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Knee joint loading in forward versus backward pedaling: implications for rehabilitation strategies

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Abstract

Objective. To use forward dynamic simulations of forward and backward pedaling in order to determine whether backward pedaling offers theoretical advantages over forward pedaling to rehabilitate common knee disorders.

Design. A comparison of knee joint loads was performed during forward and backward pedaling.

Background. Pedaling has been shown to be an effective rehabilitation exercise for a variety of knee disorders. Recently, backward gait has been shown to produce greater knee extensor moments and reduced patellofemoral joint loads compared to forward gait. But to date, no study has examined the efficacy of backward pedaling as a safe alternative to forward pedaling in rehabilitation programs.

Methods. A musculoskeletal model and optimization framework was used to generate simulations of forward and backward pedaling. Tibiofemoral and patellofemoral joint reaction forces were quantified.

Results. Lower tibiofemoral compressive loads, but higher patellofemoral compressive loads, were observed in backward pedaling. Lower protective anterior–posterior shear force was observed in backward pedaling near peak extension.

Conclusions. Backward pedaling offers reduced tibiofemoral compressive loads for those patients with knee disorders such as menisci damage and osteoarthritis, but higher patellofemoral compressive loads. Therefore, backward pedaling is not recommended for patients experiencing patellofemoral pain. Further, backward pedaling should not be recommended after anterior cruciate ligament injury or reconstruction.

Relevance

The results of this study indicate that the design of rehabilitation programs including pedaling exercises should be injury specific with particular attention paid to the mechanics of the task.

Keywords: Pedaling; Knee joint rehabilitation; Simulation; Musculoskeletal modeling; Biomechanics

1. Introduction

Pedaling a stationary ergometer has been used for rehabilitation of a variety of knee disorders such as anterior cruciate ligament (ACL) reconstruction and patellofemoral pain (PFP). Pedaling has been used due to reductions in ACL strain and compressive loading of the knee joint compared to full weight bearing rehabilitative exercises (e.g. walking, running). Another commonly used rehabilitation technique has been backward gait [1,2], since it provides several advantages similar to pedaling when compared to forward gait. Recently, investigators have suggested that similar to gait, backward pedaling might offer similar advantages over conventional forward pedaling [3]. However, these recommendations have been made largely by extrapolating the results found in gait and applying these results to pedaling, rather than basing these recommendations on an understanding of the mechanics of backwards versus forward pedaling.

For example, conventional (forward) pedaling has been shown to reduce ACL strain values as compared to those values recorded during other rehabilitation exercises such as stair climbing, leg extension and squatting exercises [4]. One of the primary goals of ACL rehabilitation is to provide dynamic stability of the knee and protection for the ACL graft. The hamstrings are the primary providers of knee stability by preventing ante-

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rior translation of the tibia relative to the femur in response to extensor loads generated by the quadriceps muscle [5,6]. Considering the importance of the hamstrings to protect the ACL and that forward pedaling is a safe and viable rehabilitation exercise, researchers have examined non-traditional pedaling exercises to strengthen the hamstrings during functional rehabilitation. Eisner et al. [7] examined backward pedaling and hypothesized that the hamstrings would demonstrate greater activation duration during backward pedaling that would provide greater increases in strength and a basis for varying pedaling direction during knee rehabilitation. Their results showed an increase in hamstring activity during the recovery phase and led them to conclude that backward pedaling is a viable alternative for ACL rehabilitation. But recent studies have shown that the total duration of hamstring activity [3,8] and power output [9] is similar in both pedaling directions, and therefore, backward pedaling may not provide a rehabilitative advantage over forward pedaling. In addition, the task mechanics of forward pedaling requires co-activation of the quadriceps and hamstrings muscles during that latter portion of knee extension in the propulsive phase, and therefore, ACL strain is reduced when the knee is approaching full extension [4]. But co-activation of the hamstrings and quadriceps does not occur in backward pedaling [3,8]. Instead, the task mechanics of backward pedaling requires a primarily anti-phase activation of the hamstrings and quadriceps groups with the quadriceps active in isolation near full extension when the ACL is most vulnerable to anterior translation [5,6]. Therefore, the ACL may be more susceptible to anterior shear force and strain during backward pedaling.

Forward pedaling has also been used to rehabilitate patients with PFP to strengthen the quadriceps muscle group while reducing the compressive loading in the patellofemoral joint. Gait studies have shown that greater knee extensor moments [10,11] and reduced patellofemoral joint loads [12] are elicited in backward versus forward walking and running, which has caused backward gait to become a common component of functional knee rehabilitation [2]. These results have led others to speculate that the same may hold true in backward pedaling. Bressel et al. [3] observed heightened quadriceps activity during the downstroke in backward pedaling that would be useful for strengthening the knee extensor moment, but the effect of increased quadriceps activity on the patellofemoral joint loads was not quantified. The mechanics of backward pedaling require quadriceps activity during regions of greater knee flexion compared to forward pedaling [3,8]. Increased patellofemoral loads are associated with extensor activity at increased flexion angles [13], and therefore, higher patellofemoral joint loads may be experienced during backward pedaling.

Reduced joint loading is also desired in the rehabilitation of tibiofemoral disorders such as meniscus damage and osteoarthritis. As a result, pedaling has been used as a rehabilitation exercise for these disorders since it provides reduced joint loading compared to full weight bearing gait. However, the different task requirements between forward and backward pedaling may provide different joint loading patterns and it is not known whether one pedaling direction may offer advantages over the other in rehabilitating these disorders.

This study uses forward dynamic simulations of forward and backward pedaling to determine whether backward pedaling offers theoretical advantages over forward pedaling in the rehabilitation of common knee disorders. Forward dynamic simulations of human movement have been shown useful in examining joint loading during human movement [14–16]. But to date, no study has theoretically examined knee joint loading during forward and backward pedaling. Therefore, the goal of this study was to use a musculoskeletal model and simulation of forward and backward pedaling to quantify the sensitivity of the tibiofemoral and patellofemoral joint loads to pedaling direction. These results will provide the data necessary for choosing the appropriate pedaling exercises when designing specific rehabilitation programs for different knee disorders. As a result, the clinician will have a theoretical basis for selecting such programs.

2. Methods

2.1. Simulation model

A planar two-legged bicycle-rider musculoskeletal model was generated using SIMM (MusculoGraphics, Evanston, IL, USA) (Fig. 1). This simulation model was previously described in detail [17] and will be reviewed briefly here. Each leg consisted of three rigid-body segments (thigh, shank and foot) with the hip joint center fixed and foot rigidly attached to the pedal. The hip and ankle joints were modeled as revolute while the tibiofemoral joint was modeled with three degrees of freedom with a moving center of rotation for flexion-extension specified as functions of knee flexion angle [18]. The patella was constrained to move along a prescribed trajectory relative to the femur as a function of knee flexion angle [18].

The dynamical equations-of-motion for the model were derived using SD/FAST (Symbolic Dynamics, Mountain View, CA, USA) and a forward dynamic simulation was produced by Dynamics Pipeline (MusculoGraphics, Evanston, IL, USA). The model was driven by 14 individual Hill-type musculotendon actuators that were combined into nine muscles groups with muscles within each group receiving the same excitation

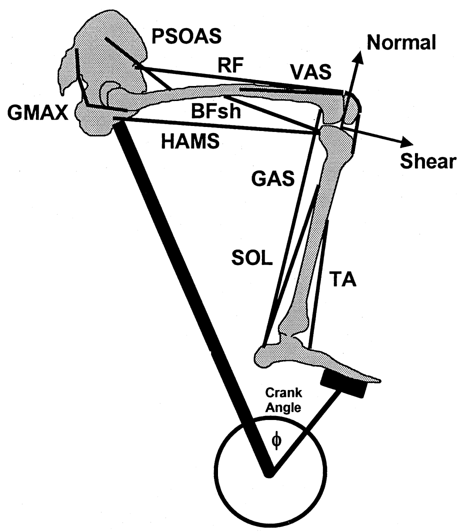

Fig. 1. Right leg of the bicycle-rider musculoskeletal model. Muscles are shown as straight lines for illustration purposes. The 14 muscles included in the model were further combined into muscle groups, with each muscle within each group receiving the same excitation signal. The muscle groups were defined as PSOAS (iliacus, psoas), GMAX (gluteus maximus, adductor magnus), VAS (3-component vastus), HAMS (medial hamstrings, biceps femoris long), SOL (soleus), BFsh (biceps femoris short), GAS (gastrocnemius), RF (rectus femoris) and TA (tibialis anterior). The tibiofemoral joint coordinate system is shown. The patellofemoral joint coordinate system is aligned with the patella with the normal axis directed anteriorly and the shear axis directed superiorly in the anatomical position.

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The individual muscle excitation patterns were modeled as block patterns defined by excitation onset, offset and magnitude and the activation dynamics were modeled by a first-order differential equation [19]. The excitation patterns between the left and right legs were considered symmetric and 180° out-of-phase. The effective crankload dynamics was modeled by an equivalent inertial and resistive torque applied about the center of the crankarm [20].

Forward and backward pedaling simulations were produced using an optimization framework to identify the muscle excitation patterns that reproduced the essential features of group kinetic, kinematic and EMG data. A global optimization simulated annealing algorithm [21] was used to solve a tracking problem by finding the muscle excitation patterns (muscle excitation onset, offset and magnitude) that minimized the performance criterion:

$$J = \sum_{j=1}^{m}\sum_{i=1}^{n}\frac{\left(Y_{ij} - \hat{Y}_{ij}\right)^2}{\mathrm{SD}_{ij}^2}, \qquad (1)$$

where Y_{ij} is the experimentally measured data, $\hat{Y}_{ij}$ the corresponding simulation data, SD_{ij} the inter-subject standard deviations, n the number of data points and m is the number of variables evaluated. The tracking quantities Y_{ij} included the radial and tangential pedal force components and pedal angle. Simulations were performed over four revolutions to assure that initial start-up transients had decayed. The performance criterion (Eq. (1)) was evaluated during the fourth revolution when the simulation had reached its steady-state and was considered to be independent of the initial conditions. An inequality constraint was enforced on the final time to assure the simulations pedaled at the average experimentally measured pedaling rate ± 2 rpm.

To quantify the sensitivity of the tibiofemoral and patellofemoral joint loads to pedaling direction, the joint reaction forces were computed. The reaction forces were expressed in the tibial and patella reference frames, respectively. In both reference frames, compressive forces act normal and the shear forces act tangential to their respective surfaces (Fig. 1). In the tibiofemoral reference frame, positive shear force indicates ACL protection; negative normal force indicates joint compression. In the patellofemoral reference frame, positive shear force is directed superiorly in the anatomical direction; positive normal force indicates joint compression.

2.2. Experimental data

To provide data for the tracking problem and validation for the simulation, the group average radial and tangential pedal force, pedal angle trajectories and EMG data from Ting et al. [8] were used and will be briefly described here. Data were collected on a stationary ergometer during forward and backward pedaling from 16 healthy subjects (8 male, 8 female; age = 24(7) yr (mean, SD); height = 1.74(0.1) m; mass = 70(9) kg) at 60 rpm with a frictional workload of 120 J/cycle. These pedaling conditions are similar to those used in the advanced stages of a typical graded rehabilitation programs [4]. The subjects used cleated cycling shoes that rigidly attached their feet to the pedal. The normal and tangential pedal forces were measured with a pedal dynamometer [22] and the pedal and crank angles were measured with optical encoders. The pedal force and encoder data were collected at 200 Hz and filtered with a zero-lag Butterworth filter at 10 Hz. The pedal forces were then transformed into the crank coordinate system to provide normal and radial forces relative to the crank arm.

Bilateral surface electromyography data were collected from the vastus medialis (VAS), rectus femoris (RF), biceps femoris long head (BF), semimembranosus (SM), tibialis anterior (TA), soleus (SOL) and the medial gastrocnemius (GAS) at 1000 Hz. To characterize the EMG patterns of each muscle, the crank cycle was divided into 16 equally spaced intervals of 22.5° and the integrated EMG was computed within each interval.

Data were collected during two 60 s trials in each pedaling direction. All calculations were performed on a cycle-by-cycle basis, averaged within each trial, and then averaged across the two trials for each subject in each pedaling direction. An ensemble average was then generated by averaging the data across subjects. Further details about the data collection and processing can be found in Ting et al. [8].

3. Results

The pedaling simulation kinetic, kinematic and muscle excitation data closely matched the group averaged experimental data (Fig. 2). The simulated pedal forces and pedal angle in both pedaling directions were almost always within 1 SD of the experimental data and there was also close agreement between the muscle excitation and group EMG data (Fig. 3). The average simulation pedaling rates were 62 and 60 rpm for the forward and backward directions, respectively.

The tibiofemoral compressive and anterior–posterior shear joint forces yielded similar patterns in both pedaling directions, although there were substantial differences in magnitude (Fig. 4). The peak and average compressive joint forces were 47% and 56% higher in forward pedaling, respectively (Table 1). The peak and average anterior–posterior shear forces were 35% and 26% higher in backward pedaling, although at peak knee extension, the protective anterior shear force in backward pedaling was 63% lower.

Similar to the tibiofemoral joint, the loading patterns in the patellofemoral joint were similar in both pedaling directions, although both the shear and compressive joint forces were higher in backward pedaling (Table 1). The largest differences occurred in the peak and average compressive forces that were 33% and 18% higher, respectively.

4. Discussion

The objective of this study was to determine whether backward pedaling offers theoretical advantages over forward pedaling to rehabilitate common knee disorders. To this end, simulations of forward and backward pedaling were generated and the tibiofemoral and patellofemoral joint reaction forces were quantified.

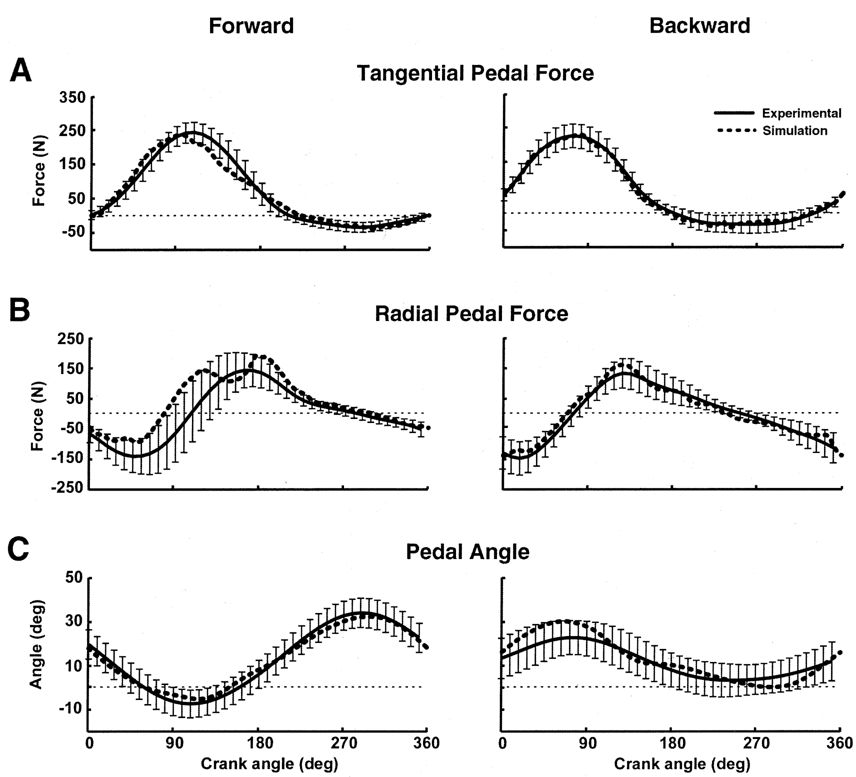

Fig. 2. Comparison of the simulation and group average experimental pedal force and pedal angle data of the right leg. The crank angle is 0° when the crank arm is parallel with the seat tube and the limb is in its most flexed position. The crank angle is positive in the direction of pedaling. Positive radial force is directed away from the crank center and positive tangential pedal force accelerates the crank. The pedal angle is relative to horizontal and positive when the front is pointed down (foot is typically plantarflexed). The simulations reproduced the salient features of the group average data.

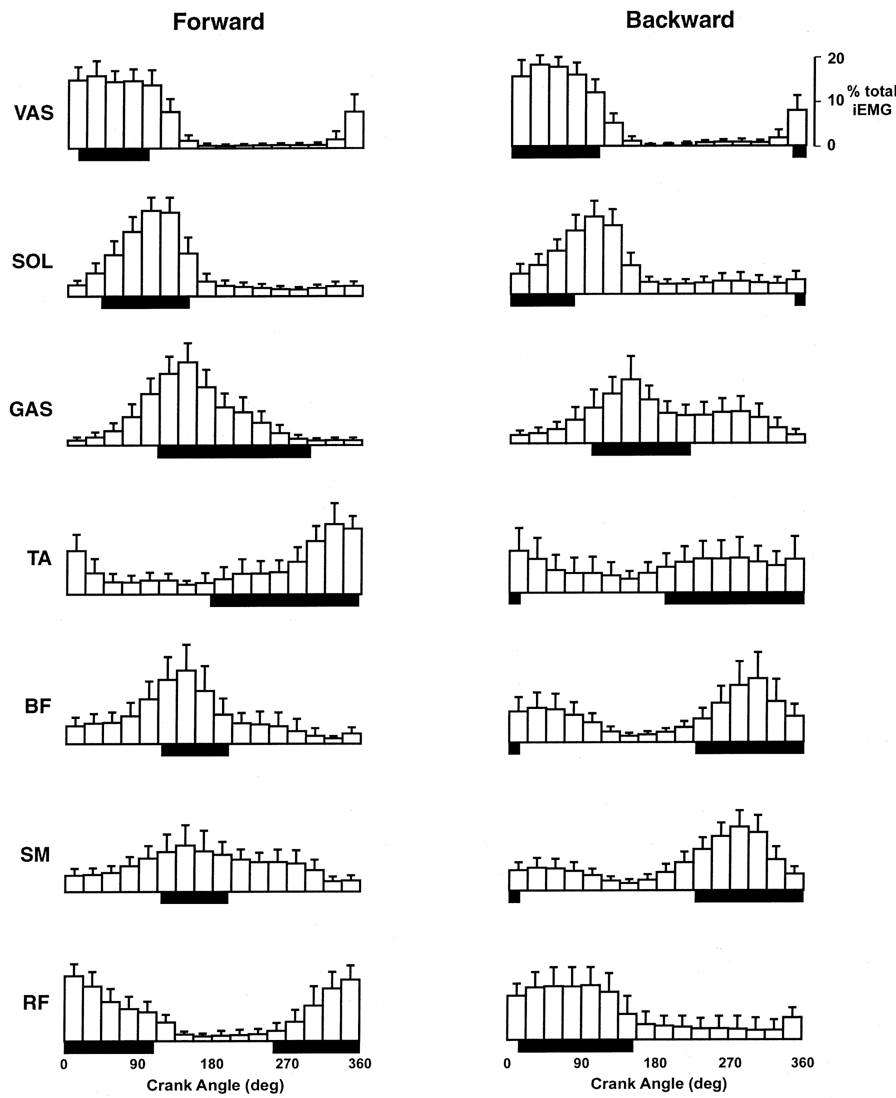

Fig. 3. Comparison of the simulation and group average experimental muscle excitation data. The solid horizontal bars indicate the simulation onset and offset timing. The experimental data represent the mean (S.D.) normalized iEMGs averaged over all subjects (adapted from Ting et al. [8]).

A potential limitation of the study is the mathematical model of the knee joint. This model has been successfully used in a variety of simulation studies [17,19,23] since it provides appropriate kinematics in the sagittal plane. But the model does not include axial rotation of the tibia relative to the femur (and of particular importance, internal rotation) that affects the loading of the ACL. However, since the foot is rigidly attached to the pedal, axial rotation in pedaling is primarily limited to that which naturally occurs during flexion-extension movements, and therefore, is kinematically the same in both pedaling directions. In addition, the knee model is comprised of rigid bodies, and therefore, contact occurs at a single point. In reality, the joint reaction forces are distributed over finite areas in the menisci, cartilage and ligaments. But the goal of the present study was to make a qualitative comparison between forward and backward pedaling. Regardless of how the total reaction force is distributed over the menisci, cartilage and ligaments, the total reaction force transmitted to the bones is nearly identical and will occur similarly in both pedaling directions since the kinetics and kinematics are nearly identical (Fig. 2). Furthermore, pedaling has fewer degrees of freedom than there are muscles in the musculoskeletal system, and therefore, the simulation's muscle coordination pattern that reproduced the experimental data is not unique. Therefore, it is possible that the subjects used muscle coordination patterns that produced lower joint loads than the loads observed in the pedaling simulations. Nevertheless, the simulation results provide valuable insight into the relationship between pedaling mechanics and joint loading. Finally, while the absolute joint loads calculated by the model should be interpreted with caution, the difference between the loading in the two directions should be a more robust result.

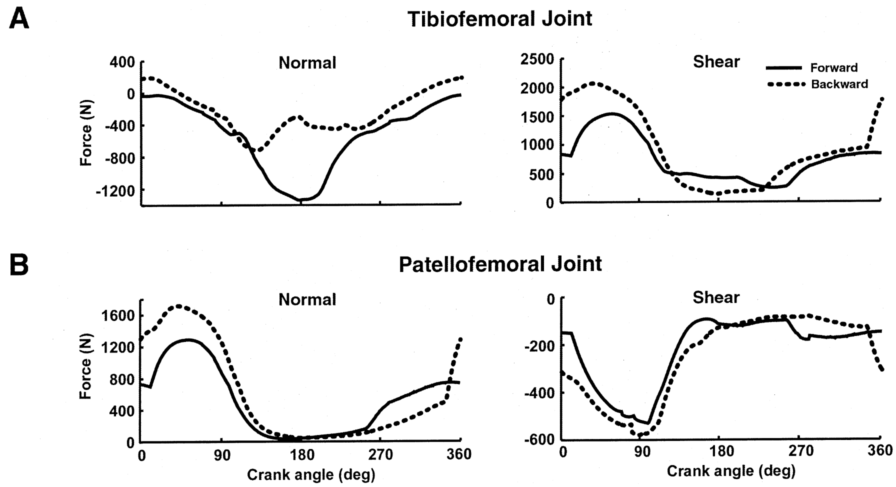

Fig. 4. Joint reaction forces during forward and backward pedaling: (A) tibiofemoral and (B) patellofemoral joint loads expressed in their respective joint coordinate systems. For the normal forces, compressive loads are negative and positive in the tibiofemoral and patellofemoral joints, respectively. In the tibiofemoral joint, positive shear is directed anteriorly and provides a protective mechanism for the ACL. In the patellofemoral joint, positive shear is directed superiorly in the anatomical position.

Table 1
Average and peak tibiofemoral and patellofemoral joint loads[a]

	Forward	Backward	% Difference
Tibiofemoral anterior–posterior shear force			
Peak (N)	1529.7	2061.9	−34.8
Average (N)	745.6	938.5	−25.9
Shear force at peak knee extension			
Force (N)	423.0	152.9	63.9
Tibiofemoral compressive force			
Peak (N)	−1335.7	−715.1	46.5
Average (N)	−528.5	−15.8	56.0
Patellofemoral anterior-posterior shear force			
Peak (N)	−530.7	−580.3	−9.4
Average (N)	−226.6	−253.4	−11.8
Patellofemoral compressive force			
Peak	1283.9	1709.1	−33.1
Average	516.8	610.4	−18.1

[a] Average values were computed over the entire crank cycle. For the compressive joint forces, only negative values were considered. Percent differences were computed relative to forward pedaling.

The joint loading magnitudes were lower than estimates of joint loading during other dynamic movement tasks. Peak tibiofemoral joint compressive loads have been estimated at two to four times bodyweight (BW) during normal walking [24], four times BW during stair climbing [25] and three to seven times BW during rising from a chair [26]. The peak tibiofemoral compressive loads were 1.8 and 1.0 times BW during forward and backward pedaling, respectively. Lower tibiofemoral shear and compressive loads were observed in Ericson and Nisell [27] during similar pedaling conditions as the present study. But the biomechanical model to compute the joint loads used by Ericson and Nisell [27] was based on inverse dynamics and did not include muscle co-contraction contributions between the knee extensor and flexor muscle groups to the total joint loads.

The magnitudes of the joint reaction forces are dependent on the muscle excitation patterns used to generate the simulations. To assess the sensitivity of the joint reaction forces to excitation magnitudes, a post-hoc sensitivity study was performed on the hamstrings and quadriceps. The magnitudes were independently varied ±10% and the difference in joint loading was quantified. Of particular interest was the tibiofemoral shear force at peak knee extension when the ACL is most vulnerable to anterior strain [5,6]. The results showed that the difference in shear force between forward and backward pedaling averaged 65% and was independent of the excitation magnitude variations. These results suggest that the difference in shear force is governed primarily by the mechanics of the pedaling task, not the specific muscle coordination pattern used.

The reduced magnitude of the shear force, which is functionally protective, suggests that backward pedaling is not an appropriate rehabilitation exercise for ACL patients. In forward pedaling, the hamstring activity is centered over the region of maximum knee extension that provides a protective mechanism for the ACL. Fleming et al. [4] reported that in forward pedaling, ACL strain values were minimal. However, the task mechanics of backward pedaling requires prolonged quadriceps activity during knee extension without the protective co-activation of the hamstrings to reduce anterior strain in the ACL (Figs. 3 and 4). Therefore, forward pedaling is the safer pedaling exercise for patients with ACL injuries. However, backward pedaling may be appropriate for individuals with tibiofemoral disorders such as meniscus damage and osteoarthritis since backward pedaling reduced the peak and average tibiofemoral compressive loads by 47% and 56%, respectively. Pedaling exercises have been used in the rehabilitation of joint disorders because pedaling offers reduced compressive joint loading compared to gait. The results of the present study suggest that even greater reductions in joint loading can be achieved during backward pedaling.

Patellofemoral pain is one of the most common knee disorders, comprising nearly one in four clinical diagnoses [28]. The success of gait studies showing that greater knee extensor moments [10,11] and reduced patellofemoral joint loads [12] are elicited in backward versus forward gait have led others to speculate that the same may hold true in backward pedaling [3]. However, these speculations were made without a biomechanical analysis of the backward pedaling task. Recently, such an analysis revealed that the quadriceps are one of the primary power producers in backward pedaling, and that they generate their power when the knee is in a more flexed position than in forward pedaling [9]. Since the quadriceps generate their power in a more flexed position, large compressive loads in the patellofemoral joint are produced (Fig. 4). The sensitivity analysis of the joint loading to quadriceps excitation showed that the patellofemoral joint load responded fairly linearly with the quadriceps activity. These results are consistent with other studies showing increased patellofemoral loads are associated with extensor activity at high flexion angles [13]. Although backward pedaling may be useful in developing an increased extensor moment due to the increase in extensor activity over that in forward pedaling [3,8], these results suggest that backward pedaling may be detrimental for those patients with PFP.

In summary, pedaling has previously been shown to be an effective rehabilitation exercise for a variety of knee disorders and backward pedaling has been suggested to have some advantages over forward pedaling. In order to improve rehabilitation of several common knee disorders, we examined the potential of backward pedaling as a safe alternative for forward pedaling. Our results indicate that the design of rehabilitation programs including pedaling exercises should be injury specific due to differences in pedaling mechanics between forward and backward pedaling. Backward pedaling offers reduced tibiofemoral compressive loads, but higher patellofemoral compressive loads. For those patients with ACL injuries, forward pedaling offers increased protective mechanisms against anterior strain.

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