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Biomechanical joint loading mechanism of Tai Chi gait in individuals with knee osteoarthritis: A pilot simulation study

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Abstract

Objectives: Tai Chi (TC) has shown beneficial effects on joint function in knee osteoarthritis (OA). Biomechanical mechanisms of knee joint contact load (JCL) and muscle activations during TC are less understood. The purpose of this biomechanical simulation study was to examine JCL of TC gait, the most common used TC from and its causal interactions with muscle activations in knee OA.

Design: Six knee OA and five healthy participants were recruited. Their full body motion of TC gait was collected. The JCL and muscle forces were quantified using a musculoskeletal simulation approach based on collected kinematics and kinetics. The JCL and muscle activations were compared between knee OA and healthy control group. In addition, the muscle contributions to the JCL were determined and compared between the two groups.

Results: Knee OA subjects had lower peak anterior-posterior shear forces and higher lateral shear forces than healthy control subjects during TC gait. Knee OA subjects also showed higher activations of knee flexor muscles than control subjects. Both knee extensor and flexors of the knee OA group were contributing to JCL and in the control group mainly the knee extensors.

Conclusions: Our simulation results showed the JCL, muscle forces profiles, and muscle contributions to the JCL during TC gait in knee OA. The findings of this study provided a direct scientific link between JCL and muscle forces during TC gait in knee OA. This would allow us to develop more effective TC interventions for knee OA in the future.

Keywords

Tai chi gait; Walking; Knee joint contact force; Knee osteoarthritis; Simulation

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Credit author statement

Dr. Liu contributes to the conception and design of the study and takes responsibility for the integrity of the work, from inception to finished article. Authors Dr. Roos, Mituniewicz, and Liu take responsibility of the integrity of the data and the accuracy of the data analysis. Study concept and design: Liu. Acquisition of data: Liu. Computer simulation design: Roos. Analysis and interpretation of data: Mituniewicz, Roos and Liu. Drafting of the manuscript: Mituniewicz, Roos and Liu. Critical revision of the manuscript of important intellectual content: Mituniewicz, Roos and Liu. Statistical analysis: Mituniewicz.

Declaration of competing interest

The authors have no conflicts.

1. Introduction

Knee osteoarthritis (OA) is a leading cause of adult disability in the United States [1]. Knee OA is associated with impairments at the body structure level that directly affect the medial compartment articular cartilage of the knee with a heterogeneous local mechanical response [2]. Knee OA is characterized by consistent pain [3], joint stiffness [4], and swelling [5]. Dysfunction resulting from knee OA affects daily activities such as walking and can adversely affect independent functioning and quality of life.

Tai Chi (TC) is aimed at improving balance, muscle strength and flexibility, and would therefore appear to be a particularly well-founded intervention for people with knee OA. The most common TC practice is the simplified yang style 24 forms [6] consisting of a series of sequential movements, these movements incorporate uninterrupted, slow, and rhythmic components using gait phases such as single support, double support, and swing but with what is believed to be relatively low loading and unloading patterns. TC has shown beneficial promise in modifying knee OA pain symptoms. However, other studies have not supported the claim that standardized TC may be an effective therapeutic option for knee OA [7,8]. The main factor accounting for such inconsistent findings is that individual TC forms are arbitrarily selected among previous studies with unknown mechanism of joint contact load (JCL) acting on the knee and muscle contributions to the JCL during TC.

The presence of JCL and its magnitude have mainly been reported in studies evaluating walking [9]. As biomechanics of TC among individuals with knee OA remains completely unexplored, quantifying the biomechanical characteristics of TC would aid in our understanding of the biomechanical mechanism used in TC, provide direct evidence of the neuromuscular and musculoskeletal insights of the knee during TC, and lead to the development of more effective TC treatment in knee OA. Therefore, the purpose of this pilot study is to quantify JCL profiles and muscle contribution to the JCL during TC gait, the most common TC form [6]. By doing so, our findings will fill the knowledge gap of understanding biomechanical mechanism of TC in knee OA.

2. Methods

Six knee OA and five healthy control participants were recruited for participation in the study: (OA: 1 male and 5 females, age = 62.5 years, height = 1.67 m, weight = 83.1 kg; control: 1 male and 4 females, age = 52.4 years, height 1.69 m, weight = 66.5 kg). Knee OA participants met the American College of Rheumatology criteria for symptomatic knee OA. Knee OA participants with a history of neurological and cardiovascular conditions were excluded. Healthy control participants without a known history of musculoskeletal, neurological diseases, orthopedic and cardiovascular conditions were enrolled. Prior to participation, each subject signed an informed consent document approved by the Institutional Review Board.

The TC course in this study was delivered in 3 sessions per week for 2 weeks. The TC lessons were delivered on a one-on-one basis between the TC instructor and each participant. Participants were required to perform TC gait independently before each session. Biomechanical measurements including the full-body kinematics and kinetics, were measured while performing the TC gait. All participants performed five trials of TC gait. Full-body kinematics were collected from 37 reflective markers placed on the body's bony landmarks using a motion capture system (Vicon, UK) sampling at 100 Hz[10]. Ground reaction force (GRF) data were recorded synchronously with the kinematic data at a sampling frequency of 1,000 Hz using two force plates (AMTI, MA).

A custom MATLAB (The Mathworks Inc., MA) script was used to rotate the marker and GRF data into the CoBiDyn (CFD Research Corporation, AL) [11], as well as to convert the marker data from the.c3d file format to the.trc file format. The Hamner lower extremity model [12] was scaled for each subject using the marker data from the subject's static calibration trial and markers were adjusted automatically. After scaling and marker placement was complete, CoBiDyn was used to run inverse kinematics (IK). The average root mean square marker error during IK was no greater than 2.5 cm for all trials of all subjects. The IK motion was then filtered using a 12 Hz low-pass recursive Butterworth filter. The GRF data was filtered using a 20 Hz low-pass recursive Butterworth filter. A threshold value of 30 N in vertical GRF was used to determine when a foot was in contact with a force plate. Both kinematics and GRF motion files were trimmed to begin 0.2s before the right foot stepped on the force plate and end 0.2s afterwards. All OA subjects had OA in their right knee. Musculoskeletal simulations were then performed in CoBiDyn [11].

Simulation data were analyzed using custom written analysis in GNU Octave 5.1.0. The peak JCL, normalized to subject body mass, occurring in the right knee were determined for each trial. The simulated muscle activations and muscle contributions [13] from the simulations were examined at the times of peak JCL in four directions. These times include peak loads in the anterior-posterior direction ($JCL_{Anterior}$), the superior-inferior direction ($JCL_{Compressive}$), lateral direction ($_{JCLLateral}$) and the medial direction (JCL_{Medial}). If data violated the normal distribution, a logarithm transform was executed. Independent t-tests were performed in R (R Core Team, Austria) to determine significance between the groups. The results of the t-tests were deemed significant if the p-value was less than 0.05.

3. Results

3.1. Joint contact load (JCL)

The OA group's average right knee peak $JCL_{Anterior}$ was significantly less that the control group's ($p = 0.005$), while their average peak $JCL_{Lateral}$ was significantly greater ($p < 0.001$) (Fig. 1).

3.2. Muscle activations

At the time of the peak $JCL_{Anterior}$ (Fig. 2a), the OA group showed significantly greater muscle activation than the control group for the following muscles: the biceps femoris long head (BFLH; $p = 0.002$), lateral gastrocnemius (LG; $p = 0.016$), medial gastrocnemius (MG; $p = 0.022$) and semimembranosus (SM; $p = 0.001$) muscles. Activation of the rectus femoris (RF; $p = 0.003$) was significantly lower in the OA group compared to the control group at this time. When the peak $JCL_{Compressive}$occurred (Fig. 2b), the OA group showed significantly higher activation than the control group in the biceps femoris short head (BFSH; p = 0.001), LG (p = 0.010), MG (p < 0.001), sartorius (SAR; = 0.002), and SM (p = 0.015). Activation in the vastus intermedius (VI), vastus lateralis (VL) and vastus medialis (VM) were significantly lower in the OA group compared to the control group at this time point (all p-values < 0.001). The activation at the time of the peak $JCL_{Lateral}$ (Fig. 2c) was similar to that of the compressive force, where the BFSH (p = 0.011), LG (p = 0.015), MG (p = 0.002), and SAR (p = 0.004) were all significantly greater in the OA group than the control group. While the VI (p = 0.003), VL (p = 0.002), and VM (p = 0.002) activations were all significantly larger in the control group at this point. At peak JCL_{Medial}, the OA group showed significantly greater activation in the BFLH (p < 0.001) and the SM (p < 0.001) and significantly less activation than the control group in the RF (p = 0.009) (Fig. 2d).

3.3. Muscle contributions

At the peak $JCL_{Anterior}$ (Fig. 2e), the BFLH and SM contributed significantly more to force anterior shear force production in the OA group than the control group (p < 0.001). The LG and MG contributed significantly more for the OA group as well (p < 0.001), but in the posterior shear direction. The RF (p = 0.002) contributed significantly less at peak $JCL_{Anterior}$ in the OA group than the control group. The significance of the muscle contributions at the time of the peak $JCL_{Compressive}$ (Fig. 2f) was similar to the activation where the BFSH, LG, MG, SAR, and SM were all significantly greater in the OA group (p-values for all but SM < 0.001, SM p = 0.001). The VI, VL, and VM all contributed significantly more in the control group than the OA group at this time (p < 0.001). The only significantly different flexor muscle at peak $JCL_{Lateral}$ was the BFSH (p = 0.015), while the VI, VL, and VM all showed significant differences (p < 0.001) (Fig. 2g). At peak JCL_{Medial}, the SM (p < 0.001) produced a significantly greater force contribution in the lateral direction for the OA group (Fig. 2h).

4. Discussion

To our best knowledge, this is the first study to directly explore the relationship between JCL and muscle contributions during TC gait in knee OA. Results of peak JCLs (Fig. 1) for both the OA and control group were comparable to those reported for walking [9,14]. This simulation study showed that OA subjects had lower peak anterior-posterior shear forces and higher lateral shear forces at their affected knee joint than healthy control subjects (Fig. 1). The lower anterior-posterior shear forces in the OA group could be due to the smaller peak knee flexion at the time of the peak JCLs. At the time of peak $JCL_{Anterior}$, the control subjects mainly activated their knee extensor muscles (RF, VI, VL, and VM), while the OA subjects also showed activation of knee flexor muscles (BFLH, MG, LG, and SM) (Fig. 2a). This is likely due to a difference in kinematics between the OA and Control groups, possibly due to knee joint related pain or fear of pain during TC gait. The simulations did not account for potential muscle co-contraction as a result from pain in OA subjects. As expected, the muscle contributions to the $JCL_{Anterior}$ show a similar pattern to the muscle activations (Fig. 2e), with both knee extensors and flexors of the OA group contributing to the shear force and mainly the extensors in the control group. This may explain the control group's larger shear force; activation of mainly extensor muscles will pull on the tibia in an anterior direction relative to the femur. While in the OA subjects, the flexor muscle forces counter this pulling force, reducing the overall force. It was also noted that the control group showed significantly higher muscle activations and contributions of RF at the time of $JCL_{Anterior}$ (Fig. 2e). This muscle is typically weakened in OA subjects and this has been linked to an increased risk of falls [3,10].

At peak $JCL_{Lateral}$, both the OA and control groups showed similar activation patterns to the time of peak $JCL_{Compressive}$. The OA group showed significantly higher activation of flexor muscles than the control group (BFSH, LG, MG, SAR, and SM), while the control group activated their vastus muscles (VI, VL, VM) significantly more (Fig. 2b and c). These differences are mirrored in muscle force contributions (Fig. 2f and g) at peak $JCL_{Compressive}$ and $JCL_{Lateral}$, except by the flexors at peak $JCL_{Lateral}$ where only the BFSH showed a significantly greater contribution in the OA group than the control group. Potentially, this may be because the posture taken by the OA group positioning the vastus muscles in a way that they could not affect frontal plane knee loading like in the control group. At peak $JCL_{Lateral}$, the control group's muscle contributions were in both the medial and lateral directions, possibly lowering the overall contact force. While in the OA subjects the contributing percentages were mostly in the medial direction or did not make a meaningful contribution to the peak $JCL_{Lateral}$ (Fig. 2g). At peak JCL_{Medial}, the OA group showed knee flexor muscles (BFLH, SM) had higher activations than control group (Fig. 2d). Interestingly, the SM contributed to lateral shear force at JCL_{Medial}. This result suggests that during TC gait, knee flexor muscles may play an important role in reducing the medial compartment knee joint load (Fig. 2h).

Our simulation study has limitations. The relatively small sample size of our study may lead to a large variation in simulation outcome measurements, which would affect our findings. Individuals with knee OA are known to have higher levels of muscle co-contraction resulting in increased joint forces. Our simulations were not based on electromyography driven approach, thus any co-contraction observed during simulations was a direct result of the experimental kinematics and kinetics. All limitations warrant further investigations with a larger sample size.

This study demonstrated a simulation framework that can provide insights at the joint and muscle level during TC motions that cannot be uncovered using experimental data alone. While TC has been demonstrated to be beneficial to OA patients, it is not fully understood why and by what mechanisms TC benefits OA patients. A specified knee model including anatomical knee joint regions over the medial and lateral condyles is needed to further understand these differences that were shown in the frontal plane between OA and control subjects.

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Significance and innovations

- This study aimed to quantify the knee joint contact load of Tai Chi (TC) gait in knee OA. This would provide insight into our understanding of the biomechanical mechanism of TC.
- Our results provided direct evidence of the muscle contributions to the knee joint contact load of TC gait and demonstrated causal inference between knee joint contact load and muscle forces during TC gait in knee OA.
- Our simulation study established a solid scientific foundation for future efforts to determine the most effective TC forms for knee OA rehabilitation.

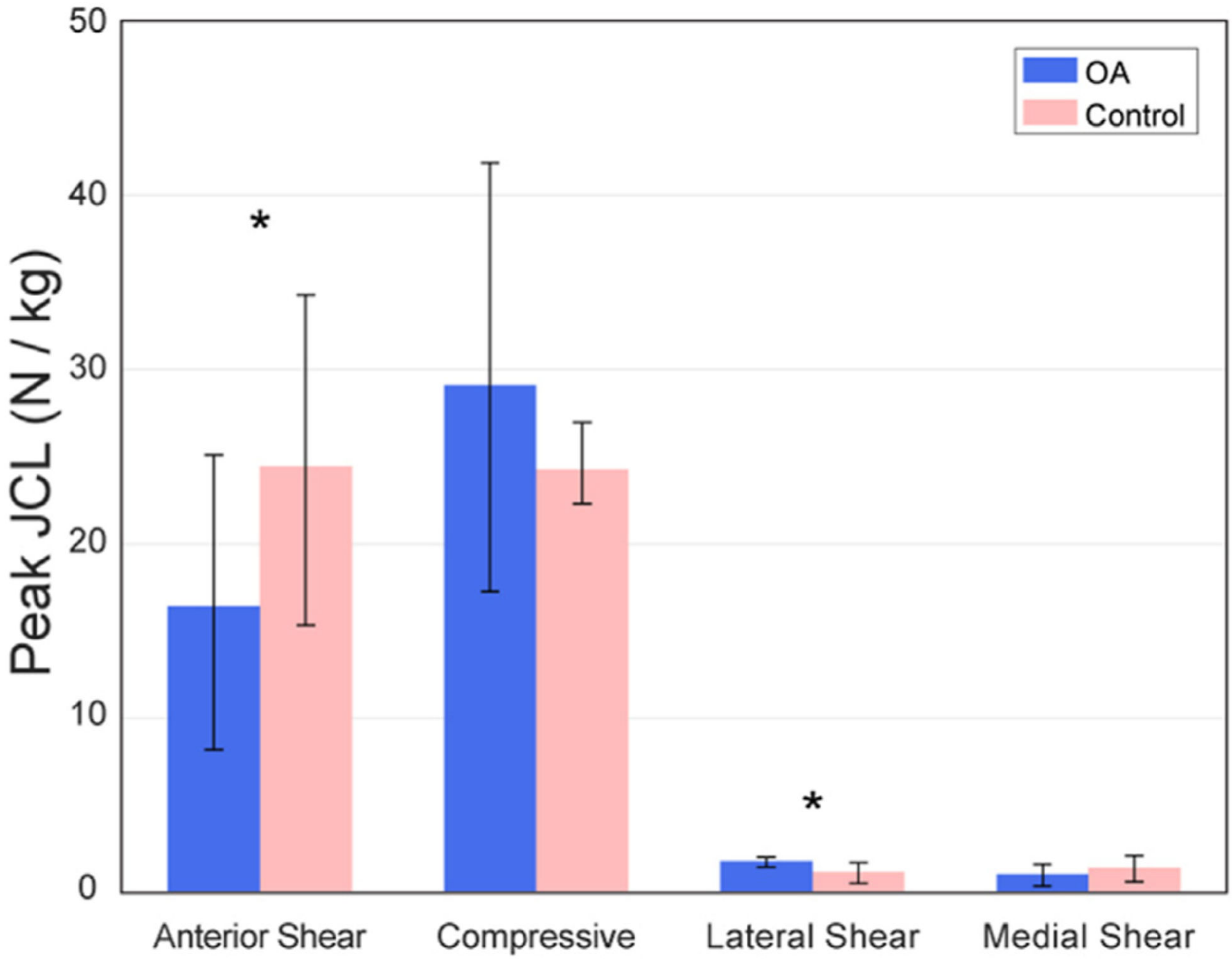

Fig. 1.

Comparisons of peak $JCL_{Anterior}$, $JCL_{Compressive}$, $JCL_{Lateral}$ and JCL_{Medial} between OA and control group during TC gait. The JCL is normalized to the body mass. *Significant difference between OA and Control ($P < 0.05$).

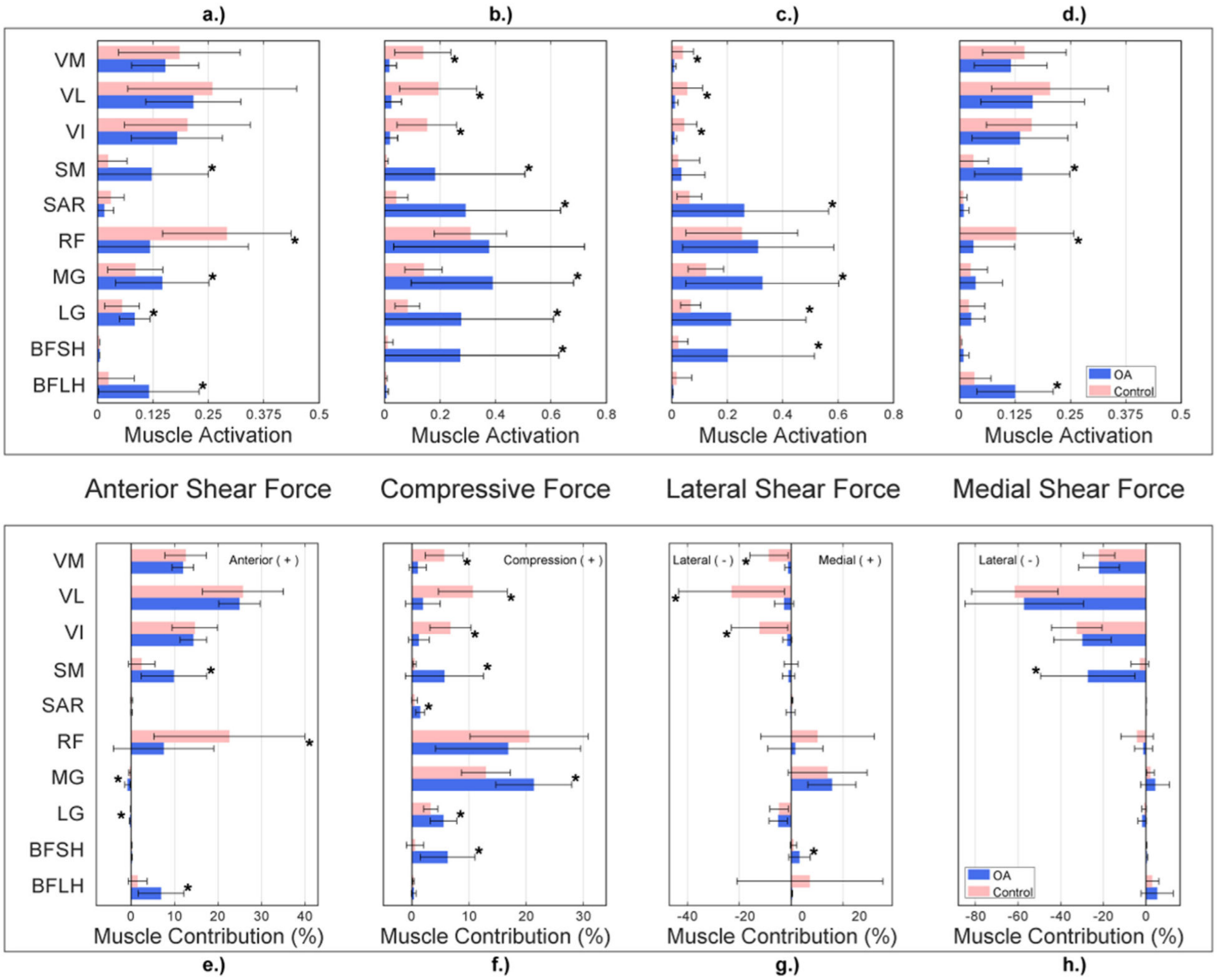

Fig. 2.

Comparisons of muscle activations between OA and control group at peak $JCL_{Anterior}$ (a), $JCL_{Compressive}$, (b), $JCL_{Lateral}$ (c) and JCL_{Medial} (d) during TC gait; Comparisons of muscle contributions to the JCL between OA and control group at peak $JCL_{Anterior}$ (e), $JCL_{Compressive}$, (f), $JCL_{Lateral}$ (g) and JCL_{Medial} (h) during TC gait. *Significant difference between OA and Control ($P < 0.05$).