A PHYSIOLOGICALLY BASED CRITERION OF MUSCLE FORCE PREDICTION IN LOCOMOTION*

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Abstract—The inversely-nonlinear relationship of muscle contraction force and the possible contraction duration (i.e., endurance) is utilized in a method to mathematically predict individual muscle forces. The method uses a nonlinear optimization scheme to solve the redundant distribution problem at the joints of interest. The method is demonstrated at the elbow during isometric contraction and in the lower extremity during locomotion. During gait, the observed muscle activity pattern in the lower extremity (as determined by EMG) shows substantial agreement with that activity pattern predicted when endurance is used as the optimization criterion. The importance of selecting muscle prediction criteria based on sound physiological bases rather than on an arbitrary or mathematically convenient basis is emphasized.

INTRODUCTION

Musculoskeletal function results from complex neurological, muscular and skeletal interactions. Nearly all joints of the body are crossed by a large number of musculotendinous units. In many instances, these musculotendinous units possess both an apparent diversity and a redundancy of function. The rationale by which the muscles act at any instant during body function, is not readily apparent. Considering the diversity of musculoskeletal function, it is likely that a number of distinctly different criteria for muscle selection may be utilized for different activities. Interest in basic locomotion mechanics, musculoskeletal pathomechanics and the mechanics of treated musculoskeletal problems have resulted in many efforts to quantitatively predict muscle activity (Seireg and Arvikar, 1973 and 1975; Chao and An, 1976; Crowninshield, 1978; Crowninshield et al., 1978; Hardt, 1978; Pedotti et al., 1978). Although not explicitly stated in these reports, linear optimization was used perhaps more for reasons of mathematical convenience than for reasons of physiological requirement. Pedotti et al. (1978) reported a nonlinear optimization technique which produced patterns of muscle activity that agreed more closely with EMG recordings than did contrasting linearly optimum solutions. Pedotti et al. did not, however, offer any physiologically based argument for their reported nonlinear optimization criteria.

All optimization procedures, whether linear or nonlinear, require the assumption that the body selects muscles for a given activity according to some criterion (e.g. minimization of muscle force). The previously mentioned investigations used a variety of criteria ("objective functions" or "cost functions") with muscle activity predictions of varying degrees of reasonableness.

This paper presents a quantitative method of muscle activity prediction which uses a criterion of maximum endurance of musculoskeletal function. The method is based on the inversely nonlinear relationship of muscle force and contraction endurance. We suggest the muscle selection, so as to maximize activity endurance, is physiologically reasonable during many normal activities, particularly prolonged and repetitive activities, such as normal gait. The criterion is not applicable to all forms of locomotion (e.g. activities occurring to maximize speed or to minimize pain). The results obtained through the use of this criterion agree more closely with known patterns of muscle activity, as revealed by EMG, than most linear optimization techniques.

METHOD

Grosse-Lordemann and Muller (1937) first proposed a quantitative relationship between a muscle's

* Received 26 December 1980; in revised form 12 June 1981.
contractile force and the maximum time for which the contraction could be held. This muscle force–endurance relationship was proposed to be of the form

$$\log T = -n \log f + c$$

(1)

where $T$ is the maximum time of contraction, $f$ is the contractile force, and $n$ and $c$ are experimentally obtainable constants. More recently, several experimental efforts characterized the constants in the proposed muscle force–endurance relationship (Gunnar, 1963; Kajiser, 1970; Herrmiston and Bonde-Petersen, 1975; Dons et al. 1979). These experiments involved a variety of human muscular activities in both static and dynamic situations. The estimates of the constant $n$ resulting from these studies vary from 1.4 to 5.1 with an average of about 3.0. It is thus the conclusion of these reports that muscle contraction endurance time is inversely related to contraction force raised to a power.

Most of the experiments conducted to estimate the constants in equation 1 have involved subjects possessing a variety of physical capabilities. If the force ($f$) expressed in equation 1 is the force (in an absolute sense) that an individual exerts during an activity, then individuals of considerably different force exertion capabilities will correspondingly yield different values for the constants in equation 1. Dons et al. (1979) characterized this force–endurance relationship with the value of force in equation 1 normalized to each individual's maximum force exertion capability. In this "relative strength" endurance expression the values of $n$ (equation 1) ranged from 2.54 to 3.14. This normalized endurance relationship presumably applies to individuals, independent of their maximum force exertion capability.

Fick (1910) presented the concept that a muscle's physiologic cross-sectional area is linearly related to the muscle's maximum force exertion capability. Subsequent researchers agreed that a muscle's gross size and force exertion capability must, in some manner, be related. Although not precisely defined, physiologic cross-sectional area is now generally taken to be the muscle's volume divided by its length (an average cross-sectional area). The constant of proportionality between muscle maximum force exertion capability and cross-sectional area has been reported by Fick (1910), Recklinghausen (1920), Morris (1948), Iki and Fukunaga (1968). This constant of proportionality, with estimates ranging from about 0.4 to 1.0 MN/m², is thus a maximum stress that the muscle tissue is capable of generating.

Dons et al. (1979) showed that the muscle force–endurance relationship (equation 1) proposed by Grosse-Lordemann (1937) can be normalized to an individual's force exertion capability. Fick (1910) and others have reported that individual muscle force exertion capability can, at least in an approximate manner, be related to muscle cross-sectional size through a constant of proportionality with the units of stress. Based on these reports, we assume that in an approximate manner the muscle force–endurance relationship of equation (1) is a basic property of muscle tissue. We suggest that the maximum endurance of a muscle contraction is thus related to the magnitude of the average stress within the muscle tissue. At an arbitrary joint within the body, where ligament and articular surface contact forces are assumed to insignificantly contribute to the intersegmental resultant moment ($M$), a maximum endurance of function will occur when

$$\bar{M} = \sum_{i=1}^{m} (f_i \times \bar{f}_i)$$

(2)

such that the Euclidean (or $L_2$) norm

$$u_i = \sqrt{\sum_{j=1}^{n} (f_j/A_i)^2}^n$$

(3)

is minimized. This norm is minimized when the radicand (summation of the $n$th power of muscle stresses) is minimized in equation 3. In equations 2 and 3, $m$ is the number of muscles crossing the joint, $f_i$ is the tensile force in the $i$th muscle, $A_i$ is the average cross-sectional area of the $i$th muscle, and $n$ is the exponent in the muscle endurance–force relationship. Normalization is not theoretically required but is carried out for 2 practical reasons: (1) Normalization of the objective function causes the objective function to have the same units (Pa) as the state variables (muscle stresses). (2) Normalization greatly reduces the magnitude of the objective function and avoids some numerical problems in large scale optimization. For simplicity the norm of the objective function will not be used in the following discussion. The determination of muscle force during body function may then be formulated as a nonlinear optimization problem with an objective to minimize the summation of muscle stress to the $n$th power. Muscle forces predicted in this manner will tend to keep individual muscle stresses low. Low individual muscle stresses are achieved by predicting force activity in numerous muscles and preferentially predicting force in muscles with large cross-sectional areas and long moment arms. Precipitating muscle forces to minimize this objective or cost function (equation 3) coincides with achieving maximum endurance of activity as defined by equation 1 (i.e., individual muscle stresses will be low and thus their potential for prolonged contraction will be high). Although the appropriate power ($n$) is not known exactly, experimental studies clearly show that it is not unity (in which case the force and endurance would be a simple inverse relationship, thus making linear optimization appropriate). The actual value of $n$ may vary between individual subjects and individual muscle (due to fiber type, fiber orientation, etc.). If accurate and detailed experimental data were available (unfortunately it is not), it would be possible to specify a value of $n$ for each muscle. At this juncture the use of $n = 3.0$, the average value reported in the literature, has been selected as a reasonable value. The relationship of muscle stress and...
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Fig. 1. The general form of the proposed muscle stress-endurance relationship. Relatively high muscle stresses can be sustained for only short durations while lower muscle stresses can be sustained longer. In a cost function sense, high muscle stresses are costly.

maximum endurance of contraction is thus assumed to be of the form of Fig. 1.

This criterion for muscle force prediction was employed in two examples of musculoskeletal function. The first was a simple planar model of the elbow. Muscle forces were predicted at the elbow such that the muscle action generated the elbow resultant moment

\[ M = 10 \text{ Nm} = \sum_{i=1}^{3} r_i \times f_i, \quad (4) \]

and

\[ u = \sum_{i=1}^{3} (f_i/A_i)^n \quad n = 1, 2, 3, 4 \text{ and } 100. \quad (5) \]

was minimized. At approximately 90 degrees of flexion, the moment arms of three elbow flexors (\( r_i \) in equation 4) were estimated from a cadaver arm X-ray. The average muscle cross-section area (\( A_i \) in equation 5, muscle volume/muscle length) was calculated from muscle measurement in situ and water emersion volume measurement (Table 1). Additional muscles are obviously involved in elbow function and all muscles crossing the elbow have three-dimensional function. We chose this simple model not to comment upon elbow mechanics but rather to provide a simple example of various optimum solution characteristics. Restricting this model to three muscles permitted convenient three-dimensional graphic illustration.

The second example considered was a complex musculoskeletal model of the lower extremity during gait, similar to that reported by Crowninshield et al. (1978). The muscle model contains 47 musculotendinous elements, each with an origin, insertion, and average cross-sectional area determined by cadaver study. The anatomic data was obtained from the dissection and analysis of six cadavers' extremities representing a range of adult body sizes. Through normalization to anthropometric measures the cadaveric data was tailored to individual gait analysis subjects. A sagittal plane midswing view of the muscle model is shown in Fig. 2. Gait kinematics and kinetics were obtained by methods previously reported (Crowninshield et al., 1978). Similar to other anatomic models (Seireg and Arvikar 1975; Crowninshield et al. 1978; Hardt 1978; Pedotti et al. 1978), ligament function at the hip is assumed to be negligible, while at the knee and ankle, the ligaments and articular surfaces are assumed to constrain varus-valgus and internal-external rotation. The model's 47 muscle elements are thus required to generate three orthogonal components (\( j \)) of intersegmental resultant moment at the hip and 1 component (flexion-extension) at the knee and at the ankle. Muscle forces during gait were predicted, such that

\[ \bar{M}_j = \sum_{i=1}^{21} r_i^j \times f_i, \quad j = 1, 2, 3 \]

\[ \bar{M}_3 = \sum_{i=25}^{37} r_i^3 \times f_i, \]

\[ \bar{M}_3 = \sum_{i=36}^{47} r_i^3 \times f_i, \quad (6) \]

and

\[ u = \sum_{i=1}^{47} (f_i/A_i)^3 \]

was minimized. \( \bar{M}_h, \bar{M}_h, \bar{M}_e \) are respectively three components of intersegmental resultant moment at the hip, and one component each at the knee and ankle and \( \bar{r}_j, \bar{r}_j, \bar{r}_j \) are muscle moment arms about the hip, knee and ankle respectively. In this model the hip is crossed by 31 muscles, the knee by 13 muscles, and the ankle by 12 muscles.

The determination of individual muscle forces requires the solving of a specially structured nonlinear optimization problem. The objective functions utilized are of a general nonlinear form (equation 3) and the solution space is defined by linear constraint functions (i.e. Equation 2). The solutions to the examples in this paper were obtained using Rosen's gradient constraint algorithm for linear constraints (Haug and Arora, 1979). In general, nonlinear optimization convergence on a global minimum is not assured. However, the present problem due to the continuous convex character of the objective function (equation 3) and the linear constraints, falls into the category of convex programming. This convexity assures that the only minimum is a global or absolute minimum (Haug and Arora, 1979).

Table 1. Elbow muscle data

<table>
<thead>
<tr>
<th>Muscle</th>
<th>Moment arm</th>
<th>Average cross-sectional area</th>
</tr>
</thead>
<tbody>
<tr>
<td>Biceps</td>
<td>0.052</td>
<td>( 7.8 \times 10^{-4} )</td>
</tr>
<tr>
<td>Brachialis</td>
<td>0.025</td>
<td>( 8.4 \times 10^{-4} )</td>
</tr>
<tr>
<td>Brachioradialis</td>
<td>0.064</td>
<td>( 4.7 \times 10^{-4} )</td>
</tr>
</tbody>
</table>
RESULTS

The simple planar model of the elbow with 3 potentially active flexor muscles demonstrates the behavior of optimization solutions obtained for indeterminant problems of muscle force prediction. Individual muscle forces were predicted using equations 2 and 3 ($M = 10 \text{ Nm}$), and incorporating values of $n$ of 1, 2, 3, 4 and 100 (100 numerically approximating infinity). The solution space for this problem can be conveniently illustrated as a plane in three-dimensional space, with each axis representing the stress in one of the three muscles (see Fig. 3). All feasible solutions or possible combinations of tensile muscle stresses acting to generate the required joint moment, are contained on the planar solution space surface. The linearly optimum solution, of the form of equation 3 with $n = 1.0$, lies on the solution space corner where only one elbow flexor muscle, the biceps (that muscle with the largest product of moment arm and cross-sectional area), is predicted to be active. As the limit of $n$ approaches infinity, equation 3 becomes the maximum norm and the resulting optimum solution approaches a point on the solution space where a line from the origin which forms equal angles to each axis intersects the solution space. This optimum solution, $n$ equal to infinity, has equal stresses acting in each muscle, that is, muscles acting proportional to their average cross-sectional area. All other optimum solutions corresponding to $n$ greater than 1 and less than infinity, fall along a straight line between these solutions. All nonlinearly optimum solutions result in the prediction of simultaneous activity in more than one elbow flexor muscle. Other optimization criteria, either linear or nonlinear, will result in solutions lying elsewhere on the solution space. Linear solutions will reside at solution space corners and nonlinear solutions will occur on the interior of the solution space.

The optimum solution path shown in Fig. 3 contains only the optimum solutions of the form of equation 3.

The occurrence of an optimum solution can be conveniently demonstrated graphically in a planar problem where 2 unknown muscle forces act to generate a single component of moment. One such problem can be reasonably approximated by considering the fully pronated forearm during elbow flexion. The forearm pronation effectively prevents significant activity in the biceps (Basmajian and Latif, 1957), requiring that forced elbow flexion be produced by the brachialis and brachioradialis alone. In Fig. 4 the horizontal axes represent the stress present in these 2 muscles while the vertical axis represents values of the objective or cost function (equation 3 with $n = 3.0$). A plot of the cost for any combination of muscle stresses produces a curved cost-solution surface. The constraint that a combination of muscle force must generate a particular moment at the elbow, equation 2, can be illustrated as a vertical plane, i.e., a constraint plane. All feasible solutions (i.e., combinations of muscle forces satisfying the constraint) lie along the curved line of intersection between the cost-solution surface and the constraint plane. The optimum solution (i.e., minimum cost) occurs at the lowest point on this intersection line.

In a lower extremity model this optimization procedure (equations 6 and 7 with $n = 3.0$) is used to quantitatively determine muscle activity in 47 muscle elements during gait (Fig. 5). The lower extremity muscle model subdivides several muscles (adductor magnus, gluteus maximus, gluteus minimus, and gluteus medius) into separately acting muscle elements. The activity predicted in these muscle elements are combined to represent a total muscle function in Fig. 5.

Rectified and integrated (time constant equal to 20 msec.) EMG was collected on the same subject whose gait kinematic and kinetic data were utilized in the muscle force prediction. The EMG was measured using surface electrodes at locations suggested by Delagi et al. (1980). A relative insensitivity to $n$ values greater than 2 is demonstrated in Fig. 3. There is a large change in muscle stresses between $n$ equal to 1.0 and $n$ equal to 2.0. A similar occurrence can be demonstrated in the more complex lower extremity model. Figure 6 demonstrates changes in the articular surface contact force resultant at the hip calculated from muscle forces.
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Solution

Ti = \frac{T_i x_i}{10NM}

Fig. 3. Solution space for three flexors acting to generate 10 Nm moment at the elbow. The linear constraint (moment at the elbow) causes the solution space to be planar. The optimum solutions shown are those associated with the minimization of the summation of muscle stresses raised to a power.

predicted with \( n \) equal to 1, 2, 3, and 5. The optimum solution for \( n \) equal to 1.0 occurs when total muscle stress is minimized by predicting relatively high forces in a few large muscles with long moment arms. The resulting contact force at the hip is relatively low. When the objective function involves the sum of muscle stresses raised to a power (2, 3, and 5) an optimum solution is predicted with lower individual muscle stresses and a higher number of active muscles, some with relatively short moment arms. The resulting contact force at the hip is higher.

DISCUSSION

Several optimization techniques to predict muscle forces have been suggested. These techniques were most commonly applied to lower extremity function (Seireg and Arvikar, 1973 and 1975; Crowninshield et al., 1978; Hardt, 1978; Pedotti et al., 1978). An examination and comparison of these techniques reveals several characteristic traits and limitations of optimum solutions.

Seireg and Arvikar first reported the use of optimization in determining force in lower extremity muscles. They presented several candidate linear optimization criteria (Seireg and Arvikar, 1973) and applied a preferred criterion to quasi-static locomotion (Seireg and Arvikar, 1975). This work employed a lower extremity muscle model of 31 elements. During walking, simultaneous activity was predicted in only five to eight muscle elements. The prediction of few simultaneously active muscles resulted from the requirement, inherent with linear optimization, that the solution resides at a solution space corner. The maximum force predicted in an individual muscle occurring gait was about 2100 N (in the tibialis anterior, a small muscle), illustrating that, unless restrictions are imposed on maximum muscle forces, linear optimization will predict unrestricted and possibly unreasonably large forces in the single most advantageous muscle. Figure 6 demonstrates that joint contact forces calculated by these linear optimization techniques are unrealistically low.

In an effort to predict muscle activity which was more reasonable, these authors (Crowninshield, 1978; Crowninshield et al., 1978) reported a procedure in which inequality constraints were imposed on linearly optimum solutions. The solutions were more realistic than previously reported solutions from 3 standpoints: (1) unreasonably large or small muscle force predictions were avoided. (2) more simultaneous muscle activity was predicted, (3) the muscle activity predictions temporally correlated much better with EMG activity. However, no physiological argument
Fig. 4. The occurrence of an optimum solution to a model of two elbow flexors satisfying a constraint moment of 10 Nm.

was presented for the imposition of the inequality constraints. The constraints were simply empirically imposed to improve the solution and not imposed because the constraints had a physiological analog.

Hardt (1978), in a review of optimization procedures applied to the lower extremity during gait, demonstrated some of the inherent limitations of linear optimization. For example, the number of muscles with some predicted activity would be limited to the total number of constraints (equality plus inequality), a clearly artificial restriction. The possibility of predicting exceedingly large force in inappropriately small muscles (tensor fascia lata) and little or no force in large muscles (adductor magnus, gastrocnemius, gluteus medius) was demonstrated with a minimum force criterion. Furthermore, poor agreement between EMG and force predictions was demonstrated with several criteria. Hardt concluded that attention needed to be directed at criteria with known physiological analogs and suggested the use of thermodynamic descriptions of muscle function.

These initial efforts to predict muscle forces through the use of optimization were restricted to the use of linear optimization criteria. The techniques were used perhaps more for numerical convenience than for any compelling belief in a physiologic basis for linear optimization. There are well-established algorithms producing efficient and stable solutions for linear optimization. General nonlinear optimization algorithms are less common and the convergence of the solution on a global minimum is not assured (Haug and Arora, 1979). The nonlinear optimization criterion of Pedotti et al. (1978) was a significant advance. These investigators compared several linear optimization criteria with nonlinear criteria in a sagittal plane gait analysis. The preferred objective function, on a basis of temporal correlation with EMG, was the sum of muscle stresses squared. Although the results of this nonlinear optimization were more reasonable (based on EMG activity) than previous results, no physiological argument was presented for the criterion.

Since the Weber brothers (1836) suggested that we walk in some manner that minimizes "energy" (or conversely to maximize endurance) researchers have generally accepted this idea. While the belief is based more on intuition than scientific proof, it remains a reasonable hypothesis (and in any case has not been disproven). The criterion suggested here is based on a known physiological principle (muscle force-endurance relationship). While the specific quantitative characteristics of this relationship may be
debated, the general nonlinear characteristic has been established (Grosse-Lordemann et al., 1937). The use of this criterion predicts muscle forces during gait which are far below the likely maximum attainable forces in each muscle, since endurance is increased at low contraction levels (Fig. 1). Through the use of this criterion, intuitively reasonable muscle force predictions are obtained for the activity of normal gait. Furthermore, there is substantial agreement between the temporal aspects of muscle activity prediction and EMG pattern (Fig. 5). In interpreting the EMG data of this report, it is inappropriate to associate EMG signal magnitude in any specific way to muscle force magnitude. Since signal strength in EMG activity is related to many factors (e.g., skin preparation, subdermal fat, subdermal muscle depth, and contractile velocity) its presence should only temporally be associated with muscle force predictions.

Caution must be exercised in drawing conclusions about the validity of an optimization criteria based solely on muscle activity predictions even when temporally correlated with EMG. We have demonstrated that nonphysiological criteria (and in fact arbitrary mathematical criteria) (Crowninshield, 1978) can predict reasonable muscle activity. Therefore, good agreement in muscle force predictions and EMG activity cannot alone be considered proof that the optimization criteria selected is the criteria the body uses. However, it seems more reasonable to draw such inferences when there is good agreement in force predictions and EMG, when the force predictions generally correlate with muscle size, and when the criterion is based on recognized, experimentally proven principles (i.e., when the criterion has a physiological analog). A physiological criterion is necessarily one which requires no arbitrary upper limits of individual

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**Fig. 5.** Predicted muscle forces during gait and rectified and integrated surface EMG activity for a normal subject. EMG activity is reported only for temporal correlation with muscle forces. EMG signal magnitude should not specifically be compared to predict muscle force magnitude. The EMG signal shown corresponding to the semimembranosus and semitendinosus is medial hamstring EMG recording and is most likely generated by the action of both muscles.
muscle force. Even so, the ultimate proof that a selected objective function is indeed the one used by the body would be difficult even if the technologies to directly validate muscle force predictions (such as tendon or muscle force transducers or quantitative EMG) were available, which they are not.

It must also be emphasized that while the criterion of maximum endurance might be reasonable for an activity such as walking at a comfortable pace (in which endurance is great), it might not be reasonable for other activities such as climbing stairs, sprinting, or pathologic gait. In these cases, the body may select very different criteria. For example, patients with painful arthritic weight-bearing joints exhibit an antalgic gait in which little time is spent in weight bearing. If joint loading is related to the degree of pain, then we could hypothesize that muscles are selected to minimize the vector sum of muscle forces since the resultant muscle force vector contributes significantly to the joint contact force. This vector sum is a nonlinear function of muscle force magnitudes.

Using general nonlinear procedures, a variety of hypothesized objective criteria can be tested in many differing normal and pathologic situations. It must be emphasized, however, that the criteria for a specific situation is best selected on a sound physiological basis rather than selected arbitrarily or for mathematical convenience. It is proposed in this report that muscle force prediction such as to minimize the sum of muscle stresses to the third power will result in predicted muscle action that maximizes activity endurance. The power of 3.0 was chosen because it appeared to represent the consensus of values reported in the literature; however, it is arguable that another similar value may be more appropriate (e.g., 2.0 or 4.0). Although it is our best estimate that a power of 3.0 is appropriate, the patterns of muscle force prediction are not very sensitive to small changes in this value. In the example of elbow muscle function, changing $n$ from 2.0 through 4.0 had no effect on the number of predicted active muscles (Fig. 3). The changes did, however, change slightly the predicted values of individual muscle forces. The resulting individual muscle force predictions with $n$ equal to 2, 3, and 5 causes a small change in predicted hip contact force during gait (Fig. 6). The use of a power of 2.0, as reported by Pedotti et al. (1978), may for many purposes be adequate, and offers the advantage of permitting the use of quadratic programming in place of more general nonlinear programming.

Acknowledgement—Supported in part by NIH grants AM14486 and AM00712; Zimmer, U.S.A.; and the Hearst Family Foundation.

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