Geometrical and functional parameters of human lower extremity muscles obtained by ADAMS simulation

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Abstract

Background. Several authors used computer simulation to obtain biomechanical data of the human hip musculature, but there are few three-dimensional studies including all hip muscles. This study was conducted to obtain data on muscle length, muscle function and lever arms of hip and knee muscles from a representative model of the human lower extremity.

Methods. ADAMS was used to model a three-dimensional human body representing the 50th percentile of the male population. Muscle attachments of 31 muscles and muscle parts were measured in previous studies, and muscles were represented by straight lines. Muscle lengths were requested, and torque components relative to flexion, abduction and rotation axes were calculated. Lever arms and arm components relative to the axes were computed by dividing torque magnitude and torque components by force magnitude and force components relative to the motion axes.

Results. In this paper, results of the piriformis muscle are presented graphically in order to demonstrate methods and some applications of the data.

Discussion. Muscle functions assessed by length changes and relative torque components demonstrate good agreement with functions previously published; muscle lengths and lever arms could only be compared with few publications and showed reasonable qualitative agreement. Application of the data in biomechanical engineering and physical therapy is discussed.

Introduction

Although several biomechanical computer simulations have been published (e. g. Delp et al. 1990, van Soest et al. 1992), only a few studies are based on multibody programmes (McGuan et al. 1994).

The capability of a muscle to generate torque is determined by the force generated by the actuator and the lever arm. To assess the muscle function, length changes and torque components relative to the motion axes can be used. All these parameters depend on musculoskeletal geometry. However, in some cases a static approach is more appropriate to calculate these values in ADAMS.

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Material and methods

Computer model

Using the ADAMS/Android database, a computer model of the human body was developed representing the 50^{th} percentile of the male adult population. The knee joint was modeled as a revolute joint despite its more complex kinematics; however, this behaviour can be modeled by a curve-to-curve constraint if necessary.

The hip joint can be described by a spherical joint. As spherical joints could not be controlled by MOTION functions in the ADAMS/Solver version employed here (7.0), the hip was represented by three superimposed revolute joints connecting pelvis and thigh via two massless parts; the joint axes were perpendicular to each other in neutral position and intersected in the hip joint centre (i.e. centre of femoral head).

Two lower extremities of cadaver specimens of dimensions similar to the ADAMS model were dissected to measure muscle attachments of 31 muscles and muscle parts crossing the hip and the knee: Circumferences of muscle origins and insertions were measured and centres of attachment areas were calculated; the data were scaled to fit the geometry of the ADAMS/Android model (Lengsfeld et al. 1993, 1994a, 1994b).

The muscles were represented by straight lines connecting origin and insertion points. Where the curve of the muscle belly differs clearly from a line in living subjects, additional wrapping points were measured so that these muscles were comprised of two lines.

With the aid of this model and ADAMS/Solver, several parameters can be calculated.

The model was represented graphically by lines and frustums. Muscles were displayed by outline statements.

Preprocessor software

A preprocessor computer programme was developed using Turbo Pascal (Version 6.0). The programme loads a complete ADAMS data set into memory; muscles and joints which are to be incorporated into the model can be easily selected and parameters including MOTION and SFORCE functions can be edited. After input of simulation time and number of steps, an ADM file containing the new parameters is created along with an ADAMS command file and a batch file which makes possible the automatic execution of analysis.

Motion axes

Anatomical motion axes may differ from the global coordinate system. For example, abduction without flexion is performed within the frontal plane; abduction at 90° flexion is performed in a horizontal plane. Therefore, lever arm and relative torque components had to be calculated with respect to markers which changed their orientation according to hip joint position. Z-axes of j-markers defining the three revolute joints of the hip were used.

Joint positions

The range of motion of the hip joint was taken from the anatomical literature (e. g. Fick 1911, v. Lanz and Wachsmuth 1972, Gray and Clemente 1985, Fick et al. 1987, Schiebler and Schmidt 1991).

The model was moved using MOTION functions; simulation time and step size were adjusted to achieve the range of motion. As range of motion comprises angles in both directions from neutral position, negative angles were subtracted. The joint positions were set in steps of 10°.

Extension / flexion:

MOTION/, rotation, function=10D * time - 20D

Adduction / abduction:

MOTION/, rotation, function=10D * time - 30D

Internal / external rotation:

MOTION/, rotation, function=10D * time - 40D

Range of motion depends on hip joint position; abduction and rotation range of motion as a function of hip flexion angle was estimated on the basis of data published in the literature (e. g. v. Lanz and Wachsmuth 1972).

Combined motions were controlled by an ADAMS command file. (A) Rotation was performed by MOTION functions; after each rotation movement, (B) the abduction angle was increased or decreased by 20° until the maximum or minimum abduction angle was reached. This sequence was performed (C) at all flexion angles from 20° extension up to 140° flexion. Abduction and flexion angles were directly selected by motion functions. As direct changes of joint angles beyond 60° are not possible in ADAMS, direction of movement was changed after each step (A) or (B):

Different directions of rotation were simulated by using two different motion equations which were applied alternately:

function=20D * (time - rotation_start_time) - 40D (increasing angle)

function=20D * (rotation_end_time - time) - 40D (decreasing angle)

Muscle lengths

Displacement of the origin marker with respect to the insertion marker was calculated; x, y and z components of this vector are included in the tabular output file written out by ADAMS/Solver.

REQUEST/ , displacement, i=origin, j=insertion

, comment=displacement of ... muscle

Physiological ranges of motion around the three axes of the hip joint were combined and muscle length of each actuator was requested. A computer programme was developed which determined minimum and maximum muscle lengths and the hip joint position, respectively.

Lever arms

As position of the muscles and of the underlying bones changes during motion and differs between the actuators, a static approach was used.

A muscle force of 1 N acting between origin and insertion markers was assumed and generated by a SFORCE statement:

SFORCE/, i=origin, j=insertion, translational, function=1

The lever arm was calculated by dividing the torque at the hip joint by the muscle force (1 N):

$L = T / F$

The lever arm components were calculated likewise; torque components about the motion axes had to be divided by the force component acting perpendicularly on the axis:

F_{xy} = sqrt(fx(origin, insertion, joint_j_marker)**2 + fy(origin, insertion, joint_j_marker)**2))

The lever arm components are calculated by

$L = T_z + F_{xy}$

Relative torques

Torques generated by a muscle depend on

1. muscle force and
2. lever arm

Muscle force is a function of muscle length and is also determined by other properties of a muscle, e.g. cross-sectional area and pennation angle (Kaufman et al. 1991). As these properties are not considered here, torque components were normalized by dividing them by the torque magnitude acting at the hip joint:

T_{rel} = tz(joint_j_marker, joint_i_marker, joint_j_marker) / tm (joint_j_marker, joint_i_marker)

Graphical presentation of results

ADAMS tabular output files were edited in order to match the Microsoft Excel text file conventions. The files were imported into Microsoft Excel 5.0 (Microsoft 1994), and charts were generated.

Results

As most of the results are more important for anatomy, biomechanics and physical therapy than for engineering, the results of only one muscle will be presented here. Important conclusions resulting from the model analysis are discussed below. Piriformis muscle was chosen due to its many functional abilities and its clinical importance. Details can be seen from the charts shown in Appendix A.

The piriformis muscle lengthens during hip flexion and is shortened by abduction and by external rotation. The resting length of the muscle is 121 mm. When hip joint inplane motions are combined, the muscle vector reaches its maximum of 162 mm at 60° flexion, 60° adduction and 40° internal rotation and becomes minimum at 60° flexion, 90° abduction and 40° internal rotation. The lever arm of M. piriformis is 53 mm in neutral position; flexion, abduction and external rotation components are -19 mm, 25 mm and 52 mm.

The lever arm shortens during flexion, is lengthened by abduction and by external rotation until 30°.

Discussion

Model

At first it has to be stated that the model presented here can not be compared with the sophisticated multi-body dynamical models which are employed in engineering.

However, the objectives of this work can be reached successfully with the methods presented here.

In contrast to other investigations conducted previously (e. g. Seireg and Arvikar 1973, Jensen and Davy 1975, Dostal and Andrews 1981, Brand et al. 1982, Nemeth and Ohlsen 1985, White et al. 1989), the model presented here is based on a representative investigation of anthropometrical data. However, muscle attachment points have been measured on two cadaver specimens only, but the error has been minimized by choosing appropriate specimens and by a scaling procedure.

The limitations of representing muscle lines-of-action by straight lines must always be considered. Some authors have shown that representation by centroid lines (Jensen and Davy 1975) should be more accurate (Santaguida and McGill 1995). As centroid lines differ with respect to joint position, each muscle has to be measured at all possible joint configurations (Kaufman et al. 1991). When many actuators have to be incorporated in a model for motion analyses, this data collection is extremely time-consuming. Considering the relatively small differences in force calculations (1% - 12%) when the straight line approach and centroid line approach are compared (Jensen and Davy 1975), the straight line model seemed to be acceptable for the objectives of this study.

In general, muscle lengths are smaller in the model than in vivo because a straight line is defined to be the shortest distance between two points; lever arms are usually smaller in living subjects than the values measured in the model (Santaguida and McGill 1995).

Muscle lengths

There were only few publications which could be used in order to be compared to the muscle lengths presented here. Although many papers deal with muscle lengths, many authors measured the length of muscle tissue without tendons or tendons only, so that these data can not be compared with our results (Wickiewicz et al. 1983, Spoor et al. 1989, Friederich and Brand 1990). In other cases, regression equations for muscle length as a function of the hip flexion angle were published, but some essential values for calculation were not precisely given in the text (Visser et al. 1990, Hawkins and Hull 1990).

Comparison of the graphical presentation of results in several works (e. g. Hawkins 1992, Samuelsen et al. 1993) with the results of this study showed good qualitative agreement; however, as model size was different in all of these studies, validity of our data could not be completely checked.

Muscle functions

Muscle functions can be assessed from two parameters obtained in the simulation presented here:

1. Length changes caused by motion; only the function in the analysed plane can be assessed.
2. Relative torques; torque components acting about motion axes of the hip joint provide valuable data on muscle functions.

The functions obtained by these two parameters show reasonable agreement and were compared to muscle functions described in several anatomical textbooks (v. Lanz and Wachsmuth 1972, Gray and Clemente 1985, Kapandji 1985, Fick et al. 1987; Rauber et al. 1987, Schiebler and Schmidt 1991). The muscle functions calculated with the ADAMS model provide in all cases a larger data base than published data in the reviewed literature. The publications agree on the main functions of all muscles. In most of the cases one or more of the authors acknowledge results obtained by the ADAMS model. Function changes of several muscles during inplane motions agree in most of the muscles with the data published in the literature (e. g. Kapandji 1985). The difference in hip joint position where the change of function occurs compared to the literature usually lies within the range of 30°. Good agreement with published muscle functions show that the ADAMS model is capable to represent the human lower extremity muscles.

Lever arms

Similar to muscle lengths, only a few publications were suitable for comparison of the data. The lever arms computed here demonstrated good qualitative agreement with Hawkins (1992).

Application

This study provides data on biomechanical properties which depend on musculoskeletal geometry.

All biomechanical data calculated could serve as an input to other studies. Lengths and relative torque components are valuable in further assessment of muscle functions.

The hip positions where maximum or minimum length occurs are important for physical therapy; maximum length positions suggest improvement in stretching techniques, whereas minimum length positions can be used to relax injured muscles. Exact joint positions to achieve muscle stretch are not available yet even in scientific publications (e. g. Evjenth and Hamberg 1991); hence, this work could contribute to stretching techniques when single muscles have to be specifically treated.

Outlook

In contrast to many other biomechanical simulations, the ADAMS model can be easily linked to other mechanical devices, e. g. cars.

Evaluation of different hip joint prosthesis designs is planned; with further improvement in computer processing of imaging data, mechanical analysis of individual biomechanics could make mechanically customized hip joint implants possible.

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