EMG-DRIVEN INVERSE DYNAMIC ANALYSIS OF KNEE CONTACT FORCES DURING GAIT AND CYCLING USING OPENSIM

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INTRODUCTION

Joint contact forces determine the loading experienced by cartilage tissue and, thus, may be used to predict risk of cartilage tissue damage and osteoarthritis (OA). Participating in low impact and/or non-weight bearing activities such as cycling may help reduce knee OA risk by limiting forces exerted during exercise [1]. Cycling is a common recommendation for rehabilitative or fitness sustainment exercise for select patients [1]. Although knee joint contact forces have been directly measured in gait and cycling using instrumented knee implants [2,3] and calculated in gait using EMG-driven analysis [4]; they have not been calculated in cycling using EMG-driven analysis.

The long-term goal of this study is to identify weight control exercises for overweight (OW) and obese (OB) subjects that minimize OA risk. This current study tests the hypothesis that knee joint contact forces are significantly lower during cycling than gait. The objectives are to: (1) conduct motion analysis experiments and EMG-driven OpenSim analyses for gait and cycling, (2) compare predicted tibiofemoral (TF) contact forces to published values, and (3) test for significant differences in maximum TF compressive forces in gait and cycling.

METHODS

Equipment. Kinematic data was collected using a 10-camera motion analysis system with Cortex software (Motion Analysis Corp., Santa Rosa, CA, USA). Gait experiments were conducted using 3 ground force plates (Accugait, AMTI, Watertown, MA, USA). Cycling experiments used a stationary bike (Lifecycle GX, Life Fitness, Schiller Park, IL, USA) retrofitted with custom pedals containing 6-axis load cells (AMTI, Watertown, MA, USA). EMG data was collected using 4 wireless EMG sensors (Trigno, Delsys, Natick, MA, USA).

Experimental Studies. Six subjects (average body mass index (BMI) = 25.0) aged 18-26 years and with no previous knee injury history participated. Protocols were approved by Cal Poly's Human Subjects Committee and were designed to minimize risk to human subjects. EMG sensors were placed on the following dominant leg muscles: lateral gastrocnemius, vastus lateralis, vastus medialis and semimembranosus. An enhanced Helen Hayes marker set with 32 retroreflective markers was used to track kinematics. First, the subjects performed 10 gait trials, 5 analyzing each leg, at self-selected walking speeds. Subsequent analyses used the middle three trials of the dominant leg. Then, each subject performed 3 cycling trials at a moderate machine resistance level (10) and 70 RPM. A static trial was captured at the end of the experiments for determining reference knee angles and scaling procedures in OpenSim (Stanford University, Palo Alto, CA, USA). Finally, body mass, height and Q-angle measurements were recorded.

Data Processing and Analysis. Trials were processed using Cortex to identify markers and create virtual markers to generate vectors of body segments. Kinematic and kinetic data were filtered (4th order Butterworth filter, cutoff frequency = 6 Hz) and exported to Matlab (MathWorks, Natick, MA, USA). EMG data were also exported to Matlab and time synced with the kinematic data. A bandpass filter of 20Hz to 450Hz [5] was applied to the time synced EMG data to create a Control Constraints file for muscle activation input to OpenSim [4].

OpenSim analyses proceeded with the following steps. (1) A full body musculoskeletal model was scaled to each subject [6]. (2) The Inverse Kinematics (IK) and Residual Reduction Algorithm (RRA) tools were used to output joint kinetic data and a corrected kinematic file. (3) An EMG-driven Computed Muscle Control (CMC) analysis was used to obtain muscle activations. (4) CMC results were used in Joint Reaction (JR) analysis to calculate knee joint contact forces. Results were normalized by body weight (BW) and trimmed to one full gait cycle of the dominant leg (0% = 1st heel strike, 100% = 2nd heel strike), and one full crank revolution (0 deg. = 1st top dead center, 360 deg. = 2nd top dead center).

Statistics. A paired t-test was used to compare the knee joint compressive contact forces for the two types of exercises (gait vs. cycling). Significance levels were defined by $p = 0.05$.

RESULTS

Results for TF compressive forces were more similar to publications for gait [4] than for cycling [2, 3]. For gait, average TF compressive contact forces (Fig. 1) closely resembled published results when subjects walked at 1.25 m/s [4]; the 6 subjects of this current study had an average walking speed of 1.32 m/s. For cycling, average TF compressive contact forces (Fig. 1) resembled published results [3], which reported knee forces during cycling at select power outputs. In particular, in [3] the reported maximum TF contact force at a power of 25 W was 0.6xBW, whereas in this study the maximum TF compressive contact force at an average power of 26 W was 0.75xBW. Also for cycling, our maximum TF contact forces were 27% lower compared to those reported in [2]. Peak compressive forces experienced during gait were 2.9xBW while for cycling they were 0.75xBW, a difference that was found to be significant (Fig. 2).

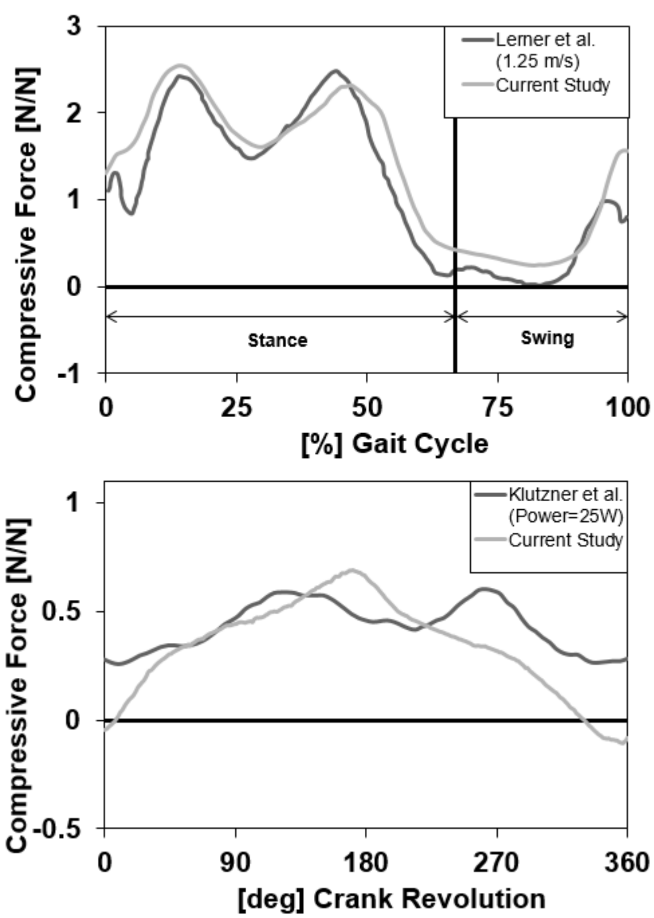

Fig. 1: Tibiofemoral joint compressive contact forces normalized by BW and averaged over 6 subjects during one full gait cycle and one full crank revolution.

DISCUSSION

These findings reinforce the hypothesis that cycling results in relatively low TF joint compressive contact forces. A novel finding of this study is that OpenSim may be used to calculate knee contact forces during cycling as the results generally agreed with published results.

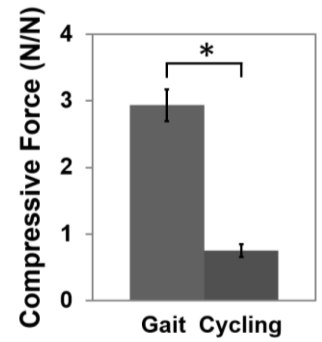

Fig. 2: Maximum TF compressive force for gait and cycling (n=6). Mean ±1 S.D. shown. *Significant difference (p<0.0001).

The difference in the graphs for compressive forces during cycling, observed in Fig. 1, might be due to the difference in setup of the subject on the stationary bike, the type of stationary bike, and the pedal design (i.e. clipped vs strapped). For this study, an upright stationary bike was used and the seat height was positioned such that at 180^o crank angle the subject's leg was almost straight. It should be noted that the published results in Fig 1. were recorded at a cadence of 40rpm while our subjects kept a 70rpm constant cadence. Also, the subjects used in [4] were older and were suffering from OA.

It is emphasized that in OpenSim's RRA, a pelvic residual is introduced for dynamic consistency. For gait, that pelvic residual is minimized to a value close to zero, but for cycling the pelvic residual is minimized to a relatively large value that should represent the effect of seat and handlebar forces. A future study should directly measure the seat and handlebar forces and, thus, provide an experimental target for the pelvic residual.

This study calculated TF joint contact forces for cycling and gait non-invasively using an EMG-driven inverse dynamics OpenSim analysis. The results suggest that cycling may be the preferred exercise for limiting OA risk in populations at high risk for knee OA. In order to better identify exercises that minimize the OA risk, future studies may include other potential weight control and fitness sustainment exercises and populations that are at high risk for knee OA such as obese, amputee, and ACL reconstructive surgery subjects.

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REFERENCES

[1] McLeod, W et al., *Am J Sports Med*, 8:175-180, 1980.
[2] D'Lima, D et al., *Clin Orth Rel Res* 469:2953-2970, 2011.
[3] Kutzner, I et al., *J Orth Sports Phys Ther*. 42:1032–1038, 2012.
[4] Lerner, Z. F et al., *Gait & Posture*. 39:978–84, 2014.
[5] De Luca, C et al., *J Biomech*, 43:1573-1579, 2010.
[6] Anderson F et al., *J Biomech Eng*, 123:381-390, 2001.