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Comparative study between two elbow flexion exercises using the estimated resultant muscle force

Estudo comparativo entre dois exercícios de flexão do cotovelo utilizando a estimativa da força muscular resultante

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Abstract

Objective: The present study sought to compare the estimated resultant muscle force required throughout the range of motion in two variations of elbow flexion exercise: the Scott exercise, performed with the aid of a Scott bench, and the unsupported exercise, performed with the upper arm simply resting on the leg. **Method**: Eight healthy individuals performed each exercise eight times, with the same 4kg load. The biceps brachii and triceps brachii muscles were monitored using surface electromyography and the elbow joint movement was measured using an electrogoniometer. A mechanical model of the situation was proposed to evaluate the resultant muscle force acting throughout the range of motion of the exercise. **Results**: Comparisons between the two exercise variations presented significant differences ($p<0.01$) in practically all of the angles. **Conclusion**: Analysis of the model suggests that greater muscle demand is not necessarily associated with higher resistance torque values in the exercise as it is fundamental to take into consideration the moment arm of the muscles involved.

Key words: exercise; torque; load-bearing.

Resumo

Objetivo: O presente estudo objetivou comparar a força muscular resultante estimada necessária durante toda a amplitude do exercício de flexão de cotovelo, executado em duas variações: exercício Scott, realizado com auxílio do banco Scott, e exercício sem suporte, realizado com o braço apenas apoiado na perna. **Método**: Oito indivíduos saudáveis realizaram oito execuções de cada exercício, com uma carga fixa de 4kg. Os músculos bíceps braquial e tríceps braquial foram monitorados com eletromiografia de superfície e o movimento da articulação do cotovelo acompanhado com eletrogoniometria. Um modelo mecânico da situação foi proposto para avaliar a força muscular resultante atuante durante toda a amplitude do exercício. **Resultados**: Comparações entre os exercícios apresentaram diferenças significativas em praticamente todas as angulações ($p<0,01$). **Conclusão**: A análise do modelo sugere que a maior exigência muscular não necessariamente está associada aos maiores valores do torque de resistência do exercício, sendo fundamental levar em consideração a distância perpendicular da musculatura envolvida.

Palavras-chave: exercício; torque; suporte de carga.

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Introduction

Exercise-based muscle stimulation is widely used by health professionals, from physiotherapists who seek to rehabilitate patients recovering from injury, to personal trainers who want to induce hypertrophy in the muscles of a healthy individual. Activation of a specific muscle group occurs voluntarily through the firing of the respective alpha motor neurons under the control of the central nervous system (CNS) and modulated by external stimuli[1]. These external stimuli, normally referred to as "external load"[2] because they act externally on a system of levers, i.e. in the human body, can be objectively classified according to the torque they produce. This torque, known as resistance torque, can be quantified (scalar quantity) by the product between the resistance force and the moment arm, which is defined as the shortest distance between the line of action of a force and its rotation axis[3].

Many different types of equipment are commonly used to adjust the external load of an exercise to the demand required for muscle stimulation[4]. Free weights[5,6] (dumbbells, barbells, plates and ankle and wrist weights), elastic resistance[7,8] (bands, rubber bands, tubes and springs), machines[9,10] (devices with pulleys and/or torque transmission bars) and equipment for aquatic exercise[11,12] (paddles, fins and buoys) are some of the items used while performing exercises. Different items have distinct resistance force characteristics due to the physical laws that cause these forces. The line of action of the forces generated through the use of this equipment is also dependent on the design of the exercise.

When free weights are used as auxiliary exercise equipment, there is a tendency to consider only the force magnitude originating from the equipment and ignore the moment arm of the force or even the inertial factors originating from occasional acceleration in the evaluation of the "exercise load". As a result of this approach, the mechanical stimulus may be less than what is necessary for the established objectives of the chosen exercise, or may even overload the stimulated structures, thereby incurring the risk of injury[13]. While it is important to know the external torque, it is equally important to evaluate its repercussions for the target musculature[4]. Initially, the demand from the muscles by a specific exercise can be evaluated by balancing it with the external torque. Accordingly, high external torque magnitudes are associated with high internal muscle torque magnitudes. Similarly, when an increase or decrease in external torque occurs at any given point of the range of motion (ROM) of the exercise, this variation is accompanied by a corresponding increase or decrease in internal muscle torque. Nevertheless, an increase in muscle torque does not necessarily represent a proportional increase in muscle force. When seeking to discover the magnitude of the muscle force associated with a specific muscle torque, one should take into account the variation in the muscle moment arm (which is a mechanical characteristic)[3,14,15].

Specifically in regard to elbow flexion exercises performed with free weights, some variations are possible even if the external load is maintained, such as the positioning of the upper arm and forearm segments. These variations produce differentiated external resistance, and consequently require distinct muscle efforts in each case. Although the organization of different activation strategies for muscle groups is highly complex and is not fully understood[16], mechanical analysis has recently been used to understand exercises[10], and even to choose one exercise over another[17,18]. Hence, in an attempt to understand the elbow flexion exercise from a mechanical perspective, the objective of the present study was to compare the estimated resultant muscle force required throughout the full ROM in two distinct elbow flexion exercises.

Material and Methods

Eight healthy subjects, with a mean age of 22.4 years (±2.2), mean height of 1.75m (±0.05), body mass of 75.0kg (±2.8), and mean forearm length (elbow axis to ulnar styloid) of 0.24cm (±0.01), who exercised regularly, performed two variations of elbow flexion exercise in the seated position, with the radio-ulnar joint in supination: (a) Scott exercise – upper arm and chest supported on a Larry Scott Bench with shoulder flexion maintained at 60º; (b) unsupported exercise – trunk leaning forward with the upper limb hanging and vertically touching the thigh, forming a shoulder flexion angle of around 60º. The subjects performed eight repetitions of each variation in random order. They were asked to perform the exercises at a slow and constant speed, working the muscles to the full extent of the movement. The load used by all the subjects in all the situations was a 4kg dumbbell.

With the objective of the study in mind, and in order to reduce the variability of the outcomes, a nonrandom sample was used in which the selection criteria were similarity of anthropometric variables and muscle capacity of the subjects. All the subjects had practiced weight training for at least five years, and were familiar with the exercises analyzed in this study. The load used in the normal training sessions was far greater than the 4kg used in the present study (ranging from 14 to 16kg). The load was chosen, like the number of repetitions, in order to minimize the possibility of fatigue while exercising. All the participants signed a consent form, and the project was approved by the Ethics Committee of the Universidade Federal do Rio Grande do Sul (approval report number 2007752) where the study was carried out.

Instrumentation

For the purpose of evaluating the proposed method, the subjects were monitored using a four-channel Miotec system linked to a notebook computer (HP Pavilion ZV5000). While the exercises were being performed, measurements of the elbow angle were made using an electronic goniometer and the electrical activity of the biceps brachii and triceps brachii muscles were registered through surface electromyography (EMG). Disposable surface electrodes (Medtrace) were placed longitudinally, along the supposed direction of the muscle fibers in a bipolar configuration, according to the Surface Electromyography for the Non-Invasive Assessment of Muscles (SENIAM)[19]. The data acquisition rate used was 2,000Hz per channel.

Mechanical analysis

A free-body diagram (FBD)[1], drawn in the plane of movement execution, was used to represent all the torques and forces acting on the forearm-hand segment. The movement of the wrist joint was not considered, and the forearm-hand segment was considered to be a single solid segment. The torques were represented on the diagram around the axis where they acted, and each force was represented according to the point on which it acted and its line of action. In general, single plane exercises, executed around a single joint, can be represented by four vectors: the external force, the weight of the human body segment, the muscle torque and the reaction joint force (Figure 1A). A specific muscle torque is produced to counterbalance the resistance torque produced by the exercise.

Based on the FBD, the situation will be described by the movement equations that orchestrate the movement, both from the rotational and translational points of view[20]. Given that the exercise is executed in a single plane, the forearm-hand segment represented in Figure 1A can be modeled using equations (1) and (2), which refer to translation and rotation movements respectively:

$$\sum F = m.a \quad (1)$$

$$\sum T = I.\alpha \quad (2)$$

where:

$\sum F$ refers to the sum of all the forces acting on the forearm-hand segment

m is the mass of the forearm-hand segment

a is the linear acceleration of the center of mass of the forearm-hand segment

$\sum T$ refers to the sum of all the torques acting on the forearm-hand segment

I is the inertial moment of the forearm-hand segment

α is the angular acceleration of the forearm-hand segment

When a movement is performed by a human being, there is no possibility for the speed to be constant during the movement unless there is external help, such as an isokinetic machine[21]. However, considering the typical range of human motion around 90º, an average angular speed of 30º/seconds will correspond to a peak acceleration of less than 0.1g. In this case, the acceleration involved will be very slight and the inertial effects negligible (lower than 10% of the amount). In other words, the linear and angular accelerations (a and α) can be assumed to be zero. Hence, it can be assumed that there is a

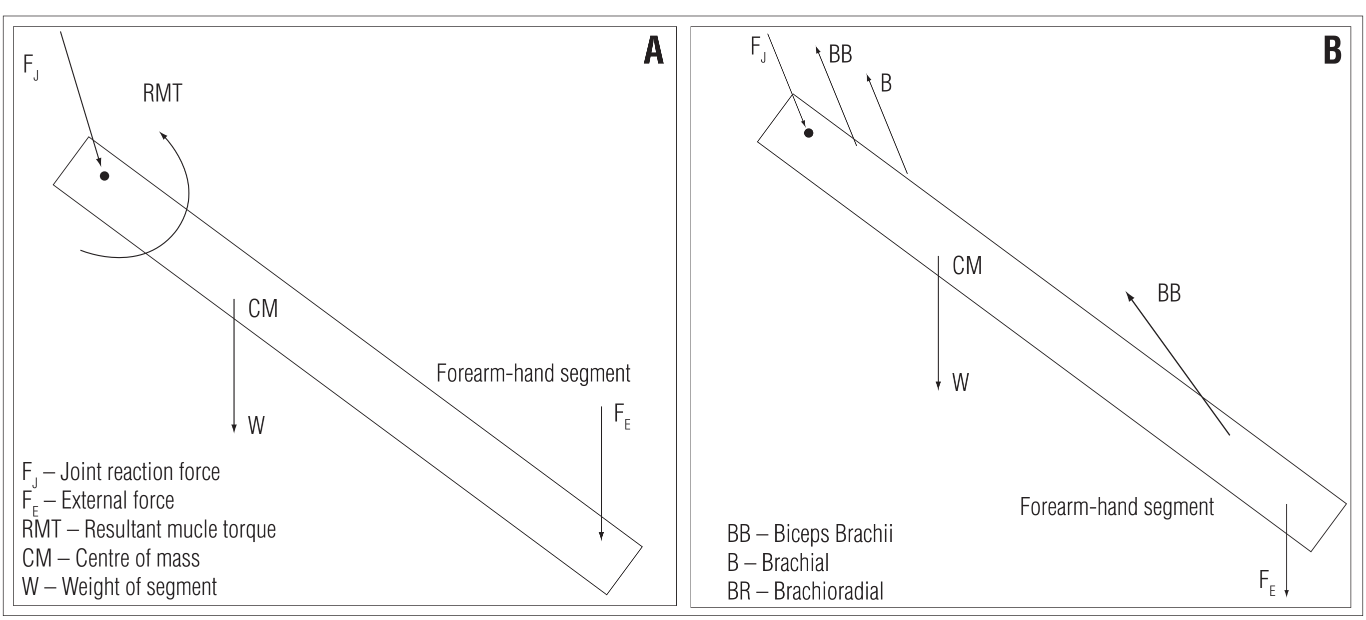

Figure 1. Free-body diagram of the forearm-hand segment during an intermediate position of the elbow flexion exercise: (A) muscle forces represented by their torque (Resultant muscle torque – RMT), (B) muscle forces represented by the main agonist muscles.

balance between the torques throughout the duration of the exercise, and therefore equation (2) can be rewritten to represent this situation, as shown in equation (3):

$$\vec{T}_M+\vec{T}_E+\vec{T}_W=0 \qquad (3)$$

where,
$\vec{T}_M$ is the muscle torque
$\vec{T}_E$ is the torque exerted by the external force
$\vec{T}_W$ is the torque exerted by the weight force

The resultant muscle torque (RMT) represents all the muscles activated during the movement (both the agonist and the antagonist groups), and this was represented around the joint in the direction of the agonist muscles[1]. Based on equation 3, it was possible to isolate the RMT so that the other torques formed the resistance torque (T_R). Once the torques involved had been identified, the respective forces together with their respective moment arms were specified, and the RMT was deduced based on this knowledge.

By identifying the forces acting on the forearm-hand segment (Figure 1B), and assuming the participation of only the main agonist muscles involved in the movement, i.e. by ignoring the involvement of accessory muscles and the action of any antagonist muscles acting concomitantly, equation 1 can be rewritten as shown in equation 4:

$$\vec{F}_{BB}+\vec{F}_{BR}+\vec{F}_{BC}+\vec{W}+\vec{F}_J+\vec{F}_E=0 \quad (4)$$

where,
$\vec{F}_{BB}$ is the biceps brachii muscle force
$\vec{F}_{BR}$ is the brachioradialis muscle force
$\vec{F}_{BC}$ is the brachialis muscle force

Equation 4 clearly shows the indeterminate nature of the musculoskeletal system and has no single solution. In order to solve this indetermination, all the muscle forces were grouped into a single RMF. Hence, the FDB was redrawn to represent this new approach (Figure 2).

The RMF represents the active forces originating from muscle contractions and the stretching of the passive structures that compose the muscle (fascias and tendons). It was arbitrarily located at the insertion point of the main agonist, and with the direction of action in line with the direction of the tendon of this agonist in the region of the insertion. The joint reaction force (F_J) represents the forces originating from the contact between ligaments, cartilage and the passive structures not considered among the active forces, and it was arbitrarily located at the center of the joint. As a mere representation of the effects on the FBD, the resultant joint force can be approximately located in parallel to the RMF, but acting in the opposite direction. As F_J does not exert torque (in relation to the axis of the joint), its direction is irrelevant in evaluating the balance of the torques involved. Once the solution to the equations is known, and the components of each force have been identified, the true direction can be established.

The weight force of the human body segment (W), which is gravitational in origin, was represented at the center of the mass of the forearm-hand segment, acting vertically downwards[22]. The external force (F_E) acting on the distal part of the forearm-hand segment was represented at the center of the contact region between the segment and the equipment used. The origin of this force is based on the interaction between the objects involved (forearm-hand segment and equipment), and its magnitude can be calculated from Newton's Third Law. The amount of force on the equipment is equal to the amount of the force on the forearm-hand segment, and the direction of the force on the equipment is opposite to the direction of the force on the forearm-hand segment. The force acting on the equipment can be calculated from the weight of the equipment plus the inertial effect, which can contribute towards increasing or decreasing the amount of this force, in accordance with the acceleration involved[5]. Because the inertial effects were not taken into consideration, this force was considered equal to the weight of the dumbbells.

Based on this perspective, equation 4 was rewritten as shown in equation 5:

$$\vec{F}_M+\vec{W}+\vec{F}_J+\vec{F}_E=0 \quad (5)$$

A new FDB representing this situation can be seen in Figure 2, in which the RMT has been substituted by a force acting at some distance from the rotation axis. This force is

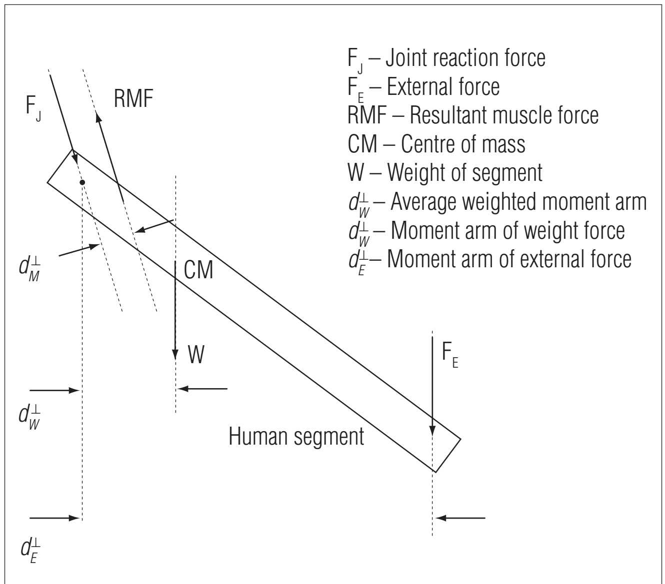

Figure 2. Free-body diagram of the forearm-hand segment during a representative exercise position showing the "virtual muscle".

the RMF and, together with its respective moment arm, can be understood as a "virtual muscle".

The moment arm of the RMF will be represented by an "average value", calculated from the moment arm of each of the agonist muscles analyzed[18,23]. The physiological cross-sectional area will be used to weigh the moment arm, because the muscles with the largest number of fibers will be liable to produce more force and, consequently, their participation in the RMT will be more significant, as shown in equation (4):

$$d_M^{\perp} = \frac{\sum_{i=1}^{n} d_i^{\perp} PCSA_i}{\sum_{i=1}^{n} PCSA_i} \qquad (4)$$

where,

$d_M^{\perp}$ is the weighted mean moment arm

$d_i^{\perp}$ is the moment arm of the i^{th} muscle

$PCSA_i$ is the physiological cross-sectional area of the i^{th} muscle

n is the number of muscles included

Thus, the RMF can be expressed by equation (5):

$$RMF = \frac{T_R}{d_M^{\perp}} \qquad (5)$$

where,

RMF is the resultant muscle force

T_R is the resistance torque

The moment arm or the weight and external force were obtained by trigonometric deduction. Considering that these forces act downwards, and with knowledge of the flexion angle, the moment arms were obtained as the product of the distance of the application point of each force (weight and external) from the axis rotation and the sine of the flexion angle.

Signal processing

Following data collection, the signals were digitally filtered using a low-pass third-order Butterworth filter at a cutoff frequency of 5Hz, for the electronic goniometer signal, and a band-pass third-order Butterworth filter between cutoff frequencies of 20 and 500Hz, for the EMG signals. The EMG signals were then submitted to a smoothing process using a mobile RMS window (envelope), in one-second windows with weighted Hamming. Prior to performing the exercises, maximum voluntary contractions (MVC) of the elbow flexors and extensors were performed at 90-degree flexion in order to normalize the EMG signal. There was a five-minute interval between each exercise in order to avoid muscle fatigue effects.

With the aid of the electrogoniometer, the EMG signal curves for each muscle, previously normalized and smoothed, were sliced from full elbow extension to maximum elbow flexion, focusing on the concentric phase of the elbow flexors. Mean values and standard errors of the 64 performances (eight repetitions by eight subjects) were plotted against the elbow flexion angle. Extreme values from each performance were eliminated in order to maintain an area of analysis common to all the subjects in all the performances. The data referring to the mass of the segments involved (forearm and hand)[24], the physiological cross-sectional area[25] and the moment arm of the muscles involved[26] were obtained from the literature.

No type of biofeedback was used to aid in controlling the execution speed of the exercises, and the experience of the subjects was considered sufficient for this purpose. Accordingly, after the signal processing, the average angular speed values were calculated. No execution speed was higher than 30º/seconds, and therefore no executions were discarded.

Statistical analysis

Two-way ANOVA including the factors of exercise and angle was used to evaluate the difference between the results for muscle force. The angle was analyzed every 10º, starting from the 20º position of flexion and going as far as 90º. The main effects were identified using the Bonferroni *post hoc* test. Normality was confirmed using the Lillifors test and homogeneity variance was confirmed by means of the Levene test. Pearson's correlation coefficient was used to evaluate the relationship between the muscle force results obtained from the model with the EMG signal from the biceps brachii. The level of significance adopted for all the tests was 1%.

Results

The torques of weight force (forearm-hand) and external force (dumbbell) values were obtained with the aid of trigonometric comparisons and, together, these formed the resistance torque. Figures 3 and 4 show these torques for the Scott and unsupported exercises, respectively. Based on equation 3, the muscle torque is equal to the resistance torque, but with opposite (negative) signs. Figure 5 shows the moment arm for the main forearm muscles, as reported in the literature[26], together with the respective weighted mean moment arm calculated in equation 4. The RMF was calculated using equation 5 and the values presented throughout the ROM are shown in Figure 6 for the Scott and unsupported exercises.

Figures 7 and 8 show the mean value and standard error of the electromyographic activity from the muscles evaluated, throughout the flexion movement for the Scott and unsupported exercises, respectively.

Table 1 shows the results from the statistical analysis of the estimated muscle force values for each exercise. There was a significant difference in the RMF between the exercises at all the angles, except at 70º. When the angles for the same exercise were compared, the RMF was different at all the angles in the Scott exercise, and at several angles in the unsupported exercise.

The correlation between RMF values and the respective EMG values from the biceps brachii muscle, obtained over the ROM, was significant ($p<0.01$) with high Pearson's coefficients, from both the Scott (0.932) and the unsupported (0.793) exercises.

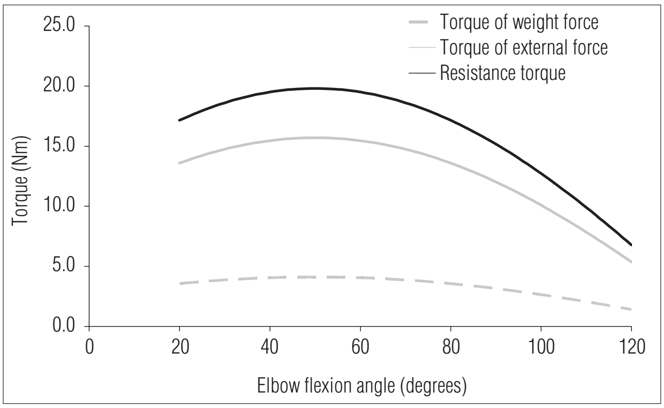

Figure 3. Resistance torque and its components throughout the ROM in the Scott exercise.

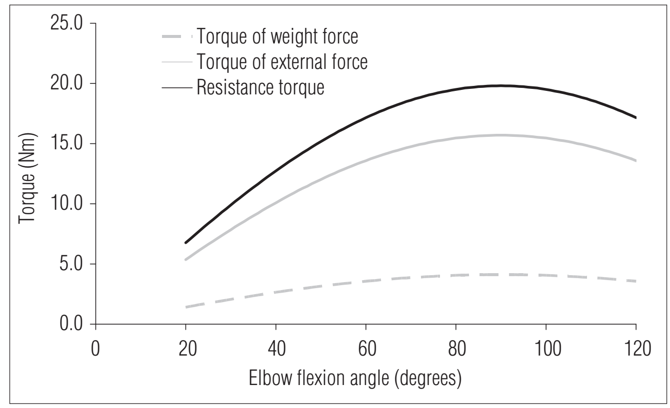

Figure 4. Resistance torque and its components throughout the ROM in the unsupported exercise.

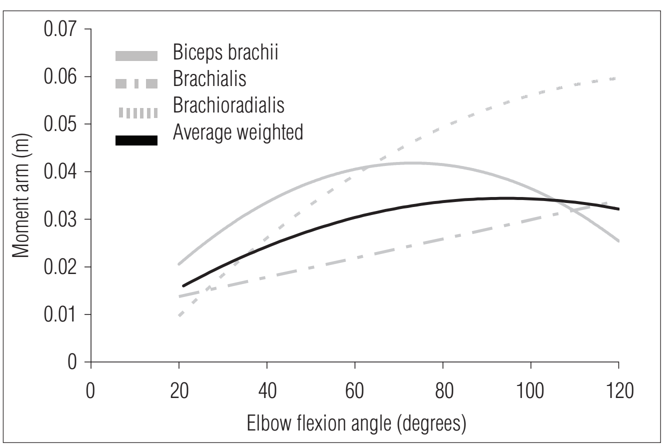

Figure 5. Moment arm for main forearm[26] muscles and respective weighted mean moment arm.

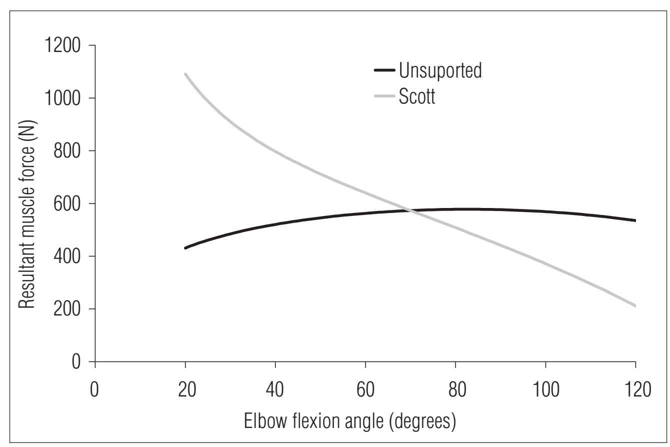

Figure 6. RMF throughout the ROM in Scott and unsupported exercises.

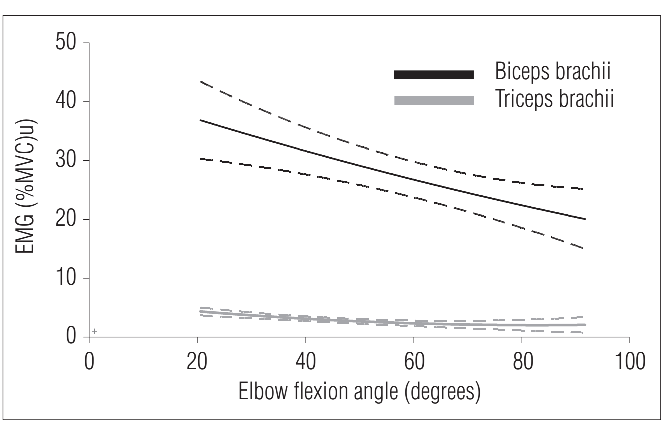

Figure 7. Mean EMG activity and standard error from the biceps brachii and triceps brachii throughout the flexion movement in the Scott exercise.

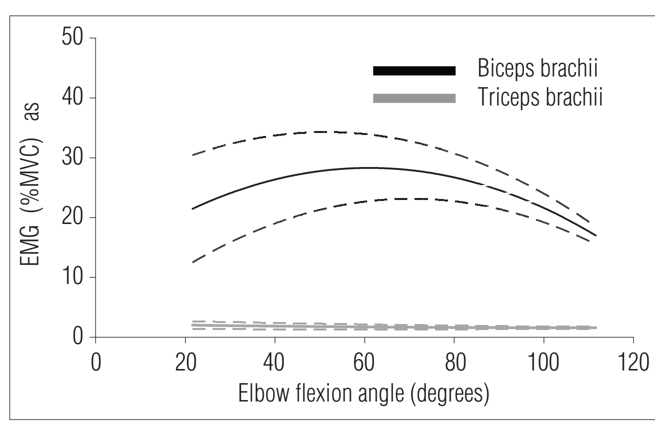

Figure 8. Mean EMG activity and standard error from the biceps brachii and triceps brachii throughout the flexion movement in the unsupported exercise.

Discussion

The proposed analytical method was based on a mechanical analysis of an exercise situation in which a resultant muscle force would be acting on a "virtual muscle" representing the muscle forces exerted by all the muscles involved in the movement. Given that the mechanical characteristics of the main agonist muscles composing the flexor group are similar among them, and the behavioral pattern of the changes to moment arm values during the ROM is the same, it is therefore possible to see the RMT in terms of a force (RMF) and a moment arm (weighted mean moment arm). This approach is useful when attempting to understand the mechanical demand imposed on flexor muscles. The absolute force values found for the "virtual muscle" are compatible with the mechanical characteristics of the problem, in which the forces composing the RMT have low values but are applied far from the rotation axis, while the RMF is very close to the rotation axis. Validation of these values would only be possible by direct measurement, which would require an invasive procedure and consequent ethical and technological complications. Hence, only an assessment of the proposed model was carried out, by comparing the results obtained using the model with those from other noninvasive instruments, as proposed by Nigg and Herzog[27]. There are some limitations to the use of surface EMG because some important agonist muscles, such as the brachialis muscle in the evaluated exercise, along with several auxiliary muscles, cannot be monitored. It is also important to note that EMG does not represent the force[28]. Nevertheless, greater normalized EMG activity indicates a greater number of recruited motor units, which is a consequence of a greater external demand required from the evaluated muscle. The agreement between the RMF and EMG curves suggests that the proposed method facilitates evaluation of the muscle demand required throughout the ROM of the exercise. The results obtained for the Scott exercise in the Pearson correlation test were higher than the results for the unsupported exercise. This was probably because the RMF and EMG curves obtained for the unsupported exercise were almost flat as there is only a small variation during the ROM.

Table 1. Mean values (± standard deviations) of muscle force in the Scott and unsupported exercises p values refer to comparisons between exercises.

Angular position (°)	Resultant muscle force (N)		p
	Scott exercise	Unsupported exercise	
20	1093.0 (±38.4)[a]	430.6 (±10.8)[a]	<0.001
30	911.5 (±21.7)[b]	484.5 (±6.8)[b]	<0.001
40	797.5 (±15.1)[c]	520.3 (±4.7)[c]	<0.001
50	712.3 (±12.2)[d]	545.5 (±3.3)[d]	<0.001
60	640.1 (±10.8)[e]	562.8 (±2.2)[e]	<0.001
70	573.6 (±10.3)[f]	573.5 (±1.2)[ef]	0.990
80	495.2 (±10.4)[g]	578.0 (±0.2)[f]	<0.001
90	441.7 (±10.8)[h]	576.6 (±0.7)[f]	<0.001

Note: different letters refer to significant differences in the same exercise.

However, the behavior of the RMF throughout the ROM is more important than the absolute values obtained, as seen in Figure 6. In the Scott exercise, the behavior of the RMF declines notably with increasing ROM, with significant differences at each 10º of amplitude (Table 1). In the unsupported exercise, the behavior of the RMF remains almost constant throughout the ROM, and the significant differences found were not important from a physiological point of view. These differences may occur due to the small standard deviation of the measurements, which is characteristic of data extracted from models.

It is noteworthy that maximum muscle force is not necessarily required when external torque reaches its maximum. In the Scott exercise, the maximum external torque was reached at 50º of flexion (Figure 3), while the maximum muscle force was reached at the beginning of the exercise (Figure 7). In contrast, in the unsupported exercise, the maximum external torque was reached at 90º of flexion (Figure 4), while the maximum muscle force was reached around 60º of flexion (Figure 6). The divergence between the required maximum muscle torque and the muscle force is due to the weighted mean moment arm, in which there is increased behavior relating to increased angular flexion.

Contraction of the antagonist muscle groups occurring simultaneously with the action of agonist muscle groups (whether in order to control movement or as a result of lack of coordination by the individual making the movements) will increase the resistance torque, thereby increasing the muscle torque. When only the agonist muscles are taken into consideration, the method tends to underestimate the results of both RMF and F_J. This was not the case in the evaluated exercises because it was possible to confirm the low activation of the triceps throughout the ROM in both exercises (Figures 7 and 8). Nevertheless, even with this limitation in regards to absolute values, the behavioral pattern of the RMF during the movement can be used to obtain a wider perspective on the effect of exercise on the musculature.

From a therapeutic point of view, knowledge of external torque behavior and its influence on the musculature is fundamental to the success of any treatment. While the proposed method does not allow an objective statement of the total force acting on each muscle, knowledge of which region within the ROM gives rise to the greatest demand contributes towards the safety of the chosen exercise, thereby helping to keep the effects of the external load on the structure under recovery within safe limits. Considering the adaptation of the muscle to the increasing load as a criterion, an exercise sequence might be established. The findings from the present study suggest that the ROM of the Scott exercise should be divided into two parts, with the latter half performed first, followed by the full unsupported exercise, and finishing with the first half of the Scott exercise. It is important to remember that the magnitude of the F_J is closely associated with the magnitude of the RMF (equation 5). Accordingly, the most extended positions of the Scott exercise should be avoided in those cases in which the joint is also undergoing a process of recovery. Similarly, the most flexed positions of the Scott exercise can be used to work on the joint ROM, thus minimizing the risk of overloading the joint and muscles.

It is important to note that the mechanical properties of muscles, such as pennation angle or force-length and force-velocity relationships, were not taken into consideration, because the focus of this study was not to identify when the muscle was best able to exert force, but at which point in the ROM the highest demand was made on the muscles, due to the external load. This aspect is particularly important when dealing with the recovery of injured athletes. The use of the proposed method in such cases could be a useful aid when selecting and evaluating the exercises performed by individuals following rigorous training programs that involve specific requirements from the muscles relating to that particular sport.

In summary, the resistance torque behavior is not necessarily reflected in the musculature involved. An increase in resistance torque may not necessarily represent an increase in the muscle force involved in the exercise. The final effect on the muscles involved should be taken into consideration when choosing an exercise and its variances. Exercises performed for therapeutic purposes are normally carried out at low speeds, and in this case, the inertial effects may be negligible. These inertial effects have not been evaluated in the present study, and their quantification and possible therapeutic repercussions require further study.

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