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Technical note

Calculation of muscle forces during normal gait under consideration of femoral bending moments

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ABSTRACT

This paper introduces a new approach for computing lower extremity muscle forces by incorporating equations that consider "bone structure" and "prevention of bending by load reduction" into existing optimization algorithms.

Lower extremity muscle and joint forces, during normal gait, were calculated and compared using two different optimization approaches. We added constraint equations that prevent femoral bending loads to an existing approach that considers "minimal total muscular force". Gait parameters such as kinematics, ground reaction forces, and surface electromyographic activation patterns were examined using standardized gait analysis. A subject-specific anatomic model of the lower extremities, obtained from magnetic resonance images of a healthy male, was used for the simulations. Finite element analysis was used to calculate femoral loads.

The conventional method of calculating muscle forces leads to higher rates of femoral bending and structural stress than the new approach. Adding equations with structural subject-specific parameters in our new approach resulted in reduced femoral stress patterns.

These findings show that our new approach improves the accuracy of femoral stress and strain simulations. Structural overloads caused by bending can be avoided during inverse calculation of muscle forces.

1. Introduction

Structural simulations of femoral stress patterns supported by experimental tests indicate bending at the femoral diaphysis [1–3] that may lead to structural overload [4,5]. Compression provokes negative deformations on the medial side of the *Femur* whereas tension induces positive deformation laterally [6,7]. Pauwels [8] noticed possible overloads due to bending, and proposed four mechanisms to reduce bending forces, achieving "economy of material". According to Wolff [9], muscle and joint forces play an especially important role in forming bone shape. Roux [10] postulated a minimum amount of bone material to withstand external loads.

Several studies demonstrated a correlation between muscle function (loads) and bone mass (shape), addressed as "muscle-bone unit" [11]. Interdependencies within the "muscle-bone-unit" were described in professional racket ball players [12]: the playing arm showed greater bone mass and cross sectional areas than the nonplaying arm. Patients with pathologies causing immobilization of lower extremities experience loss of bone mass on the impaired side [13]. How the muscle forces influence changes in bone mass is not understood. In vivo strain measurements [14] indicate that femoral stress and strain values applied in current simulation models are generally too high.

Accurate simulations of femoral strain patterns and bone modeling require detailed, subject-specific data on joint and muscle forces, as well as information about mechanical properties of bones [15,16]. Inverse optimization permits the calculation of muscle and tendon forces, obviating the need to measure single muscle forces in vivo [17]. Conventional methods (CM) mainly consider given joint moments. To ensure equilibrium, CM compute

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muscle forces based on minima of total force, fatigue or metabolic requirements [18–21].

Only Krieg [22] and Sverdlova et al. [23] considered a possible reduction of bending in their calculations. Krieg demonstrated the applicability of bending reduction as a criterion in a nonlinear object function, however, he does not provide detailed information about the applied methods and subject-specific data. Sverdlova et al. [23] employed an algorithm without providing details.

Aim of the present study was to introduce a new approach (NA) for calculation of muscle forces that minimize femoral bending. We hypothesize that calculating muscle forces with NA improves the quality of femoral stress patterns or force-induced bone transformations around hip prostheses. We compared forces calculated with CM and NA with in vivo data.

2. Methods

2.1. Gait analysis method

Kinematic data were collected by motion capture system (VICON® Motion Systems, Oxford, UK) operating with 200 fps. A previously described protocol [24] was applied to determine joint centers and gait parameters.

An AMTI® force plate (Advanced Mechanical Technology Inc., Watertown, MA, USA) was used to collect ground reaction forces at 1000 Hz. Muscle activities of five lower limb muscles (left leg) were recorded by 16-channel surface electromyography (EMG) (myon AG, Schwarzenberg, Switzerland) and processed via software ProEMG (v1.3.07, prophysics, Kloten, Switzerland). Following skin preparation, one pair of bipolar Ag/AgCl electrodes (Ambu Blue Sensor N) was placed according to SENIAM conventions [25] above *M. gluteus medius* (GLMe), *M. vastus lateralis* (VL), *M. vastus medialis* (VM), *M. rectus femoris* (RF), and *M. semitendinosus* (SE). Raw EMG signals (1000 Hz) were zero lag band pass filtered (Butterworth, 40 Hz–400 Hz), full wave rectified, and smoothed using a root mean square filter (window size 50 ms) [26]. Data were time normalized (% of gait cycle duration). To determine individual muscle activities, an ensemble average mean (EAM) of 18 gait cycles was calculated for each muscle [27]. Amplitude threshold for each muscle was set as the mean 3 SD of the “most quiet” 50 ms period of each EAM [28–30]. A muscle was “on” if the corresponding EAM exceeded the threshold for 50 ms.

The subject was thoroughly familiarized with the study design before giving written informed consent to participate in this study, as approved by the local medical ethics committee and in accordance with the 1964 Helsinki Declaration and its later amendments. We focus our method-oriented technical report on one healthy subject (29 years, male, body height 175 cm, body mass 68.6 kg). He walked barefoot at a self-selected speed on a 15 m level walkway. Subject-specific maximal muscle ($F_{Muscle,ACSA}$) and tendon force ($F_{Tendon,ACSA}$) per anatomical cross-sectional area (ACSA) were calculated based on an adapted mechanical model by Kummer [31]. Force and motion measurements during hopping on one leg [32] provided input data. Additional data of the mechanical model and the applied muscles and tendon forces within the optimization are available online.

2.2. Magnetic resonance tomography measurements

The subject was scanned using a 1.5-T Philips ®Achieva® magnetic resonance (MR) scanner (Royal Philips, Amsterdam, NL). In order to superimpose the body model after reconstruction from MR images onto data from gait analysis, the subject was fitted with fluid capsules during the MR scan positioned in the same locations as the external markers used in the gait tests. The field matrix of 528×528 pixel with a pixel size of 0.757 mm^2 defined a window sufficient to capture the lower extremities and the pelvis. To reduce scan time, 5 mm increments were chosen for scans performed from the foot sole to 20 mm above *Trochanter minor*. Scans were performed with an increment of 3 mm. Total scan time was approximately 45 min.

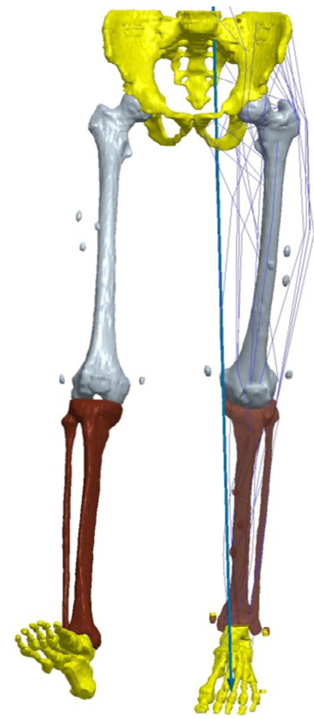

Fig. 1. Subject in gait lab in position 3 for the left leg.

2.3. Musculoskeletal body model

Considering the potential drawbacks of generic body models [33], we created a subject-specific musculoskeletal model of the volunteer’s lower extremities. Based on the MR scan, the geometrical parameters of our subject’s bone anatomy and ACSAs were reconstructed using Mimics® V14.1 (Materialise, Leuven, Belgium), and assembled in Pro/Engineer® (ProE) Wildfire™ 4 (PTC, Inc., Needham, MA, USA). Since the positioning procedure of the computer model is not automated, only three positions within the gait cycle where chosen. Based on the classification of normal gait [34], muscle forces were calculated for three events: position 1 - beginning of mid stance (12.6% of gait cycle), position 2 - end of mid stance (27.5%), and position 3 - end of terminal stance (46.4%).

First, the marker positions from the gait analysis were imported into ProE for the first event. Second, the body model was imported and the extremities were positioned in such a way that the gait lab markers and the segmented fluid capsules were superimposed with a maximal discrepancy allowance of 15 mm. The computer model muscle lines of action were oriented along the center of each corresponding ACSA. Fig. 1 shows the body model for the left leg in position 3.

Coordinate systems (CS) of the body model where chosen according to Witte et al. [35]. Due to variable activation pattern of posterior, middle, and anterior parts the ACSAs of GLMe [36], GLMe and *M. gluteus minimus* (GLMi) were subdivided into three parts each.

2.4. Finite element model

Finite element (FE) simulations were used to determine whether inversely calculated muscle forces provide physiologically