten subjects exhibited the pattern in Fig. 4 where an internal axial moment was developed in the downstroke and the other half exhibited no consistent pattern. In contrast, the study by Ruby and Hull (1993) included only three of eleven subjects with an M''_z pattern similar to Fig. 4.

To appreciate the impact that subjects with an M''_z pattern similar to Fig. 4 would have on the results of the statistical analysis, notice that the computed quantities of the positive portion of the curve were higher for AD/AB than FIX. Consequently the effect was to increase all positive quantities plus the total area and the RMS. With 50% of the subjects contributing to these increases, the corresponding average values increased for the subject sample (Table 2). However, in the study by Ruby and Hull (1993), only 27% of subjects would have had a similar effect. Evidently, this percentage was not large enough to drive the averages upward.

The inability of AD/AB to reduce the internal axial knee moments for subjects with a M''_z pattern similar to that in Fig. 4 can be explained through a moment subcomponent analysis (Ruby and Hull, 1993). This analysis shows that the intersegmental knee moments result from a superposition of the moments created by each of the pedal loads. In the case of M''_z the form of the equation is

$$M''_z = M''_z\,(F_x) + M''_z(F_y) + M''_z(F_z) + M''_z(M_x) + M''_z(M_z) + M''_z(g)$$

where each of the terms is the moment contribution by a pedal load component plus a gravity term which is negligible. Of the remaining five terms, the subcomponent due to the pedal force F_y is dominant (Fig. 5). Since the force applied by the foot to the pedal is laterally directed during the downstroke, the $M''_z(F_y)$ term contributes a large internal axial moment. This is negated to some degree by the external axial moment contributed by

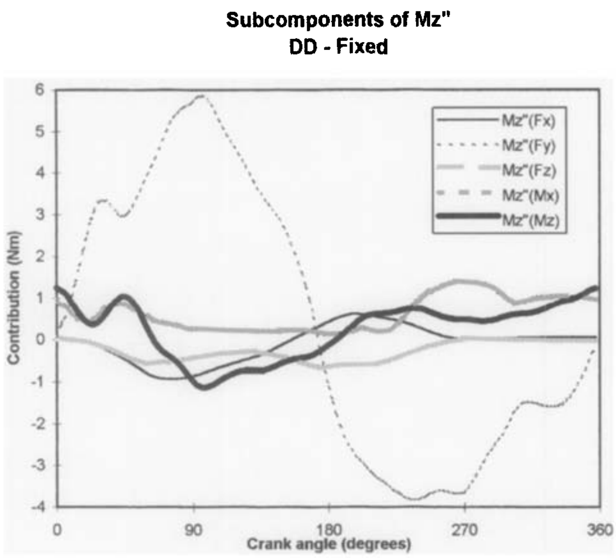

Fig. 5. Sample moment subcomponents for the axial knee moment developed with the fixed pedal platform. Data are for subject 1 averaged over 14 cycles.

the M_z'' (M_z) term. In decreasing the magnitude of the moment, AD/AB fails to achieve this negation so that the net effect is an increase in the internal axial moment during the downstroke. This amplifies the point made earlier that the benefits of pedal platform degrees of freedom must be assessed on a case-by-case basis.

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