◈ Research Paper

A Biomechanical Model of Lower Extremity Movement in Seated Foot Operation

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Abstract

A biomechanical model of lower extremity in seated postures was developed to assess muscular activities of lower extremity involved in a variety of foot pedal operations. The model incorporated four rigid body segments with the twenty-four muscles to represent lower extremity. This study deals with quasi-static movement to investigate dynamic movement effect in seated foot operation. It is found that optimization method which has been used for modeling the articulated body segments does not predict the forces generated from biarticular muscles and antagonistic muscles reasonably. So, the revised nonlinear optimization scheme was employed to consider the synergistic effects of biarticular muscles and the antagonistic muscle effects from the stabilization of the joint. For the model validation, three male subjects performed the experiments in which EMG activities of the nine lower extremity muscles were measured. Predicted muscle forces were compared with the corresponding EMG amplitudes and it showed no statistical difference. For the selection of optimal seated posture, a physiological meaningful criterion was developed for muscular load sharing developed. For exertion levels, the transition point of type F motor unit of each muscle is inferred by analyzing the electromyogram at the seated postures. Also, for predetermined seated foot operations exertion levels, the recruitment pattern is identified in the continuous exertion, by analyzing the electromyogram changes due to the accumulated muscle fatigue.

1. Introduction

Biomechanical models are a simple and safe alternative to the use of invasive measurements for determining the distribution of forces within the musculoskeletal system. A number of models have used optimization schemes to solve the indeterminate problem of predicting muscle forces(Seireg and Arvikar, 1973; Hardt, 1978; Crowninshield, 1978; Pedotti et al., 1978; An et al., 1984; Olney, S. J. and Winter, D. A. 1985; Pedersen et al., 1987; Bean et al., 1988). The purpose of these studies was typically related to gaining insight into the mechanisms of movement control(Pedotti et al., 1978) or to calculating forces acting on the musculoskeletal system, such as joint contact forces. An et al.(1984) reported that linear optimization-based models predicted muscular forces that did not agree well with the corresponding EMG signals when the summation of muscle forces was used as the objective function. They also noted that only a few muscles were predicted to be active when a weighted linear combination of muscle forces was adopted as the objective function.

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On the other hand, Pedersen et al.(1987) reported that nonlinear methods allow more active muscles in the solution without the formulation of additional constraints. These authors also found that the EMG signals and the corresponding muscle force predictions were in better agreement as compared to the data predicted using a linear objective function. Dul et al.(1984) used the nonlinear criterion of minimum muscular fatigue to obtain load sharing during static-isometric knee flexion. Brand et al.(1986) minimized the sum of the cubed stresses, in order to obtain the sensitivity of muscle force solutions to possible variations in PCSA(physiological cross sectional area) during a gait cycle. Using three sets of PCSA's including those reported by Pierrynowski (1982), it was shown that a muscle force solution was very sensitive to PCSA, whereas peak hip contact force solution was much less so. Bean et al.(1988) has advocated the use of a double linear optimization scheme to predict forces in the spinal structures. The idea of minimization of the co-contractions between agonistic and antagonisic muscles is implied by the objective functions use in the double linear optimization scheme. This resulted in predictions of zero force in some biarticular muscles. This is because the biarticular muscles have a dual role in establishing equilibrium at both the associated proximal and distal joints. So, the model failed to predict co-contractions in their cases. Most other investigation into processed EMG-muscle force relationships have assumed that there is no antagonistic and synergistic activity during isometric contraction. It was important that optimization model developed to predict antagonistic and synergistic activities of muscle groups.

One goal of this paper, therefore, is developed the revised a nonlinear optimization model of lower extremity movement in seated foot operation to predict synergistic and antagonistic muscle activity. It is employed to consider the synergistic effects of biarticular muscles, assuming that the muscle forces are distributed proportionally based on their physiological cross sectional area and moment arm. Also, this model should predicted the force generated from antagonistic muscles at the stabilization of the joint. Another goal is, for the selection of optimal seated posture, to propose a physiological meaningful criterion for muscular load sharing. The other goal is, for exertion 25%MVC, 50%MVC, 75%MVC, 100%MVC, the transition point of type F motor unit of each muscle is inferred by analyzing the electromyogram at the seated postures. Also, for predetermined seated foot operations and exertion levels, the recruitment pattern is identified in the continuous exertion, by analyzing the electromyogram changes due to the accumulated muscle fatigue.

2. Materials and Methods

2.1 Biomechanical Model Formulation

The human body was represented by a sagittaly articulated system of four rigid links connected by frictionless hinge joints. The 47 lower extremity muscles were idealized as a system of 24 single equivalent muscles and the patellar tendon. These muscles were selected by lumping synergistic muscles into a single equivalent muscle or by excluding those muscles which had less functional importance to joint flexion-extension or small cross-sectional areas. Thus, the lower extremity muscles modelled were the iliopsoas(PSO), the iliacus(CUS), the gluterus maximus(GMA), the gluterus medius(GME), the gluterus minimus(GMI),the tensor fasciae latae(TFL), the piriformis(PIR), the adductor longus(ADD), the adductor magnus(ADM), the gracilis(GRA), the sartorius(SAR), the rectus femoris (RFM), the vastus(VAS), the biceps femoris longus(BFL), the biceps femoris short(BFS), the semiten- dinosus(STN), the semimembramosus(SMM), the tibialis anterior(TAN), the extensor digitorum longus (EDL), the gastrocnemius medialis(GAM), the gastrocnemius lateralis(GAL), the soleus(SOL), the tibialis posterior(TPO), and the peroneus longus (PER)(Fig. 1). The model muscles were assumed to produce all the internal joint moment required for equilibrating the external moments caused by gravitational and inertial forces.

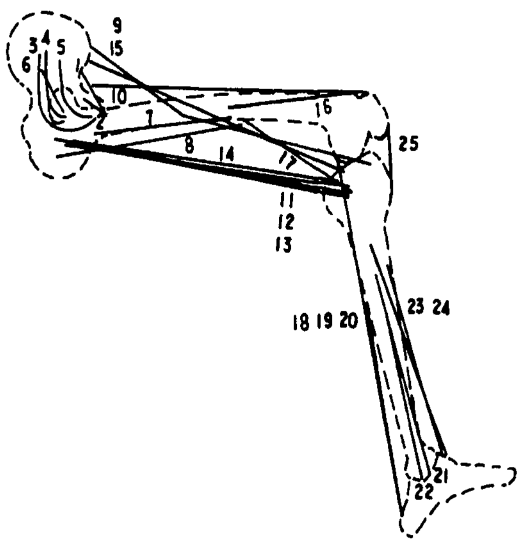


Fig. 1. Schematic representation of the lower extremity model with major muscles and muscle equivalents. These represent the PSO(1); the CUS(2); the GMA(3); the GME(4); the GMI(5); the PIR(6); the ADD(7); the ADM(8); the TFL(9); the SFM(10); the BFL(11); the SMM(12); the STN(13); the GRA(14); the SAR(15); the VAS(16); the BFS(17); the GAM(18); the GAL(19); the SOL(20); the TPO(21); the PER(22); the TAN(23); the EDL(24); and the patellar tendon(25).

Passive tissues such as ligaments and joint capsules were assumed to supply negligible moment resistance at joints. A muscle's capability as a moment generator was represented by its cross- sectional area and lever arm about a moment center. The moment centers in the trunk and lower extremity were located at the L3 trunk level, the hip joint center, the tibio-femoral contact point of the knee joint, and the ankle joint center. The PCSA of each lower extremity muscle was calculated by scaling data reported by Brand et al.(1986) to the body weight of the average subjects. Most lower extremity muscles were described with a straight line-of-action midel. A lever arm was calculated by the appropriate coordinate transformation of muscle origin and insertion coordinates given by Brand et al.(1982). GMA and TEL were exceptions. The GMA, represented by a curved-line model, was assumed to have a similar lever arm to that reported of hip flexion (Nemeth and Ohlsen 1985). The TFL, represented by a two-segment straight-line model, was scaled from anatomical literature (Sobbotta and Figge 1974). Data from Nisell et al.(1986) were used for the location of the knee tibio-femoral contact point and the patella lever arm about that point.

2.2 Optimization Schemes

Using the above data set, a nonlinear optimization model was formulated in the following form:

$$Minimize \quad : \quad [\ \sum \{ (F_i/A_i) - (\sigma) \}^2 \ ]^{1/2} \qquad \text{......................... (1)}$$

$$subject\ to \quad : \quad M_{fj} = \sum r_{ij} F_{ij} \qquad \text{......................... (2)}$$

$$F_i/F_j = (r_i/r_j)^{1/2} (A_i/A_j)^{3/2} \quad (i \neq j) \qquad \text{......................... (3)}$$

$$F_i/A_i \leq \sigma \qquad (i = 1\ to\ 24) \qquad \text{......................... (4)}$$

$$F_i \geq 0 \qquad \text{......................... (5)}$$

where, F_i = force in the i-th muscle

A_i = physiological cross sectional of area i-th muscle(㎠)

M_{fj} = moment in the j-th joint

σ = maximum muscle stress

r_i = moment arm in the i-th muscle

The first constraint(equation 2) arise from the moment equilibriume equation. Synergistic muscle action can be enforced by the formation of an additional the second constraint(equation 3) that relates the stresses in two synergistic muscles. The constraint does guatantee that all the synergistic muscles are predicted to carry the same stress at the this model. So, model does predict the forces generated from biarticular muscles. The third constraint(equation 4) relates the actual stress in muscles with the maximum permissible stresses. The fourth constraint(equation 5) is the nonnegativity constant: a muscle can only develop a tensile force. A objective function(equation 1) that minimizes sum of the squares of the difference between the actual muscle stress and the maximum permissible muscle stress for various muscles.

2.3 Experimental Procedures

Experiments were performed on three healthy young male subjects of a Judo expert. Subjects ages, statures, and weights had with average 24 years, 175cm, 75.5kg, respectively. The seated postures in this study were measured 24 postures in consideration of Henry dreyfuss's literature(Fig. 2). The postures are different with respect to angle between seat and backrest(90°, 120°), angle between seat and upper-leg(0°, 15°), Knee angle(105°, 120°), Ankle angle(90°, 100°, 110°). The chair and pedal equipment are specially constructed to provide experimental postures of seated foot operation in this study(Fig. 3). Force plate (Kistler co. 9281B type) was used in the measurement of foot-pedal reaction forces, and the pedal equipment mounted to a face plate. EMG signals were measured in order to analyze instantaneous right lower extremity muscles activity during the maximum voluntary contraction in the seated pedal operations and to analyze electromyogram change of the accumulated muscle fatigue due to the continuous exertion for 25% MVC, 50% MVC, 75% MVC, 100% MVC. The most accessible 9 muscles were selected on the right leg. These muscles were: GMA, SAR, RFM, BFL, STM, VAS, TAN, GAL, SOL. Electrodes were of a surface bipolar type, composed of two 10mm diameter Ag-Agcl disks spaced 20mm center-to-center, used with electrolytic gel(Fig. 4). Electrode signals, amplified from microvolt to volt levels using a differential preamplifier stage with gain of 300 and main amplifier stage with band-width of 10~2000Hz, were sampled at 1000Hz for three seconds of the maximum voluntary contraction using a IBM-PC and were recorded on a hard disk for a back-up purpose.

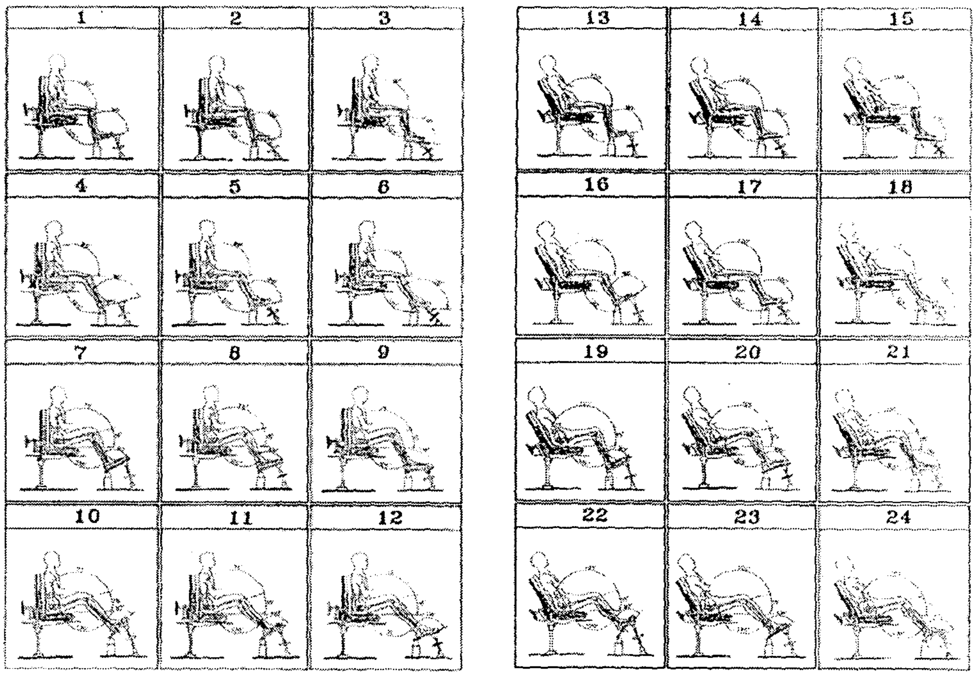
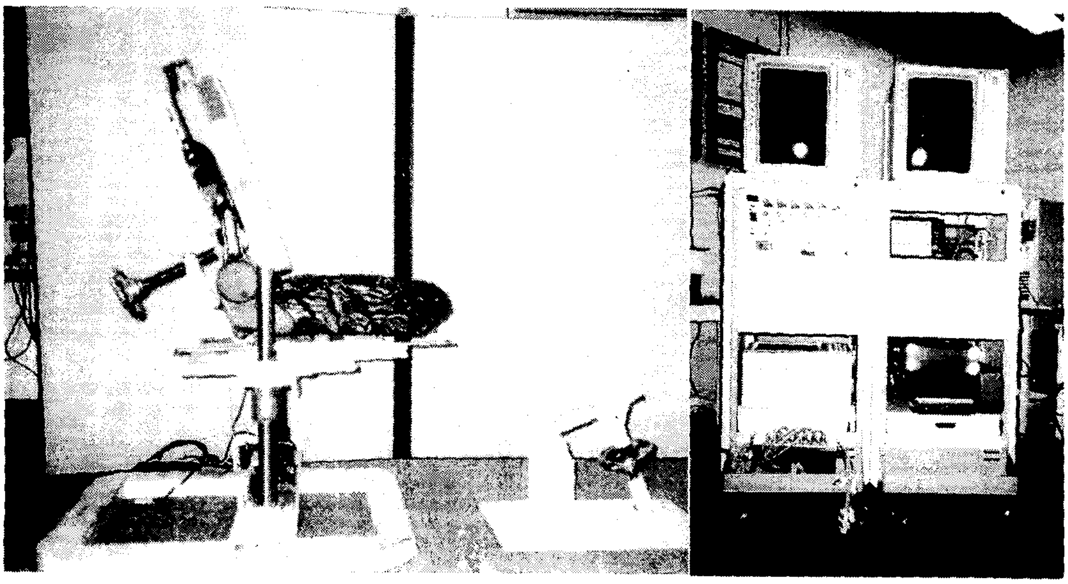


Fig. 2. Diagram of the seated posture in the experiment.



(a) (b)

Fig. 3. Photography (a) The seat and pedal equipment , (b) The EMG equipment.

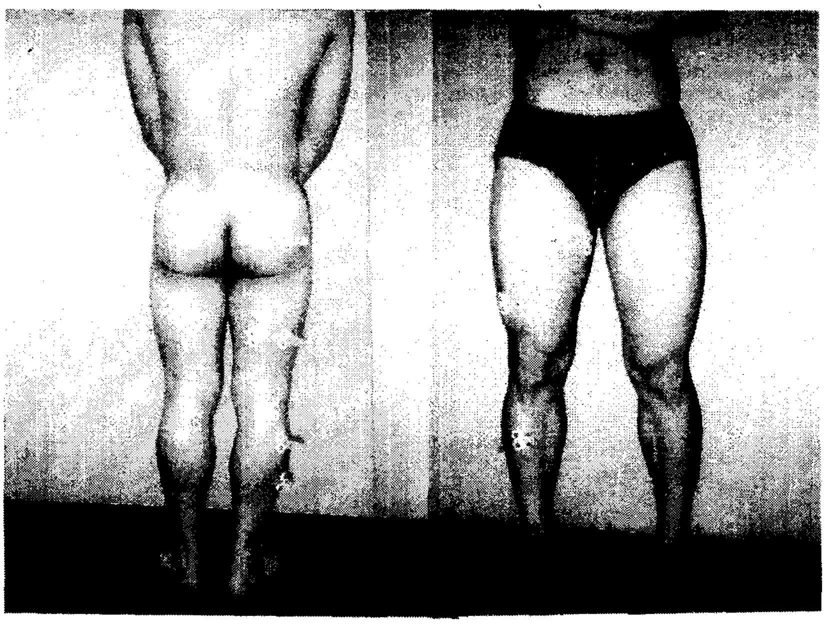


Fig. 4. Photograpy of EMG electrode placement.

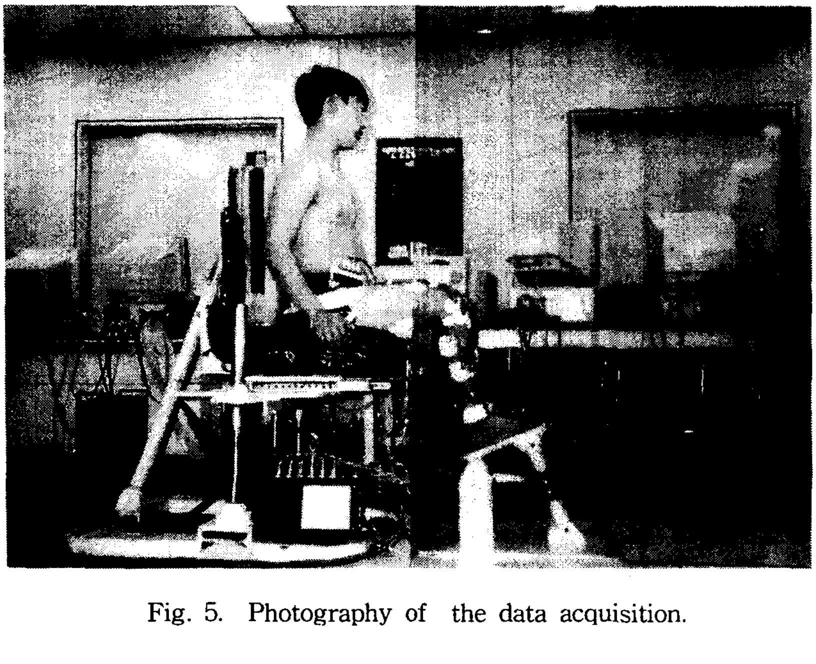


Fig. 5. Photography of the data acquisition.

3. RESULTS

The range of exertion forces of the 24 seated postures in the seated foot operation were presented 130.5~337.1kg. Analysis of variance of exertion forces for subjects and seated postures were showed statistical difference and no interactions between subjects and seated postures. The exertion force of lower extremity become larger in the order of ankle's angle 110° , 100° , 90° . EMG signal pattern of each postures according to change of ankle's angle 90° , 100° , 110° (the plantar flexors) showed that

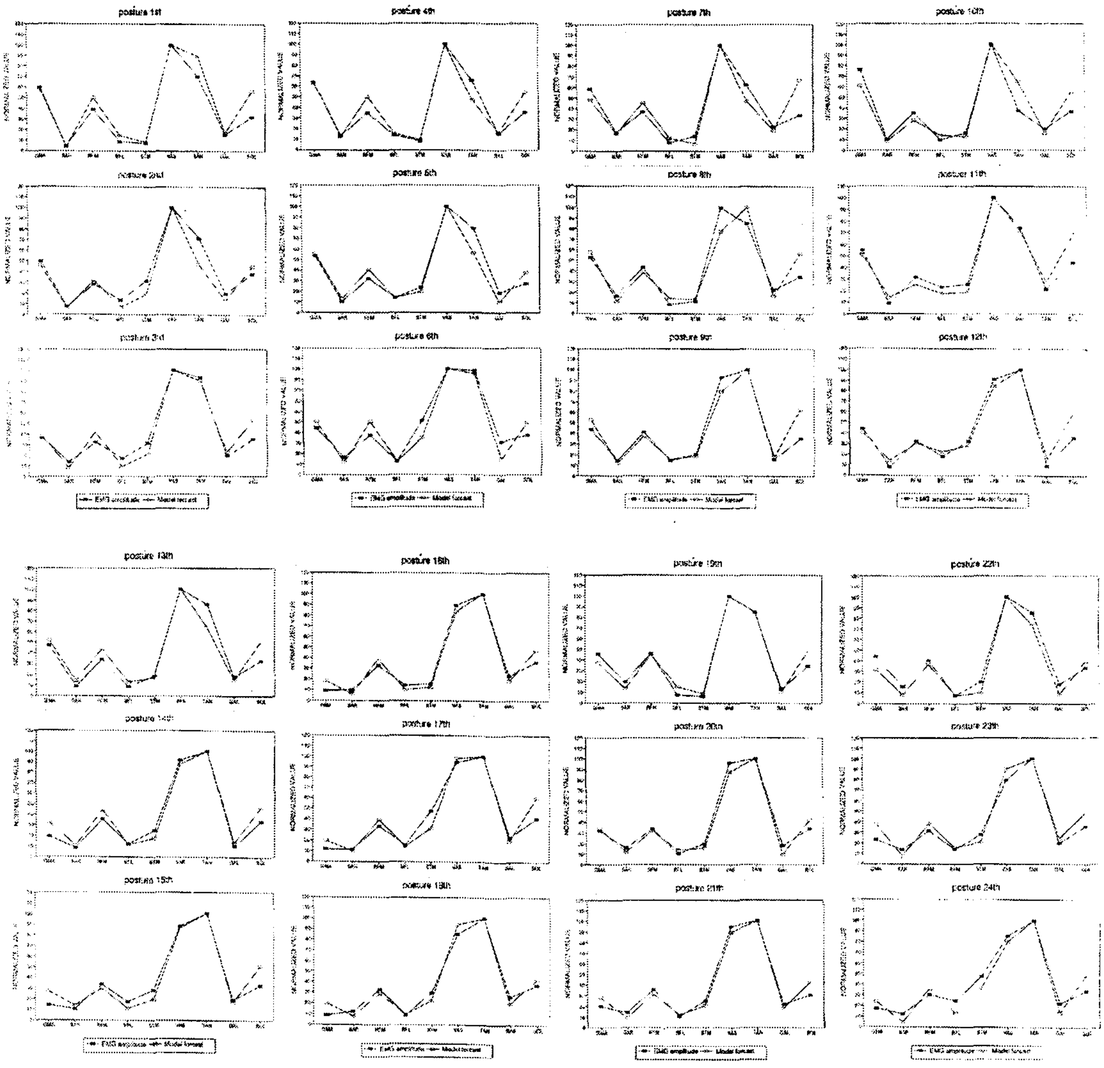


Fig. 6. Comparison between the mean EMG amplitude and the predicted muscle force by the model.

while quadricep muscles force were on the decrease, hamstring muscles were on the increase. For ease of comparison, the amplitude of each EMG signal was normalized to its maximum value. Predicted muscle forces of the 24 seated postures for subjects were compared with the corresponding EMG amplitudes(Fig. 6) and it showed no statistical difference($p<0.05$). The T-test results of each muscle between EMG signal and predicted muscle force for all seated postures showed statistical difference for GMA and SOL muscle and no statistical difference for the other muscles. In the present paper, for the selection of optimal seated posture, a physiological meaningful criterion for muscular load sharing developed. The criterion is based on the hypothesis that muscular fatigue is minimized during seated foot operation and it is the coefficient of variation(C.V.) of muscle stress(predicted muscle force per unit area). The value of C.V. showed larger in order of postureⅡ(posture 4, 5, 6), postureⅣ(posture 10, 11, 12), postureⅠ(posture 1, 2, 3), postureⅢ(posture 4, 5, 6) and the other postures. For exertion 25% MVC, 50% MVC, 75% MVC, 100% MVC during endurance activities in the 1th posture, the recruitment pattern of each muscle due to the accumulated muscle fatigue showed as Fig. 7. The predicted endurance time showed 182 seconds for 25% MVC, 52 seconds for 50% MVC, 26 seconds for 75% MVC.

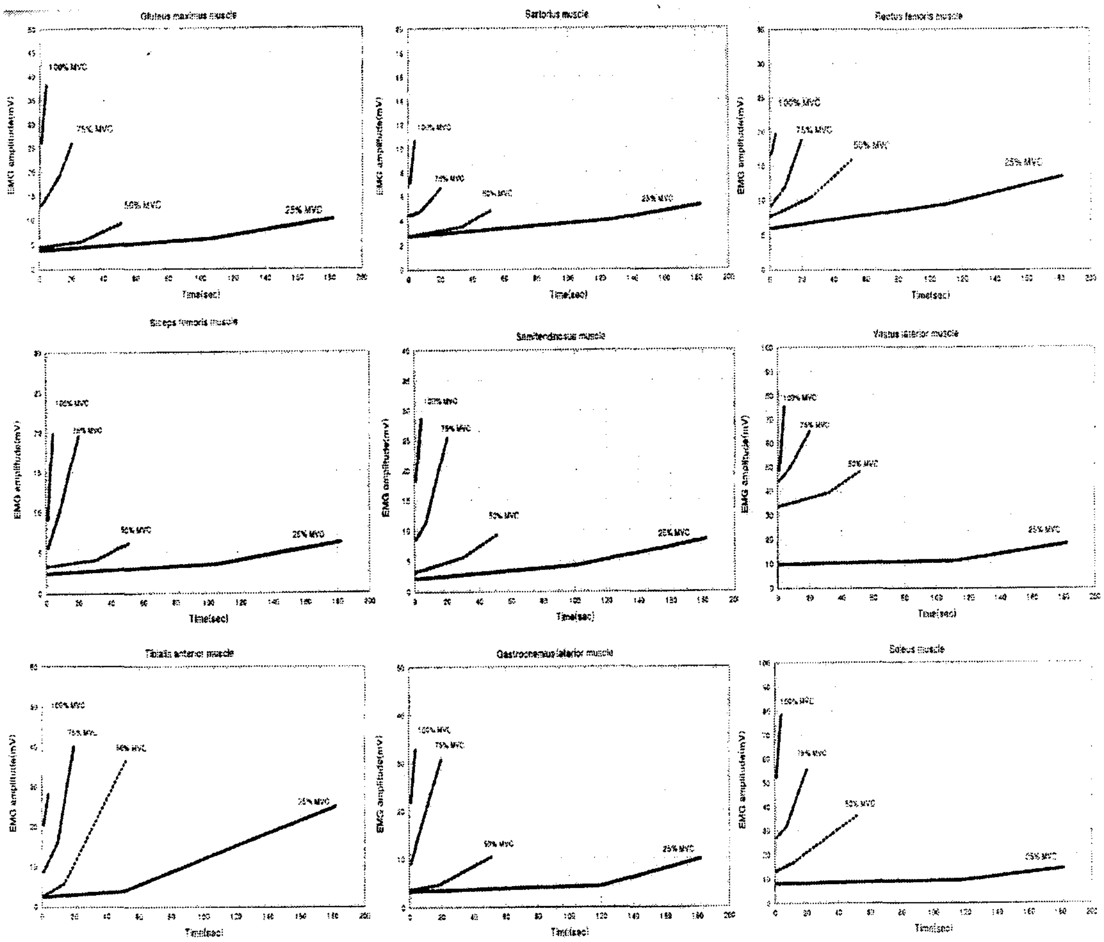


Fig. 7. EMG amplitude variation for maximum voluntary contraction as sustained contraction.

4. Discussion

Optimization methods have traditionally been used to solve the redundancy problem associated with predicting muscle forces in musculoskeletal models. The major problem with the optimization approach in general is that it assumes the human body follows the criteria given in the objective function under various conditions. There is a growing realization, however, that the neuromuscular system may modify the criteria it uses to perform a task in various unpredictable ways. Most other investigation into processed EMG-muscle force relationships have assumed that there is no antagonistic or synergistic activity during isometric contraction. Dul(1985a) presented a nonlinear optimization model that could predict synergistic activity. Antagonistic activity, however, was not model. It is important that optimization model developed to predict antagonistic and synergistic activities of muscle groups. The cocontraction of antagonistic muscle is created to balance torques produced in other directions around the joint. The forces produced would have to be balanced by some equal opposing forces, thus antagonistic forces occur. This would have have the benefit of reducing the potential strain on the ligaments surrounding the joint. Nonlinear objective function in this paper is based on the least square concept; the depature of the actual stress from the optimum surface should be minimum and it has enabled to predict realistic the force generated from antagonistic muscles at the stabilization of the joint. It was found that the action of the TAN created larger amounts force in seated foot operation. In this study, synergistic muscle action can be enforce by the formation of an additional equality constraint that the muscle forces are distributed proportionally base on their physiological cross sectional area and moment arm, such as constraint 2. No formulation included another important factor of load sharing: fiber type coposition. Therefore, nonlinear optimization schemes in this paper can adequately predict the forces generated of biarticular and antagonistic muscles in seated foot operation. The T-test results of GMA and SOL muscle for all steated postures showed statistical difference. EMG amplitude of GMA show smaller compare with predict muscle force. It should suppose that attached electrodes pressured by the seat from a characteristic of posture. In the SOL case, it should suppose that this is due to the difference between subject's PCSA and the literature data although it is inaccurated model. For the selection of optimal seated posture, a physiological meaningful criterion for muscular load sharing uniformly is to propose the C.V. of variation be obtained to divide predicted muscles force in PCSA, hence it is minimized muscular fatigue. The transition point of type F motor unit of TAN muscle was early recruited compared with the other muscles and even for exertion 25% MVC, the TAN muscle was a limited muscle due to reach EMG amplitude of 100% MVC on the last contraction. It is shown that TAN is important muscle to design the pedals and foot-related equipments.

5. Conclusion

In this paper, nonlinear optimization schemes can adequately predict the forces generated of biarticular and antagonistic muscles in seated foot operation. It was found that the action of the TAN created larger amounts force in seated foot operation. The exertion force of lower extremity become larger in the order of ankle's angle 110°, 100°, 90°. EMG signal pattern of each postures according to change of ankle's angle 90°, 100°, 110°(the plantar flexors) showed that while quadricep muscles force were on the decrease, hamstring and TAN muscles were on the increase. For the selection of optimal seated posture, a physiological meaningful criterion for muscular load sharing developed. it is the C.V. of muscle stress. Optimal seated posture by C.V. were in sequence of postureⅡ, postureⅣ, postureⅠ, postureⅢ and the other posture. The transition point of type F motor unit of TAN muscle was early recruited compared with the other muscles and even for exertion 25% MVC, the TAN muscle was a limited muscle due to reach maximum value EMG amplitude on the last contraction. TAN muscle is important to design the pedals and foot-related equipments.

Future formulations of the optimization should include variables related to the muscle fiber type composition for muscular load sharing and antagonistic behavior a more in depth anatomical model of musculature.

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