Examining the Role of the Gastrocnemius During the Leg Curl Exercise

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This study examined the moment-producing capabilities of the gastrocnemius during isokinetic knee flexion tasks. Nine healthy men were tested using a Biodex isokinetic dynamometer. Each one completed 3 maximum repetitions at 3 angular velocities, 30, 75, and 150°/s, with his ankle braced in either full dorsiflexion or full plantar flexion. A computer model was used to simulate the experimental tasks. Experimentally, the moment produced at the knee joint with the ankle dorsiflexed was significantly higher than the moment with the ankle plantar-flexed at all 3 angular velocities, \( p < 0.05 \). This suggests that lengthening the gastrocnemius allowed for greater contribution of the gastrocnemius to the total moment produced at the knee during isokinetic knee flexions. The simulations supported the experimental data and suggested that, with the ankle dorsiflexed, the gastrocnemius acts on a more favorable part of the muscle’s force-length curve compared with the plantar-flexed condition. The results of the experimental work, along with the simulations, demonstrated that lengthening the gastrocnemius significantly increased the moment produced at the knee joint during isokinetic knee flexion tasks. These results have implications for instructions given to persons who perform leg curls for muscle strengthening, and for the design of knee flexion exercise machines.

Key Words: biarticular, muscle, knee flexion

Introduction

Many exercises used for strength training isolate muscle groups by attempting to allow motion at one joint only. With free weights such an isolation can be hard to achieve, but many resistance training machines are designed so that the muscles crossing one joint are the only ones that can contribute to the motion. It has been demonstrated that the force a muscle can produce changes with its length (Gordon, Huxley, & Julian, 1966). In vivo, these length changes are associated with changes in the joint angles of the joints which the muscles cross.

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A number of muscles in the human body cross more than one joint; typically these are biarticular muscles and therefore their length is influenced by two joint angles. The moment generated by a biarticular muscle at the exercising joint will in part depend on the angle of the nonexercising joint. For example, knee angle can influence the plantar-flexion moment due to the biarticular gastrocnemius (Sale, Quinlan, Marsh, McComas, & Belanger, 1982). Adjustments of the latter angle can have an influence on the execution of the exercise. The aim of this study was to examine the influence of the nonexercising joint angle on the role of a biarticular muscle during a strength training exercise.

The biomechanics of strength training exercises have been the focus of attention in studies ranging from an examination of the kinematics of the exercise (e.g., Lander, Bates, Sawhill, & Hamill, 1985) to the estimation of the loads acting at the joint during their execution (e.g., Cappozzo, Felici, Figura, & Gazzani, 1985). A detailed examination of the literature did not reveal any studies that examined how biarticular muscles contribute to the execution of single-joint exercises. With the plethora of specialized equipment that is designed to isolate muscle groups, it could prove beneficial to understand better how the entire system, exercising and nonexercising joints, work together to create a moment about a given joint.

Leg curl machines are a popular way of training the knee flexors (hamstrings and gastrocnemius). Most of these leg curl machines do not fix the ankle joint during execution of the exercise, so the exerciser is free to select the ankle joint angle. Theoretically the exerciser can adjust his or her ankle angle and change the contribution of the gastrocnemius to the knee flexion moment. It was the purpose of this study to examine the role of the gastrocnemius during isokinetic leg curls. Through experiments, we examined whether the contribution of the gastrocnemius to the moment at the knee was changed by manipulating its length, that is, by having the ankle either fully plantar-flexed or dorsiflexed. The task was simulated using a model to examine whether any differences in the moments could be explained by changes in the kinematics of the gastrocnemius during knee flexions with the ankle angle varied.

**Methods**

To examine the contribution of the gastrocnemius to the leg curl exercise, the participants performed leg curls with their ankle joints at two different angles. By changing the ankle angle, it was assumed that the length of the gastrocnemius would be changed and therefore its ability to generate force. The following describes the experimental protocols as well as the development of a model for further examining the activity.

Nine men volunteered to participate in the study (age 26.0 ± 4.4 yrs, mass 97.0 ± 11.7 kg, height 1.851 ± 0.047 m). All were actively involved in resistance training programs, which were discontinued for at least 48 hours prior to the beginning of testing. The participants reported no past history of injury to the lower extremities. They all provided informed consent, and the Institutional Review Board of The Pennsylvania State University approved all procedures.

The participants performed a series of isokinetic leg curls using a Biodex dynamometer (Model II) beginning with the knee fully extended. The dynamometer was calibrated before and after each data collection session. All dynamometer
output was corrected for the moment due to gravity (Herzog, 1988). Prior to testing, each participant was allowed to warm up on a cycle ergometer until he felt prepared to undergo testing, at which stage he was free to perform as many warm-up trials on the dynamometer as he felt were needed. During all trials, the participants received verbal encouragement and visual feedback. At least 2 minutes of rest was provided between trials to reduce any influence of fatigue.

Isokinetic knee flexion was performed at three angular velocities: 30, 75, and 150º/second. Participants performed 3 maximal repetitions at each velocity with their dominant leg. The range of motion was set for each individual and corresponded to his comfortable range of motion. The participants were secured to the dynamometer to eliminate any effects of extraneous movements on the results (Weir, Evans, & Housh, 1996). The dynamometer position was manipulated so that the axis of rotation of the knee joint was aligned with the axis of rotation of the dynamometer. All data were collected with the hip at 105º of flexion (see Figure 1 for the definition of all joint angles).

To change the potential moment contribution of the gastrocnemius at the knee, we manipulated its length by having the ankle either fully plantar-flexed or dorsiflexed. Therefore, with the ankle plantar-flexed throughout the isokinetic knee flexion, the muscle was shorter than during the trials with the ankle dorsiflexed. Due to the force-length properties of the gastrocnemius (Out, Vrijkotte, van Soest, & Bobbert, 1996), the assumption is that the muscle will have different force-producing capacity under the two conditions, thus producing different contributions to the knee flexion moment. The ankle was secured in either full plantar flexion or full dorsiflexion using aquaplast splinting materials (Smith & Nephew Inc., Germantown, WI) that were fitted to each participant immediately prior to the testing period for that specific ankle angle.

To ensure minimal movement of the ankle joint during the testing, each aquaplast splint was melted in water at 180 ºF until the material became transparent and could be easily molded. The splint was fitted to the anterior of the lower leg from the proximal joint of the metatarsals to approximately 3 inches below the patella, for each testing angle. It was then taped to the participant’s lower leg, which was then immediately placed in an ice bath to solidify the mold. The ankle
position to be tested first was randomly assigned among participants to minimize
the influence of fatigue and growing familiarization with the protocols on the group
data. The splinted ankle angle was measured using a goniometer.

For the tests of the knee flexors under each condition, the trial with the high-
est peak moment was selected for further analysis. As joint angle changes, so does
the length and moment arm of the muscle; therefore the peak knee moment at a
knee angle of 130° was used for obtaining criterion moment data. At this knee
angle, the joint had achieved the required isokinetic angular velocity for all test
conditions for all participants.

To evaluate the statistical differences between the isokinetic knee flexion
with the ankle either plantar-flexed or dorsiflexed, we used a two-way repeated-
measures analysis of variance (ANOVA). We did this to compare the moments
produced at a knee angle of 130° under the 6 conditions (2 ankle positions and 3
angular velocities). Post hoc comparisons, where appropriate, were made using a
Bonferroni test. For all statistical comparisons, a significance level of 0.05 was
used. Prior to performing the ANOVA, we used a Bartlett test to confirm homoge-
neity of variance for the data.

A model of the gastrocnemius was used to elucidate the role of this muscle
in isokinetic knee flexions. The muscle model is phenomenological in nature, but
broadly speaking, the contractile component represents the action of the muscle
fibers while the series elastic component represents the tendon. The components
of the model and the simulation process is a slight variant of that presented by
Challis and Kerwin (1994). The force produced by the muscle model ($F_M$) is de-
pendent on four factors which are represented in the following equation,

$$F_M = q \cdot F_{\text{max}} \cdot F_L(L_F) \cdot F_V(V_F)$$

(1)

Where

$q$ = current active state of muscle model,
$F_{\text{max}}$ = maximum isometric force possible by muscle model,
$F_L(L_F) = \text{fraction of normalized force-length curve that the model can produce at its current fiber length ($L_F$)},$
and $F_V(V_F) = \text{fraction of normalized force-velocity curve that the model can produce at its current fiber velocity ($V_F$)}.$

The normalized force-length properties, Figure 2a, of the muscle model were rep-resented by

$$F_L(L_F) = 1 - \left( \frac{L_F - L_{\text{OPT}}}{wL_{\text{OPT}}} \right)^2$$

(2)

Where

$L_{\text{OPT}} = \text{optimum length of muscle fiber, which is the length at which the fibers can produce their maximum force},$
and $w$ is a parameter indicating width of the force-length curve.

The force-velocity properties, Figure 2b, were based on the equation pre-sented by Hill (1938) for a concentric muscle action, and by Fitzhugh (1977) for an eccentric muscle action:
In series with the contractile component is an elastic component. The model of this element assumes that the tendon has a linear stress-strain curve (Figure 2c). The force-extension curve of this element is represented by

\[
L_T = L_{TR} + \frac{c}{F_{MAX}} \cdot L_{TR} \cdot F_M
\]  

Where

- \( L_T \) = current length of the tendon,
- \( L_{TR} \) = resting length of the tendon,
- \( c \) = the extension of tendon under maximum isometric force as a fraction of tendon resting length.

\[ F_V(V_F) = \frac{(V_{FMAX} - V_F)}{(V_{FMAX} + k \cdot V_F)} \quad V_F \geq 0 \quad (3) \]

\[ F_V(V_F) = 1.5 - 0.5 \frac{(V_{FMAX} + V_F)}{(V_{FMAX} - 2 \cdot k \cdot V_F)} \quad V_F < 0 \quad (4) \]

Where

- \( V_{FMAX} \) = maximum muscle fiber velocity,
- \( k \) = the model shape parameter.

In series with the contractile component is an elastic component. The model of this element assumes that the tendon has a linear stress-strain curve (Figure 2c). The force-extension curve of this element is represented by

\[ F_V(V_F) = \frac{(V_{FMAX} - V_F)}{(V_{FMAX} + k \cdot V_F)} \quad V_F \geq 0 \]  

\[ F_V(V_F) = 1.5 - 0.5 \frac{(V_{FMAX} + V_F)}{(V_{FMAX} - 2 \cdot k \cdot V_F)} \quad V_F < 0 \]

Figure 2 — Key properties of the muscle model. (a) normalized force-length properties; (b) normalized force-velocity properties; and (c) tendon force-length curve. Note that fiber shortening, concentric muscle action, has been designated a positive velocity.
The active state is a function of the neural excitation of the muscle \((u)\). The relationship between the neural excitation and the active state is represented by a first-order differential equation presented by Pandy, Anderson, and Hull (1992), and modified here to account for increases in active state only,

\[
\dot{q} = \left( \frac{l}{t_{\text{rise}}} \right) \cdot (l - q) \cdot u
\]  

(6)

Where

\[
\dot{q} = \text{rate of change of the active state of a muscle model with respect to time},
\]

\(q = \text{active state of the muscle model after time interval (} 0 \leq q \leq 1\))

\(t_{\text{rise}} = \text{constant associated with the increase in active state},\)

and \(u = \text{current level of neural excitation (} 0 \leq u \leq 1\)).

The muscle’s active state represents the recruitment as well as the firing rate, or rate coding, of the \(\alpha\)-motorneurons. The active state is normalized so that zero represents no activation, and one represents maximum activation.

To simulate the action of the muscle model, it was first necessary to know the muscle model’s active state. For all simulations it was assumed that neural excitation was maximal throughout the movement, and active state was zero at the start and increased from there on; this corresponds with the instructions given to the participants. The simulation process was based around the iterative solving of the following equation:

\[
\frac{dF_M}{dt} = K(MT - F) = k \cdot V_T
\]  

(7)

Where the rate of change of the muscle force is equal to the product of the tendon stiffness \((K)\) and the velocity of the tendon \((V_T)\), and the velocity of the tendon is equal to the difference between muscle-tendon velocity \((V_{MT})\) and muscle fiber velocity \((V_F)\).

The tendon stiffness is just the gradient of the slope of the force-extension curve for the model of the series elastic component presented in Figure 2c. This first-order ordinary differential equation, (7), was solved using a variable step size Runge-Kutta technique (Press, Flannery, Teukolsky, & Vetterling, 1986).

For input into the model, it was necessary to know the length \((L_{MT})\) and velocity of the muscle-tendon complex. To compute the moment produced at the knee joint, it was necessary to know the moment arm of the gastrocnemius at the knee joint. The lengths and moment arms of muscles vary with joint angle, while the muscle-tendon complex velocity varies with joint angular velocity. Grieve, Pheasant, and Cavanagh (1978) measured the length change of the gastrocnemius with changes in joint angle on a set of cadavers. These were modeled by expressing change in length of the muscle-tendon complex as a function of joint angle. The first derivative of the change in length of the muscle-tendon complex with respect to the joint angle gives the moment arm of the muscle.

Based on the equations presented in Grieve et al. (1978), the length and moment arm of the biarticular gastrocnemius could be computed given both ankle and knee joint angles. The ankle angle was fixed at either 70° of dorsiflexion or
Role of the Gastrocnemius

130° of plantar flexion throughout the trials, and the knee angle data were obtained from the output of the isokinetic dynamometer. First-order finite difference equations were used to obtain the rate of change in length of the muscle-tendon complex.

The two heads of the gastrocnemius were combined into one equivalent muscle; the model parameters for this muscle model are presented in Table 1. For isolated fibers, the width of the force-length curve is around 0.56, but here it has been increased to 0.70 to reflect the influence of nonhomogeneity of the muscle fiber length (Challis, 2000).

### Table 1 Parameters for the Model of the Gastrocnemius

<table>
<thead>
<tr>
<th>Model parameter</th>
<th>Gastrocnemius</th>
<th>Source</th>
</tr>
</thead>
<tbody>
<tr>
<td>Force-Length (Equ. 2)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>$L_{F,OPT}$</td>
<td>0.059 m</td>
<td>Out et al. (1996)</td>
</tr>
<tr>
<td>$w$</td>
<td>0.70</td>
<td>Challis (2000)</td>
</tr>
<tr>
<td>Force-Velocity (Equ. 3 &amp; 4)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>$V_{FMAX}$</td>
<td>$6 \cdot L_{F,OPT}$</td>
<td>Faulkner et al. (1986)</td>
</tr>
<tr>
<td>$k$</td>
<td>3.0</td>
<td>Faulkner et al. (1986)</td>
</tr>
<tr>
<td>Tendon model (Equ. 5)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>$L_{TR}$</td>
<td>0.387 m</td>
<td>Out et al. (1996)</td>
</tr>
<tr>
<td>$c$</td>
<td>0.04</td>
<td>Out et al. (1996)</td>
</tr>
<tr>
<td>$F_{max}$</td>
<td>1407 N</td>
<td>Out et al. (1996)</td>
</tr>
<tr>
<td>Active State model (Equ. 6)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>$\tau_{rise}$</td>
<td>20 ms</td>
<td>Pandy et al. (1992)</td>
</tr>
</tbody>
</table>

The knee flexion moments produced by the participants were assessed with the ankle joints both plantar-flexed and dorsiflexed. The participants all had different ranges of motion at the ankle joints, which is reflected by the angles at which their ankles were braced in plantar flexion or dorsiflexion, as follows:

- Dorsiflexed ankle position = ankle angle of 70.8 ± 2.8°
- Plantar-flexed ankle position = ankle angle of 132.4 ± 4.3°

Generally, with the ankle dorsiflexed the knee moments were larger than those recorded with the ankle plantar-flexed (Figure 3). At each tested angular velocity, there was a statistically significant greater moment produced at the knee joint when the ankle was dorsiflexed compared with the plantar-flexed condition (Table 2).

The simulation model was used to examine the influence of ankle angle changes on the knee moment-generating capabilities of the gastrocnemius. The
Figure 3 — Time plots of the resultant joint moments at the knee during isokinetic knee flexions, with the ankle either plantar-flexed or dorsiflexed. (a) angular velocity of 30º/s; (b) angular velocity of 75º/s; (c) angular velocity of 150º/s. Peak moments measured at a knee angle of 130º are marked (■ = dorsiflexed, ○ = plantar-flexed).

Table 2 Knee Flexion Moments During Isokinetic Knee Flexions with Knee at 130º Angle

<table>
<thead>
<tr>
<th>Angular velocity</th>
<th>Ankle dorsiflexed Mean ± SD</th>
<th>Ankle plantar-flexed Mean ± SD</th>
</tr>
</thead>
<tbody>
<tr>
<td>30º/s *</td>
<td>148.7 ± 19.9</td>
<td>135.0 ± 16.2</td>
</tr>
<tr>
<td>75º/s *</td>
<td>137.7 ± 19.4</td>
<td>126.8 ± 16.1</td>
</tr>
<tr>
<td>150º/s *</td>
<td>114.2 ± 22.8</td>
<td>103.2 ± 12.7</td>
</tr>
</tbody>
</table>

* Significant difference between the two conditions, $p < 0.05$. 
moment arm of the gastrocnemius at the knee decreases with increasing knee flexion (Figure 4a). The length of the gastrocnemius also decreases with increasing knee flexion (Figure 4b). Dorsiflexing the ankle joint increased the length of the gastrocnemius by an average of 0.046 m (Figure 4b). To examine the static properties of the gastrocnemius, we computed the moment at the knee joint for knee angles from 90 to 180º with the ankle either dorsiflexed to 70º or plantar-flexed to 130º. These results show that as knee flexion angle varies, the maximum isometric moment produced at the knee by the gastrocnemius depends on ankle joint angle (Figure 5). These moments are larger with the ankle dorsiflexed versus plantar-flexed.

The experimental tasks were simulated with the model to examine the contribution of the gastrocnemius. The model estimated the moment produced at the knee joint for isokinetic motions at angular velocities of 30, 75, and 150º per second (Figure 6). The graphs show that the moments generated by the gastrocnemius are much greater when the ankle is dorsiflexed compared to plantar-flexed. At the reference knee angle of 130º, the differences between the gastrocnemius-produced moments at the knee joint, when the ankle was either dorsiflexed or plantar-flexed, were 13.7, 10.9, and 11.0 N.m for the angular velocities of 30, 75, and 150º per second, respectively. These differences compare favorably with the mean differences from the experimental data (13.8, 10.6, and 11.0 N.m).
Discussion

The primary goal of this investigation was to examine the influence of the gastrocnemius on the performance of a knee flexion task. Two questions were addressed. The first was, Did the moment at the knee joint during flexion vary as ankle joint angle was varied? Statistical analysis confirmed that a greater moment was generated at the knee joint with the ankle fixed in the dorsiflexed position compared with those produced when the ankle was plantar-flexed. The second question was, Could any differences in the moments be explained by changes in the kinematics of the gastrocnemius during knee flexions with the ankle angle varied? The length of the gastrocnemius was different when the ankle was dorsiflexed compared with when the joint was plantar-flexed.

The simulations showed that this change in length placed the muscle on a different part of its force-length curve and therefore changed its potential for contributing a moment to knee flexion. In these simulations, the gastrocnemius produced a much greater moment at the knee when it was dorsiflexed versus when it was plantar-flexed. The difference in moment produced in these simulations, at the specified knee joint angle of 130º, compared favorably to the difference found in the experimental data. The results of the simulation suggest that the moment-producing capabilities of the gastrocnemius under the two ankle joint conditions was responsible for the difference seen in the experimental data on knee flexion.

One limitation of this study was the inability to obtain electromyogram (EMG) data from the gastrocnemius during the isokinetic knee testing. This inability was
due to the method of securing the aquaplast splint to the participant’s lower leg. To secure the ankle in either dorsiflexion or plantar flexion, the splint was placed on the lower leg, making it impossible to place electrodes over the muscles of interest. EMG data would have assisted in confirming that the gastrocnemius was equally active under all conditions regardless of whether it was plantar-flexed or dorsiflexed. The analysis assumed that the participants were attempting to maximally activate their knee flexor muscles; this is a reasonable assumption for these well-trained individuals (Enoka & Fuglevand, 1993). There is evidence that during submaximal activity, depending on the nature of the task, different motor units are recruited (e.g., Grimby, 1984); therefore, these results should be carefully extrapolated to submaximal knee flexor activity.

As with any model, a number of simplifying assumptions were made concerning the system under study; the primary one concerning the muscle model will be highlighted here. The muscle model did not include a parallel elastic element. This model element would have been associated with passive moments generated at the knee joint. Measurements made in vivo indicate that throughout the midrange of motion, the passive moment at the knee is so small that it can be ignored (McFaull & Lamontagne, 1998).
The series elastic component was assumed to have a linear stress-strain curve. It was presumed that this model component essentially represents the properties of tendon. Outside the toe region of the stress-strain curve of tendon, there is research to support the assumption of a linear stress-strain curve (Baratta & Solomonow, 1991; Bennett, Ker, Dimery, & Alexander, 1986). In the present study, muscle forces were high enough to ensure that tendon was operating outside this toe region. It should be pointed out that while the model has a number of deficiencies, it does represent a physiological system not unlike the one being studied.

In the simulation model the other knee flexors were not modeled, as their kinematics should have been the same regardless of whether the ankle was dorsiflexed or plantar-flexed. Therefore, the contribution of the other knee flexors was assumed to be constant for all trials performed with the same angular velocity knee flexions, regardless of the ankle joint angle.

The present study has demonstrated how the contribution of the gastrocnemius as a knee flexor varies with ankle angle. This contribution is a function of anatomy and therefore, while the results were obtained for isokinetic leg curls, it seems reasonable to extrapolate these results to maximal nonisokinetic leg-curl machines. The results of this study provide some guidance on the use of the leg-curl machine for strengthening the knee flexors. Exercisers and trainers should be aware that different ankle angles influence the movement, regardless of whether these angle changes take place within a set of repetitions or between sets. A leg-curl machine could be designed in which the ankle joint angle is fixed by some appropriately placed pads.

In summary, this study demonstrated that a statistically significant greater knee moment was produced during isokinetic knee flexions when the ankle was braced in dorsiflexion versus when it was braced in plantar flexion. This trend was consistent throughout the three angular velocities used in testing. By lengthening the biarticular gastrocnemius, a greater moment was produced during knee flexion. A simulation model of the gastrocnemius demonstrated that this difference was due to plantar flexion of the ankle moving the gastrocnemius to a less favorable part of its force-length curve throughout the movement, compared with the ankle being dorsiflexed. These results have implications for the way in which leg curls are executed and the design of leg-curl machines.

References


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