Gait & Posture 40 (2014) 341–345

Contents lists available at ScienceDirect

Gait & Posture

journal homepage: www.elsevier.com/locate/gaitpost

Knee adduction moment and medial knee contact force during gait in older people

Shinya Ogaya [a,b,*], Hisashi Naito [b], Akira Iwata [a], Yumi Higuchi [a], Satoshi Fuchioka [a], Masao Tanaka [b]

[a] Division of Physical Therapy, Department of Comprehensive Rehabilitation, Osaka Prefecture University, 3-7-30 Habikino, Habikino-shi, Osaka 583-8555, Japan

[b] Division of Bioengineering, Department of Mechanical Science and Bioengineering, Graduate School of Engineering Science, Osaka University, 1-3 Machikaneyama, Toyonaka, Osaka 560-8531, Japan

ARTICLE INFO

Article history:
Received 19 September 2013
Received in revised form 11 April 2014
Accepted 27 April 2014

Keywords:
Adduction moment
Medial knee contact force
Gait
Simulation
Older people

ABSTRACT

External knee adduction moment has been studied as a surrogate for medial knee contact force. However, it is not known whether adduction moment is a rational measure for predicting medial knee contact force. The aim of this study was to investigate the correlation between knee adduction moment and medial knee contact force in older people, using musculo-skeletal simulation analysis. One hundred and twenty-two healthy older subjects participated in this study. Knee moment and medial knee contact force were calculated based on inverse dynamics analysis of normal walking. Muscle force and joint reaction force were used to determine the medial knee contact force during stance phase. The results showed that the maximum medial knee contact force was moderately correlated to the maximum knee adduction ($r = 0.59$) as well as the maximum extension moment ($r = 0.60$). The first peak of medial knee contact force had a significant strong correlation with the first peak of adduction moment and a moderate correlation with the maximum flexion moment. The second peak of medial knee contact force had a significant moderate correlation with both the second peak of adduction and the maximum extension moment. These results implied that the maximum adduction moment value could be used, to some extent, as a measure of the maximum medial knee contact force.

1. Introduction

Osteoarthritis (OA) is a major joint disease that restricts activities of daily living. The risk of developing OA increases with age. Adverse mechanical loading of the knee, such as repetitive and high-magnitude joint load, is believed to contribute to the development of knee OA [1]. Knee OA is developed predominantly in the medial compartment of the tibiofemoral joint [2]. Measurement of medial knee contact force, i.e., contact force on the medial compartment of the tibiofemoral joint, would be valuable for identifying the risk of knee OA development. However, thus far, it has been impossible to measure medial knee contact force in a noninvasive manner, and the number of studies on medial knee contact force measurement is limited [3–5].

It is considered that the maximum external knee adduction moment is positively correlated to the maximum medial knee contact force [6,7] because the external knee adduction moment acts as compressive force on the medial compartment of the knee joint. Indeed, the maximum value of external knee adduction moment during gait was found to be higher in OA patients [8], and external knee adduction moment can be regarded as a risk factor for OA progression [9]. In contrast, Walter et al. reported that a decrease in the maximum external knee adduction moment did not always guarantee a decrease of the maximum medial knee contact force [3]. This is because an increase in external knee flexion or extension moment leads to an increase in quadriceps or hamstring tension and, consequently, to an increase in the medial knee contact force even if the external knee adduction moment decreases. It is unknown whether the maximum external adduction moment could be a real measure to predict the maximum medial knee contact force.

Musculo-skeletal model-based computational analysis works as a powerful tool to investigate the medial knee contact force because it can be applied for calculating joint contact forces in

* Corresponding author at: Osaka Prefecture University, Department of Physical Therapy, Division of Comprehensive Rehabilitation, 3-7-30 Habikino, Habikino-shi, Osaka 583-8555, Japan. Tel.: +81 72 9502111; fax: +81 72 9502130.
E-mail address: ogaya@rehab.osakafu-u.ac.jp (S. Ogaya).

http://dx.doi.org/10.1016/j.gaitpost.2014.04.205
0966-6362/

noninvasive manner by kinematic and kinetic data. With the use of musculoskeletal model–based simulation analysis, some studies have shown that there is a correlation between the external knee adduction moment and the medial knee contact force [10,11]. In these studies, inverse dynamic analysis was performed to calculate the medial knee contact force using knee joint model with muscles and ligaments. These studies analyzed the gait data of young subjects, but no study involved older people.

Gait pattern with ageing are individual and could be characterized by short stride, wide base, slow speed, flexed knee and limited ankle dorsiflexion angle, all of which are usually depended on decreased physical functions or malalignment [12]. Because of such large individual differences, a large number of gait data must be analyzed to investigate the relationship between the external knee adduction moment and the medial knee contact force in older people. Thus, the aim of this study was to investigate the correlation between the external knee adduction moment and the medial knee contact force during gait for a large number of older people using musculo-skeletal model-based simulation analysis.

2. Methods

2.1. Measurement protocols

Community-dwelling 122 older people (31 male and 91 female) participated in this study. The subjects were 73.8 ± 6.3 years old, 153.9 ± 7.5 cm in height, and 51.1 ± 7.4 kg in weight. They had (1) no cognitive impairment insufficient to understand instructions, (2) no serious neuro-muscular impairment insufficiently to prevent measurements, and (3) no difficulty in ambulating independently and thus no need for daily life assistance. Written informed consent was obtained from each subject. The study was approved by the Ethical Review Board of Osaka Prefecture University.

Three-dimensional motion analyses and lower extremity muscle activity measurements were performed during standing and gait conditions. In the assessment for the standing condition, the subjects were instructed to stand on a force plate in a relaxed posture with both legs slightly apart, and this posture was captured for 1 s. For gait assessment, the subjects were instructed to walk as usual at their normal speeds while passing over the force plate. The trials in which the subjects failed to record a complete step on the force plate were rejected.

2.2. Kinetic and kinematic data

Three-dimensional coordinates of reflective markers and ground reaction force were measured in standing and in gait using a VICON MX motion analysis system (Vicon, Oxford, UK), which is a three dimensional motion capture system. This system consists of six infrared cameras with a sampling rate of 100 Hz and two force plates (Kistler, Switzerland) with a sampling rate of 1000 Hz. Seventeen markers were attached to each subject skin at anatomical landmarks: anterior superior iliac spines, sacrum, greater trochanters, medial and lateral knees, medial and lateral ankles, heels and second metatarsal heads.

Markers were processed using Vicon Nexus (Vicon, Oxford, UK). Low-pass Butterworth filters were applied to the data of reflective marker coordinates and ground reaction force with cutoff frequencies of 5 and 15 Hz, respectively. The center of the ankle joint was set as the midpoint between the lateral and medial malleolus markers, and the center of knee joint was defined as the midpoint between the medial and lateral knee markers. The center of the hip joint was calculated from the pelvis markers [13]. Joint angles were calculated from the positions of the adjacent joints' centers following the International Society of Biomechanics' recommendations for definitions joint coordinate systems [14]. The repeatability of knee joint angle was examined in the preliminary test. Trials assessing seven subjects were carried out in separate two days by the same investigator. Intraclass correlation coefficient of knee flexion angle at initial contact between days was 0.94. Each segment length was calculated from the captured data for standing, and the segment mass and moment of inertia values were calculated by referring to a previous report [15]. Inverse dynamic analysis was performed to calculate the joint moments and joint reaction forces at ankle, knee and hip joint from the data of segment motions and ground reaction forces using the Newton–Euler equation. The inverse dynamics programs are implemented in C language and run on a personal computer.

2.3. Musculo-skeletal model

A previously described musculo-skeletal model, of which the validity has been verified [16], was used in this study. This musculo-skeletal model consists of four segments, i.e., pelvis, thigh, shank, and foot. The ankle, knee, and hip joints have two, one, and three degrees of freedom, respectively. The model has 42 muscle tendon units, the anatomical position, optimal length, tendon slack length, and pennation angle were decided based on the data from Delp [17]. The segment lengths were used to scale the simulation model to the subject which included scaling of the anatomical position, optimal length and tendon slack length. The physiological cross-sectional area of each muscle was determined on the basis of Horsman's older people's muscle data [18]. Maximum muscle stress was extracted from the literature [19]. Hill's model [20] was used for calculating muscle force considering the effects of velocity and length in accordance with a previous study [21].

The net joint moments were decomposed into individual muscle forces by solving a minimization problem of the cubic sum of muscle activations at each sampling instance in stance phase [22,23].

The knee contact force was calculated as a point load acting on the tibial plateau. The moment vector $\vec{m}_{\text{muscle}}(j)$ generated by the jth muscle was expressed as follows:

$$\vec{m}_{\text{muscle}}(j) = \vec{f}_{\text{muscle}}(j) \times \vec{r}_{\text{muscle}}(j) j = 1, 2, \cdots, N \tag{1}$$

where $\vec{f}_{\text{muscle}}(j)$ and $\vec{r}_{\text{muscle}}(j)$ denote the muscle tension force vector and the moment arm vector corresponding to jth muscle, respectively, and N denotes the number of muscles crossing the knee joint. There were 13 muscles crossing knee joint in this model; rectus femoris, vastus medialis, vastus lateralis, vastus intermedialis, medial and lateral gastrocnemius, semimembranosus, semitendinosus, biceps femoris long and short head, tensor fascie latae, gracilis, and sartorius. Muscle tension force vector $\vec{f}_{\text{muscle}}(j)$ was decomposed into $F_{\text{muscle}}(j)$ (knee contact forces resulting from the muscle tension forces); parallel components along the long axis of the femur. The net knee contact force F_{net}, which is the sum of medial and lateral knee contact forces F_{m} and F_{l}, were calculated as the sum of the knee joint reaction force F_{ext} and the knee contact forces resulting from the muscle tension forces $F_{\text{muscle}}(j)$.

$$F_{\text{net}} = F_{\text{m}} + F_{\text{l}} = \sum_{j=1}^{N} F_{\text{muscle}}(j) + F_{\text{ext}} \tag{2}$$

That is, the equilibrium of the adduction/abduction moment of the knee joint is written as follows:

$$F_{\text{m}} d_{\text{m}} - F_{\text{l}} d_{\text{l}} = \sum_{j=1}^{N} M_{\text{muscle}}(j) + M_{\text{ext}} \tag{3}$$

where d_{m} and d_{l} denote the medio-lateral moment arm lengths, i.e., the distances from the center of the knee joint to the point of contact forces acting on the medial and lateral compartments,