Influence of the Fascicle Length and Physiological Cross-sectional Area of M. soleus on Ankle Joint Kinetics

Akinori Nagano*, Shinsuke Yoshioka**, Taku Komura*** and Senshi Fukashiro**

*Institute of Medical Sciences, University of Aberdeen
Foresterhill, Aberdeen AB25 2ZD United Kingdom
a.nagano@abdn.ac.uk

**Department of Life Sciences (Sports Sciences), University of Tokyo
3-8-1 Komaba, Meguro, Tokyo 153-8902 Japan

***School of Informatics, University of Edinburgh
IPAC JCMB, Kings Building, Mayfield Rd, Edinburgh EH9 3JZ United Kingdom
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The purpose of this study was to examine which of the following exhibited greater advantageous in achieving greater mechanical outputs, on the condition that muscle mass (thus the muscle volume) is identical: (1) to have a greater fascicle length, or (2) to have a greater physiological cross-sectional area (PCSA). (1) Corresponds to a thin and long muscle, whereas (2) corresponds to a thick and short muscle. Three simulation models of the human m. soleus in the region of the ankle joint were constructed. (A) The default model muscle parameter values were derived from the literature. (B) The optimal length of the contractile element (CE) was changed to 1.1 times and the maximal isometric force of the CE was changed to 1/1.1 times the original values. (C) The optimal length of the CE was changed to 1/1.1 times and the maximal isometric force of the CE was changed to 1.1 times the original values. The total muscle volume was unchanged through these modifications. Maximal-effort isokinetic plantarflexion/dorsiflexion activities were simulated at a variety of angular velocities. In concentric actions excluding those at very slow angular velocities, greater mechanical outputs were obtained with model (B). During eccentric actions, the mechanical output was consistently greater with model (C). The findings of this study raise suggestions related to assessing the aptness of individuals for athletic activities in which the competitors are classified by their body mass (such as weight lifting and judo). Results suggest that it is more advantageous to have a greater fascicle length (tall-and-slender physique) for concentric activities, whereas it is more advantageous to have a greater PCSA (short-and-thick physique) for eccentric activities.

Keywords: simulation, isokinetic, plantarflexion, dorsiflexion

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1. Introduction

It is crucial for athletes and coaches to assess type of movements for which their physique (e.g., tall-and-slender or short-and-thick) is suited. This assessment is especially valuable for sports activities in which competitors are classified by body mass (such as weight lifting, boxing, wrestling, judo, etc.), as the amount of muscle mass that the athletes can gain through strength training is limited in these cases.

The maximal isometric force development capability of a muscle is a linear function of its physiological cross-sectional area (PCSA). The coefficient of the

linear function is referred to "specific tension" (Close 1972; Fukunaga, Ito, et al., 1996). This relation can be expressed as $F_{\text{isom}} = \text{SPT} * \text{PCSA}$, where F_{isom} is maximal isometric force and SPT is specific tension. In general, the greater the PCSA is, the greater the expected isometric force development. However, when considering the mechanical output of muscles during movement, another factor, the length of muscle fascicles, exerts an important influence. This is due to the force-length-velocity relation of muscles (Edman 1988; Lieber 2002).

When considering a situation in which the muscle volume (thus the muscle mass) is given, it is not clear which of the following is more advantageous: to have

Figure 1 Schematic diagram of the musculoskeletal model. The model of the m. soleus had force-length-velocity characteristics. A positive ankle joint angle represents a dorsiflexion and a negative ankle joint angle represents a plantarflexion. CE: contractile element, SEE: series elastic element.

a greater PCSA or to have a greater muscle fascicle length. In other words, when considering sports activities in which the competitors are classified by body mass, it is not obvious whether it is more advantageous to have a tall-and-slender physique or a short-and-thick physique. It is difficult to perform this type of quantitative evaluation through experimental procedures. Therefore the purpose of this study was to evaluate the mechanical output of a muscle as a function of its PCSA and fascicle length using computer simulation.

2. Methods

A computer simulation model of the human ankle musculoskeletal system, which consists of two rigid bodies connected with a frictionless hinge joint and a Hill-type musculotendon (Hill 1938), was constructed using MATLAB (The MathWorks, Inc., Natick, MA, USA) (Nagano and Komura 2003) (**Figure 1**). The musculotendon model consisted of two elements, i.e., a contractile element (CE) representing fascicles and a series elastic element (SEE) representing all series elastic structures. The mathematical representation of the properties of the Hill-type musculotendon (e.g., force-length-velocity relations) was adopted from preceding studies (van Soest and Bobbert 1993; Nagano and Gerritsen 2001).

The ankle joint and the m. soleus were modeled, as the muscle path of the m. soleus can be represented well with two points (an origin and an insertion). The parameter values of the musculotendon, i.e., maximal

isometric contraction force ($F_{\max}$), optimal length (L_{CEopt}) and pennation angle of the CE, slack length of the SEE, and the coordinates of the origin and insertion were adopted from a preceding study (Delp, Loan, et al., 1990). With these parameter values, the m. soleus attained the maximal isometric force development at approximately 20 deg of ankle joint plantarflexion (**Figure 4, 5**).

The m. soleus was maximally activated, as the ankle joint was rotated at constant angular velocities. Kinetic (force, moment, power and work) output around the ankle joint were evaluated for variations of angular velocities, i.e., from -210 deg/s (plantarflexion) to $+210$ deg/s (dorsiflexion). Plantarflexion motions were initiated from the ankle joint angle = 0 deg (neutral position), whereas dorsiflexion motions were initiated from the ankle joint angle = -60 deg (60 deg plantarflexed position). The range of -60 deg to 0 deg covers the range of active force development of this muscle with the parameter values utilized in this study (Delp, Loan, et al., 1990) (**Figure 4, 5**). These correspond to isokinetic measurements on a muscle testing machine.

Thereafter, the same evaluation was repeated, with the muscle parameter values modified in two ways, maintaining the muscle volume constant. In one case (noted " $1.1 L_{\text{CEopt}}$ " model), L_{CEopt} was increased by 10% (1.1 times the original value), whereas $F_{\max}$ was decreased by 9.09% ($1/1.1$ times the original value) (**Figure 2**). 10% of variations were found to be reasonable by referring to the experimentally

Figure 2 Schematic diagram of the manipulations of the muscle parameter values applied in this study. For the "1.1 F_{max} " model, the F_{max} value was increased to 1.1 times the original value, whereas the L_{CEopt} value was decreased to 1/1.1 times the original value. For the "1.1 L_{CEopt} " model, the L_{CEopt} value was increased to 1.1 times the original value, whereas the F_{max} value was decreased to 1/1.1 times the original value. The total muscle volume remained constant through these manipulations. F_{max} : maximal isometric force of the contractile element, SPT: specific tension, PCSA: physiological cross-sectional area of the contractile element.

Figure 3 The shape of the force-length curve. For the "1.1 L_{CEopt} " model, the active range of force development was increased and the maximal force development was decreased. For the "1.1 F_{max} " model, the maximal force development was increased and the active range of force development was decreased. These modifications kept the area under the force-length curve constant. F_{CE} : force development of the contractile element; L_{CE} : length of the contractile element.

collected muscle architecture data summarized in a preceding study (Yamaguchi 1990). This corresponds to a muscle model whose fascicle length is greater than the original model. As F_{max} is assumed to be proportional to PCSA, this manipulation kept the total volume of the muscle constant. This

manipulation changes the shape of the force-length relation of the contractile element such that the range of active force development is increased by 10% and the peak force is reduced by 9.09%, while maintaining the area under the force-length curve constant (**Figure 3**).

In another case (noted "1.1 F_{max} " model), F_{max} was increased by 10%, whereas L_{CEopt} was decreased by 9.09% (**Figure 2**). This corresponds to a muscle model whose PCSA is greater than that of the original model, while keeping the muscle volume constant. This manipulation reduces the range of active force development by 9.09% and increases the peak force by 10%, while maintaining the area under the force-length curve constant (**Figure 3**).

3. Results

For concentric actions, the muscle force development was greater for the "1.1 F_{max} " model than for other two models when the angular velocity was relatively low. At a relatively high angular velocity (> 90 deg/s), the muscle force development was greater for the "1.1 L_{CEopt} " model than for other models (**Figure 4**, **Table 1**). Similarly, at a relatively low angular velocity, the joint moment output was greater for the "1.1 F_{max} " model than for other two models. At a relatively high angular velocity (> 120 deg/s), the joint moment output was greater for the

Figure 4 Muscle force development during concentric actions. The solid line represents the data for the default model. The dashed line represents the data for the "1.1 L_{CEopt} " model. The dotted line represents the data for the "1.1 F_{max} " model.

Table 1 The mechanical outputs of the m. soleus during concentric actions. Variations of ankle joint angular velocities were tested. According to the definition described in Figure 1, a negative angular velocity stands for a plantarflexion motion. Greater mechanical outputs, except for work outputs, were obtained with the "1.1 F_{max} " model at a relatively small angular velocity. Greater mechanical outputs were obtained with the "1.1 L_{CEopt} " model at a relatively large angular velocity (indicated by bold letters).

		-210 deg/s	-180 deg/s	-150 deg/s	-120 deg/s	-90 deg/s	-60 deg/s	-30 deg/s	Isometric
Peak Force (N)	1.1 L_{CEopt}	511.0	662.5	850.8	1079.9	1366.8	1735.9	2222.7	2580.3
	Default	413.4	575.6	777.7	1027.0	1345.3	1767.3	2349.1	2838.5
	1.1 F_{max}	308.5	476.2	687.8	955.4	1303.9	1777.3	2459.1	3122.0
Peak Plantarflexion Moment (Nm)	1.1 L_{CEopt}	29.3	38.0	48.3	60.8	76.3	95.7	120.2	137.1
	Default	24.5	33.9	45.2	59.4	77.3	100.5	131.1	154.9
	1.1 F_{max}	18.5	28.4	40.9	56.7	76.8	103.6	141.0	174.3
Peak Power (W)	1.1 L_{CEopt}	107.4	119.3	126.5	127.3	119.9	100.2	62.9	-
	Default	89.6	106.5	118.4	124.5	121.5	105.2	68.6	-
	1.1 F_{max}	67.8	89.2	107.1	118.8	120.7	108.5	73.8	-
Work (J)	1.1 L_{CEopt}	9.1	12.1	15.8	20.3	25.9	33.1	43.1	-
	Default	6.5	9.3	12.8	17.1	22.7	30.0	39.9	-
	1.1 F_{max}	4.2	6.8	10.0	14.2	19.6	26.8	36.9	-

"1.1 L_{CEopt} " model than for other two models (**Table 1**). Furthermore, at a relatively low angular velocity, the power output was greater for the "1.1 F_{max} " model than for other two models. At a relatively high angular velocity (> 120 deg/s), the power output

was greater for the "1.1 L_{CEopt} " model than for other two models (**Table 1**). For all angular velocities, the concentric work output was consistently greater for the "1.1 L_{CEopt} " model than for other two models (**Table 1**).

Figure 5 Muscle force development during concentric actions. The solid line represents the data for the default model. The dashed line represents the data for the "1.1 $L_{C_{Eopt}}$ " model. The dotted line represents the data for the "1.1 F_{max} " model.

Table 2 The mechanical outputs of the m. soleus during eccentric actions. Variations of ankle joint angular velocities were tested. According to the definition described in **Figure 1**, a positive angular velocity represents a dorsiflexion motion. As the motions were eccentric, negative power and work outputs were calculated. Greater mechanical outputs were consistently obtained with the "1.1 F_{max} " model (indicated by bold letters).

		Isometric	30 deg/s	60 deg/s	90 deg/s	120 deg/s	150 deg/s	180 deg/s	210 deg/s
Peak Force (N)	1.1 $L_{C_{Eopt}}$	2580.3	3547.0	3784.3	3904.2	3976.5	4021.6	4059.4	4076.1
	Default	2838.5	3950.3	4205.9	4331.8	4405.2	4449.0	4479.7	4517.2
	1.1 F_{max}	3122.0	4395.7	4669.9	4800.9	4871.7	4922.6	4962.7	4985.1
Peak Plantarflexion Moment (Nm)	1.1 $L_{C_{Eopt}}$	137.1	183.7	195.1	201.0	204.4	206.9	208.6	209.4
	Default	154.9	210.4	223.2	229.5	233.1	235.6	237.4	238.6
	1.1 F_{max}	174.3	239.7	253.9	260.7	264.5	267.4	268.5	270.3
Peak Negative Power (W)	1.1 $L_{C_{Eopt}}$	-	96.2	204.3	315.7	428.0	541.7	655.4	767.5
	Default	-	110.2	233.7	360.5	488.2	616.7	745.8	874.7
	1.1 F_{max}	-	125.5	265.9	409.5	554.0	700.0	843.5	990.7
Negative Work (J)	1.1 $L_{C_{Eopt}}$	-	80.9	86.5	89.6	91.4	93.2	94.2	95.4
	Default	-	82.7	88.2	91.4	93.3	95.2	96.7	98.0
	1.1 F_{max}	-	84.6	90.3	93.3	95.7	97.6	99.3	100.8

For eccentric actions, the force development of the muscle, joint moment output, negative joint power output and negative work output were consistently greater for the "1.1 F_{max} " model than for other models (**Figure 5**, **Table 2**).

4. Discussion

The purpose of this study was to examine which of the following, a greater fascicle length or a greater PCSA, exhibits a greater advantage when an identical muscle volume is assumed. During

concentric actions, there was an overall tendency for greater mechanical output for the "1.1 $F_{\max}$ " model than for other models at a relatively small angular velocity (**Figure 4, Table 1**). However, at greater angular velocity, the mechanical outputs were greater for the "1.1 L_{CEopt} " model than those for the other two models. This is true for all force (> 90 deg/s), moment (> 120 deg/s) and power (> 120 deg/s) outputs (**Figure 4, Table 1**). This is due to the major role played by the non-linear drop of force development resulting from the force-velocity relation in the higher angular velocity region (Edman 1988; Nagano and Komura 2003). Having longer fascicles (greater L_{CEopt}) means that the contractile element experiences lesser relative shortening at a certain ankle joint angular velocity. Therefore, there was a smaller drop of force development of the CE with a greater L_{CEopt} .

For eccentric actions, all force, moment, (negative) power and (negative) work output values were greater for the "1.1 $F_{\max}$ " model than for other models (**Figure 5, Table 2**). In the "1.1 $F_{\max}$ " model, a greater $F_{\max}$ was advantageous for force development. Likewise, with a shorter L_{CEopt} , relative lengthening speed of the CE at a certain ankle joint angular velocity was greater for this model than for the other two models. This also enhanced the muscle force development due to the eccentric part of the force-velocity relation. Therefore for eccentric actions, a greater PCSA (thus $F_{\max}$) seems to be more advantageous.

Peak plantarflexion angular velocity during squat jumping has been reported to be ~ 900 deg/s (Bobbert and van Ingen Schenau 1988); peak plantarflexion angular velocity during walking has been reported to be ~ 250 deg/s (Winter 1990); and peak plantarflexion angular velocity during sprinting push-off has been reported to be ~ 950 deg/s (Jacobs, Bobbert, et al., 1996). This implies that the non-linear effect of the fascicle length/PCSA relation on joint kinetic outputs investigated in this study comes into play in many cases of human locomotion. When the required speed of concentric joint motion is *very* slow (< 120 deg), it was found to be advantageous to have a greater PCSA. However, in most types of daily and sports movements, concentric angular velocity of joint motions easily enters into the "fast" angular velocity region (> 120 deg/s) investigated in this study. Therefore, in practical terms, it would be safe to state that it is more advantageous to have

longer fascicles to enhance concentric mechanical outputs. This finding seems to be consistent with what has been reported in a preceding study (Abe, Fukashiro, et al., 2001). These researchers investigated the correlation between the muscle fascicle length and the sprinting performance of elite sprinters. The fascicle length was experimentally obtained using ultrasonography. It was concluded that a longer fascicle length is associated with a greater sprinting performance. Although a great care needs to be taken when comparing the findings of this simulation study with the experimental findings (Abe, Fukashiro, et al., 2001), the consistent results suggest the importance of muscle fascicle length in concentric activities. It would be valuable to perform more rigorous comparisons between simulated and experimentally obtained data in future studies.

The m. soleus was modeled in this study, as the origin, insertion and line of action of this muscle can be modeled in a simple form (Delp, Loan, et al., 1990). In addition, a constant pennation angle was assumed for the sake of simplicity. This approach is similar to that taken in a preceding study investigating the relation between the length of moment arm and the mechanical outputs of the musculoskeletal system (Nagano and Komura 2003). Although the current study yielded clear and reasonable results, it is necessary to investigate whether the same behaviors are observed for other muscles for which the pennation angle changes dynamically. This modification is technically demanding due to the three-dimensional configuration of the joints and muscles; however, it is likely that the results obtained in this study would be valid in such cases as well, as the results are derived from the basic force-velocity relation of muscles (Edman 1988; Lieber 2002).

It is widely recognized that the cross-sectional area of muscles can be increased through strength training (McDonagh and Davies 1984; Narici, Hoppeler, et al., 1996; Hakkinen, Newton, et al., 1998). However, it is not likely that the muscle fascicle length can be lengthened substantially through normal (non-invasive) training protocols (Lieber 2002). Therefore, while the results of this study are not likely to suggest a direction of athletic training, they do raise suggestions related to assessing the aptness of individuals for athletic movements. It was generally suggested that an individual with a tall-and-slender physique is more suited to

concentric movements, whereas an individual with a short-and-thick physique is more suited to eccentric movements, although it goes without saying that numerous other factors (such as the relative length of body segments and so on) should be taken into consideration when applying this finding to actual sports activities.

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Name:

Akinori Nagano

Affiliation:

Institute of Medical Sciences, University of Aberdeen

Address:

Foresterhill, Aberdeen AB25 2ZD United Kingdom

Brief Biographical History:

1996 B.L.A., University of Tokyo.

1998 M.A., University of Tokyo.

2001 Ph.D., Arizona State University.

2001-2002 researcher, Hitachi, Ltd.

2002-2003 post-doc, Boston University.

2003-2006 post-doc, RIKEN.

2006-present lecturer, University of Aberdeen.

Main Works:

- Nagano, Yoshioka, Hay, Fukashiro, 2006. Light finger touch on the upper legs reduces postural sway during quasi-static standing. *Motor Control*.
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