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Muscle contributions to knee extension in the early stance phase in patients with knee osteoarthritis

Shinya Ogaya[a,*], Ryo Kubota[b], Yuta Chujo[b], Eiko Hirooka[c], Kim Kwang-ho[b], Kimitaka Hase[b]

[a] Service Robotics Research Team, Robot Innovation Research Center, National Institute of Advanced Industrial Science and Technology, Japan
[b] Department of Physical Medicine and Rehabilitation, Kansai Medical University Hospital, Japan
[c] Department of Orthopaedic Surgery, Kansai Medical University Hospital, Japan

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ABSTRACT

The aim of this study was to analyze individual muscle contributions to knee angular acceleration using a musculoskeletal simulation analysis and evaluate knee extension mechanics in the early stance phase in patients with knee osteoarthritis (OA). The subjects comprised 15 patients with medial knee OA and 14 healthy elderly individuals. All participants underwent gait performance test using 8 infrared cameras and two force plates to measure the kinetic and kinematic data. The simulation was driven by 92 Hill-type muscle-tendon units of the lower extremities and a trunk with 23 ° of freedom. We analyzed each muscle contribution to knee angular acceleration in the 5%–15% and 15%–25% periods of the stance phase (% SP) using an induced acceleration analysis. We compared accelerations by individual muscles between the two groups using an analysis of covariance for controlling gait speed. Patients with knee OA had a significantly lesser knee extension acceleration by the vasti muscles and higher knee acceleration by hip adductors than those in controls in 5–15% SP. In addition, knee OA resulted in significantly lesser knee extension acceleration by the vasti muscles in 15–25% SP. These results indicate that patients with knee OA have decreased dependency on the vasti muscles to control knee movements during early stance phase. Hip adductor muscles, which mainly control mediolateral motion, partly compensate for the weak knee extension by the vasti muscles in patients with knee OA.

1. Introduction

Knee osteoarthritis (OA) is a major musculoskeletal disease that causes the decline of physical and locomotor function. The weakness of the quadriceps muscles is a common clinical sign associated with knee OA [1,2]. The quadriceps acts in the early stance phase and produces knee extension moment. The quadriceps and other muscles generate a force to support the body's center of mass against the downward acceleration of the body due to gravity. The strength of quadriceps is related to the knee extension moment in early stance phase during gait in patients with knee OA [3]. Therefore, knee extension moment generally decreases in these patients [4,5].

Patients with knee OA also have a kinematic feature of knee joint excursion in early stance. In the stance phase, the excursion of the knee angle in these patients differs from that in healthy individuals. In normal gait, the knee is gradually flexed after an initial heel contact and turns to extended during early and mid-stance phases [6]. In many patients with OA, however, the knee is slightly and continuously flexed during early stance phase [7], and there is no clear peak knee flexion angle. This kinematic feature of knee flexion in OA may influence knee extension moment, but it is unclear how quadriceps weakness directly related to the generation of knee extension acceleration in the gait of patients with knee OA.

The contraction of the quadriceps generates a knee extension moment; however, knee motion is not only controlled by muscles crossing the knee joint. A muscle that spans one joint has the potential to accelerate other joints. Analysis of forward dynamics simulations of walking that are driven by individual muscles can identify how each muscle contributes to knee angular accelerations [8]. For example, in studies on normal walking in healthy young individuals, the gluteus maximus strongly contributes to the control of the knee joint to generate intensive extension acceleration in early stance, similar to the quadriceps [9,10]. Another simulation study showed that the gluteus maximus and soleus have a similar role to that of the quadriceps in knee extension during normal walking [11].

A musculoskeletal simulation model can reveal the direct relationship between excursion of the knee joint and muscle work in the early stance phase. The application of this analysis to gait in patients with knee OA may clarify how they control their knee in the early stance phase, even with quadriceps weakness. Therefore, the objective of this

* Corresponding author at: Tsukuba Central 2, 1-1-1 Umezono, Tsukuba, Ibaraki 305-8568, Japan.
E-mail address: shinya.ogaya@aist.go.jp (S. Ogaya).

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study was to determine the individual muscle contributions to knee angular acceleration in the early stance phase in patients with knee OA. A musculoskeletal simulation analysis was used to evaluate the knee extension mechanics during gait in patients with knee OA. The muscle contribution to knee extension was obtained with adjustment for gait speed to control for unusual OA gait patterns, slight and continuous flexed knee motion.

2. Methods

2.1. Participants

Fifteen patients with medial knee OA and 14 healthy elderly control subjects were recruited from the community. All subjects provided written consents, and the study was approved by the institutional review board. Subjects with knee OA were included if they had radiographic changes with Kellgren-Lawrence (KL) grade of 2 or higher in the medial tibiofemoral compartment. Exclusion criteria included a history of other orthopedic injury in the lower extremities, neurological injury, rheumatoid arthritis, joint surgery in lower extremities, lateral knee OA, or use of an assistive device. If a patient had bilateral knee OA fitting the criteria, the more involved knee, as identified by the patient, was used for analysis. Control subjects were included if they reported no history of knee dysfunction or previous lower extremity injury.

2.2. Experimental data

Three-dimensional (3D) coordinates of reflective markers and ground reaction force were obtained during standing and gait using a 3D motion analysis system (Locus 3D MA-300, Anima, Chofu, Japan). This system consists of eight infrared cameras and two force plates (MG-1190, Anima, Chofu, Japan), each with a sampling rate of 100 Hz. Nine markers were attached to the skin of each subject at anatomical landmarks: acromion, C7, first metatarsal head, fifth metatarsal head, and heel. In addition, three markers were attached to a pelvis and two shank devices. Distances between these three markers were fixed because they were firmly fixed on the inflexible hard devices.

The devices were used for defining the positions of imaginary markers [12]. The shank device was used to define four imaginary markers on each leg, medial/lateral malleolus, and medial/lateral knee. This device, shaped like a leg guard, was fixed to the shank and firmly strapped by a hook and loop fastener. Before measurement, imaginary markers were defined using a pointing device connected to the 3D motion analysis system. The tip of the pointing device was located at a body landmark to set the imaginary marker, and the relative position of the imaginary marker was defined by the position of the three markers on the shank device. The pelvis device, shaped like a belt, was also used to define imaginary markers of the anterior and posterior superior iliac spines. In a preliminary trial, the root mean difference between imaginary and real marker positions attached at the same location was < 19.8 mm during gait measurement. The subjects were instructed to stand on the platform for capturing static standing data, and to walk twice on the platform at their preferred speed for capturing gait data. Gait data were low-pass filtered using a Butterworth filter with a cut-off frequency of 6 Hz. The walking speed was calculated by C7 position during 4 steps.

The center of the ankle joint was defined as the midpoint between the lateral and medial malleolus markers, and the center of the knee joint was defined as the midpoint between the medial and lateral knee. The center of the hip joint was calculated from the pelvis markers [13]. The Euler angle in each joint was calculated from adjacent joint center positions as recommended by the International Society of Biomechanics for definitions of a joint coordinate system [14]. The knee varus angle was measured from the static standing data.

2.3. Modeling and simulations

Subject-specific simulations were created using OpenSim software [15]. A simulation model was created based on a generic musculoskeletal model of lower extremities and trunk with 92 Hill-type muscle-tendon units [16]. The deformation of the knee joint was achieved with the assumption that patients with knee OA usually have varus knees. To include varus alignment in the analysis, a degree of knee adduction–abduction was added to the model. The subtalar and metatarsophalangeal joints were locked at 0°. Therefore, a model with 21 ° of freedom was scaled to match each subject's anthropometry using static standing data. The dimensions of each body segment in the model were scaled based on relative distances between pairs of markers obtained from a motion-capture system and the corresponding virtual marker locations in the model. The adduction angle of the model’s knee was carefully adjusted to match the captured knee varus angle.

Because the knee adduction–abduction motion during gait is very small despite a static alignment change in patients with knee OA, this motion was locked after the scaling process. The positions of model markers during gait were calculated to minimize the difference between experimental and model markers, thus simulating the experimental kinematic motion. Dynamic inconsistency between measured ground reaction force data and model kinematics was reduced using the residual reduction algorithm by applying residual forces and torques to the pelvis and adjusting the mass properties and kinematics of the model [17] (Supplementary Figs. S1, S2). A computed muscle control was used to estimate muscle excitation patterns to track the gait motion [18]. Because measurements were made on a walkway equipped with two force plates, the analysis was restricted to the double stance phase after heel contact and the single stance phase, which correspond to early and mid-stance phases, respectively.

2.4. Induced acceleration analysis

An induced acceleration analysis was used to compute the contributions of forces of individual muscles and total contributions of lower limb muscles to knee angular acceleration in the early stance phase [10]. A rolling without slipping constraint was used to model the foot–floor interaction in contact constraint equations [16]. The early stance was divided into two phases: 5–15% stance phase (% SP) and 15–25% SP. The average contribution of muscle forces to knee angular acceleration was calculated to assess the muscle contributions to knee acceleration.

2.5. Statistical analysis

Differences in demographic data, knee varus alignment, and gait speed between the groups were evaluated by *t*-test. Gender difference was examined by chi-square test. We integrated some muscles into groups: hip adductors (adductor brevis, adductor longus, adductor magnus, and pectineus); hamstrings (biceps femoris, semitendinosus, and semimembranosus); and vasti (vastus medialis, vastus intermedialis, and vastus lateralis). Analysis of covariance (ANCOVA) adjusted by gait speed was used to evaluate the differences in the contribution of averaged individual muscle forces to knee angular acceleration during 5–15% SP and 15–25% SP to assess data controlled for gait speed. Significance level was set at 0.05.

3. Results

There were no significant differences in age, height, weight and gender between the patients with knee OA and the control subjects (Table 1). The knee varus angle was significantly larger in patients with knee OA than that in control subjects (9.28 ± 7.60° vs 0.82 ± 3.34°, $p < 0.001$), and patients with knee OA walked significantly slower than control subjects (0.90 ± 0.16 vs 1.11 ± 0.21 m/s, $p < 0.001$).