Published in final edited form as:

Gait Posture. 2013 September ; 38(4): 1051–1053. doi:10.1016/j.gaitpost.2013.03.014.

Minimum detectable change for knee joint contact force estimates using an EMG-driven model

Emily S. Gardinier^{a,b,*}, Kurt Manal^{a,b,c}, Thomas S. Buchanan^{a,b,c}, and Lynn Snyder-Mackler^{a,b,d}

^aBiomechanics and Movement Science Program, University of Delaware, Newark, DE, United States

^bDelaware Rehabilitation Institute, University of Delaware, Newark, DE, United States

^cDepartment of Mechanical Engineering, University of Delaware, Newark, DE, United States

^dDepartment of Physical Therapy, University of Delaware, Newark, DE, United States

Abstract

Adequate test–retest reliability of model estimates is a necessary precursor to examining treatment effects or longitudinal changes in individuals.

Purpose—The purpose of this study was to establish thresholds for minimal detectable change (MDC) for joint contact forces obtained using a patient specific EMG-driven musculoskeletal model of the knee.

Design—A sample of young, active individuals was selected for this study, and subjects were tested on 2 separate days. Three-dimensional motion analysis with electromyography (EMG) was used to obtain data from each subject during gait for model input. An EMG-driven modeling approach was used to estimate joint contact forces at each session.

Results—MDC's for contact force variables ranged from 0.30 to 0.66 BW. The lowest MDC was for peak medial compartment force (0.30 BW) and the highest was for peak tibiofemoral contact force (0.66 BW). Test–retest reliability coefficients were also reported for comparison with previous work.

Conclusions—Using the present model, changes in joint contact forces between baseline and subsequent measurements that are greater than these MDCs are greater than typical day-to-day variation and can be identified as real change.

Keywords

Contact force; Electromyography; Musculoskeletal model; Minimal detectable change; Reliability

1. Introduction

Clinical applications of musculoskeletal models are broad because they allow investigation of constructs that are not feasibly measured *in vivo*. The most common use in movement science has been in estimating muscle and joint loads in healthy and patient populations [1]

© 2013 Elsevier B.V. All rights reserved.

*Corresponding author at: 301 McKinly Laboratory, Newark, DE 19716, United States. Tel.: +1 302 831 3613. smack@udel.edu (L. Snyder-Mackler).

Conflict of interest statement

No conflict of interest to disclose.

and in rehabilitation science many applications also exist for assessing the efficacy of interventions. When examining longitudinal changes, however, the test–retest reliability of model estimates must be established. An estimate of typical measurement variation is clinically useful for identifying real and measurable change.

The minimal detectable change (MDC) is the smallest measured change that can be interpreted as a true difference. When calculated using a 95% confidence level, the MDC is the greatest change score that would be demonstrated by 95% of stable patients when tested on multiple occasions. MDC's are useful in identifying individuals in a cohort who achieve true change and in assessing the efficacy of interventions that have heterogeneous effects.

The purpose of this study was establish MDC's for joint contact forces derived using an EMG-driven musculoskeletal model of the knee [2,3].

2. Methods

Nine subjects (4M, 5F) who were physically active (Tegner [4] score = 5) participated in this study. Exclusion criteria were symptomatic lower extremity musculoskeletal injury or history of surgery. Subjects completed two testing sessions at least 24 h apart, and one limb was randomly selected for analysis. This work was approved by institutional Human Subjects Review Board and all participants provided informed consent.

Gait analysis was performed using an eight-camera video system (Qualysis Medical AB, Gothenburg, Sweden; sampling rate = 200 Hz) and force platform (Advanced Mechanical Technology, Inc., Watertown, MA, USA; sampling rate = 1000 Hz) embedded in a 20-m walkway. Joint centers were identified by placing retroreflective markers over lower extremity bony landmarks and segment motion was tracked using 4-marker clusters attached to rigid shells. Five gait trials were recorded in which there was no evidence of visual targeting and gait speed was within 5% of the subject's established self-selected speed. Two photoelectric beams (Brower Timing Systems, Draper, Utah) were used to capture speed. Stance phase kinematics and kinetics were computed using Visual 3D software (C-motion, Inc., Germantown, MD).

Surface electromyography (EMG) was recorded using two MA-300 EMG systems with MA-420 preamplifiers (Motion System Labs, Baton Rouge, LA). Skin was cleaned and lightly abraded with isopropyl alcohol prior to electrode application (disposable bipolar electrodes; 10 mm gel well diameter, 22 mm inter-electrode spacing). Electrode placement followed SENIAM recommendations for the following muscles: medial/lateral vastus (VM, VL), rectus femoris (RF), medial/lateral hamstring (SM, BFL), medial/lateral gastrocnemius (MG, LG). Subjects performed maximum volitional isometric contractions to elicit maximal EMG. Gastrocnemius was tested with knee extended and neutral ankle. Quadriceps and hamstrings were tested in 90° hip flexion and 60° knee flexion. EMG data were band-pass filtered (20–500 Hz) prior to sampling at 1000 Hz, and then high-pass filtered (30 Hz), full-wave rectified, low-pass filtered (6 Hz) and normalized to the maximum EMG during isometric or walking trials to create linear envelopes. Because EMG was unavailable for all muscles in the model, linear envelopes for semitendinosus (ST) and short head of biceps femoris (BFS) were set equal to SM and BFL, respectively; vastus intermedius (VI) was equal to the average of VM and VL.

Muscle and joint contact forces were estimated using an EMG-driven model of the knee [2,3]. The anatomical model (SIMM 4.0.2, Musculographics, Chicago, IL) [5] included 10 musculotendon actuators for the knee. The activation dynamics model included a recursive filter [6] (characterized by τ_1 and τ_2) in which activation was estimated from EMG and which allowed up to 70 ms of electromechanical delay (d). A piecewise logarithmic function

[7] (characterized by A_{len}) modeled the non-linear EMG-to-force relationship at low activation levels. The contraction dynamics model was a Hill-type muscle model with a muscle fiber and parallel elastic component in series with a tendon [8]. The tendon was characterized by a non-linear function normalized to slack length (l_{st}). The muscle fiber consisted of a contractile element in parallel with elastic and damping elements, characterizing passive force-length and viscoelastic properties of muscle, respectively [8]. Force in the contractile element was governed by a generic force-length curve, which was normalized to maximal isometric force, optimal fiber length (l_m^0) and maximum contraction velocity [6].

For model calibration, six parameters per muscle ($\alpha_1, \alpha_2, d, A_{len}, l_{st}, l_m^0$) and two global strength multipliers (G_f for flexors, G_e for extensors) were allowed to vary within limits specified previously [2]. Muscle parameters were iteratively adjusted so that the internal flexion/extension muscle moment and sagittal plane moment from inverse dynamics were in best agreement.

After subject-specific calibration for each testing session, muscle forces were predicted for 3 walking trials. Medial compartment contact force was calculated by balancing the external adduction moment about the lateral contact point with muscle forces and with the medial contact force [9]. Contact points were placed at 25% of tibial plateau width as measured medially/laterally from the joint center. Lateral contact force was correspondingly calculated, and tibiofemoral contact force equaled the sum of medial and lateral forces. Contact forces were normalized to body weight (BW) and the average of 3 trials was used for analysis.

Loading variables of interest were: peak medial compartment force, peak tibiofemoral force, lateral compartment force at peak knee flexion angle and percentage of tibiofemoral force borne by medial compartment at peak knee flexion angle. Reliability was assessed by calculating intraclass correlation coefficients (ICC (3, k); two-way mixed, average measure) using IBM SPSS Statistics 19 (Armonk, NY). Typical error was estimated as:

$$\text{Standard error of the measure} = \text{SD} \times \sqrt{1 - \text{ICC}}$$

where SD is the pooled standard deviation of measurements from all subjects. Minimal detectable change was estimated as:

$$\text{MDC} = \text{Standard error of the measure} \times 1.96 \times \sqrt{2}$$

3. Results

Time between sessions ranged from 1 to 42 days (mean = 10.4, SD = 13.3). Subjects were 22.1 (SD = 2.6) years old and 1.70 (SD = 0.12) m in height, with BMI of 24.3 (SD = 2.9) kg/m². Gait speed was 1.47 (SD = 0.16) m/s. Subject-specific calibration resulted in acceptable convergence of the internal muscle moment to the inverse dynamics moment ($R^2 = 0.857$, SD = 0.047; relative RMSE = 10.6, SD = 1.7%).

MDC's for contact force variables ranged from 0.30 to 0.66 BW (Table 1). Test-retest reliability coefficients were also reported for comparison with previous work.

4. Discussion

Our patient-specific EMG-driven model demonstrated MDC's ranging from 0.30 to 0.66 BW. These are clinically useful thresholds (for example) in athletes with ACL injury [9], who demonstrate interlimb differences in knee contact forces greater than these MDC's. Our model demonstrated similar test–retest reliability for the loading variables examined (ICC's 0.733–0.869) when compared to the generalized reliability coefficient of 0.84 reported by Alkjaer et al. [10] for their knee model.

In this study, gait mechanics were highly repeatable and this was an important prerequisite to obtaining repeatable contact force estimates. The standard errors of measurement reported for gait mechanics capture natural day-to-day gait variability as well as measurement error from camera recalibration and marker replacement. Relatively greater standard errors for loading variables were expected, reflecting the additional variability introduced by MVIC task performance, electrode replacement and model recalibration. The MDC's calculated here for healthy, active individuals are lower bounds for these quantities in older or impaired populations, which may exhibit greater variability in neuromuscular strategies and movement patterns.

Using the present model, changes in an individual's loading over time greater than these MDCs are greater than typical day-to-day variation and can be identified as real change.

Acknowledgments

Role of funding source

The study sponsor had no role in study design, collection/analysis/interpretation of data, writing of the manuscript or in the decision to submit the manuscript for publication.

Funding support

National Institutes of Health: P30-GM103333, R01 AR046386, R01 AR048212.

We would like to acknowledge Stephen Suydam and Amelia Lanier for their assistance with data collection and Kristen Stump for her assistance with data processing. This work was supported by the National Institutes of Health (P20 RR016458, R01 AR046386, R01 AR048212).

References

1. Pandy MG, Andriacchi TP. Muscle and joint function in human locomotion. *Annual Review of Biomedical Engineering*. 2010; 12:401–33.
2. Gardinier ES, Manal K, Buchanan TS, Snyder-Mackler L. Gait and neuromuscular asymmetries after acute anterior cruciate ligament rupture. *Medicine and Science in Sports and Exercise*. 2012; 44(8):1490–6. [PubMed: 22330021]
3. Buchanan TS, Lloyd DG, Manal K, Besier TF. Neuromusculoskeletal modeling: estimation of muscle forces and joint moments and movements from measurements of neural command. *Journal of Applied Biomechanics*. 2004; 20(4):367–95. [PubMed: 16467928]
4. Tegner Y, Lysholm J. Rating systems in the evaluation of knee ligament injuries. *Clinical Orthopaedics and Related Research*. 1985; (198):43–9. [PubMed: 4028566]
5. Delp SL, Loan JP, Hoy MG, Zajac FE, Topp EL, Rosen JM. An interactive graphics-based model of the lower extremity to study orthopaedic surgical procedures. *IEEE Transactions on Biomedical Engineering*. 1990; 37(8):757–67. [PubMed: 2210784]
6. Lloyd DG, Besier TF. An EMG-driven musculoskeletal model to estimate muscle forces and knee joint moments in vivo. *Journal of Biomechanics*. 2003; 36(6):765–76. [PubMed: 12742444]

7. Manal K, Buchanan TS. A one-parameter neural activation to muscle activation model: estimating isometric joint moments from electromyograms. *Journal of Biomechanics*. 2003; 36(8):1197–202. [PubMed: 12831746]
8. Zajac F. Muscle and tendon: properties, models, scaling, and application to biomechanics and motor control. *Critical Reviews in Biomedical Engineering*. 1989; 17(4):359–411. [PubMed: 2676342]
9. Gardinier ES, Manal K, Buchanan TS, Snyder-Mackler L. Altered loading in the injured knee after acute ACL rupture. *Journal of Orthopaedic Research*. 2012; 31(3):458–64. [PubMed: 23097309]
10. Alkjaer T, Simonsen EB, Jorgensen U, Dyhre-Poulsen P. Evaluation of the walking pattern in two types of patients with anterior cruciate ligament deficiency: copers and non-copers. *European Journal of Applied Physiology*. 2003; 89(3–4):301–8. [PubMed: 12736838]

Table 1

Mean values, typical variance, reliability coefficients and MDC's are reported for stance phase gait mechanics variables (top) and loading variables (bottom).

Variable	(Units)	Mean	SD	SEM	ICC	MDC
Peak knee flexion angle	(deg)	-21.1	3.7	1.0	0.922	2.9
Peak knee adduction angle	(deg)	2.6	3.2	0.6	0.963	1.7
Peak knee flexion moment	(Nm/kg*m)	0.40	0.08	0.03	0.808	0.09
Peak knee adduction moment	(Nm/kg*m)	-0.28	0.10	0.02	0.946	0.06
Peak medial compartment force	(BW)	2.48	0.30	0.11	0.869	0.30
Peak tibiofemoral contact force	(BW)	3.61	0.48	0.24	0.761	0.66
Lateral compartment force at peak knee flexion	(BW)	0.90	0.43	0.22	0.733	0.61
Medial load share at peak knee flexion	(%)	73.3	11.4	5.8	0.740	16.1