Pedaling Forces Normalized to Body Weight in cyclists with a Uni-lateral Transtibial Amputation

W. Lee Childers, Karen L. Perell-Gerson, Robert Kistenberg and Robert J. Gregor

School of Applied Physiology, Georgia Institute of Technology, Atlanta, GA USA
Dept. of Kinesiology, California State University, Fullerton, Fullerton, CA, USA

Abstract. Persons with lower limb amputation may be on weight-bearing restrictions following amputation. Cycling provides a method for rehabilitation within a limited body weight environment. Pedal force data obtained from studies on intact cyclists and cyclists with uni-lateral trans-tibial amputation were used to calculate forces normalized to body weight on the shank or residual limb segment and then related to power output. Linear regression was used to develop models describing this relationship. Results show that patients operating within power outputs typical in rehabilitation will experience compressive forces below 30% body weight. Therefore, we conclude that cycling may be safely utilized as a rehabilitation modality for patients on weight-bearing restrictions.

Keywords. Cycling, Amputee, Rehabilitation, Prosthesis, body weight

1. Introduction

While movement is encouraged early in the rehabilitation plan to improve patient outcomes, weight-bearing restrictions may be necessary to reduce the forces imposed on the healing residual limb [1,2]. Cycling serves as a rehabilitation modality within a limited weight-bearing environment that encourages rhythmic, reciprocal movement [3] and may provide an intermediate step to encourage neuromuscular learning and weight bearing on the residual limb before the patient can progress to standing balance and gait training. While much is known about the forces produced during intact cycling [4], the relationship between power outputs and forces imposed on the lower limb relative to body weight has not been published. The purpose of this study was to discuss the relationship between different power outputs and forces experienced by the residual limb relative to body weight with comparison to intact cyclists.

2. Methods

Regression analysis was performed on 42 datasets from ten CTA (eight males and two females; average age 38.8 +/- 13.1 yrs, height 1.76 +/- 0.08 m, and mass 82.5 +/- 13.5 kg) and 28 datasets from twelve intact cyclists (eleven males and one female; average age 39.5 +/- 11.8 yrs, height 1.81 +/- 0.06 m, and mass 73.8 +/- 6.4 kg). These datasets

1 Corresponding Author. W. Lee Childers, School of Applied Physiology, Georgia Institute of Technology, 281 Ferst Ave., Atlanta, GA 30332, USA; lee@gatech.edu. +001-404-894-7658
were derived from previous studies [5, 6]. All subjects signed separate written consent for IRB approval. The subjects rode on a stationary bicycle adjusted to their preferred position with a centripetal resistance unit (1-UP USA Inc.) and adapted with dual piezoelectric element force pedals [7]. The subject’s personal prosthesis was modified by removing the pylon and foot section and replacing it a stiff aluminum pylon/foot section (STIFF) for all ten CTAs [5, 6] or a flexible carbon fiber pylon/foot section (FLEX) for eight CTAs [5].

One study [5] utilized eight CTA subjects and nine intact subjects pedaled for six minutes at loads corresponding to 70 and 90% of their age-predicted maximum heart rate at preferred cadence. The CTA group pedaled at both intensities with a stiff and flexible prosthetic foot. An additional eight CTA datasets and were obtained from a separate study [6] where two CTA and two intact cyclists pedaled at 100 watts at 60 and 90 rpm as well as loads corresponding to 70 and 90% of their age-predicted maximum heart rate. The remaining two CTA and two intact datasets were obtained during pilot testing at 200 watts and 90 rpm.

Both orthogonal and shear components of the pedal reaction force [7] were recorded over the final one minute at each randomly selected load condition. Five complete pedal cycles from each trial were averaged together for analysis. The maximum orthogonal component of force was normalized to the subject’s body weight (%BW) for each limb. The maximum orthogonal component of force occurs at approximately 90 degrees of crank rotation [4]. Linear regression formulas were employed relating power output to %BW of the limb. Pearson’s product moment correlation coefficient was used to test whether %BW was significantly correlated to power output. Fischer’s Z-Transformation was used to test if relationships between limbs or groups were significantly different. Significance was defined as p < 0.05.

3. Results
Percent BW versus power outputs are plotted as linear regression lines for both dominant (DOM) and nondominant (nonDOM) limbs in each group (Fig 1). The relationship between %BW and power was not significantly different between the DOM and nonDOM limbs of the intact group, therefore, those limbs were combined into one dataset (n = 56). Power correlated significantly with %BW for all groups and both limbs within each group (p <0.001) (Fig. 1). The relationship between %BW and power for the intact group was significantly different from the sound limb but not the amputated limb of the CTA group.

4. Discussion
There is a strong, positive relationship between %BW and power output during cycling for intact cyclists and CTA. This relationship is not different between the amputated limb of CTA and the intact group indicating that the clinician may use the same linear model to predict %BW for both groups. The y-intercepts for all linear models were greater than zero and averaged 21.5% (Fig. 1). A non-zero %BW measurement when power output is low may be explained by inertial forces developed at the foot/pedal interface due solely to the reciprocal movement of the limbs. Orthogonal pedal forces remained below 30% body weight for power outputs similar to those selected by cyclists with lower limb impairment (20 – 75 watts) [8].

In conclusion, these findings suggest that cycling may serve as a rehabilitation modality, even within the confines of post-surgical weight bearing restrictions, to promote cardiovascular conditioning, range of motion, and strength gains for an
individual with lower limb loss. The models presented here may also be used to describe the lower limb weight bearing forces achieved by a patient prior to progression into balance or gait training activities.

Figure 1. Percent body weight for the shank vs. power output for both limbs combined in the intact group (solid line), amputated (dashed) and sound (grey) limb for the CTA group. Equations for the linear model are given below the respective group. The linear model for the intact group and the amputated limb of the CTA group was not significantly different.

The authors gratefully acknowledge Beth Brown and Kate McDonald for their review and edits of the manuscript, and Laura Jones for her help with data collection. We would also like to thank Ossur, Prosthetic Design Inc. and Serotta Bicycles for their donations.

References