# THE INFLUENCE OF THE BIARTICULARITY OF THE GASTROCNEMIUS MUSCLE ON VERTICAL-JUMPING ACHIEVEMENT

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Abstract—Hypotheses concerning the influence of changes in the design of the human musculoskeletal system on performance cannot be tested experimentally. Computer modelling and simulation provide a research methodology that does allow manipulation of the system's design. In the present study this methodology was used to test a recently formulated hypothesis concerning the role of the biarticularity of the gastroenemius muscle (GAS) in vertical jumping [Bobbert and van Ingen Schenau, J. Biomechanics 21, 249-262 (1988)]. This was done by comparing maximal jump heights for a model equipped with biarticular GAS with a model equipped with a monoarticular GAS. It was found that jump height decreased by 10 mm when GAS was changed into a monoarticular muscle. Thus, the hypothesis formulated by Bobbert was substantiated, although quantitatively the effect is small. Our result differs from that of Pandy and Zajac [J. Biomechanics 24, 1-10 (1991)], who performed similar model calculations. It is shown that the results described by these authors can be explained from the moment-arm-joint-angle relation of GAS at the knee in their model.

#### NOTATION

CE	contractile element
F	force
GAS	m. gastrocnemius
GLU	mm. glutei
HAM	hamstring muscles
L	length
М	joint moment
MTC	muscle-tendon complex
PEE	parallel elastic element
REC	m. rectus femoris
SEE	series elastic element
SOL	m. soleus
STIM	muscle stimulation
V	velocity
VAS	mm. vasti

#### INTRODUCTION

In the study of the musculoskeletal system, a distinction can be made between design questions and analysis questions. Analysis aims at understanding how the system should be used, given the design of the system. Design aims at understanding why the system is built as it is. A design question that has puzzled scientists for more than a century (Cleland, 1867) is why the human musculoskeletal system is equipped with biarticular muscles. Recently a hypothesis concerning the role of the biarticularity of the gastrocnemius muscle (GAS) in vertical jumping was formu-

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lated by Bobbert and van Ingen Schenau (1988). This hypothesis can be summarized as follows. It is well known that subjects performing maximum-height jumps extend their joints in a proximodistal sequence. This sequence requires high net joint moments that must be produced at high extension velocities at the distal joints, specifically at the ankle joint. The instantaneous power output (equalling the product of net joint moment and joint angular velocity) around the ankle joint reaches values as high as 2500 W. It was argued that biarticular muscles such as GAS are specifically suited for delivering this power: because the knee is extending, GAS length remains close to optimum and its contraction velocity remains relatively low, even when the ankle is extending fast. As a result, it can generate a large force and a large power output around this joint. A related advantage of biarticular muscles such as GAS concerns the fact that joint angular velocities must be low at full joint extension to preclude damage; in order to comply with this 'anatomical constraint' (van Ingen Schenau et al., 1987), either the knee extensors must be deactivated prematurely, or antagonists must be activated in order to reduce extension velocity of the joints. By activating GAS, a biarticular antagonist, knee extensors can remain active until take-off and the work produced by the knee extensors is not dissipated, but is used in part to extend the ankle joint [see van Ingen Schenau et al. (1990) for a definition of this 'transport mechanism'].

Unfortunately, these ideas cannot be tested experimentally; it would require surgery on experimental subjects, which is not done for obvious ethical reasons. Alternatively, a modelling approach can be used. First, a physical model called 'Jumping Jack' was constructed to elucidate the principle described above

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(Bobbert et al., 1987). In this model, the knee extensors were represented by a linear spring and GAS was represented by a stiff wire. By setting the wire length, the knee angle at which the wire became taut could be manipulated. It was shown that jump height depends strongly on wire length. 'Jumping Jack' served well to exemplify the principle involved. However, its properties differ too much from those of the human musculoskeletal system to generalize model results to human vertical jumping. To name a few, the actuator properties (linear spring and stiff wire) are very different from muscle, the moment arm of the wire at knee and ankle is not realistic, and the model does not suffer from balance problems, since the hip can only move in the vertical direction. The construction of a more realistic physical model is impractical. Therefore, it was decided to turn to mathematical modelling and simulation.

A model of the musculoskeletal system was constructed, including relations governing the behavior of the muscle-tendon complexes. After modelling the interaction with the environment, the dynamical equations of motion were derived. From these the position of the system can be calculated as a function of time using numerical integration, given the independent neural inputs to the muscles. To perform a model test of the hypothesis that the biarticularity of GAS contributes significantly to vertical-jumping achievement. the following simulation experiment was designed: first, it was established how high the model can jump with intact GAS, when push-off is started from a prescribed static squatted position. Next, GAS was changed into a monoarticular ankle plantarflexor, and maximal jump height was established for this model as well. The difference between these jump heights indicates the specific contribution of the biarticularity of GAS to vertical jumping.

#### METHODS

### Skeletal model

The skeletal system was modelled as a planar mechanism consisting of four rigid segments connected in three frictionless hinge joints. At the toe, the four-segment skeletal model was connected to the ground by a fourth frictionless hinge joint. The initial position of the skeletal system was derived from experimental data. It was found that in the initial static squatted position, the heel is off the ground. Possible ground contact at the heel during push-off was modelled by a slightly damped very stiff spring; further details are omitted, since in the optimal solutions ground contact at the heel did not occur. Segmental parameter values were derived from Clauser et al. (1969). These were scaled in order to make the dimensions of the model identical to the average of the experimental subjects whose jumps were used for comparison with simulation results. Segmental parameters and initial joint angles are presented in Table 1.

The skeletal model was defined in SPACAR, a software system for simulation of the kinematics and dynamics of multibody mechanical systems. SPACAR was developed at the Technical University of Delft and has been described elsewhere (van der Werff, 1977; van Soest et al., 1992). The SPACAR model represents a human subject performing a maximal vertical jump [Fig. 1(a)]. It consists of four BEAM elements and four HINGE elements [Fig. 1(b)]. The deformations of the HINGE elements correspond to joint angles in the hip, knee, ankle and metatarsophalangeal joints. These deformations are used as the degrees of freedom of the system. Thus, joint angles are used as state variables; angular accelerations are calculated from the angles, angular velocities, gravitational forces and forces acting at the degrees of freedom, i.e. net joint moments [see van Soest et al. (1992) for a more comprehensive description of this model].

# Muscle model

Hill-type models of muscle dynamics yield a satisfactory description of muscle behavior in the context of simulations of gross motor behavior (e.g. Winters, 1990). A Hill-type general purpose muscle model (GPMM) has been described elsewhere (van Soest, 1992). This model consists of a series elastic element SEE, a contractile element CE and an elastic element PEE parallel to the CE. SEE and PEE behavior is governed by nonlinear force-length characteristics. CE behavior is more complex: force depends on the active state,  $L_{CE}$  (CE length) and  $V_{CE}$  (CE velocity). The active state as defined by Ebashi and Endo (1968) is related to *STIM* through a first-order process,

Table 1. Mean segmental parameter values and initial joint angles (INIT. ANGLE) (n=6).  $I_{CM}$  = moment of inertia, relative to the segment's center of mass. CMPOS = position of the segment's center of mass, expressed as a fraction of segment length, measured from the cranial end of the segment. Segment angles are relative to the right horizontal

	Length (m)	Mass (kg)	I <sub>CM</sub> (kg m <sup>2</sup> )	CMPOS	INIT. ANGLE (rad)
Foot	0.165	2.5	0.02	0.27	2.28
Lower leg	0.458	7.1	0.14	0.43	0.84
Upper leg	0.485	16.9	0.42	0.43	2.59
Trunk	0.920	55.7	3.90	0.68	0.73



Fig. 1. (Left) Human subject performing maximal vertical jump. (Right) Schematic representation of the planar SPACAR model.

where *STIM* represents the level of stimulation of the muscle, ranging between zero and one. This first-order process is modelled as described by Hatze (1981). When *STIM* is taken as the independent input, the muscle model is of second order. The concentration of free calcium (which is algebraically related to the active state) and  $L_{CE}$  are used as its state variables. The time derivatives of these variables are calculated at any instant from their values and *STIM* and  $L_{MTC}$  (muscle-tendon complex length), which can be calculated from the position of the skeletal system. The dependent variable muscle force is calculated as follows: by subtracting  $L_{CE}$  from  $L_{MTC}$ ,  $L_{SEE}$  is obtained; SEE force depends solely on  $L_{SEE}$ ; as the SEE has no elements in parallel, SEE force equals muscle force.

Muscle groups incorporated in the model are glutei (GLU), hamstrings (HAM), vasti (VAS), rectus femoris (REC), soleus (SOL) and gastrocnemius (GAS). Where possible, parameter values are derived from morphological data reported in the literature [see van Soest (1992) for a definition of the parameters]. A number of parameters were given identical values for all muscles: the stretch of SEE at maximum isometric force was set to 0.04 times the SEE slack length;  $L_{CE}$  at which the active force is zero was set to 0.44 times  $L_{CE(OPT)}$  or 1.56 times  $L_{CE(OPT)}$ ; Hill's force-velocity parameters  $a/F_{MAX}$  and  $b/L_{CE(OPT)}$  were set to 0.41 and 5.2. The values of the muscle specific parameters are listed in Table 2. CE optimum lengths were derived from sarcomere numbers (Huijing, personal communication); relative values of maximal isometric force were based on muscle cross-sectional areas, whereas absolute values were chosen in such a way that realistic maximal isometric moments were obtained: SEE slack lengths were chosen in such a way that the joint angle at which the isometric joint moment is maximal corresponds to the experimentally obtained isometric moment-angle relations.

# Connection between skeletal model and muscle model

To intertwine skeletal and muscle models as described above, two related problems must be solved. The first involves a calculation of  $L_{\text{MTC}}$ , which is an input of the muscle model, from the position of the skeletal system; the second involves the representation of the influence of the muscle forces on the acceleration of the skeleton.

One possible approach is based on modelling of the path followed by the muscle; once this path is known for every position,  $L_{MTC}$  is easily calculated. The influence of muscle force on skeletal acceleration is represented by applying muscle force to the skeleton at all points where contact exists; the direction of these forces depends on the muscle path. Modelling of a muscle path is straightforward if the muscle runs along a straight line from origin to insertion. However, muscles such as VAS and GAS are known to curve around bony protrusions at the knee. Unfortunately, for these muscles reliable data on muscle paths as a function of joint angle are not available. For an example of this approach see Pandy and Zajac (1991).

An alternative approach can be used that builds on two basic facts. The first is that it is not necessary to directly apply the muscle forces to the skeletal system: the influence of a muscle on skeletal movement depends solely on the muscle moment M, equalling the muscle force times the moment arm relative to the axes of the joints it spans. The second is that this moment arm is equal to the derivative of  $L_{MTC}$  with respect to joint angle. Thus, it suffices to know  $L_{MTC}$  as a function of joint angle. A straightforward way to obtain such data was first described by Grieve et al. (1978) and used by others (Visser et al., 1990; Spoor et al., 1990). A disadvantage of this approach is that the contribution of muscle forces to reaction forces in the skeleton cannot be obtained. Given the fact that this study focusses on the relation between neural input to muscles and the resulting skeletal movement, the second approach is the most appropriate. Moment arm data used in this study were derived from experimental data reported by Grieve et al. (1978) and Visser et al. (1990), as well as from Huijing (personal communication).

A third problem concerning the connection between the skeletal model and the muscle model is of a different kind: it is desirable to integrate the differential equations describing the skeletal movement and those describing the muscle dynamics concurrently. Although this is no problem using SPACAR, in many software systems for multibody mechanics it is impossible. A schematic representation of the coupling between the GPMM and the SPACAR skeletal model is presented in Fig. 2.

# **Optimization** of STIM pattern

The movement of the model as defined above can be calculated given STIM(t) for the six muscle groups and given the initial conditions for the 20 state vari-

Table 2. Values of muscle-specific parameters. D = average moment arm,  $L_{CE(OPT)} = CE$ length at which maximal force can be delivered, LSLACK = SEE slack length,  $F_{MAX}$ = maximal isometric CE force. Values for  $F_{MAX}$  are given for two legs

	D <sub>HIP</sub> (m)	D <sub>KNEE</sub> (m)	D <sub>ANKLE</sub> (m)	L <sub>CE(OPT)</sub> (m)	LSLACK (m)	F <sub>MAX</sub> (N)
GLU	0.062			0.200	0.150	5000
HAM	0.077	0.026		0.104	0.370	4000
VAS		0.042		0.093	0.160	9000
REC	0.035	0.042		0.081	0.340	3000
SOL			0.046	0.055	0.246	8000
GAS		0.017	0.046	0.055	0.382	4000



Fig. 2. Simplified block diagram of connections between the mechanical SPACAR model and the muscle model. Blocks marked \* indicate the connections between the muscle model GPMM and SPACAR, as discussed in the text. Block marked INT represents the integration algorithm.

ables (two per muscle and two per mechanical degree of freedom). To achieve the goal of this study, it must be established which STIM(t) leads to a maximal vertical jump, i.e. to a maximal height of the body center of mass. This is not a simple problem: six STIM signals as a function of time must be optimized with respect to jump height. This dynamic optimization problem was studied in its full complexity by Pandy et al. (1990) and Pandy and Zajac (1991). In this study a more restricted form of dynamic optimization is used which, partly based on the results obtained by Pandy et al., is believed to render comparable results in case of maximum-height vertical jumping (see below). The following constraints were imposed on STIM: first, the initial STIM level was set in such a way that a static squatted position was maintained. Second, STIM was allowed to take on either this initial value or the maximal value of 1.0. Third, STIM was allowed to switch to its maximal value once, and thereafter remained maximal until take-off. The constraint on the lowest level of STIM was imposed to ensure that no countermovement could occur. This was done to allow a comparison with experimental jumps where no countermovement was allowed. Under these constraints, STIM(t) of each muscle group can be described by a single parameter: the instant of time at which STIM switches from the initial value to the maximal value. Thus, the optimization problem is reduced to finding the point in the sixdimensional control space that results in maximal jump height. Thus formulated, the optimization problem can be solved using standard algorithms. NAG subroutine E04UCF, a sequential quadratic programming algorithm, was used (NAG Fortran Library Manual Mark 13, Numeric Algorithms Group Ltd., Oxford, U.K.). To reduce the calculation time, the simulations were terminated at the instant of take-off, which was detected by monitoring vertical ground reaction force: as soon as this force became zero, simulations were terminated and jump height was calculated.

The numerical optimization routine used in this study was found to converge to an 'optimal' solution. A well-known problem of application of numerical optimization algorithms is that it is not guaranteed that the obtained solution is optimal in a global sense; the obtained solution may well represent a local optimum. It is not possible to prove formally that the solution found represents the global optimum. However, the optimization can be started from different points in the six-dimensional control space. It was found that optimizations started from highly differing starting points resulted in identical solutions. Therefore, it is likely that this solution represents a global optimum.

Severe constraints were imposed on STIM in this study in order to keep calculation time manageable and to ensure that a well-defined optimum could be found. To make sure that these constraints did not influence the performance significantly, optimizations were performed where each muscle was given more 'freedom': each was allowed to switch on two times and to be deactivated in between. In other words, the dimension of the control space was increased from six to 18. It was no longer clear that the obtained 'optimum' was optimal in a global sense. Although jump height improved, this improvement was only of the order of 2 mm. Apparently, when the number of constraints on STIM is reduced, several STIM patterns exist that yield virtually identical jump heights. Although such a flexibility is highly advantageous for the human being, it constitutes a nightmare to the scientist attempting to find the optimal solution numerically. It is concluded that by reducing the dimension of the control space to six, as is done in this study, the optimization problem becomes well-behaved while the reduction in jump height is small.

# Experimental data

Six elite male volleyball players performed a number of jumps starting from a freely chosen static squatted position. The subjects were instructed to jump as high as possible, not to make any countermovement prior to push-off, and to keep their hands on their back. The highest successful jump of each subject was selected for further analysis. During jumping, kinematic data were gathered using a 100 Hz VICON system and ground reaction force was measured using a force platform (Kistler 9281B). These data were used for an inverse dynamical analysis using standard procedures.

## **RESULTS AND DISCUSSION**

The optimal STIM pattern for the reference model is depicted in Fig. 3. Qualitatively, it corresponds well with the experimentally observed proximodistal sequence in muscle activation in vertical jumping (Bobbert and van Ingen Schenau, 1988). Such a sequence was also found by Pandy *et al.* (1990), using a simulation approach comparable to ours.

In order to gain confidence in the model, simulation results concerning the optimal jump were compared with the experimental data. As an example of kinematics, in Fig. 4 joint angular velocities obtained through simulation are compared with those obtained experimentally. Until shortly before take-off, correspondence is good; just before take-off, however, the simulated angular velocities continue to rise, whereas the experimental subjects show a decrease in angular



Fig. 3. Optimal *STIM(t)* for the six muscle groups for the standard model (upper line of each pair) and for the model with monoarticular GAS (lower line of each pair). Drawn lines indicate the period of maximal *STIM*. Time is expressed relative to the take-off. See text for details.

velocities. Considering the corresponding joint angles, this deceleration cannot be caused by a stretch of passive structures, but must be a result of muscle activity. This decrease is functional, in that it protects the passive structures around the joints from excessive loading ('anatomical constraint'). An attempt was made to incorporate such a constraint in the simulation model by adding to the optimization criterion a penalty on the summed rotational energy of the segments. Such an approach proved not to be fruitful. In our view, the observed anatomical constraint in the real system is likely to be the result of hard-wired



Fig. 4. Joint angular velocities versus time for hip (top), knee (middle) and ankle (bottom) joints. Solid curves: simulation results; dashed curves: experimental data. Time is expressed relative to take-off.



Fig. 5. Net joint moments versus joint angles for hip (top), knee (middle) and ankle (bottom) joints. Solid curves: simulation results; dashed curves: calculated from experimental data. Zero angles represent full flexion.

constraints on the form of *STIM* (i.e. inhibitory connections to alpha motoneurons). In the simulation model used in this study, such constraints were not incorporated. Consequently, it is not surprising that the simulated angular velocities did not decrease.

In Fig. 5 the experimentally obtained joint angle versus joint moment diagrams for the hip, knee and ankle joints are compared with those obtained through simulation. The degree of correspondence is good. The major difference is that the experimental subjects seem to be able to reach a more fully extended posture before take-off occurs. The area under these curves represents the amount of work produced. This integral yields 267, 211 and 134 J for the hip, knee and ankle joints for the experimental subjects; the corresponding values for simulation are 278, 254 and 125 J. As expected from Fig. 5, the correspondence is reasonably close.

Of the 125 J produced around the ankle joint in the model, 27 J is due to 'transport' by GAS. In a relative sense, this is comparable to the results for one-legged countermovement jumps obtained by Bobbert *et al.* (1986) using a combination of inverse and forward dynamics. When it is realized that for our model a 1 cm jump height corresponds to 8.1 J of effective energy, it is obvious that the effect of biarticularity of GAS is, at most, of the order of centimeters.

The maximal height reached by the body center of mass using optimal *STIM* was 1.480 m. Relative to upright standing, the jump height was 0.392 m. For the experimental subjects, the jump height, as obtained directly from positional data, amounted to 0.447 m. The vertical velocity of the body center of mass at take-off is very close for model and experiment (2.59 vs 2.67 m s<sup>-1</sup>). The difference in the jump height is caused, for the largest part, by a difference in the position of the center of mass at the instant of take-off.

The question arises as to how the experimental subjects can jump 0.055 m higher than the model, whereas the amount of work delivered to the skeleton is higher for the model than for the experiment. At first thought, this might be attributed to a higher 'effectivity' of the real system: it might be that a larger percentage of energy delivered to the skeleton contributes to jump height. However, for the model this effectivity already amounts to 87%. This cannot be significantly improved upon, since at a given position the vertical velocity of the body center of mass is directly related to angular velocity of the segments; in other words, vertical kinetic energy is directly related to rotational kinetic energy. A closer analysis of the experimental data revealed that higher jumping of our subjects using an apparently smaller amount of energy is due to the fact that the assumption of rigidity of segments does not hold during the final 50 ms of the push-off, especially for the trunk segment. The assumption of rigidity does not result in serious errors in joint moments, since these are calculated from the feet upward on the basis of measured reaction forces. However, it does result in an underestimation of the total amount of work, as the work done in extending the trunk is not taken into account. Nonrigidity of segments must also be assumed to explain how in the last part of push-off ground contact is maintained while at that time all joint angular velocities are decreasing (i.e. flexing joint accelerations): as can be seen from simulation results, in the case of rigid segments even large extending angular accelerations do not suffice to postpone take-off until full extension of the joints. This is, of course, due to well-known geometrical laws (van Ingen Schenau et al., 1990). We conclude from our comparison of simulation and experimental data that, although there are a few areas of concern, the degree of correspondence between simulation and experimental data is highly encouraging, especially if it is realized that wherever possible muscle parameter values were derived independently of the experimental data.

Turning to the central question of this study, we present the optimal STIM pattern found for the model with monoarticular GAS in Fig. 3. The most striking difference with the optimal STIM for the standard model is the earlier stimulation of GAS. This can be explained as a functional adaptation of the control to the new 'handicapped' system. To understand this, note that contraction velocity in the biarticular GAS is lower than that in the monoarticular GAS, because the knee is extending. Due to the force-velocity relationship, monoarticular GAS forces are lower. If the STIM pattern would not be changed, GAS would not contribute fully to the push-off, resulting in impaired plantarflexion. By stimulating GAS earlier in the push-off, this is partly prevented. Thus, functional adaptation of the control reduces the effect of potentially detrimental changes in the design of the system.

The maximal jump height with monoarticular GAS was 1.470 m. Relative to upright standing, this amounts to 0.382 m. The total energy delivered by the muscles was 654 J, of which 560 J (86%) contributed to jump height. The decrease in jump height which results from changing GAS into a monoarticular muscle was 10 mm. Although this difference is small, it is definitely larger than the resolution with which the optimization problem was solved. In the final iteractions of the optimization process, changes in jump height were of the order of 0.1 mm. Thus, our results support the hypothesis of Bobbert and van Ingen Schenau (1988).

In contrast, Pandy and Zajac (1991) claimed, on the basis of comparable simulation data, that making GAS monoarticular leads to a slight increase of jump height. In our view their results can be explained from the fact that they model GAS by a straight line connecting origin and insertion (see Fig. 1 of Pandy and Zajac, 1991). As a result, the moment arm of GAS at the knee approaches zero as the knee approaches full extension (see their Fig. 4). Since a moment arm at the knee equalling zero, in fact, indicates a monoarticular GAS, and considering that GAS is active only in the last phase of the push-off, where the knee is close to full extension, in the model of Pandy et al. GAS effectively behaves as a monoarticular muscle. Therefore, it was to be expected that jump height was hardly affected when GAS was made formally monoarticular. A similar argumentation was recently forwarded by Spoor and van Leeuwen (1992).

To test this argumentation, we changed our model so that the moment arm of GAS decreased to zero at the knee as the knee approached full extension, while the average moment arm was kept identical to that in our standard model. The optimal jump height for this Pandy-like model was slightly (2 mm) lower than that of the model with monoarticular GAS, just as Pandy and Zajac reported. This supports our view that differences in the moment arm at the knee between our model and that of Pandy and Zajac are responsible for the difference in results. We are convinced that the moment arm used in this study, which is based on cadaver measurements, is closer to reality. According to Spoor et al. (1990), the moment arm at the knee increases even more strongly than is assumed here as the knee is extended. Based on the results presented so far, it was expected that a larger increase of moment arm near full extension would lead to an increased effect of biarticularity of GAS. To test this expectation, we changed our model to comply with GAS moment arm data reported by Spoor et al. (1990). Confirming our expectation, the optimal jump height for this model was 18 mm higher than that of the model with monoarticular GAS. It is concluded that the way in which the moment arm at the knee of GAS depends on knee angle determines the effect of the biarticularity of GAS on jump height.

Another point on which Pandy and Zajac (1991) seemingly disagree with earlier work of our group concerns the 'proximodistal transport of power' (Gregoire et al., 1984). This concept was introduced from within the context of a joint power approach (Aleshinsky, 1986) in order to make clear that the net power at a joint does not necessarily reflect the sum of the powers produced by the muscles crossing that joint. This is obvious from the following example. Suppose that both VAS and GAS are active while the knee and the ankle are extended simultaneously. Further suppose that the biarticular GAS contracts isometrically and, as a result, produces no power. The net joint power at the knee (equalling joint angular velocity times the net joint moment) in this case is smaller than the power of VAS, because the net joint moment is smaller than the moment exerted by VAS due to the antagonistic action of GAS. It is easily shown that part of the power delivered by VAS reappears as net joint power at the ankle. Thus, GAS can be said to 'transfer power' from the knee to the ankle joint. Once again, this line of reasoning makes sense only in a joint power approach. In a segment power approach as adopted by Pandy and Zajac (1991), most of the power produced by the muscles obviously flows to the heaviest segment, i.e. the trunk. In our view, the criticism by Pandy and Zajac of the concept of joint power transport rests on a misunderstanding, on their part, of the theoretical framework (joint power as opposed to segment power approach) from within which this concept was formulated. It should be noted that this concept of redistribution of joint power and, similarly, of joint moments through a stimulation of biarticular muscles may help to place the unique actions of these muscles in a broader perspective. Recently, it was demonstrated that this capacity of biarticular muscles contributes to the solution of the problem of the hybrid control of position and force in multijoint systems (van Ingen Schenau et al., 1992).

The simulation results described here lend support to the hypothesis forwarded by Bobbert and van Ingen Schenau (1988). However, the decrease in jump height after making GAS monoarticular is modest. Most importantly, the effect of biarticularity of GAS in vertical jumping has been shown to depend critically on the nature of its biarticularity, i.e. the way in which the moment arm of GAS at the knee is related to the knee joint angle. Nevertheless, the issue of the effects of biarticularity cannot be entirely settled until the anatomical constraint is accounted for.

## CONCLUDING REMARK

It was shown that the simulation results obtained with the model described in this study are in fair agreement with the experimental results. The same holds for the model described by Pandy and Zajac (1991). Thus, an important validity test is passed by both the models. Nevertheless, the results described here support the hypothesis of Bobbert and van Ingen Schenau (1988) on the special role of GAS, whereas Pandy's results seem to refute this hypothesis. Although an explanation for Pandy's results was offered in this study, the important lesson, in general, is that subtle differences in modelling methods can lead to diametrically opposite results when applied to design questions.

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