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Effect of foot progression angle adjustment on the knee adduction moment and knee joint contact force in runners with and without knee osteoarthritis

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ABSTRACT

Background: Knee adduction moment (KAM) is often used as a surrogate marker of knee contact force (KCF) during walking. Previous studies have reported potential benefits to reduce KAM in patients with knee osteoarthritis (OA) by foot progression angle adjustment. However, KAM is an external moment and it does not consider any muscle contribution to the joint loading, which should pose a greater influence in running than walking.

Research question: This study used a computational model to compare KAM and KCF between runners with and without knee OA during running. In addition, we evaluated the KAM and KCF when runners adjusted to an out-toe running style.

Methods: Kinematic, kinetic, and lower limb EMG data were collected from 9 runners with knee OA and 10 healthy counterparts. They were asked to run at their usual speed with standard shoes on an instrumented treadmill.

Results: We found no significant difference in the KAM during running between OA and the healthy group ($p > 0.376$). However, runners with knee OA exhibited a greater total KCF than the healthy counterparts ($p < 0.041$). We did not observe any reduction in KAM after foot progression angle adjustment ($p > 0.346$). Surprisingly, an increase in the longitudinal KCF and total KCF were found with adjustment of foot progression angle ($p < 0.046$).

Significance: Unlike the findings reported by the previous walking trials, our findings do not support the notion that foot progression angle adjustment would lead to a lower joint loading during running.

1. Introduction

Running is a popular sport and the number of runners is increasing substantially every year [1]. Running has been reported to be effective in weight control, improving exercise tolerance, and reducing the risks of developing cardiovascular diseases [2,3]. However, approximately 50% of runners of the Americans have experienced running-related injuries annually [4]. Knee joint is the most commonly injured site and knee osteoarthritis (OA) is one of the most prevalent musculoskeletal conditions in runners [5]. Since high exposure to sport is associated with increased risk of OA [6], researchers are hence interested in investigating whether running will lead to the development of knee OA [7]. It has been postulated that repeated submaximal loading may lead to excessive cartilage wearing at the tibiofemoral joint and in turn cause knee OA in runners [8–10]. However, a recent review by Timmins et al. [11] failed to observe such association between running and knee OA. Instead of a single factor problem, the pathology of knee OA in runners is believed to be multifactorial [12].

Biomechanically, knee adduction moment (KAM) is often used as a surrogate marker of the knee joint loading during gait [13]. KAM has been correlated with pain [14], severity [15], extensiveness of cartilage damage [16], and disease progression in patients with knee OA [17]. In view of the relationship between KAM and knee OA, attempts to lower KAM have been suggested. These interventions include knee brace [18], foot orthotics [19], special footwear design [20], and gait pattern modification [8,21–23], such as adjustment of the foot progression angle [24]. Although previous gait retraining studies reported successful reduction in KAM [23,24], most experiments adopted a walking

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protocol and therefore its clinical efficacy for runners with knee OA remains unknown.

More importantly, a recent study by Meireles et al. [25] reported the limitations of using KAM alone to predict joint contact force of the tibiofemoral joint in patients with early knee OA. Such inaccurate estimation can be due to the fact that KAM is an external joint moment which only considers external forces i.e. ground reaction force during gait, but not internal forces e.g. muscle forces. Compared to walking, the tibiofemoral joint sustains a much greater loading resulting from both internal and external forces during running [26]. Therefore, the estimation of joint contact force during running would be more inaccurate than during walking.

Researchers have used alternative method to measure tibiofemoral joint contact force (KCF) instead of using KAM alone. Previous studies have directly measured the KCF through instrumented prosthetic implants [27,28]. However, this method is invasive, expensive and not feasible to measure the KCF in runners. For this reason, a computational EMG-driven model has been developed to estimate the KCF [26,29,30]. The advantage of a computational EMG-driven model is that it can estimate the KCF without invasive instrumentation and muscular contribution is incorporated. It allows a reasonable estimation of KCF and can be used as an outcome measure to evaluate the effectiveness of an intervention.

At our best knowledge, there is limited scientific research scrutinizing KCF in runners with and without early knee OA. Hence, the objective of this study was two-fold. Firstly, we estimated the KCF by a computational EMG-driven model and compared both KAM and KCF between runners with and without early knee OA during running. Secondly, we evaluated the KAM and KCF when the runners adjusted their foot progression angle. We hypothesized that runners with knee OA would present greater KAM and KCF during running than their healthy counterparts. Theoretically, with a greater foot progression angle, the center of pressure shifts laterally and hence a reduction in the lever arm of KAM would be expected [31]. Based on the previous findings [8,32], we also expected that both KAM and KCF of runners would be reduced by increasing the foot progression angle to a more toe-out running gait.

2. Methods

2.1. Subjects

Nine regular runners (> 5 km/week for more than 12 months) with early knee OA in the medial compartment (OA group), which was confirmed with radiological findings (i.e. Kellgren & Lawrence grade I–II) by orthopedic surgeons, were recruited from local running clubs. Another 10 characteristics-matched healthy counterparts (healthy group), who did not have any knee injury and symptoms, were enlisted. The study protocol was reviewed and approved by the institutional review board. All the subjects provided a written consent before being tested.

2.2. Procedures

Reflective markers at specific bony landmarks were firmly affixed on the subjects according to an established model [33]. The movement trajectories of the markers were captured with an 8-camera motion capture system (T series, Vicon, Oxford UK). In order to have better estimation of the KCF [25], wireless surface EMG electrodes (Trigno, Delsys Inc., Boston, MA, USA) were also attached onto bilateral rectus femoris, vastus lateralis, vastus medialis, medial hamstrings, lateral hamstrings, medial gastrocnemius, and lateral gastrocnemius [35]. Similar to a previous study, we assumed that the short head and long head biceps femoris showed identical EMG pattern [34]. Similar assumption was also applied between semimembranosus and semitendinosus. The vastus intermedius EMG activity was defined as the mean value from the vastus lateralis and vastus medialis [34,35]. All the subjects were asked to run on an instrumented treadmill (Tandem force treadmill, AMTI, Watertown, MA, USA) at their usual running speed with standard shoes (Gelfeather Glide 3, Asics, Kobe, Japan) for five minutes [36], with and without foot progression angle adjustment, i.e. increase the foot progression angle to an out-toe running style [16]. The test sequence was randomized by an online program (www.random.org).

2.3. Data processing and computational modeling

Synchronized EMG, kinematics, and kinetics data were collected from the last minute of the running trial at 1000 Hz, 200 Hz, and 1000 Hz respectively. Kinematics and kinetics data were filtered using a fourth-order, phase-corrected, Butterworth, lowpass filter at 8 Hz and 50 Hz respectively [37,38]. Joint moments were expressed as external moments, referenced about the proximal end of distal segments. The peak value of KAM in each step was extracted and then normalized with body mass and height across all the footfalls in each trial. The raw EMG data was firstly band-pass filtered (10–450 Hz), full-wave rectified and then low-pass filtered at the cutoff frequency of 6 Hz [37]. The resulting linear envelopes were normalized to the peak processed EMG value of each muscle obtained from the entire dataset during the trials.

A generic musculoskeletal model [33] in the OpenSim platform (Version 3.3, National Center for Simulation in Rehabilitation Research, Stanford, CA, USA) was scaled to accommodate each subject's anthropometry. Virtual makers of the knee joint center were created based on the experimental makers recorded from the static standing trial and inverse kinematic algorithm was utilized to obtain joint angles. The joint moment that tracked the joint kinematics were calculated using inverse dynamics and residual reduction analyses.

The experimental EMG signals and the joint moments were then input to calibrate an EMG-driven model [39]. The goal of calibration was to account for subject-specific muscle physiology and dynamics in the modeling, which was shown to increase the accuracy of muscle force prediction [35]. The excitation patterns of muscles that could not be measured by surface EMG were constructed using optimization algorithms [35]. Muscle parameters were therefore refined by tracking of the experimental joint moment and that derived from EMG-driven musculotendon units. Since tibiofemoral contact force is the focus in this study, calibration was implemented with respect to the degree of freedom of the knee joint. After calibration, the modified model was applied for dynamic analysis. KCF was calculated as the sum of joint reaction force and muscle force that spanning the knee joint. The direction of each selected muscle force was obtained using an OpenSim plugin [40]. In addition, we also calculate the joint contact forces in the medial and lateral compartment according to a validated EMG-driven model suggested by Winby et al. [30].

2.4. Statistics

Shapiro-Wilk tests were used to check the normality of the data. If the data was normally distributed, independent *t*-tests (continuous variables) and the Chi-square tests (nominal variables) were used to compare the baseline characteristics of the runners with and without knee OA. Two (knee OA vs. healthy) by two (with vs. without foot progression adjustment) ANOVA was used to examine the interaction effect on KAM and KCF during running. When applicable, independent *t*-test was used to compare the between-group difference of KAM and KCF, while paired *t*-test was used to compare the within-group difference of KAM and KCF. Wilcoxon Signed Ranks test and Mann-Whitney test were used if normality of the data was rejected. All statistical analyses were performed using SPSS version 20 (SPSS Inc., Chicago, IL, USA) with the global alpha at 0.05. In order not to over-rely on the p value interpretation, we also used Cohen's d to quantify the effect and the cut-off values for small, medium, and large effects were 0.2, 0.5 and