# BIOMECHANICS OF SPORTS ¾
# SELECTED EXAMPLES OF SUCCESSFUL APPLICATIONS AND FUTURE PERSPECTIVES

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## Introduction

The overriding principle governing sports performance and, indeed, all forms of human activities, is the attempt of an individual, or a group of individuals, to perform a given task "in the best possible way", even if this tendency may not always be obvious. This phenomenon has its roots in the *teleological behaviour* of biological systems, as we shall discuss later. It will become obvious that a large class of problems confronting the sports biomechanist can be ultimately identified as belonging to the field of *optimization*.

The strong desire of biomechanists to simulate and analyse human motion, and thereby gain a better understanding of the laws governing the actions of the human neuromusculoskeletal system, has prompted the development of *geometrico – mathematical body models* of ever increasing sophistication. We shall devote some space to the discussion of the question of the complexity and adequacy of neuromusculoskeletal system models, with a certain emphasis placed on those developed by the author.

Closely related to the question of model development is that of *model parameter individualization*. By this we mean the determination, by means of specific methods, of the values of certain model parameters for a specific subject. The development of adequate techniques for the experimental determination of subject–specific segmental, articular, morphometric, myodynamic, and myocybernetic parameter values constitutes itself a major research field in biomechanics. Some attention will be given to this topic in the sequel, followed by a brief demonstration of the usefulness of appropriate models of the human neuromusculoskeletal system in connection with the adequacy of muscle models used in motion simulation and by the role they played in the discovery of the hyposensitivity of skeletal motions to neural control perturbations and in relation to the inverse dynamics problem.

Finally, it will be demonstrated by means of selected examples from the author's own work that biomechanics can contribute substantially to the *practical solution of a large variety of sports related problems*. The examples include the computer simulation and successful analysis of an accident involving a rock'n roll Betterini somersault; a biomechanical method for testing objectively the quality criteria of tennis rackets; and investigations into the validity and reliability of vertical jumping performance testing methods. Some thoughts on the *future perspectives of sports biomechanics* will conclude this overview.

## Teleological Biosystem Behaviour and Optimization of Sports Performance

As mentioned above, there is a close relationship between a general tendency of biosystems to behave teleologically (i.e. goal-directed) and the endeavour of athletes, to execute sports motions optimally. In Hatze (1976, p.163), the author has provided proof that the human neuromusculoskeletal system may be classified as a specific teleonomic system, that is, as a purposive, open, dynamic and adaptive subsystem of the human system. Its main purpose is the production of a sequence of transition processes which all have the property of converging to an optimal process that satisfies a certain performance criterion as well as certain constraints and boundary conditions (Hatze, 1976). Transition process is a term used in optimal control theory (Boltyanskii, 1971, p.4) and designates the combined set of neural control functions and the phase space trajectory. The latter consists of the myostates of all skeletal muscles involved, the first time derivatives of the configurational coordinates, and the coordinate functions themselves. The latter, in turn, represent the visible motion.

According to E. Mayr (in Munson, 1971), living organisms act purposively because they have been programmed. The necessary information stored in the nervous system of the organism is partly genetically fixed (Hull, 1974, p.112) and partly acquired by adaptive learning during the life span of the organism. Typical examples of inhereted programs in humans are the numerous reflexes that are present already at birth. On the other hand, most of the more complex motions executed by humans are the result of individual learning processes and can not be passed on from generation to generation. This is particularly true for sports motions. They are, in general, not essential for the survival of the individuum, or the species, and have therefore neither evolved into an optimal form nor are they deposited in the genetic code. This is why the detailed optimal pattern of a specific sports motion for a given athlete is generally not known. It is the development of appropriate optimization techniques where biomechanics of sports faces one of its greatest challenges for the future.

We shall now present the ***formulation of the myocybernetic optimal control problem*** as it pertains to sports performance optimization (Hatze, 1981, 1993). This will aid us in identifying those subject areas which need to be addressed in the subsequent discussion. Let $q=(q_1,q_2,....,q_f)^T$ denote the vector of generalized coordinates which describe the configuration of a human body model and define the state vector by $x=(q,\dot{q})^T$. Let furthermore denote $\mu=(\mu_1,\mu_2,.....,\mu_m)$ the myo-state vector of the skeletal muscle subsystem consisting of the models of the m muscles incorporated in the body model. These m model muscles are neurally controlled by the control functions $v_{ij}(t)$ and $z_i(t)$, i=1,...,m, j=1,...,9, where $v_{ij}$ are the normalized average stimulation frequencies of all active motor units of the i-the muscle, and $z_i$ is the corresponding motor unit recruitment rate. Finally, let the ***myocybernetic performance criterion*** be defined by

$$J = \Phi(x(T),\mu(T),T) + \int_0^T \left[ \sum_{i=1}^{m} \Lambda_i \dot{E}_i(x,\mu_j,v_i) + L_0(x,\mu,v,z,t) \right] dt. \qquad (1)$$

The ***general myocybernetic optimal control problem*** can then be formulated as follows (details in Hatze, 1981): For a given Lagrangian function $L_0(.)$, terminal function $\Phi(.)$, and given metabolic energy rate functions $\dot{E}_i(\cdot)$ weighted by factors $\Lambda_i$, ***find neural controls*** $v_{ij}(t)$, $z_i(t)$, i=1,...,m, j=1,...,9***, and, possibly, initial states*** $x_0$ ***and*** $m_o$ ***, that afford a minimum value to J subject to*** the differential equality constraints (the system dynamics)

$$\dot{x} - A^{-1}(x)\ \left[ B(x) + Q(x,\mu,t) \right] = 0, \qquad x(0) = x_0, \tag{2}$$

$$\dot{\mu} - g\,(\,x,\mu,v(t),z(t)) = 0, \qquad \mu(0) = \mu_0, \tag{3}$$

the control vector constraints $-\check{z}_i \le z_i \le 1, \quad 0 \le v_{ij} \le 1$, the myo-substate inequality constraints $0 \le n_i \le 1$, i=1,...,m, possible musculoskeletal-subsystem state equality constraints $S(x,\mu,t)=0$, and terminal constraints of the type $G(x(T),\mu(T))=0$. The terminal time T may be fixed or free.

From the above formulation it is clear that the successful application of any optimization procedure presupposes the existence of adequate models of the human skeletal subsystem (differential system (2)), the muscular and, to some extent, the neural subsystem (differential system (3)), the knowledge of the various subject-specific parameter values for a given athlete, and detailed information on the various constraints that are active during the execution of a specific motion. In addition, the precise formulation of the relevant performance criterion is of paramount importance.

## Geometrico-Mathematical Models of the Human Skeletal Subsystem: Morphology, Dimensionality, Skeletomechanical Dynamics

Anthropomorphic models of the human skeletal subsystem (so called hominoids, Hatze (1980a)) of varying complexity are used most extensively and for various purposes. More recent applications include planar locomotion simulation studies (Taga, 1995), and motion assessments in microgravity (Zinkovsky et al., 1996) both employing a seven-segment body model (head, trunk, arm, forearm, thigh, shank, foot), the use of a 13-segment model for investigations into the segmental energy contributions in human locomotion (Correa et at., 1997), and the use of a 14-segment body model consisting of pelvis, trunk, thorax, neck, head, two arms, forearms, thighs, legs, and feet for three-dimensional quantitative kinematic cycling analysis (Rodano and Squadrone, 1997). A more elaborate 15-segment model has recently been proposed by Miyanishi et al. (1997) for the three-dimensional analysis of baseball and discus throwers. Earlier modelling attempts are described in Hatze (1980b).

Probably one of the most complex anthropomorphic models currently in existence is the three-dimensional 17-segment hominoid of the author (Hatze, 1980b). It consists of the following segments: abdomino-pelvic segment, abdomino-thoracic segment, head-neck, left and right shoulders, arms, forearms, hands, thighs, legs and feet. Its appearance is shown in Figure 1 on the next page. The model possesses 42 skeletomechanical degrees of freedom for three-dimensional motion and offers the following advantages over similar but less sophisticated body models: it includes the shoulders as separate entities and subdivides segments into small mass elements of different geometrical structures, thus allowing the shape and density fluctuations of a segment to be modelled in detail; in general, no assumptions are made on segmental symmetry, and principal axes transformations are performed whenever necessary. The model also differentiates between male and female subjects (exomorphic differences, different density functions and mass distributions), adjusts the densities of certain segmental parts according to the value of a special subcutaneous-fat indicator, and fully accounts for the specificities of pregnancy and obesity. The set of segmental parameter values can be determined by the use of the BIOMLIB® computer program ANSEPA from a battery of 242 anthropometric measurements taken directly from the subject. The input data errors are drastically reduced by performing direct measurements rather than indirect measurements from photo images. The overall accuracy of the model is better than 3 % with a maximum error of about 5 %. The features listed above were confirmed by comparing experimentally determined parameter values (volumes, masses, coordinates of mass centroids, principal moments of inertia, coordinates of segment origins) with model predictions for a large number of subjects.

After introducing some of the anthropomorphic models (hominoids) currently in use, we shall briefly discuss the major problems encountered in creating appropriate body models. The first decision to be taken is that on the *morphology of the hominoid*, i.e., on the number and shapes of its segments, and the types of the joints. The hominoid morphology defines the number, f, of skeletomechanical degrees of freedom of the (unconstrained) model. While for some purposes extremely simple two- or three-segment models may suffice, this is certainly not the case for hominoids to be used for analysis, simulation, and optimization of sports performance. In this case, the most sophisticated models will just be adequate because of the large variety and complexity of sports motions.

Having decided on the fundamental morphology of the hominoid, the next obstacle encountered in considering body segmentation is the definition of *sharp intersegmental boundaries* which, in the real biosystem, do not exist but are fuzzy and change during segment motion. It is, however, possible to circumvent these difficulties by appropriately defining the intersegmental boundaries as surfaces generated by tracing out the average layers of the fuzzy sets of the respective boundary particles (Hatze, 1980b).

The next problem facing the modeller is the *non-rigidity of the body segments*. It is well known that parts of segments (muscles, connective tissue, organs, etc.) execute movements relative to the skeleton. In addition, other factors such as breathing, non-stationary joint axes, changes in the distribution of body fluids, etc. also contribute to the non-rigidity problem. There have been attempts to account

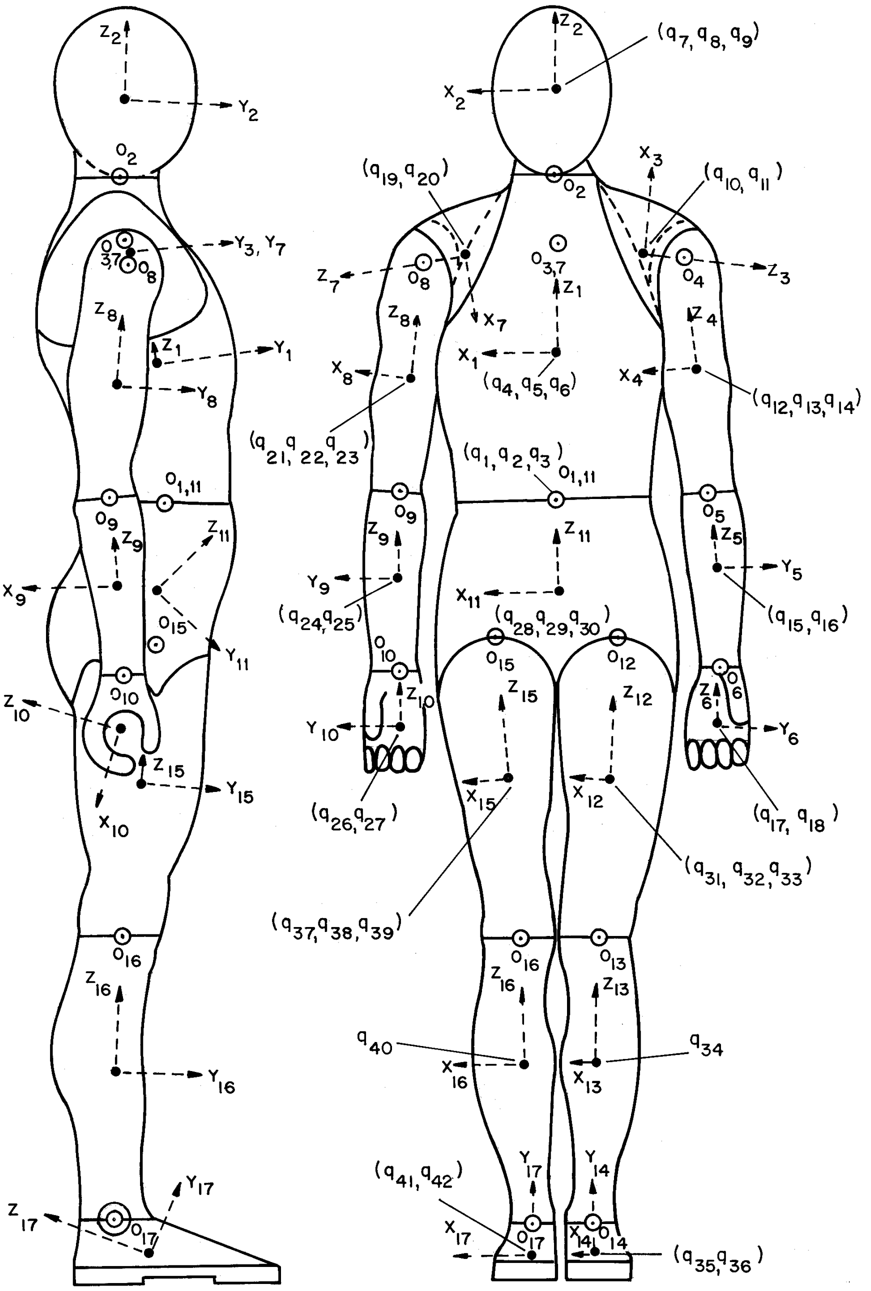

***Fig. 1:*** *Lateral and frontal view of the 17-segment hominoid of Hatze (1980b). The shapes of the segments as shown reflect the actual morphologies of the model segments. The local (segment-fixed) coordinate systems and configurational coordinates* $q_1,\ldots,q_{42}$ *are also displayed.*

for some of these effects by attaching so-called wobbling masses to rigid-rod segment model. Gruber (1987), for instance, used such a three-link wobbling-mass body model to predict reaction forces and moments in the knee and the hip during vertical-jump landings. She found tremendous differences between the predictions of the wobbling-mass and the rigid-segment model, amounting in some case to a few tousand percent ! The hip moment, for example, would show a positive peak of 1300 Nm during the first 0.01 s of the motion for the rigid-segment model, while the corresponding value for the wobbling-mass model was virtually zero (Gruber 1987, Fig. 26d). Even more incredible were the predictions of the horizontal knee reaction forces: A positive peak of about 1000 N during the first 0.01 s for the rigid-segment model and a sign reversal to a negative peak of about – 230 N for the wobbling-mass model (Gruber, 1987, Fig. 26a). No attempt was made to explain these excessive discrepancies or to validate the models in any way. Instead, the wobbling-mass model predictions were accepted as realistic and correct. Although this study again demonstrates the limitations and dangers associated with the use of oversimplified models (the 3-link model of Gruber contained no feet, among other shortcomings), an effort has at least been made to somehow incorporate into the model the soft-tissue behavior of segmental substructures. However, such endeavors are doomed to failure from the outset, for the following reason: For most segments, the soft-tissue substructures consist predominantly of muscles which change their consistence during action from very soft in the relaxed state to almost rigid in the completely contracted state. In the course of a motion, the various muscles attached to a specific segment may change their contractive state, and hence their consistence to anything between these two extremes. In addition, they also act to a certain extent on other soft tissue structures such as blood vessels and connective tissue. All of these effects combine in such a way as to continously vary the consistence of the segmental soft-tissue substructures during motion, in a virtually unpredictable manner. Moreover, the maximal excursions relative to the skeleton of the soft-tissue structures are comparatively small, even in the relaxed muscle state. Taking all of these arguments into account, the *rigid-body approximation* for segments *appears justifyable* (see also Zajac and Winters, 1990, p.125), especially since the shape fluctuations, exomorphic differences, varying density distributions, and asymmetries of individual segments are all accounted for (Hatze, 1980b).

While there is little dispute about the fact that head, upper arms, forearms, thighs, legs, and pelvis constitute, to some extent, segmental units (once the rigidity assumption has been accepted), the situation is not so clear cut for segments like the neck, shoulder, thorax, abdomen, hands, and feet. Indeed, these body parts can not really be called segments, since they consist of highly intricate bony structures connected to various types of soft tissue. Consider, for example, the thorax. The bony structure consists of the spine and the rips (the shoulders are regarded separate segments), while the inner organs, muscles, connective tissue, skin, etc. constitute the soft tissue components. A reasonable division of the thorax into subsegments is virtually impossible. Even if each of the thoracic vertebrae would be considered a disk-shaped subsegment, there would be no corresponding soft-tissue structures that could be regarded as belonging to that disk. Yet the thorax is obviously a flexible skeletal structure that can bend and rotate in all directions. Here, the modeller is in a real predictament: He can either attempt to create a complex and fairly realistic thorax model comprising some 340 interconnected hard- and soft-tissue subsegments and face the gigantic task of combining these with the remaining segments to form the complete skeletal subsystem, or he can consider the abdomino-thoracic complex as a single (rigid) segment, assuming that all rotational degrees of freedom of this pseudo-segment reside in a (pseudo-)ball-and-socket joint located in the abdomino-pelvic region. For simulating gross body motions not involving internal thorax motions, the single-segment thorax model will suffice for most purposes, while for detailed investigations into the responses of the spine to stresses during lifting tasks, the complex thorax model would be appropriate. Similar arguments apply to the segmental modelling of the neck and the abdomen.

A somewhat different approach is required for the hands an the feet. These are terminal segments especially designed for manipulating, respectively contacting, the environment. They are also special in the sense that their bony content is comparatively large and of a rather complicated structure. Because hands and feet are the segments predominantly responsible for the mechanical communication of the hominoid with its surroundings, they are the ones who most frequently experience the environmental contact forces. We may therefore postulate that these segments can be modeled as single units as far as the segmental dynamics is concerned, but should subsequently be extended to more complex structures to account more properly for their intended functional purpose.

Based on the above considerations we are lead to conclude that the *skeletomechanical model comprising 17 segments* might be appropriate for simulating a large class of gross body, in particular sports motions.

As far as the *dimensionality* of hominoids is concerned, the current trend is clearly towards the use of three-dimensional (spatial) models. The rationale behind this is the fact that there exists not a single human motion that is really planar but only motions which may be considered to possess *predominantly* planar components. Why then is the overwhelming majority of musculoskeletal models in use today still of the planar type and therefore unrealistic? A partial answer is to be found in the fact that, until fairly recently, the skeletodynamical (differential) equations of motion were much easier to derive for 2D-hominoids than for 3D-models. This has changed with the advent of computer programs for the automatic generation of the equations of motion by symbolic manipulation, such as NEWEUL, AUTOLEV, or SYMBA (see Yamaguchi, 1990). Other problems associated with the transition to three-dimensional hominoids are the requirement of more segmental parameters (at least seven instead of three) and the complexity of 3D joint structures and muscle architecture, as well as the difficulty in interpreting correctly the various passive and active torques generated at a specific joint in relation to the configurational joint coordinates. These problems are, however, not unsurmountable.

The *formulation of the skeletomechanical equations of motion* for 3D-hominoids is greatly facilitated by the availability of special computer programs, as already mentioned above. The human skeletal subystem may be regarded an interlinked multi-body system which implies that formalisms for the generation of equations of motion for 3D multi-body systems may be used for establishing the hominoid dynamics. Such formalism may be based on Newton's method, Langrange's equations, Kane's method, or D'Alembert's principle (see, for instance, Andrews, 1995; Wittenburg 1977).

A prerequisite for establishing the skeletomechanical equations of motion is the definition of appropriate spatial and segment-fixed axes systems, and skeletomechanical configuration coordinates that unambiguously determine the hominoid configuration at any point in time. For technical reasons, it is most advantageous to fix the origins of the segmental Cartesian coordinates systems at the mass centroids of the segments, and orientate the axes such that they constitute principal axes of inertia (see Fig. 1).

The number and type of *skeletomechanical configuration coordinates* depends on the geometry of the model joints that connect the model segments. Each human joint has, in principle, six degrees of freedom (three rotational and three translational). In many cases, however, the excursions in the direction of some of these six degrees are minute so that they may be ignored for most practical purposes, thereby effectively reducing the number of degrees of freedom to be considered. The general problem is to decide whether or not a certain articular degree of freedom is to be regarded irrelevant. The extremely complicated kinematic structure of the natural joints does not necessarily have to be incorporated in all detail in the inertial part of the equations of motion but must be taken into account in the part containing the kinematics of the driving torques of the system. On these premises, it turns out that the configuration of the 17-segment hominoid displayed in Figure 1 can be described by 42 generalized coordinates $q_i$, of which 3 are linear ($q_1$, $q_2$, $q_3$) and 39 angular. The skeletomechanical state space dimension of this model is therefore 84, which renders it a large-scale model the skeletodynamics of which is described by the differential system (2).

## Neuromuscular Models

In the present context it is, of course, neither possible nor desirable to indulge in extensive discussions on this topic. There exists a large body of contemporary literature on the subject to which the interested reader is referred (e.g. Winters and Woo, ed., 1990). We shall only touch upon for a few subtopics which are currently under debate among specialists and which have a direct bearing on sports biomechanics.

The creation of an adequate model of the myodynamics and myocybernetics of skeletal muscle is not a trivial task. One of the most important criteria is the complexity of the muscle model as has been convincingly demonstrated by Lehmann (1990). He has shown that the predictions of a musculoskeletal model are critically dependent on muscle model complexity, with the oversimplified model versions producing grossly erroreous results.

The most important properties of the passive and active structures of the contractile machinery to be incorporated in any realistic muscle model are: the nonlinear viso-elasticity of the passive parallel elastic components, the nonliniear stress-strain relationship of the tendinous series elasticities; the exponentially growing size of the individual motor units as well as their recruitment rank order-, size-, time-, and length-dependent excitation states; the recruitment rank order-, contraction-velocity-, and length-dependence of the contractile force; and the stretch potentiation of the contractile force which again is nonlinearly dependent on the recruitment rank order, the excitation, the contractile element length and velocity, and the contraction history. The situation is illustrated in Figure 2 below.

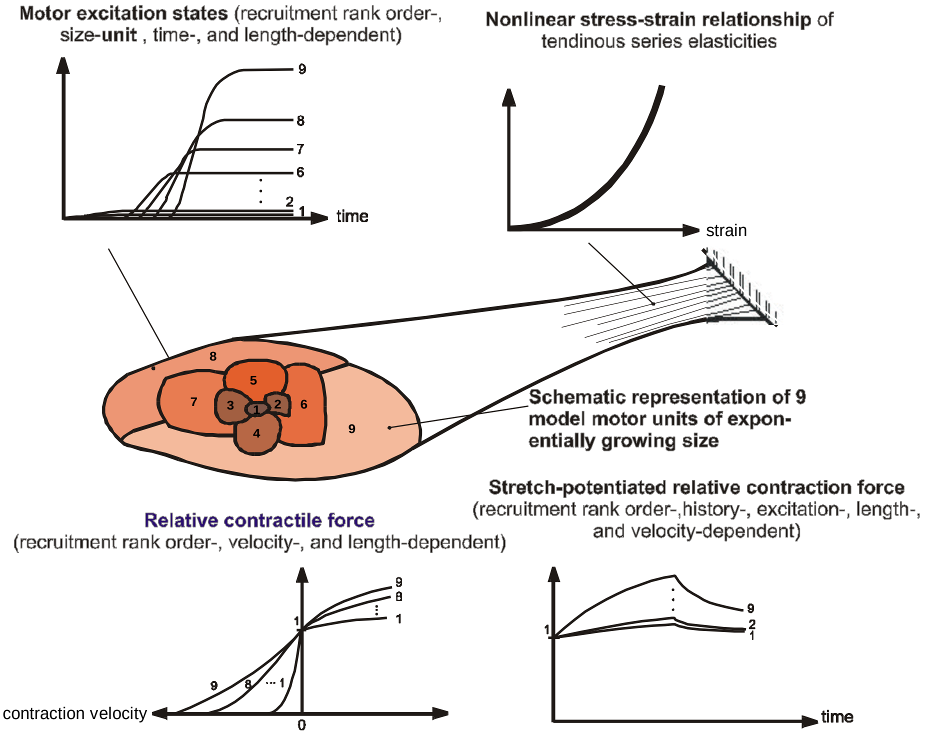

***Fig. 2 :*** *Illustration of the morphological and functional characteristics of mammalian skeletal muscle (with the exception of parallel elasticities) to be incorporated in any realistic muscle model.*

The identification of the predominant characteristics of the myostructures is one thing, the appropriate modelling of their dynamics an other. In the present section we shall restrict our attention to a (somewhat superficial) description of the salient features of the complex myocybernetic control model of a skeletal muscle of Hatze (1977, 1978, 1980c, 1981, 1990, 1997a) as used in the present large-scale 17-segment myoskeletal model. In addition, we shall confirm the findings of Lehmann as regards the influence of model complexity on prediction accuracy. In constructing the model, the following principles were observed: a) The model should be as biologically realistic as feasible and account for all nonlinearities and behavioral peculiarities inherent in the myostructures; b) it should be capable of predicting myocontractive phenomena that were not used in the construction of the model; c) it should adequately mimic the discrete motor unit structure of skeletal muscle since numerous myodynamic properties are recruitment-rankorder-dependent; and d) it should, first and foremost, contain *control parameters that reflect the actual neural controls motor unit recruiment and individual motor unit firing rates, especially with a view to possible extensions of large-scale myoskeletal to complex neuromyoskeletal models*. Because of the latter feature which emphasizes the suitability of the model for neural control applications, it has been termed a “myocybernetic control model” (see also Hatze, 1981).

A certain muscle to be modelled is regarded as consisting of *9 model motor units***,** each having its own specific excitation dynamics, size, and contractile characteristics. The common contraction dynamics is determined by the length change of the whole muscle contractile element. The ***single-muscle myostate vector*** $\boldsymbol{\mu}'$ comprises the normalized motor unit population n, the 9 normalized motor unit calcium ion concentrations $\gamma_\alpha, \alpha = 1,...,9$, the normalized contractile element length $\xi$, and the normalized stretch potentiator $\zeta$. (If fatiguing effects are to be inculded, the additional state variable $\kappa$ and is differential equation are required).

Hence $\boldsymbol{\mu}' := (n, \gamma_1, \gamma_2, ...., \gamma_9, \xi, \zeta)^T$ and the 12 ***single-muscle myostate equations*** are given by

$$
\begin{aligned}
&\dot{n} = \hat{n}\, z(t), && && n(0) = n_o, \\
&\dot{\gamma}_\alpha = m_\alpha\left[ w_\alpha(n) v_\alpha(t) - \gamma_\alpha \right], && \alpha = 1, .....,9 && \gamma_\alpha(0) = \gamma_\alpha, \\
&\dot{\xi} = f_\xi(n, \gamma_\alpha, \xi, \zeta, F^{SE}(\xi, q)), && && \xi(0) = \xi_0, \qquad (4) \\
&\dot{\zeta} = f_\zeta(n, \gamma_\alpha, \xi, \zeta, F^{SE}(\xi, q), \dot{\xi}, t), && && \zeta(0) = 1,
\end{aligned}
$$

when $z(t)$ and $v_\alpha(t)$ are the nomalized recruitment and motor unit stimulation rates respectively, $w_\alpha(n)$ are switching functions, and $f_\xi(.)$ and $f_\zeta(.)$ are highly nonlinear functions of their respective arguments. $F^{SE}(.)$ is the muscle’s output force and the remaining symbols denote constants. The first of these 12 differential equations describes the ***recruitment dynamics,*** the 9 subsequent ones the ***excitation*** **dynamics**, and the two last ones the ***contraction dynamics*** of the muscle. For a set of m muscles constituting the muscular subsystem of the complete neuromusculoskeletal model, the myodynamics is given by the differential system (3).

The predictions of the above motor unit muscle model were compared with those of a simple Hill-type muscle model, with both models incorporated in an arm-forearm skeletal subsystem (Figure 3). Figure 4 shows the model responses in a sequence of contractions that encompasses all important contractive modes: isometric from rest under maximal neural stimulation and at optimal muscle length, followed by a stretch induced by the weight of forearm and hand, which, in turn, was followed by muscle shortening accompanied by forearm flexion. As can be seen, both, the force curve of the Hill-type model and the associated trajectory of the forearm angle $\theta$ show profound deviations from the corresponding curves of the complex model, whereby the force curve of the latter model exhibits all the phenomena known to occur in living muscle (electromechanical delay, stretch potentiation, etc.). In the Hill-type model, these characteristics are absent. This confirms the result of Lehman (1990) that *oversimplified muscle models as part of neuromusculoskeletal system models may produce grossly erroneous simulation results* and should, therefore, not be used.

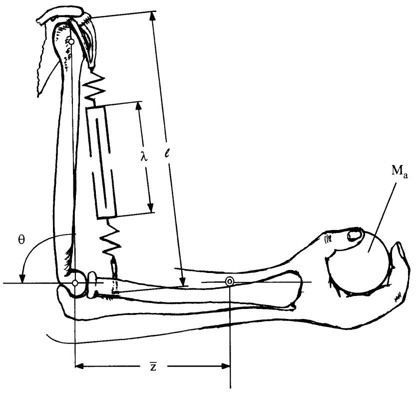

***Fig. 3:*** *Planar musculoskeletal model containing one equivalent arm flexor. The symbol l denotes the muscle length and* $\lambda$ *the length of the contractile element. The mass* $\mathrm{M}_\alpha = 1.25\,\mathrm{kg}$ *has been added and* $\bar{z}$ *denotes the center-of-mass location of forearm + hand +* $\mathrm{M}_\alpha$.

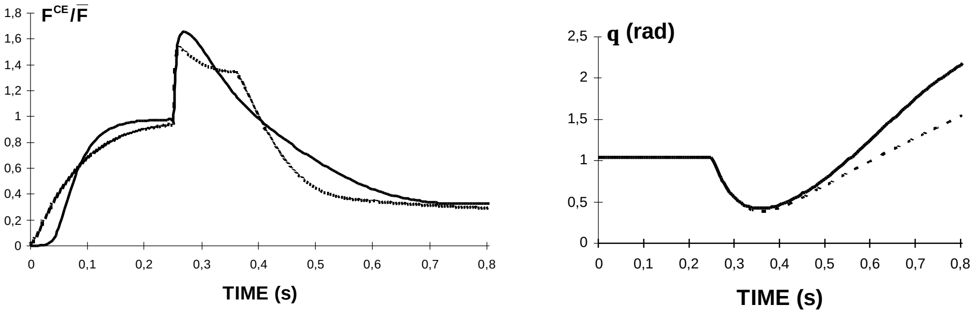

***Fig. 4:*** *Normalized contraction force* $\mathrm{F}^{\mathrm{CE}} / \bar{\mathrm{F}}$ *(left hand graph) and elbow angle* $\theta$ *(right hand graph) as simulation responses for the complex motor-unit muscle model (continuous lines) and the Hill-type muscle model (interrupted lines).*

## Identification of Model Parameter Values

In the introduction section it was mentioned that for a given individual or athlete, the values of the segmental, articular, morphometric, myodynamic, and myocybernetic parameters need to be determined by specific methods in order to individualize the general body model. The parameter value identification problem is a vast field of research in biomechanics so that in the limited space available only a small fragment of the methods in use can be presented. The emphasis will again be on the author’s work.

The set of *segmental parameter values* together with an anthropometric-computational technique (Hatze, 1980b) for determining them has already been introduced in the section on geometrico-mathematical models of the human skeletal subsystem. The technique has later been extended by recording the anthropometric dimensions *videometrically* (Hatze and Baca, 1992). A description of methods for estimating certain *myodynamic* and *myocybernetic* parameter values can be found in Hatze (1981). More recently, a new technique for generating improved estimates of maximum isometric muscle torque functions has been devised (Hatze, 1998a). The method is based on average function amplification and permits, at a sequence of distinct angular joint positions, improved estimates of the maximum isometric torque generating capacity of a group of muscles spanning the joint in question. The development of such methods is necessitated by the fact that both, model predictions and experimental observations, indicate a purely neurally based insufficiency of skeletal muscles in attaining their true isometric maximum force generating capacity in voluntary contractions. Fundamental to the present technique is the hypothesis that occasionally observed large positive deviations of voluntary maximum effort muscle torques from the experimental average maximum torque curve are comparatively unlikely events demonstrating the presence of reserves that are normally not utilized. Since the average curve is assumed to represent the shape of the basic function, the positive outliers can be used to amplify the average function and thereby obtain an improved estimate of the muscle group's real maximum torque generating capacity. With the moment arm functions and certain constants relating to the respective muscles in the group known, a fairly reliable estimate of the maximum isometric forces and the length-tension properties of the individual muscles can be obtained. The technique is demonstrated for the case of the elbow flexors in Figures 5 and 6.

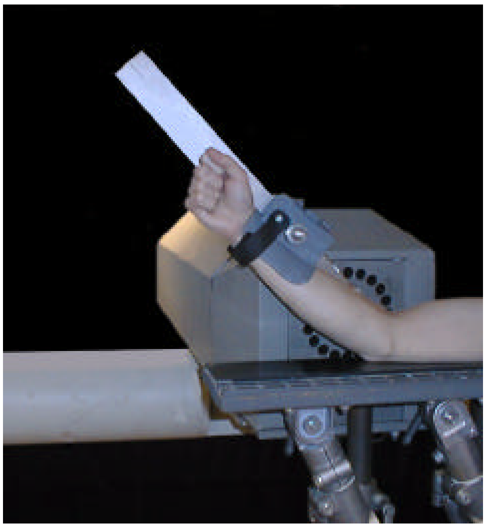

***Fig. 5:*** *Experimental arrangement utilizing a muscle dynamometer for measuring maximum-effort isometric torques of the elbow flexors.*

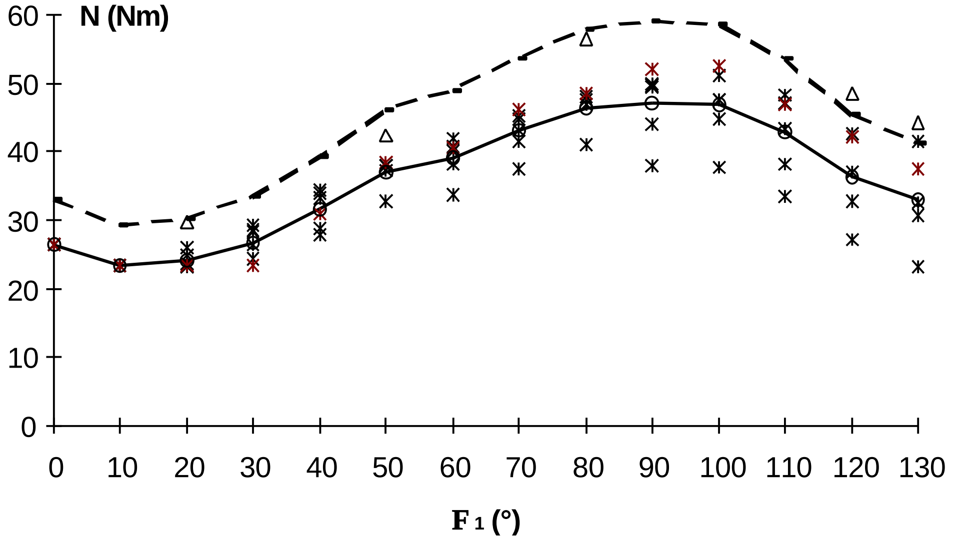

***Fig. 6***: *Isometric elbow flexor torques N resulting from six sequences of maximum voluntary contractions at distinct elbow joint angles* $\Phi_1$. *The continuous curve marks the average computed without inclusion of the ouliers (triangular symbols). The curve marked by the interupted line represents the improved estimate.*

An example for the identification of subject-specific *articular* parameter values is the determination of three-dimensional articular boundaries and passive joint torques. The system dynamical equations (2) contain joint range limitation torques $Q^L$ in the torque vector function $Q(x, \mu, t)$. These internal passive torques restrict the range of joint motion owing to the presence of skeletomechanical structures such as ligaments, connective tissue, cartilage, joint capsules, bones, and, to a significant extent, the parallel elasticities of uni- and biarticular muscles spanning the joint in question. The method developed by the author (Hatze, 1997b) permits the experimental determination of subject-specific one-, two-, or three-dimensional joint torque functions and articular boundaries, and is demonstrated by the example of the passive elbow joint torques in Figure 7 on the next page.

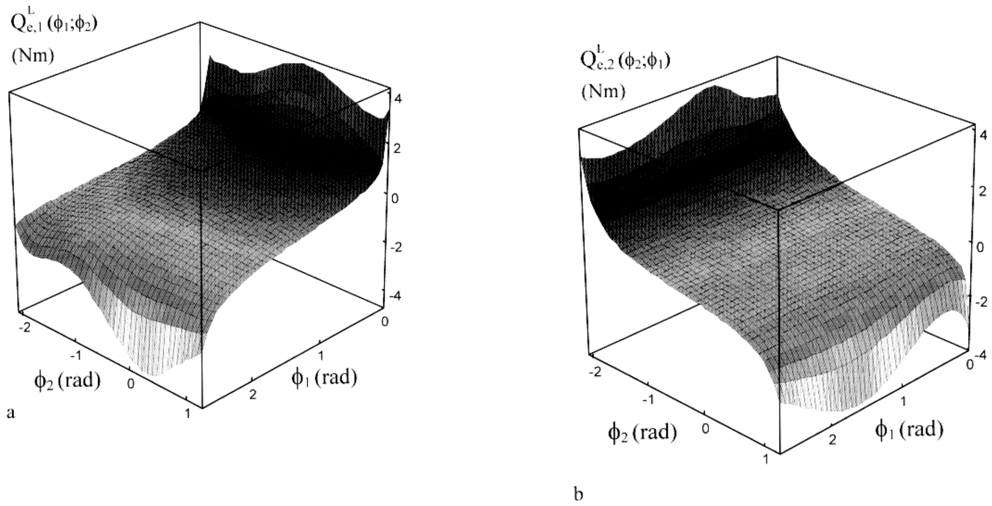

***Fig. 7:*** *Surface representations of the passive elastic torque functions* $Q^L_{e,1}(\phi_1;\phi_2)$ *and* $Q^L_{e,2}(\phi_2;\phi_1)$ *for the human elbow joint. In the graph (a), the surface was generated by varying* $\phi_1$ *through its whole range for a sequence of given fixed values of* $\phi_2$*, while in the graph (b),* $\phi_2$ *was varied through its range for a sequence of fixed values* $\phi_1$*.*

## Myoskeletal Simulation Model Responses: Hyposensitivity of Skeletal Motions to Neural Control Perturbations

The computer simulation of large-scale systems is not a trivial task. Because of space limitations it is not possible to discuss the numerous problems and pitfalls associated with the practical implementation and execution of the computerized version of the present and similar large-scale myoskeletal models. Instead, we shall proceed to the presentation of some remarkable simulation results concerning *the insensitivity of specific skeletal motions to perturbations of neural control inputs.*

The present large-scale myoskeletal model was used to shed some light on how, and to what extent, random (and deterministic) perturbations of control inputs affect the resulting motion of the system. The computerized version of the combined differential system (2) and (3) as well as all auxiliary algorithms required for its integration were used in the form of the BIOMLIB® computer program HOMYOS to perform the simulation. Eight leg muscle groups in the model were activated, four of which are bi-articular: m. iliopsoas (left and right), m. rectus femoris (l. and r.), vasti group (l. and r.), and the hamstring group (l. and r.). The segmental, articular, myodynamic, and morphometric input parameter values were determined from a healthy, 23-year old male subject. Where necessary, data for the left and right segments and muscles were equalized in order to guarantee left-right symmetry of the motion, and a mass of 10 kg was attached to each foot in order to slow down the otherwise too rapid kicking movement. The hominoid was suspended from its head and the pelvic segment was constrained in all directions by appropriately dimensioned visco-elastic external forces, as shown in Figure 8. The unperturbed (nominal) neural control input functions were of bang-bang form and chosen such as to produce the symmetric double-legged kicking motion of the vertically suspended hominoid depicted in Figure 8. Simulation time was 0.5 s.

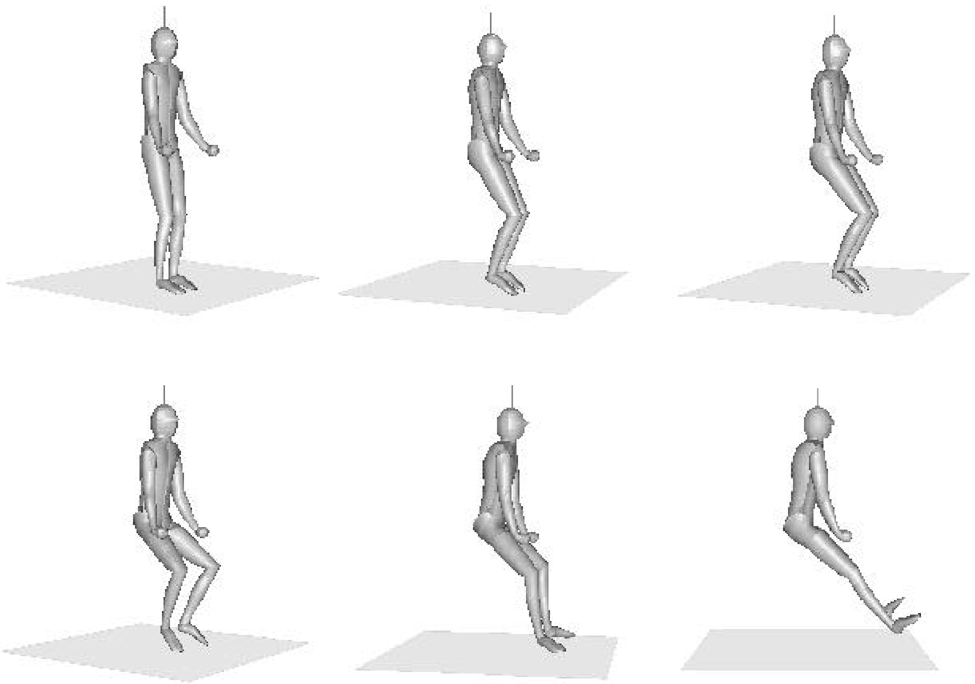

***Fig. 8**: Six configurations (left to right, top to bottom) of a kinematic animation sequence showing the figural appearance of a three-dimensional, 17-segment, 240 muscle myoskeletal model, suspended from its head and executing a double-legged kicking motion.*

Gaussian random and deterministic constant perturbations of neural interspike intervals $\tau_{\alpha j}, \alpha = 1,....,9$, and recruitment rates $z_j$ of the j-th muscle were used to produce deviations from the nominal (unperturbed) controls. The measure of perturbation used for all variables y is the ***normalized variance*** $\Delta^2 = \sigma^2 / \sigma_\omega^2 = T^{-1}\int_0^T \left[ y(t) - y^0(t) \right]^2 dt / \sigma_\omega^2$, over the simulation time interval T. The function $y^0(t)$ is the nominal value of the variable in question and $\sigma_\omega^2$ the corresponding reference variance of chaotic excursions (Hatze, 1995). The results of this simulation were remarkable indeed and are partly summarized in Figure 9.

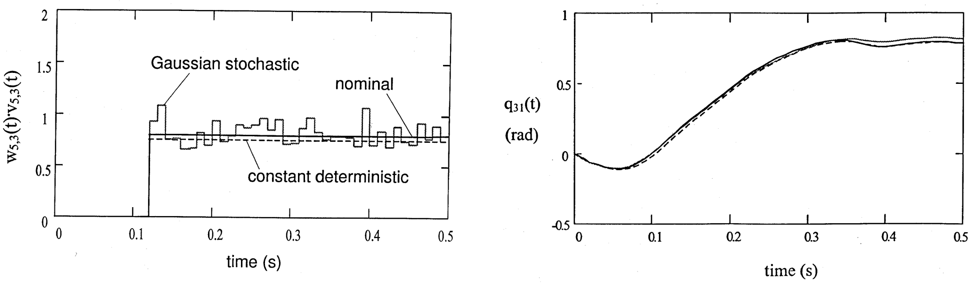

***Fig. 9:*** ***Left:*** *Nominal, Gaussian stochastic, and constant deterministic control function* $w_{5,3}$ *(t).* $v_{5,3}(t)$ *of the* 5th *motor unit in muscle 3 as a representative example of the remaining 71 controls.* ***Right****: Left hip angle* $q_{31}(t)$ *(flexion-extension) corresponding to the above controls. The second and third hip angles and the knee angle are not shown.*

As can be seen from Figure 9, *large perturbations of the neural control inputs result in comparatively small variations of the corresponding coordinate trajectories*, with standard deviation ratios for random perturbations ranging from 7.03 to as much as 23.87. It should however, be emphasized that these results pertain to this particular kicking motion only although they possess some generality.

The observed hyposensitivity of skeletal motion to neural perturbations has its origin partly in the muscular and partly in the skeletal subsystem. Among the muscular causes of this phenomenon are the comparatively sluggish reaction of the intramuscular calcium ion concentration to variations of the stimulation rate, the nonlinearity of the active-state function, and the influence of varying, recruitment rank-order dependent motor unit properties. A further reason for the occurrence of the hyposensitivity phenomenon is, of course, the smoothing action of the inertia of the skeletomechanical system components which act like a low-pass filter.

The *most important implications* of these findings are the facts that comparatively chaotic neural controls may produce well coordinated motions of the skeletal subsystem and that the *neuromusculoskeletal inverse dynamics problem is highly ill-conditioned*. It does, in fact, belong to the class of so-called physically (and mathematically) incorrectly-posed problems which do not, by definition, possess unique solutions. In practice this means that, in general, the reconstruction of neural control inputs that correspond to an observed motion is not possible with any reasonable confidence. A similar argument holds true also for the inference from motion data of muscular joint torques. These statements are strongly supported by the frequently made experimental observation of considerably varying EMG records that correspond to repetitions of one and the same stereotyped motion. Also, DeLuca and Erim (1994, Fig. 1B) have demonstrated that mean firing rates of motor units in the isometrically contracting tibialis anterior muscle may vary simultaneously by as much as 14 % without significantly affecting the muscle force level.

The present simulation result together with experimental findings convincingly demonstrates the usefulness of adequate models of the human neuromusculoskeletal system.

## Successful Application of Biomechanics to Sports Related Problems: Selected Examples

Three examples from the author's work will be presented: the biomechanical analysis of a rock'n roll accident as part of an expert opinion; the development of an objective method for testing the quality criteria of tennis rackets; and an objective evaluation of the validity and reliability of jumping performance tests currently in use.

The first example illustrates the usefulness of biomechanics in elucidating the events that lead to a *rock'n roll practice accident* with severe facial and internal injuries. The dancers practised a Betterini somersault in which the lady was thrown into the air as shown in Figure 10, performed a forward somersault, and, upon completition of the rotation, was supposed to be caught by her partner to assist her in landing.

***Fig. 10:*** *Six configurations of a computer simulation animation sequence showing various phases of an overrotated rock'n roll Betterini somersault. From top left to bottom right (pre-impact times in brackets): Initial position (1.243 s), somersault half completed (0.949 s); somersault completed, overrotation already obvious (0.466 s); female dancer passes partner (0.277 s); accident unavoidable (0.067 s); impact (0.0 s).*

However, the somersault rotation was too rapid with the result that she passed the hands of her partner in an almost prone position, making it impossible for him to catch her (see Fig. 10) . A third person, the helper in charge of assisting her in case of overrotation, did not react and she hit the floor with the chin first.

A court case ensued and the author was appointed expert. The questions to be clarified were: a) Would it have been possible for the helper to react in time and support the falling dancer, and thereby prevent or at least reduce the injuries, and b) can the helper's statement be true that the dancer completed the somersault in the air, then descended in an upright position, landed on her feet and only then fell forward and hit the floor with the chin. It was decided to perform a computer simulation of the somersault with the segmental and morphometric parameter values taken from the female dancer and her partner. In addition, the configuration of the female dancer relative to space and also relative to her partner upon completion of the somersault could be reconstructed fairly accurately from eye witness reports.

The simulation provided the following answers: a) There was sufficient time (about 0.47 s) for the helper to react and provide at least partial aid to prevent head-first impact on the floor; and b) the helper's statement on the landing configuration of the dancer could not have been correct, because the law of the conservation of angular momentum implied the continuation of the rotation prior to landing, making the landing in an upright position impossible. The helper's statement was also contradicted by eye witness testimonies.

The second application of biomechanics to sports to be discussed relates to the *objective testing of the quality criteria of tennis rackets*. Several biodynamical properties of tennis rackets such as vibrational characteristics, direction control, and the coefficient of restitution depend critically on the constraining mode of the handle. The racket response to ball impact differs fundamentally for rigid mechanical clamping of the handle and for hand-held gripping. In order to test objectively the biodynamical characteristics of tennis rackets under standardized but biomechanically realistic conditions, a mechano-electronic replica of the human hand/arm system, termed manusimulator, was developed (Hatze, 1992) and is shown in Figure 11 on the next page.

The practical implementation proved the method reliable with an exceptionally high degree of reproducibility of the test results which makes it especially suitable for prototype testing during the development of a new product. For this purpose the manusimulator testing method was used successfully in cooperation with Prince Engineering Company (Princeton, USA) and Dunlop Slazenger Internat. Ltd. (UK). The results of series testing of various types of tennis rackets revealed the superiority of this testing procedure over testing by human test players, as described in Hatze (1992).

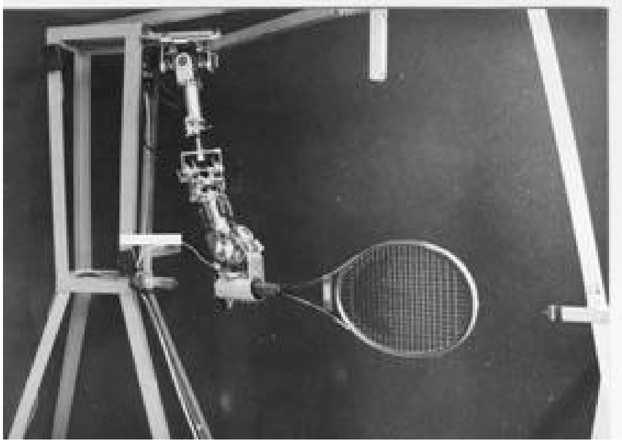

***Fig. 11:*** *The manusimulator, a mechano-electronic device for testing objectively the biodynamical properties of tennis rackets. Shown are the artificial arm mounted on its base, the test racket held by the simulator hand, and the three lasers for precise position adjustment of the racket (From Hatze, 1992)*

The final example to be presented relates to the *validity and reliability of methods used for testing vertical jumping performance.* A close scrutiny of existing procedures revealed (Hatze, 1998b) that the commonly used jumping ergometer method of Bosco, Luhtanen, and Komi (1983) is based largely on invalid assumptions which lead to unpredictable effects entailing average errors of about 5 % associated with an unacceptably large standard deviation of 4.48 % (Hatze, 1998b). This seriously calls into question the applicability of the jumping ergometer method of Bosco et al. for evaluating the jumping performance of athletes.

## Perspectives on the Future Development of Sport Biomechanics

Massive research efforts will be, no doubt, devoted in future to the *improvement of sports equipment* and increased attention will focus on sports engineering in general. Indeed, the recent launching of the new journal "Sports Engineering" is proof of the rising need for condensed information on the subject. The optimal adaptation of the equipment to the need of the customer is currently one of the dominant factors in equipment design. Another major field of sport technology is the development of appropriate methods for testing objectively the quality criteria of sport equipment, and example of which has been presented in the previous section by the testing of tennis rackets.

The biomechanics of sports offers a large variety of subtopics that are, and should be, investigated in numerous single studies. Too frequently, however, publications are encountered in which large volumes of data together with phenomenological descriptions of an observed event are presented without a clearly defined objective or any reference to the causes underlying the observed phenomenon. The mere measurement and description of the ground reaction forces during for release phase of the javelin throw, for instance, without interpreting their functional form in terms of the myoskeletal factors that determine the throwing distance, is meaningless and constitutes a futile exercise. We would be well advised to guard against this type of pseudo research.

Present international trends in biomechanics reveal an increasing use of complex large-scale neuromusculoskeletal models for simulation, prediction, and optimization. It is to be expected that these trends will also influence the future research directions in sport biomechanics.

## References

Andrews, J.G. (1995). Euler's and Lagrange's Equations for Linked Rigid-Body Models of Three-Dimensional Human Motion. In: Allard P., Stokes, I. A. F., Blanchi J. (eds.), *Three-Dimensional Analysis of Human Movement*, (p. 145-175). Champaign: Human Kinetics.

Boltyanskii, V.G. (1971). *Mathematical Methods of Optimal Control*. New York: Holt, Rinehart and Winston, Inc.

Bosco, C., Luhtanen, P., & Komi, P.V. (1983). A simple method for measurement of mechanical power in jumping. *European Journal of Applied Physiology*, **50**, 273-282.

Correa, S.C., Amadio, A.C. , Glitsch, U., & Baumann, W. (1997). Mechanical energy contribution of the segments to human locomotion. *Proceedings XIV. Congress of the International Society of Biomechanics*, Tokyo, p.135.

DeLuca, C.J., & Erim, Z. (1994). Common drive of motor units in regulation of muscle force. *Trends in Neurosciences*, **17**, 299-305.

Gruber, K. (1987). Entwicklung eines Modells zur Berechnung der Kräfte im Knie- und Hüftgelenk bei sportlichen Bewegungsabläufen mit hohen Beschleunigungen. *Ph.D.-Dissertation, Laboratory of Biomechanics, ETH Zürich.*

Hatze, H. (1976). Eine Fundamentalhypothese der Bewegungslehre des Sports. *Sportwissenschaft*, **6**, 155-171.

Hatze, H. (1977) . A myocybernetic control model of skeletal muscle. *Biological Cybernetics*, **25**, 103-119.

Hatze, H. (1978). A general myocybernetic control model of skeletal muscle. *Biological Cybernetics*, **28**, 143-157.

Hatze, H. (1980a). Neuromusculoskeletal control systems modeling – A critical survey of recent developments. *IEEE Transactions on Automatic Control*, **AC-25**, 375-385.

Hatze, H. (1980b). A mathematical model for the computational determination of parameter values of anthropomorphic segments. *Journal of Biomechanics,* **13**, 833-843.

Hatze, H. (1980c). Discrete approximation of a continuous myocybernetic model of skeletal muscle. *Special Report SWISK 18*, CSIR, Pretoria.

Hatze, H. (1981). *Myocybernetic Control Models of Skeletal Muscle*. Pretoria: University of South Africa Press.

Hatze, H. (1990). The Charge-Transfer Model of Myofilamentary Interaction: Prediction of Force Enhancement and Related Myodynamic Phenomena. In: Winters, J.M. & Woo, S.L-Y., (eds.), *Multiple Muscle Systems,* (pp. 24-45). New York: Springer.

Hatze, H. (1992). Objective biomechanical determination of tennis racket properties. *International Journal of Sport Biomechanics*, **8**, 275-287.

Hatze, H. (1993). Reasons why progress in the optimization of sports performance is painfully slow. *Proceedings XIV. Congress of the International Society of Biomechanics*, Paris, 32-33.

Hatze, H. (1995). The extended transentropy function as a useful quantifier of human motion variability. *Medicine and Science in Sport and Exercise*, **27**, 751-795.

Hatze, H. (1997a). Muskelmechanische Aspekte eines 3D Large-Scale Simulationsmodells des menschlichen neuromuskuloskelettären Systems. *Österreichische Zeitschrift für Physikalische Medizin und Rehabilitation,* **7** (Suppl. 2), 9-15.

Hatze, H. (1997b). A three-dimensional multivariate model of passive human joint torques and articular boundaries. *Clinical Biomechanics*, **12**, 128-135.

Hatze, H. (1998a). Improved estimation of maximum isometric muscle torque functions. *Proceedings of the 10-th International Conference on Mechanics in Medicine and Biology,* Honolulu, 33-35.

Hatze, H. (1998b). Validity and reliability of methods for testing vertical jumping performance. *Journal of Applied Biomechanics*, **14**, 127-140.

Hatze, H., & Baca, A. (1992). Contact-free determination of human body segment parameters by means of videometric image processing of an anthropomorphic body model. In: *Applications of Digital Image Processing XV*, SPIE-Vol. 1771, 536-546.

Hull, D.L. (1974). *Philosophy of Biological Science*. Englewood Cliffs: Prentice-Hall.

Lehman, S.L. (1990) Input Identification Depends on Model Complexity. In: Winters, J.M. & Woo, S.L-Y, (eds.), *Multiple Muscle Systems,* (pp. 94-99). New York: Springer.

Miyanishi, T., Sakurai, S., & Wakayama, A. (1997) A three-dimensional analysis of angular momentum in the Asian top discus throwers. *Proceedings on the XVI. Congress of the International Society of Biomechanics*, Tokyo, 209.

Munson, R. (ed.), (1971). *Man and Nature*, New York: Delta Books.

Rodano, R., & Squadrone, R. (1997). A procedure for quantitative kinematic analysis in cycling. *Proceedings of the XIV. Congress of the International Society of Biomechanics*, Tokyo, 154.

Taga, G. (1995). A model of the neuromusculoskeltal system for human locomotion. I. Emergence of basic gait. *Biological Cybernetics*, **73**, 97-111.

Winters, J.M. & Woo, S.L-Y., (eds.) (1990) *Multiple Muscle Systems*, New York: Springer.

Wittenburg, J. (1977). *Dynamics of Systems of Rigid Bodies.* Stuttgart: Teubner.

Yamaguchi, G.T. (1990) Performing whole-body simulations of gait with 3D, dynamic musculoskeletal models. In: Winters J.M., & Woo S.L-Y., (eds.), *Multiple Muscle Systems,* (pp. 663-679). New York: Springer.

Zajac, F.E., & Winters, J.M. (1990). Modeling musculoskeletal movement systems: joint and body segmental dynamics, musculoskeletal actuation, and neuromuscular control. In: Winters, J.M., & Woo, S.L-Y.,(eds.), *Multiple Muscle Systems,* (pp. 121-148). New York: Springer.

Zinkovsky, A.V., Sholuha, V.A., & Ivanov, A.A. (1996). *Mathematical Modelling and Computer Simulation of Biomechanical Systems*. London: World Science Publishing Company Ltd.