Journal of Biomechanics 43 (2010) 2780–2784

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

Individual muscle contributions to the axial knee joint contact force during normal walking

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ARTICLE INFO

Article history:
Accepted 3 June 2010

Keywords:
Musculoskeletal model
Forward dynamics simulation
Joint loading
Biomechanics
Gait

ABSTRACT

Muscles are significant contributors to the high joint forces developed in the knee during human walking. Not only do muscles contribute to the knee joint forces by acting to compress the joint, but they also develop joint forces indirectly through their contributions to the ground reaction forces via dynamic coupling. Thus, muscles can have significant contributions to forces at joints they do not span. However, few studies have investigated how the major lower-limb muscles contribute to the knee joint contact forces during walking. The goal of this study was to use a muscle-actuated forward dynamics simulation of walking to identify how individual muscles contribute to the axial tibio-femoral joint force. The simulation results showed that the vastii muscles are the primary contributors to the axial joint force in early stance while the gastrocnemius is the primary contributor in late stance. The tibio-femoral joint force generated by these muscles was at times greater than the muscle forces themselves. Muscles that do not cross the knee joint (e.g., the gluteus maximus and soleus) also have significant contributions to the tibio-femoral joint force through their contributions to the ground reaction forces. Further, small changes in walking kinematics (e.g., knee flexion angle) can have a significant effect on the magnitude of the knee joint forces. Thus, altering walking mechanics and muscle coordination patterns to utilize muscle groups that perform the same biomechanical function, yet contribute less to the knee joint forces may be an effective way to reduce knee joint loading during walking.

1. Introduction

The human knee joint is subjected to significant loads during walking, with peak loads well-above body weight (e.g., Anderson and Pandy, 2001; D'Lima et al., 2007; Glitsch and Baumann, 1997; Heinlein et al., 2009; Kutzner et al., 2010; Taylor et al., 2004). The high joint loading is primarily due to muscle forces (e.g., Herzog et al., 2003). Studies analyzing muscle contributions to knee joint loads during walking have focused primarily on those muscles crossing the knee joint, and found that the quadriceps and gastrocnemius are the primary contributors during early and late stance, respectively (Kim et al., 2009; Lin et al., 2010; Morrison, 1970; Schipplein and Andriacchi, 1991; Shelburne et al., 2006). Typically, joint forces are determined using the vector sum of the intersegmental joint forces calculated using inverse dynamics analysis and the compressive forces from the muscles crossing the joint. However, this method does not account for individual muscle contributions to the ground reaction forces since only the net ground reaction force is used to determine the intersegmental joint forces. Because muscles can contribute to all joint forces (even those they do not span) through their contributions to the ground reaction forces via dynamic coupling (Zajac and Gordon, 1989), it is possible for muscles spanning a joint to generate greater joint forces than the forces developed in the muscles themselves. In addition, it is possible for muscles that do not span a joint to have greater contributions to the joint force than muscles spanning the joint due to their contributions to the ground reaction forces. For example, studies have shown that the gluteus maximus and soleus have large contributions to the ground reaction forces (Anderson and Pandy, 2003; Liu et al., 2006; Neptune et al., 2004), and therefore these muscles may have significant contributions to the knee joint force even though they do not anatomically cross the joint. However, few studies have investigated how the major lower-limb muscles contribute to the knee joint contact forces during walking. Understanding how individual muscles contribute to knee joint loading has important clinical implications for developing rehabilitation strategies that focus on specific muscle groups to help reduce knee joint loads for patients with osteoarthritis and other joint disorders (e.g., Fregly et al., 2007; Mundermann et al., 2004, 2008a).

The purpose of this study was to use a muscle-driven forward dynamics simulation of normal walking to identify individual muscle contributions to the tibio-femoral joint force.

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0021-9290/$ - see front matter © 2010 Elsevier Ltd. All rights reserved.
doi:10.1016/j.jbiomech.2010.06.011

Specifically, we examined the joint force component parallel to the longitudinal axis of the tibia (i.e., axial force), which is the dominant force component (D'Lima et al., 2007). We hypothesized that (1) muscles spanning the knee joint can generate greater tibio-femoral joint forces than the forces developed in the muscles themselves and (2) muscles that do not span the knee joint can have significant contributions to the tibio-femoral joint forces through their contributions to the ground reaction forces.

2. Methods

2.1. Musculoskeletal model

A musculoskeletal model (Fig. 1) was generated using SIMM (MusculoGraphics Inc., Santa Rosa, CA), which consisted of a trunk segment (head, torso and arms) and two legs (femur, tibia, patella, calcaneus, mid-foot and toe for each leg). The model had a total of thirteen degrees of freedom in the sagittal-plane (translations and rotation of the trunk, flexion–extension at the hip, knee, ankle, mid-foot and toe joints for both legs). The motion of the patella and the tibia relative to the femur were prescribed as functions of knee flexion (Yamaguchi and Zajac, 1989). The model was driven by 25 Hill-type muscle actuators per leg, with activation–deactivation dynamics governed by a first-order differential equation (Raasch et al., 1997). The muscles were grouped into thirteen functional groups, with muscles within each group receiving the same excitation pattern (Fig. 1). Muscle electromyography (EMG) data (see Section 2.3) were used to define the muscle excitation patterns. Block patterns were used for muscles where EMG data were not available. Passive torques representing the forces applied by ligaments, passive tissue and joint structures were applied at the hip, knee and ankle joints (Davy and Audu, 1987). The passive torques for the mid-foot and toe joints were defined using the following equation:

$T = k(\text{joint angle}) + b(\text{joint angular velocity})$

where joint angle was defined as the angular displacement from the neutral anatomical position expressed in radians, constants k, b were (750, 0.05) for the mid-foot joint, and (25, 0.03) for the toe joint expressed in N m and N m s, respectively. Thirty-one visco-elastic elements were attached to each foot segment to model the foot–ground contact (Neptune et al., 2000).

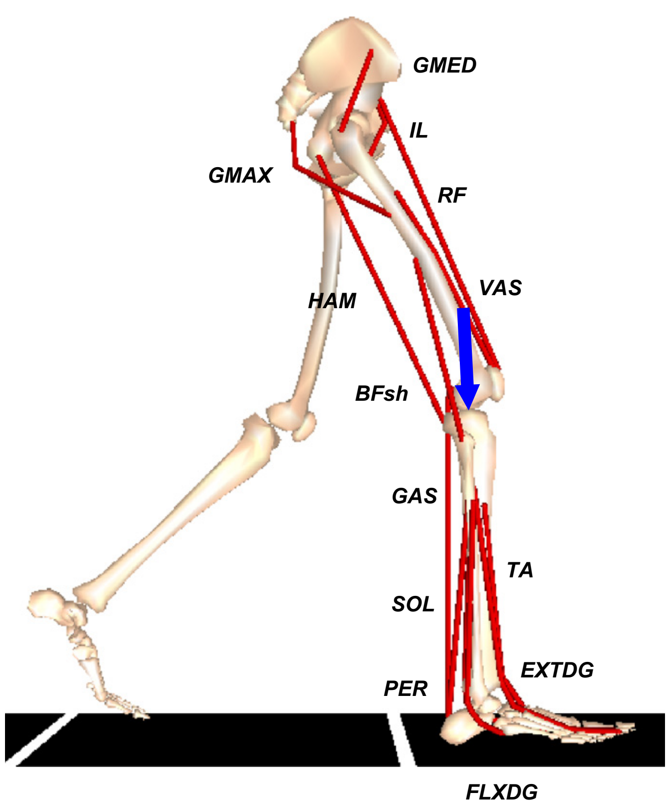

Fig. 1. The musculoskeletal model consisted of a trunk segment (head, torso and arms) and two legs (femur, tibia, patella, calcaneus, mid-foot and toe for each leg). The muscles included in the model were the GMAX (gluteus maximus, adductor magnus), GMED (anterior and posterior portions of gluteus medius), IL (psoas, iliacus), VAS (three vastii muscles), RF (rectus femoris), HAM (medial hamstrings, biceps femoris long head), BFsh (biceps femoris short head), SOL (soleus, tibialis posterior), GAS (medial and lateral gastrocnemius), TA (tibialis anterior, peroneus tertius), PER (peroneus longus, peroneus brevis), EXTDG (extensor digitorum longus, extensor hallucis longus), FLXDG (flexor digitorum longus, flexor hallucis longus). The axial knee joint force (arrow) is the force component parallel to the longitudinal axis of the tibia.

2.2. Forward dynamics simulation of walking

A forward dynamics simulation of walking was generated using Dynamics Pipeline (MusculoGraphics, Inc., Santa Rosa, CA) and SD/FAST (PTC, Needham, MA). Muscle excitation patterns were fine-tuned using dynamic optimization (e.g., Neptune and Hull, 1998), where the differences in kinematics and ground reaction forces between experimental and simulation data were minimized over a full gait cycle (from right heel-strike to the subsequent right-heel strike). In the optimization, a simulated annealing algorithm was used to minimize the following cost function:

$$J = \sum_i \sum_m w_{i,m} \frac{(Y_{i,m} - \hat{Y}_{i,m})^2}{\mathrm{SD}^2{}_{i,m}} \tag{1}$$

where $w_{i,m}$ is the weighting factor for variable m, $Y_{i,m}$ is the experimental measurement of variable m, $\hat{Y}_{i,m}$ is the simulation data corresponding to $Y_{i,m}$ and $\mathrm{SD}_{i,m}$ is the standard deviation of experimental variable m at time step i. The excitation timing for each muscle was constrained based on the EMG data to ensure that the muscles generated force at the appropriate time in the gait cycle.

2.3. Experimental data

Previously collected experimental kinematic, ground reaction force and EMG data (Neptune and Sasaki, 2005) were used. Briefly, ten able-bodied subjects (5 males and 5 females; age 29.6 ± 6.1 years old, height 169.7 ± 10.9 cm, body mass 65.6 ± 10.7 kg) walked on a split-belt instrumented treadmill (TecMachine, France) at 1.2 m/s while data were collected for 15 sec. Kinematic data collected at 120 Hz (Motion Analysis Corp., Santa Rosa, CA) using a modified Helen Hays marker set were digitally low-pass filtered at 6 Hz. Ground reaction force data were collected at 480 Hz and low-pass filtered at 20 Hz. Surface bi-polar EMG data (Noraxon, Scottsdale, AZ) were collected at 1200 Hz from the soleus, tibialis anterior, medial gastrocnemius, vastus medialis, rectus femoris, biceps femoris long head and gluteus maximus. The EMG signals were band-pass filtered (20–400 Hz), fully rectified and then low-pass filtered at 10 Hz to generate linear envelope signals. All digital filters were fourth-order zero-lag Butterworth filters. All data were normalized to the gait cycle, averaged across steps and then across subjects to obtain group-averaged data.

2.4. Muscle contributions to the tibio-femoral joint force

The axial tibio-femoral joint force was computed as the component of joint contact force parallel to the longitudinal axis of the tibia, which includes all forces acting on the joint (i.e., intersegmental joint forces and muscle compressive forces). Individual muscle contributions to the axial joint force were obtained at each time step in the simulation by (1) performing a ground reaction force decomposition to determine individual muscle contributions to the ground reaction forces (Neptune et al., 2001), (2) applying only the muscle force of interest and corresponding ground reaction forces to the system and (3) solving the equations of motion to determine the axial tibio-femoral joint contact force. This process was repeated for each muscle at each time step over the entire gait cycle.

3. Results

The simulation emulated well the experimentally measured sagittal-plane walking kinematics and ground reaction forces (Fig. 2). The mean absolute errors over the gait cycle for the hip, knee and ankle angles, and vertical and horizontal ground reaction forces were 3.9°, 4.7° and 3.2°, and 5.6% and 2.7% body weight (BW), respectively. In addition, the resulting muscle excitation patterns matched closely with the experimentally measured EMG patterns (Fig. 3).

The axial tibio-femoral joint force had two major peaks during the stance phase (Fig. 4). The first peak occurred in early stance (Fig. 4: ~15% gait cycle), reaching a magnitude of ~2100 N