

COMPARISONS OF MECHANICAL AND ELECTROMYOGRAPHICAL MUSCULAR UTILIZATION RATIOS

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ABSTRACT. The physical loading of a muscle during functional activities can be estimated by the muscular utilization ratio. This ratio is defined as the percentage of muscular involvement relative to the maximal capacity. Either mechanical or electromyographical approaches can be used to obtain the muscle utilization ratio. However, the non-linear relationship between electromyographical activity and muscle force, as well as the non-equivalence between agonist muscles, may create differences between the mechanical muscle utilization ratio calculated from joint moments and the electromyographical muscle utilization ratio calculated from electromyographical data. The aim of this study was to compare, during a squat test, the mechanical muscle utilization ratio and the electromyographical muscle utilization ratio estimated by three different methods; direct linear approximation, second order polynomial regression and linear interpolation. The knee extensor moment and electromyographical data of rectus femoris and vastus medialis of 11 subjects were recorded during both knee extension and squat. Both tests were performed with the knee maintained at 90° of flexion. The results showed that: a) the electromyographical muscle utilization ratio, calculated from the average of vastus medialis and rectus femoris, significantly underestimates the mechanical muscle utilization ratio (ANOVA, $p < 0.01$), b) the differences between the mechanical muscle utilization ratio and the electromyographical muscle utilization ratio are larger for the direct linear approximation method than for the second order polynomial regression (ANOVA, $p < 0.01$) or the linear interpolation method (ANOVA, $p < 0.01$), and c) independent of the method utilized, there is no difference between the electromyographical muscle utilization ratio predicted by the vastus medialis as compared with the rectus femoris (ANOVA, $p > 0.01$). These results indicate that the muscle utilization ratio calculated from electromyographical data underestimates the mechanical muscle utilization ratio

even after correction for the non-linearity between moment and electromyographical activity.

Key words: EMG, biomechanics, muscle function, squat test, knee joint.

INTRODUCTION

Numerous techniques are available to quantify the performance of the muscular system during functional activities. Electromyographical (EMG) and biomechanical methods are commonly used together since they are complementary. On one hand, EMG reflects the activation of a muscle and reveals the neuro-muscular strategies adopted by the nervous system to realize the intended task. On the other hand, mechanical analysis defines the physical constraints of the environment and specifies the role of muscular force in terms of kinematic and kinetic parameters (28).

These two complementary approaches are alternatively used to evaluate equivalent aspects of motor function (6, 8, 19, 20, 24). This is the case for the muscular utilization ratio (MUR) which gives the percentage of muscular involvement relative to the muscle maximal capacity. The MUR is an important muscle activity indicator because large MURs are associated with higher levels of muscular fatigue (14, 23) and a higher perception of effort (10). The MUR is often reported in ergonomic literature (11, 25) and was recently proposed as a criteria in the evaluation of motor dysfunction (22).

In the present study, two types of MUR were defined: the mechanical MUR (MMUR) and the electromyographical MUR (EMUR). The MMUR is the ratio of the moment of force produced during a functional activity to the moment of force produced during a maximal voluntary contraction (MVC) (1, 5, 16, 21, 25). The result is multiplied by 100 to obtain a percentage. The EMUR is formally defined as the

ratio of the EMG activity recorded during the functional activity to the maximal EMG recorded during a MVC (18, 22, 26). Assuming a near-linear relationship between the moment of force and EMG, the EMUR will provide a good estimate of the MMUR. However, it is well known that the EMG activity of some muscles increases more than the external moment (7, 29, 30). Therefore, as pointed out by Jonsson (11), this non-linear relationship will underestimate the MMUR when the EMG is used as a predictor. This underestimation can be minimized using a non-linear EMG-force relationship or an interpolation method.

A second problem is the prediction of MMUR based on the EMG recorded exclusively from one muscle. The prediction will be precise only under the condition of "muscle equivalence" (2). Muscle equivalence implies that EMG is modulated in the same proportion in all agonist muscles when the external moment increases or decreases. Thus, the shape of the moment-EMG relationship must be similar for these muscles. Much evidence exists, however, to suggest that the "muscle equivalence" concept must be used with caution. For example, the activation of a muscle relative to its agonists can be changed when multi-articular moments are generated (3, 4, 7, 9). Modification of the velocity of contraction (12, 15), type of contraction (15) and fatigue (17) also change the relationship between the recruitment levels of agonist muscles. Consequently, it is important that MURs be evaluated through a functional activity (squat test) which is different from the isolated knee extension used to have the moment-EMG relationship. Moreover, both mechanical and electromyographical MUR should be obtained simultaneously during the task insuring that prediction of muscle moment from EMG data could be compared to the muscle moment computed from that specific task.

The aim of this study was to evaluate, during a squat test, the difference between the MMUR and EMUR of the knee extensors as estimated by the EMG recorded from the vastus medialis and the rectus femoris. The EMUR will be computed by three EMG methods: a) a direct linear approximation assuming a linear relationship between moment and EMG, b) a non-linear regression using the moment-EMG relationship, and c) an interpolation technique using the EMG values. The exactness of these methods will be evaluated by comparing the MUR calculated by mechanical analysis (MMUR) with the estimation based on the EMG data of each individual

Table I. Characteristics of the subjects ($n = 11$)

Mean values (± 1 SD)

Sex	Age (years)	Weight (kg)	Height (m)
M = 4	25.0 (3.6)	72.6 (8.2)	1.7 (0.06)
F = 7	24.3 (2.9)	54.5 (6.3)	1.6 (0.05)

muscle (vastus medialis or rectus femoris) and on the average data of the two muscles ($(\sum(\text{vastus medialis} + \text{rectus femoris})/2)$).

METHODS

Subjects

The characteristics of the individuals ($n = 11$) who participated in this study are described in Table I. The subjects were normal and had no related conditions in the back, pelvis or lower limbs. They had never had surgery and had no ligamentous instability. They all signed an informed consent form approved by our institutional ethical committee.

Dynamometric test

The subject was seated in a specially adapted chair that stabilized the trunk and thigh segments. The force was recorded with a load cell (Gould-Statham Inc. model UC3 with coupling unit ULR-500). The cell was connected to a DC amplifier that sent the signal to a computer. The moment generated at the knee was calculated using the force data and the length of the involved lever arm. Three MVC of the knee extensors were performed with the knee joint at 90° of flexion while the hip was stabilized at 90° of flexion and the ankle free (0° = anatomical position). The subject had seven seconds to complete each contraction which consisted of a ramp force ranging from 0 to 100% MVC. A two-minute rest period was allowed between each trial (13). The average of the maximal torque values of the three trials was calculated and retained for analysis. The EMG activity of the rectus femoris, vastus medialis, biceps femoris and gluteus maximus was simultaneously recorded. The EMG activity was recorded with miniature Beckman surface electrodes (4mm) placed longitudinally to the muscle fibers with an inter-electrodes distance of 10 mm. The EMG signal was preamplified (TECA PA-62A) and amplified (TECA AA6 serie MK III: TECA Corporation, NY). The lower cut off frequency was 16 Hz and the upper cut off frequency 800 Hz. The CMRR was 50000:1 at 50 Hz. The input impedance was above 100 megohms/30 pF. The force and EMG signals were sampled at 1600 Hz by a PDP11/23 + computer and the data stored on disks for further processing. The raw digitalized EMG data of each muscle was rectified and filtered with a low pass second order numerical filter having a cut off frequency of 2 Hz (19). This filter has characteristics similar to those of the classical Butterworth filter. Windows of 50 msec of EMG data were taken at each 10% from 10% up to 100% MVC and the value of the three trials were averaged, at each percentage, to obtain the moment-EMG relationship.

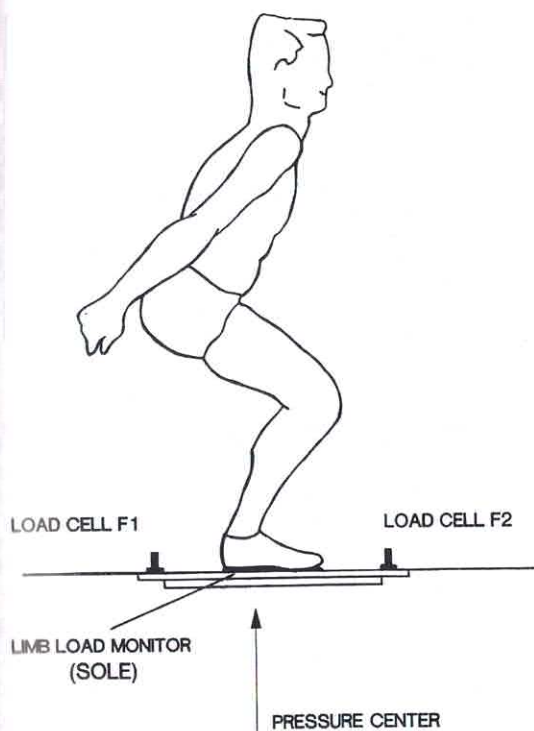


Fig. 1. Functional squat test position. The subject was standing on a plate supported by two load cells. A photographic system using markers and standard calibration metre allowed for the calculation of the knee joint moment.

Functional test

In the second part of the study, the subject had to maintain a squat position, with the knees flexed at 90° , the arms being behind the back (Fig. 1). According to Williams (27), in a squat test with a knee flexion of 90° , the hip angle reaches approximately 90° . Consequently, the knee and hip angles were about the same as for the dynamometric test allowing comparison between the dynamometric test and the squat test. The knee angle was controlled through a polycentric electrogoniometer (model 103-1, Krusen Research Center, PA). This electrogoniometer connected to two oscilloscopes, one placed in front of the subject (Tetronix type D 1011) and one facing the investigator gave a direct visual feedback of the knee angle. The subject had instructions to bend his knees until the angle signal (a moving line) was superimposed on the target line (fixed line) which had been previously calibrated for a 90° knee flexion. The subject had to maintain this position for five seconds. The subject was also constrained to support his body weight equally on each limb. The amount of weight bearing was controlled with a limb load monitor (Krusen Research Center, PA) calibrated to give an auditive feedback when the subject reached 50% of his body weight on the assessed lower limb. Each subject had to perform three squats. A one-minute period of rest was allowed between each trial. The average of the three trials was calculated and retained for the data analysis.

Skin markers were placed at the shoulder, hip, knee and ankle joint centers as well as at the head of the fifth metatarsal of the right side of the body. The center of the

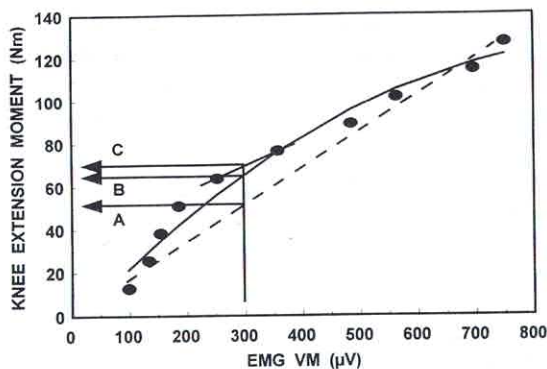


Fig. 2. Calculation of the electromyographical muscle utilization ratio (EMUR). The EMUR of the squat test was calculated from (A) Direct linear approximation, (B) Second order polynomial regression and (C) linear interpolation. The filled circles represent the electromyographical values measured during a ramp contraction for one subject. The solid line is a regression curve computed from these values. A hypothetical linear relation is given by the dotted line. The value of $300 \mu V$ on the X axis is the EMG recorded during the squat test. For each method, this EMG predicts different moments of force. The EMURs were obtained by the ratio of the predicted moment to the maximal moment time 100. Data from the vastus medialis of subject 3.

force plate was also identified by a marker. The force plate was composed of a rigid plate supported by two load cells (Gould-Statham Inc.). These load cells recorded only vertical forces applied to the side investigated. The position of the center of pressure was calculated by the proportion of the resultant vertical force measured by each load cell. Before each trial, the platform was unloaded and reset to zero. During the test, a 35-mm camera loaded with Tri-X Pan 400 ASA Kodak film was activated when the subject reached a knee angle of 90° . At the same time, a pulse voltage (35 ms) was superimposed on the force signal to indicate the exact time the picture was taken. The negative of the film was projected on a digital tablet (Jandel Scientific, CA: Numonic model 2210-1217) and the X-Y coordinates of the markers were obtained. The X-Y coordinates, along with the force data, were fed to a spreadsheet program that calculated the joint angles and joint moments of force. For all trials, the EMG activity of vastus medialis, rectus femoris, biceps femoris and gluteus maximus was simultaneously recorded with the same electrode locations as in the dynamometric test. Processing of EMG data was similar to the one used in the dynamometric tests.

Calculation of the MMURs and EMURs

In the present study, the MMUR is the ratio of the knee extensor joint moment produced during the squat test to the maximal knee extensor joint moment produced during the ramp contraction. The EMUR were calculated from direct linear approximation, non-linear regression or linear interpolation. Using a typical moment-EMG relationship, these three methods are illustrated in Fig. 2. Method A is a direct linear approximation of the EMUR calculated by dividing the EMG obtained during the squat test by the maximal EMG recorded at the time where the maximal moment was reached during the MVC. The result is multiplied by 100 and

Table II. Mean electromyographical muscle utilization ratio (%) calculated by the three methods for vastus medialis (VM) and rectus femoris (RF) and for the combined value ($(\text{vastus medialis} + \text{rectus femoris})/2$)

The average mechanical muscle utilization ratio is equal to 69%. ($n = 11$)

	VM	RF	VM + RF
Linear	45	38	42
Regression	54	55	55
Interpolation	55	56	56

expressed as a percentage. This method assumes a near-linear relationship between the EMG data and joint moments. Method B is a second order polynomial regression which is fitted to the EMG data measured at each 10% of the ramp contraction force from 10% to 100% MVC. To obtain the corrected EMUR, this regression equation was used to predict the moment associated with the EMG recorded during the squat test. This equivalent moment was then divided by the maximal moment and multiplied by 100. Method C is a linear interpolation technique. The two EMG values in the ramp contraction (the next lowest value and the next highest value, respectively) that overlap the EMG recorded in the squat test are used to interpolate the equivalent moment. As for the regression method, the result is expressed as a percentage of the maximal moment.

Statistical analysis

One way repeated analysis of variance (ANOVA) was performed on the MUR to depict the presence of differences between the methods used. Root mean square differences were calculated to evaluate the absolute differences between each EMUR and the MMUR. One way ANOVA for repeated measures was also performed to verify if a significant difference existed between the EMUR predicted by vastus medialis and rectus femoris. A 0.01 level of significance was used.

RESULTS

The results of the mean EMUR calculated separately for vastus medialis and rectus femoris and then combined (vastus medialis + rectus femoris) are shown in Table II. These values are lower than the MMUR which is equal to 69%. The ANOVAs performed on the MUR values indicate that all combined EMURs (vastus medialis + rectus femoris) are significantly lower than the MMUR ($p < 0.01$). Moreover, the EMUR values predicted by direct linear approximation are lower than those calculated from the two other methods ($p < 0.01$). However, there was no significant difference detected between the second order polynomial regression and linear interpolation methods ($p > 0.01$). The ANOVAs

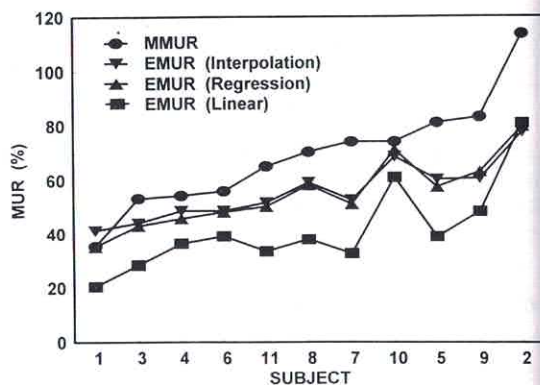


Fig. 3. Muscle utilization ratios (MUR). This figure shows the plotted MUR, both mechanical (MMUR) and electromyographical (EMUR) for all subjects. Each method is illustrated as one symbol profile. Subjects are sorted by increasing order of MMUR.

performed on the vastus medialis-EMUR and on the rectus femoris-EMUR data did not depict the presence of significant difference between the EMURs calculated with the vastus medialis and rectus femoris ($p > 0.01$). This is so even if the average rectus femoris-EMUR calculated by the direct approximation was lower than the corresponding vastus medialis-EMUR.

The individual MMUR and combined EMUR values (vastus medialis + rectus femoris) are presented in the Fig. 3. The trend of EMURs to underestimate the MMUR is clearly evident. The largest difference is for the direct linear approximation (LINEAR). In general, there is a parallel increase of both MMUR and EMUR across subjects except for subject 7 who is too low relative to subject 8 and subject 10 who is too high relative to subject 5. Fig. 4 represents the root mean square differences between the MMUR and each of the EMUR calculated by the three methods. Each column represents the average root mean square difference ($n = 11$) for one muscle (vastus medialis or rectus femoris) and one EMG method.

DISCUSSION AND CONCLUSION

The lower MUR calculated by the three EMG methods relative to that obtained by the mechanical analysis cannot be explained by a difference in the level of co-contraction of the knee flexors. In fact, for the same moment of force in knee extension, EMG activity recorded in the biceps femoris muscle was, in

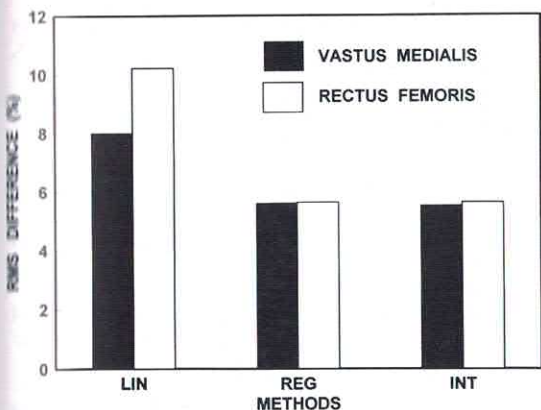


Fig. 4. Histogram of the root mean square (RMS) differences. This figure illustrates the RMS differences between the mechanical muscle utilization ratio and the electromyographical muscle utilization ratio computed by the three electromyographical methods. LIN = direct linear approximation, REG = polynomial regression and INT = interpolation.

general, higher in the squat test than in the ramp contraction. Since the knee extensors have to equilibrate both the external moment of the gravity and the moment generated by the knee flexors in the squat test, the EMUR should have been higher than the MMUR: this was not observed. A second possibility could have been the longer muscle length of the rectus femoris during the squat test because of a change in the trunk position. An increase in the rectus femoris length would increase the maximal force (force-length relationship) and less EMG would be necessary to support the same moment. However, analysis of hip joint angles reveals no systematic trend towards larger angle in the squat test than in the dynamometric test. A third hypothesis to explain the lower EMUR in vastus medialis and rectus femoris would be a larger contribution of the vastus lateralis in the squat test than in the dynamometric test. Consequently, the EMUR of this muscle would also be higher than those of vastus medialis and rectus femoris. Since the vastus lateralis muscle was not recorded in the present study, this hypothesis remains to be tested experimentally.

The EMUR computed from the direct linear approximation method clearly underestimates the MMUR as suggested by Jonsson (11). Regression and interpolation methods give the same results because second-order polynomial curve fitting of the moment-EMG relationship was very good for all subjects ($R^2 > 0.99$). Taking as a reference the MMUR, the classification of subjects in terms of

EMUR levels is better with the regression and interpolation methods than with the direct linear approximation (Fig. 3). Only subjects 7 and 10 are misclassified by the EMG methods.

There was no statistical difference between the EMUR calculated from the vastus medialis or from the rectus femoris. This result supports the muscle equivalent concept of Bouisset (2) for the squat test. The similar EMURs calculated from the vastus medialis and the rectus femoris mean that the two muscles generated an equivalent proportion of their maximal capacity. However, the absolute mechanical contribution of each muscle to the knee extension moment during the squat test is probably different. The rectus femoris, a biarticular muscle, was not less active in the squat test as it would have been expected from the work of Jacob & Van Ingen Schenau (9). These authors reported an inhibition of the rectus femoris when knee extension moments were associated with hip extension moments. Since this association of knee and hip extension moment has already been described during a squat test with a knee angle of 90° (8, 27), it would have been coherent to observe lower rectus femoris-EMUR than vastus medialis-EMUR. This may be explained by the fact that hip extensor moments were probably too low during the squat test to observe an inhibition of rectus femoris.

In conclusion, the present results indicate that MUR estimated from a second order polynomial regression or from an interpolation EMG method underestimate the MUR calculated with a biomechanical method. However, MURs computed from these EMG methods can be used to compare the MUR of different subjects since classification across subjects (in terms of relative mechanical effort) is good enough to estimate MMUR. In the present task where one precise knee angle and one velocity of contraction of the quadriceps (static condition) were investigated, there was a significant difference between the MURs as predicted from the EMG data and the ones from the mechanical data. Considering that in most functional tasks, dynamic movements take place with changes in angles and velocities of joints, EMG methods must be carefully interpreted in order to quantify the performance of the muscular system during functional activities.

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