982 *KSME International Journal, Vol. 15, No. 7, pp. 982~994, 2001*

Modelling of Ergonomics and Muscular Comfort

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Commercially available software packages permit to position human models of various geometries in practical scenarios while respecting the anatomical constraints of the skeletal joints and of the bulk of the bodies. Beyond such features, the PAM-Comfort™ software has been conceived to provide direct access to the muscular forces needed by humans to perform physical actions where muscle force is required. The PAM-Comfort™ human models are made of multi-body linked anatomical skeletons, equipped with finite elements of the relevant skeletal muscles. The hyper-static problem of determination of muscle forces is solved by optimisation techniques. Voluntary stiffening of muscles can be added to the basic contraction levels needed to perform a specific task. The calculated muscle forces obey Hill's model. The model and software have been applied in several interesting scenarios of various fields of application, such as car industry, handling of equipment and sports activities.

Key Words : Skeletal Joints, Muscular Forces, Hyper-Static Problem Optimisation Techniques, Hill's Model

Nomenclature

a_i : Activation level of a muscle
c : Voluntary contraction level of a muscle
F : Muscle force
L : Muscle length
S : Muscle physiological cross section area

1. Introduction

1.1 Comfort and fatigue

Ergonomic design and comfort of handling objects and operating of equipment are often linked to the activity of the human body, expressed in muscle action, depending on the arrangement and interaction of man and machine. Riding comfort, for example, depends on the mechanical aggression a given transport vehicle imparts on the individual, such as noise, vibration and harshness. Generally speaking, "comfort" may be linked to mechanical, acoustic, thermal, visual and psychological factors. Ergonomic and comfort design minimize the muscular energy spent to perform an action (handling, operating) and the effects of the inevitable aggression of various nature, originating from an action or an operation (driving, etc.). In both cases fatigue of the muscles plays a major role when mechanical factors are involved. Psychological fatigue, however, may also play an important role in the actions of daily life and work. Here only those aspects of ergonomics and comfort are addressed that can be linked to mechanically induced muscle action, such as from sustaining static loads.

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1.2 Muscle action and fatigue

The active forces of the skeletal muscles, for example, enable the human body to sustain a given position under static loads, or, to perform a desired motion. These conscious actions also play a role in the passive dynamic response of the human body, subjected to dynamic loads and vibrations, such as from riding a transport vehicle, or from moderate speed car accidents. However, a muscle can be kept in an activated state at a given level of activation only for a certain period of time, where after the activation level involuntarily drops due to the physiological phenomenon of fatigue. Therefore, the activation levels of the muscular system can provide direct physical information towards the evaluation of comfort or ergonomics under the given circumstances. Similar facts and criteria apply to actions that involve repeated motion (e. g., chain work), which are not discussed here.

1.3 Handling feasibility and comfort

The French Defense Agency (DGA) is designing equipment for the combatant of the next generation (program Felin) and wishes to have a predictive numerical tool for comparing the bodily implications of handling of different equipment, such as helmets, weapons, jackets, etc. Any future equipment design should be tested virtually in its realistic operating ranges, not only from the functional and geometric design point of view, but also from the "comfort" design point of view, based on muscle force analysis. Here "comfort" indicates the feasibility, effectiveness and ease of handling of a new equipment, so as to assure maximum functionality.

1.4 The H-ARB™ model

To satisfy this particular comfort issue, ESI has developed a human articulated rigid body (H-ARB™) model (Robby™) (Haug et al, 1998), based on the skeletal geometry from Viewpoint Data Labs, which closely corresponds to a 50-th percentile male human body (Robbins, D. H., 1983). In a first part of a project, the complete muscular system for the arms, shoulders and neck has been implemented into the skeleton. The muscles are represented by bars, which are connected to the bones at their anatomically correct locations (points of origin and insertion). Their anatomical cross section, which determines the force they can develop at a given activation level, has been taken from different sources found in the literature and in anatomical atlases. The so generated "muscled" skeleton of the upper body can serve to evaluate the muscle forces for tasks involving the upper body. This model is presently extended to the whole skeleton.

1.5 PAM-Comfort™

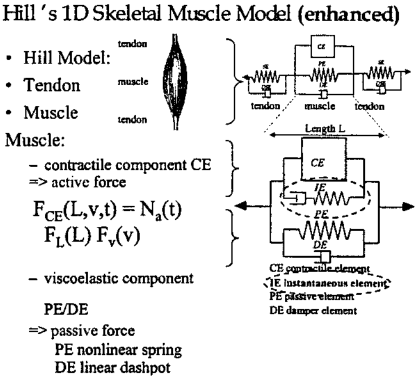
ESI has developed a prototype software, where in the present first level of implementation the active force of each modeled muscle is determined for each loaded static position as the set of muscle forces that will sustain the given skeletal position in static equilibrium, and that will also minimize the amount of spent muscular energy. Since dynamic inertia forces from imposed motions of the body can be considered as equivalent static forces, solutions can also be found in such dynamic cases. Since there are many more unknown muscle forces than skeletal kinematic degrees of freedom, the problem is statically over determined and direct solutions for the muscle forces cannot be found. The solutions are therefore determined by an optimization algorithm, which calculates the active muscle (and external contact) forces, acting on the articulated skeleton (design parameters) by minimizing the active muscle energy (objective function) under zero to full muscle activation levels (bounds) and for static equilibrium (constraints). Extra voluntary or involuntary muscle contractions beyond the levels necessary to equilibrate the imposed static loads can be taken into account by the elaborated software, when the level of extra contraction of the antagonist muscles is specified. Such bracing action may stiffen the skeletal kinematic chain, which may be beneficial in anticipation of shocks (car accidents) or imminent load peaks (weapon recoil), and others.

2. Methodology

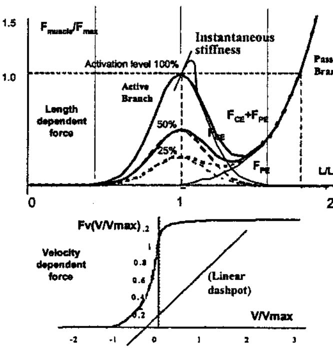
Ergonomics, in its simplest expression, deals with the feasibility and comfort of humans performing tasks of instantaneous quasi-static load carrying under prescribed zero or non-zero motions. If the prescribed motions are non-zero over time, the resulting d'Alembert dynamic forces must be added to the static loads. A procedure to evaluate such simple scenarios is described. Possible future extensions of the methodology can be to find the optimal postures for the required task, or to find optimal sequences of free motions when performing a load carrying task. In the following, the human skeleton is assumed a rigid multi-body system, with the skeletal bones linked by the anatomical joints, and with the relevant skeletal muscles represented by bar finite elements. The considered system can be loaded by external static and dynamic loads, and it can interact with the environment via contact forces. Note that any given skeletal muscle can easily be modeled with more than one bar element, either in parallel, or in series. If the muscle is a surfacic muscle (trapezius), the individual anatomical segments are each represented by a series of bars, which permits to correctly distribute the muscle forces over the skeleton.

2.1 Hill's muscle model

The active and passive skeletal muscle forces are described by the well-known Hill muscle model, see Fig. 1, which was implemented into the PAM-SCL™ Solid Core Library for bar finite element models (Wittek, 1998). This model is valid for *quasi-static* extensions and contractions of skeletal muscles. The active muscle force depends upon the ratio between the length of each muscle in the current configuration and the 'optimal' length, determined in a reference position of the skeleton, sometimes taken as the astronaut"s sleeping position. The active muscle force also depends upon the rate of length change of the muscle. In the case of suddenly applied dynamic loads to the skeleton, Hill's basic model is found inadequate, because it does not provide for the



(a) Basic mechanisms



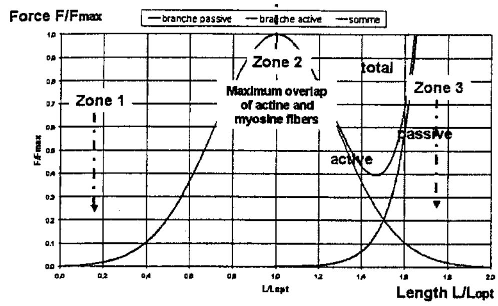
(b) Active and passive length and elongation velocity dependent forces

Fig. 1 Hill's muscle model

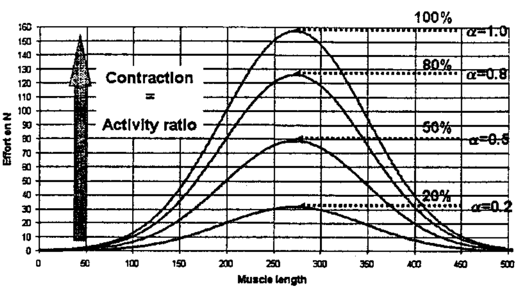
correct dynamic stiffness of activated muscles. The Hill model was therefore augmented to include an *instantaneous dynamic stiffness* under high rates of change of muscle stretch. The introduced dynamic stiffness is active for muscle stretch velocities, which are large with respect to the fastest voluntary muscle contraction velocities, see Fig. 1. This stiffness was found roughly equal to the nonlinear elastic stiffness of the muscle tendon material, spread over the length of the muscle. The dynamic stiffness is thought to result from the instantaneous locking of the cross connected bridges between the myosine and actine fibers of each muscular sarcomere. The dashpot in the Maxwell IE element, Fig. 1(a), disables the instantaneous dynamic muscle sarcomere stiffness for slow muscle stretch velocities. Figure 2 shows the application of the Hill muscle model to the biceps muscle for quasi-static muscle actions.

2.2 Basic simulation methodology

Once the 'muscled' skeleton model has been established as shown in Fig. 3 for the upper arm, positioned in the required static posture and loaded, assumptions are made on the roles of the muscles, which carry the load. For this purpose, likely "agonists" and "antagonists" ("prime movers"), and "synergizers" and "stabilizers" ("assistants"), are identified among the muscles, which participate in the investigated posture. The identified agonists are the main load carrying muscles, while the antagonists, if activated, directly counteract these muscles. The synergizers and the stabilizers play a secondary role. They hardly contribute to the principal load bearing task, but they assure overall "lateral" stability and may assist the principal agonists under applied heavy loads.



(a)



(b)

Fig. 2 Hill model output for the biceps muscle

The muscle forces are determined by the degree of voluntary muscle activation (0-100%) and are proportional to the physiological cross sectional area of the considered muscle segment. A human subject can carry a given load in a given posture under more or less extra overall voluntary muscle contraction (0-100%). This can best be illustrated by the fact that a subject can willingly tense its muscles without carrying any load at all. In the latter case, the agonist and antagonist muscles

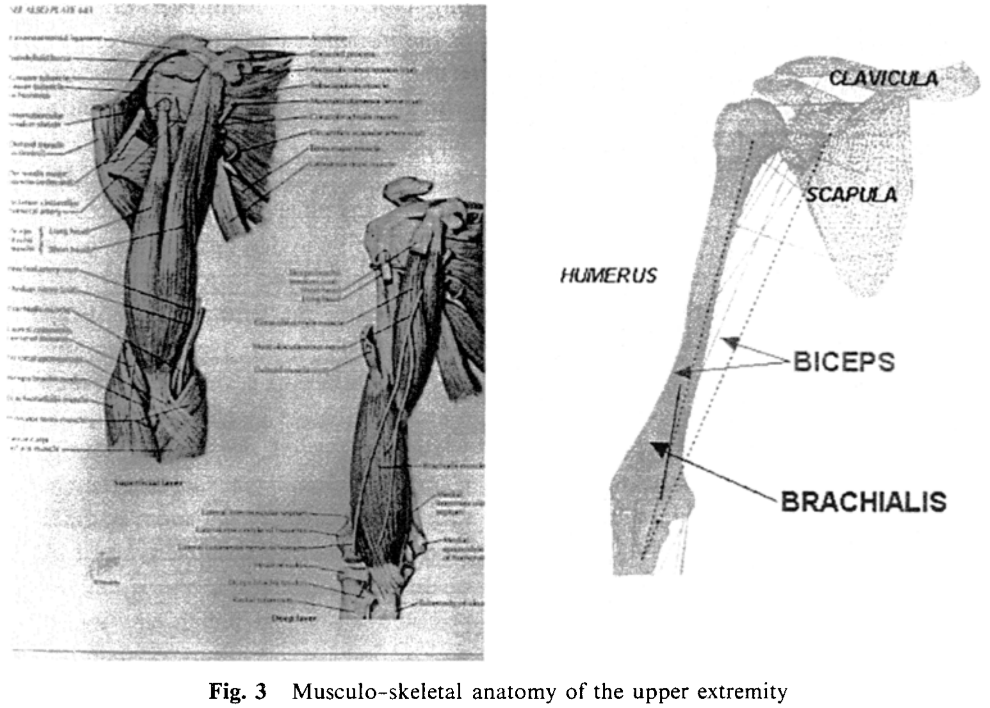


Fig. 3 Musculo-skeletal anatomy of the upper extremity

exactly balance their action on the skeleton, because otherwise the static condition of zero acceleration will not be maintained. The extra, load-independent, voluntary activation level of the muscles can therefore be considered to represent the subject's level of voluntary muscle stiffening in cases of applied static loads. This stiffening beyond the level needed to perform a given task can be beneficial to counteract sudden changes in external loads.

2.3 Over-determined system

Since the number of kinematic degrees of freedom of the skeleton (less than 100) is far less than the number of muscle segments (more than 1000), that can be activated to maintain a given static posture in equilibrium under a given static loading, the forces acting in each contributing muscle segment cannot be calculated from the mechanical conditions of equilibrium alone. For this reason it is necessary to solve an over-determined system of equations by minimizing relevant objective functions that express the optimal involvement of each muscle segment that contributes to maintain the required posture under the applied external loads.

2.4 Optimization procedure

For each instantaneous quasi-static posture, the *objective function*, assumed here to determine the likely distribution of the active muscle forces, has the form

$$F = (\sum (F_{max}(L_i, S_i)(\alpha_i - c))^2)^{1/2}, \quad \text{Eq. (1)}$$

where the sum ranges over all participating muscle segments, i, limit curve $Fmax(L_i, S_i)$ is the maximum active Hill muscle force curve of muscle segment i at 100% muscle activation with the maximal muscle activation ratio, $\alpha_i = (F_{muscle}/F_{max})_i = 1.0$, see Fig. 1(b), L_i is the instantaneous length of muscle segment i, S_i is the (constant) physiological cross section area of muscle segment i, and c is the given (average) voluntary activation ratio of muscle contraction before any load is applied $(0 \leq c < 1)$.

This function can be thought to express the useful *overall level of active internal physiological muscle energy per unit of time*, expended for a given task. Its minimum over the unknown muscle activation ratios, αi, is assumed here to correspond to the minimum of the sum of all muscle forces, minus forces due to voluntary contraction, c. While extra voluntary contraction also spends physiological muscle energy, this contribution is not thought useful for performing the given task of carrying external loads and is therefore discarded from the overall useful energy balance. This ad hoc assumption must undoubtedly be refined in future versions of the software.

The assumption that internal physiological muscle energy, or work, is proportional to the product of muscle force and the time interval over which the force acts, appears justified by the fact that a muscle spends the more internal energy the longer it sustains a force of constant magnitude. For a given time interval, this implies in a static posture that the least amount of muscle energy is spent when the sum of all activated muscle forces is minimal, because time becomes a global multiplier. In a dynamic problem with imposed skeletal motions, the activated muscle forces that enable this motion are variable over time. The instantaneous criterion is still valid, the total physiological "cost", however, will be the integral of the instantaneous muscle forces over the time of the motion.

In the case when static muscle forces must be supplied over longer periods of time, the physiological phenomenon of fatigue will gradually decrease the efficiency of the concerned muscles, so that their physiological activation levels must rise for maintaining an output of constant force. In any case, the instantaneous distribution of active muscle forces over a given time interval will be obtained by searching the minimum of the objective function given by Eq. (1).

The *constraints* for the static optimization process are given by the fact that the *skeletal joint accelerations, or moment sums*, of the links of the kinematic chain, constituted by the involved parts of the skeleton, must all be equal to *zero* in a position of quasi-static equilibrium. These skeletal joint accelerations, or moments, are due to the action of the internal muscle forces and of the externally applied loads and contact forces. For imposed dynamic motions of the skeleton, the d'Alembert forces must be added to the external loads.

The condition of zero joint moments from the muscles and the external loadings in the directions of the skeletal joint rotation degrees of freedom can be obtained by writing down the system of joint moment equations of equilibrium. The coefficients of the matrix of the equations can be calculated by applying internal unit forces to the model, muscle by muscle, and by extracting the corresponding influence vectors for joint moments for each skeletal articulation. This operation can conveniently be performed with the finite element model of the muscled skeleton. This system is therefore solved within the process of minimization of the objective function.

The *design parameters* of the optimization problem at hand are given by the *activation levels*, α_i, of the participating muscle segments, while the average voluntary contraction levels, c, are known, user specified values.

The *bounds* on the design parameters are given by $0 \leq \alpha_i \leq 1.0$, i. e., the activation level of a muscle cannot be less than zero and not greater than 100%. The outlined optimization procedure is applied to a simple one degree of freedom system.

3. One Degree of Freedom System

3.1 Test setup

Figure 4 shows an elementary one-degree of freedom model and test setup of the upper and lower arm. The single kinematic degree of freedom consists in the rotation of the lower arm about the elbow joint with all other skeletal displacements and rotations fixed. The upper arm, the shoulder and the local wrist joints are considered fixed. From the 22 muscles of the upper and lower arm with a total of 28 segments, only the 2 segments of the biceps muscle, plus the brachialis and the brachioradialis (supinator longus) muscles (4 segments) were retained as the agonists and the 3 segments of the triceps as the antagonist muscles. This very reduced set constitutes a total of 7 muscle segments for one kinematic degree of freedom, i. e., the system to determine the muscle segment forces from equilibrium is over-determined by a factor of 6.

The voluntary test subjects, Fig. 5, were asked to pose their right elbow on a padded support, to carry a load, P1, in the right hand and to volun-

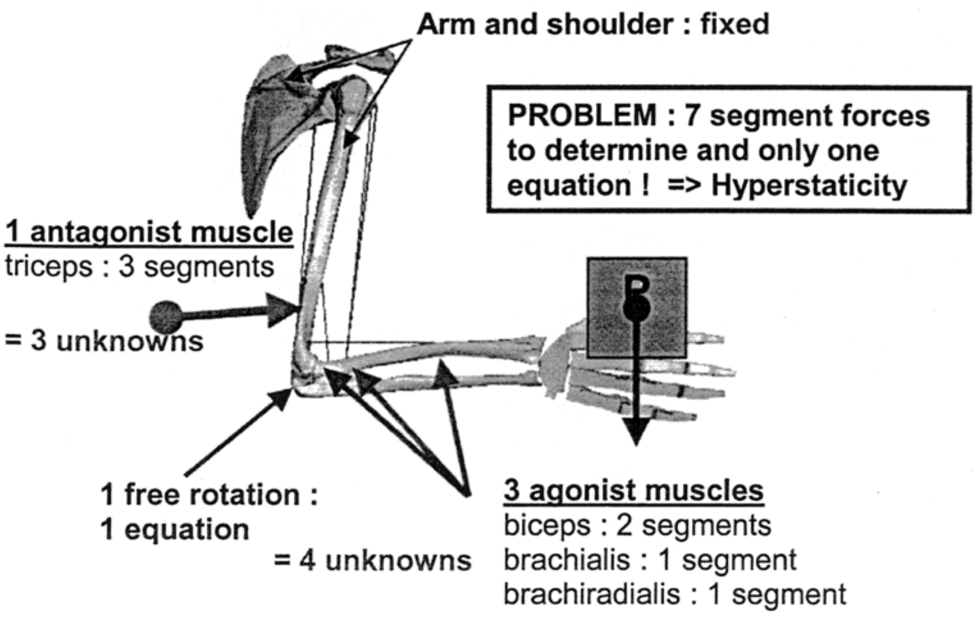


Fig. 4 Static load problem for elbow flexion

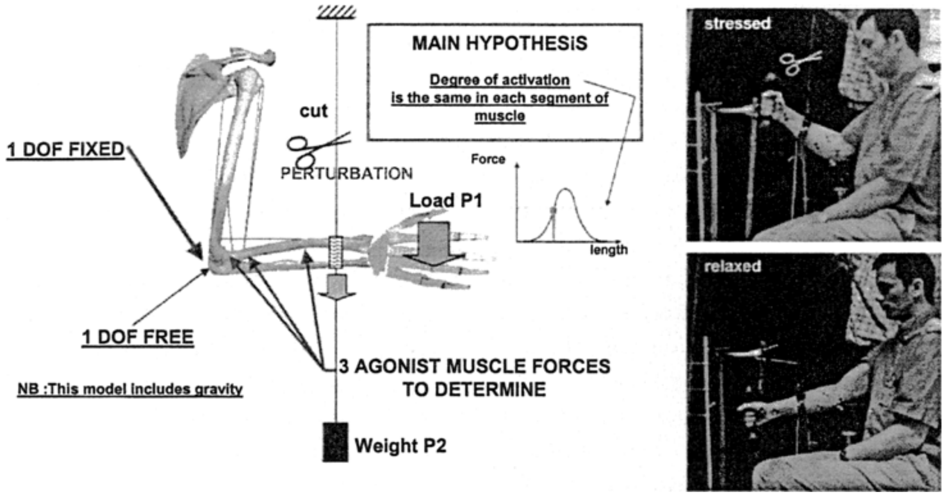


Fig. 5 Validation of model with perturbation technique

tarily pretension the arm muscles to activation levels of zero (relaxed), about 50% and 100% (stressed). At that moment a second load, P2, suspended from the ceiling, was liberated by cutting its suspension string, whereupon the load P2 suddenly came into action at about the center of the lower arm. The subject's involuntary reactions due to this suddenly applied load were video recorded. The reactions ranged from small angular responses (jolts) of the forearm for high voluntary muscle contraction (stressed) to full, uncontrollable (unstable), extension of the forearm about the elbow joint for low or zero voluntary muscle contraction (relaxed). Note that the subjects did not see when load P2 was liberated. In the short time interval during which load P2 was applied, the subjects had no time to modify their voluntary muscle activation levels.

The purpose of this test was to determine if the outlined procedure to "optimize" the muscle segment contributions in given static positions of equilibrium under applied skeleton loads can lead to plausible predictions of the forces, or activation levels, α_i, of the muscles, when the muscles undergo an average pre-tension of $c=0\%$, 50% or 100% of their maximum activation. Since direct measurements of muscle forces were not possible (no electro-myographic apparatus was available), the activation levels could only be deduced indirectly by measuring the angular perturbations of the forearm about the elbow joint under the suddenly applied loads, P2. It was assumed that, if for each applied load P1 and each level of muscular pre-stress, c, the simulation finds the same angular perturbations than were found in the tests, then the muscle force predictions can be considered accurate.

Fig. 6 Military combatant with gear

3.2 Test and simulation results

The preliminary results have shown that the test subjects' qualitative responses to the suddenly applied extra loads could be predicted correctly, ranging from small extension angles to uncontrollable extension of the forearm. Since under the applied activation levels the simulations exhibited the same angular motions of the forearm under the suddenly applied load of P2=4 kg force, it was concluded that the outlined procedure to determine the activation levels of the over-determined skeleton muscle system was realistic.

4. The Felin Project

The outlined preliminary procedure has been applied in the Felin project of the French military to evaluate the performance of the musculo-skeletal system of humans in given postures under given static loads. Such problems arise when a mechanic is asked to hold in place a piece of heavy equipment in a hard to get at place (design problem), or when a military combatant is manipulating heavy equipment, when loaded by unwieldy objects and gear, Fig. 6. Based on the

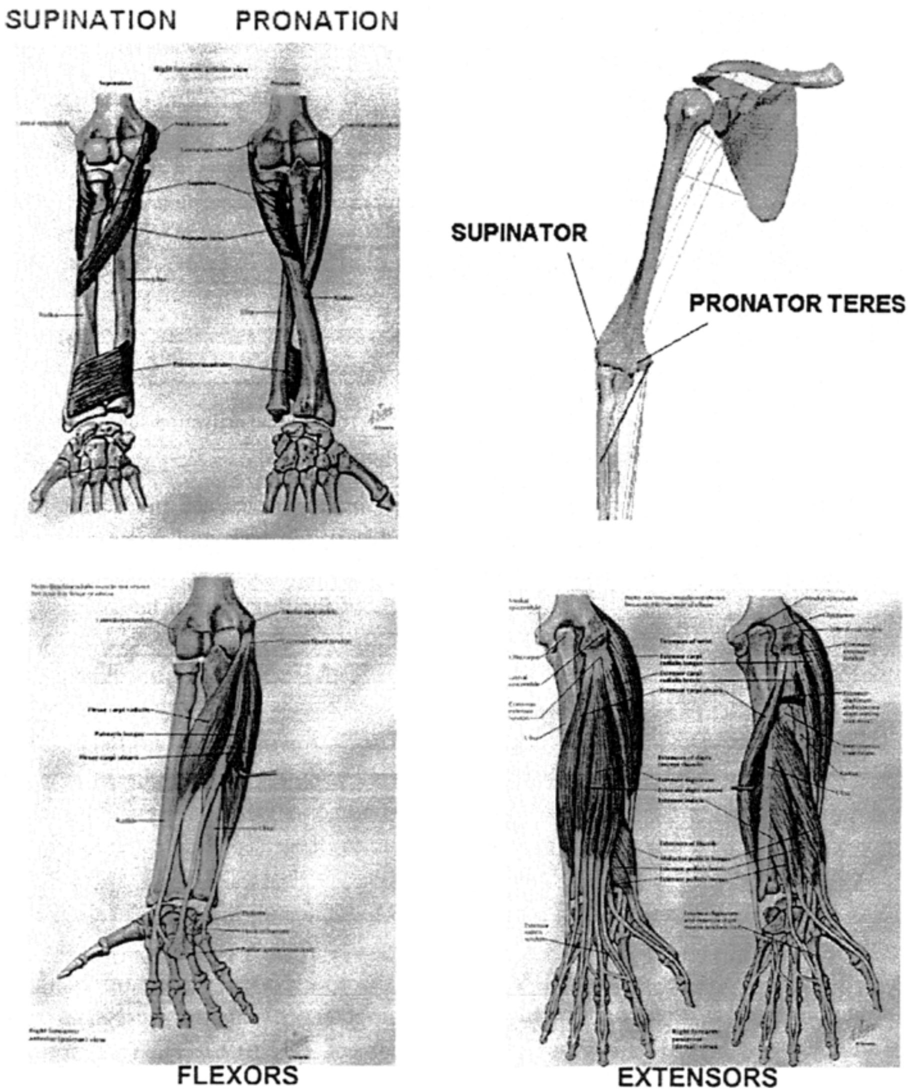
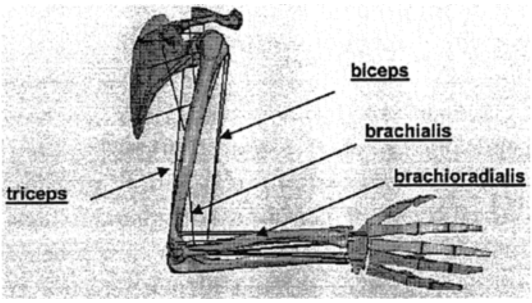
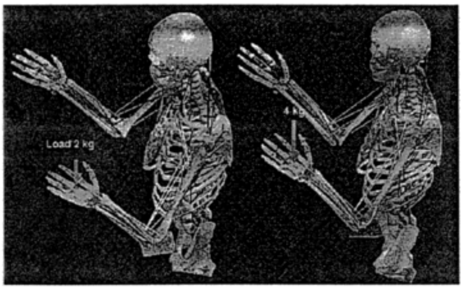


Fig. 7 Muscle anatomy of the fore arm

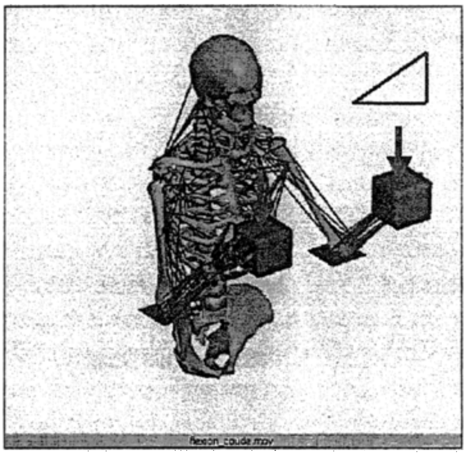


(a) Static position at about 90° elbow flexion

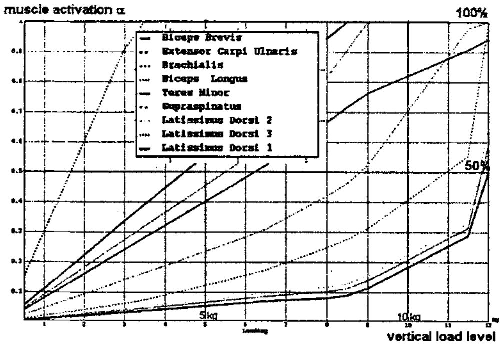


(b) Carrying loads of 2 and 4 kg

Fig. 8 Fully muscled upper skeleton (shoulder and arms)



(a) Linearly increasing load



(b) Muscle activation levels over the load

Fig. 9 Flexion of the elbow (results)

outlined procedure, criteria of "comfort" and "feasibility" may be deduced from the resulting necessary activation levels of the involved muscles.

4.1 Elbow flexion

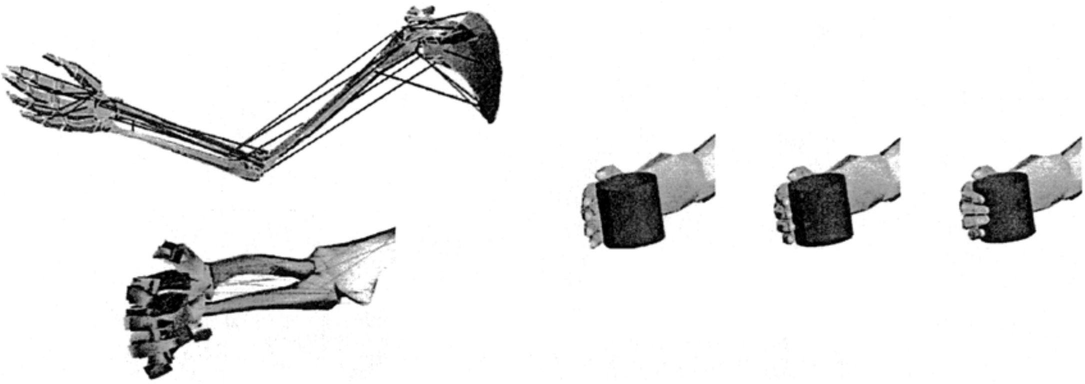
Figure 7 shows the anatomy of the lower arm, the muscles of which were added to the H-ARB™ model. Figure 8 shows the fully muscled arms and Fig. 9 gives results concerning the increase in muscle activation levels as a function of the carried loads. The curves of Fig. 9(b) plot the activation levels of the major muscles as a function of the applied load magnitudes. Some of the muscles (biceps) can be seen to saturate at 100% of their activation level at low loads, whereas some other muscles, notably the latissumus dorsi segments, are seen to activate themselves only at the highest loads. This can be ascribed to the fact that these muscles are inefficient in the present load carrying task and come to the rescue only when almost all other muscles have saturated at 100% of their activation levels.

5. Miscellaneous Applications

The following pilot applications were investigated with the emerging PAM-Comfort™ prototype numerical simulation tool in order to demonstrate its large scope of application.

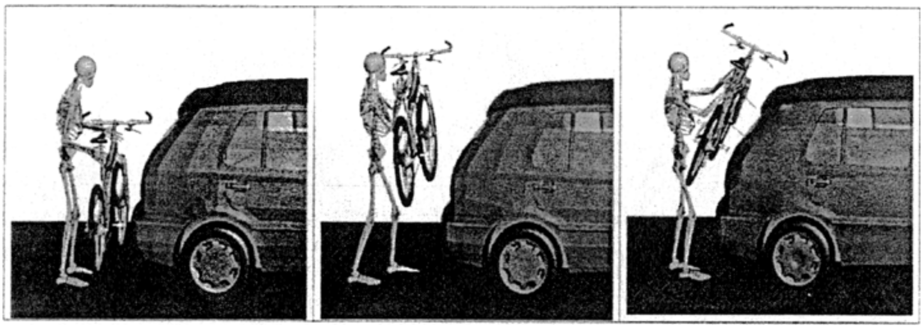
5.1 Gripping hand

Figure 10 shows the musculo-skeletal anatomy of the lower arm, wrist and hand, which was equipped with as many muscles as needed to grip an object, Fig. 10(a). The fine motion muscles of the fingers and thumb were not modelled here, because they do not play any role of importance

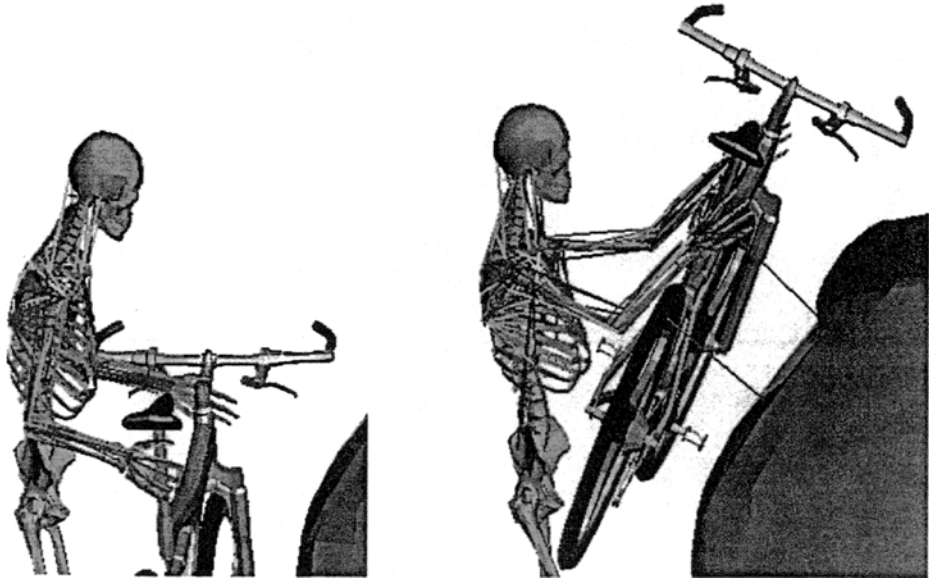


(a) Muscled skeleton (b) Gripping a cylindrical object

Fig. 10 Gripping hand model

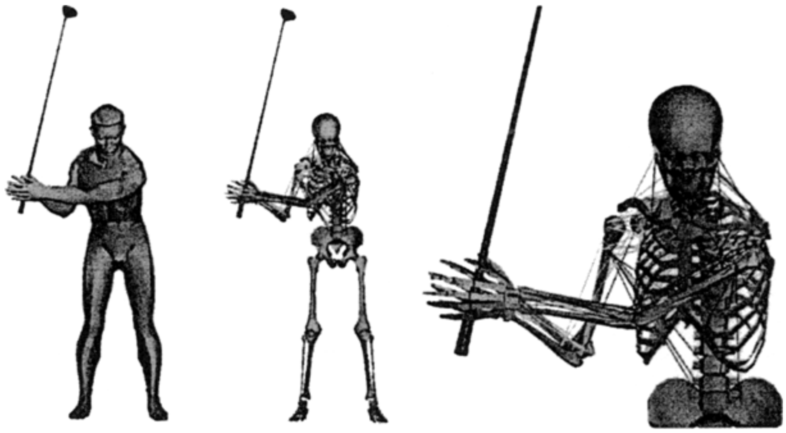


(a) Overall picture

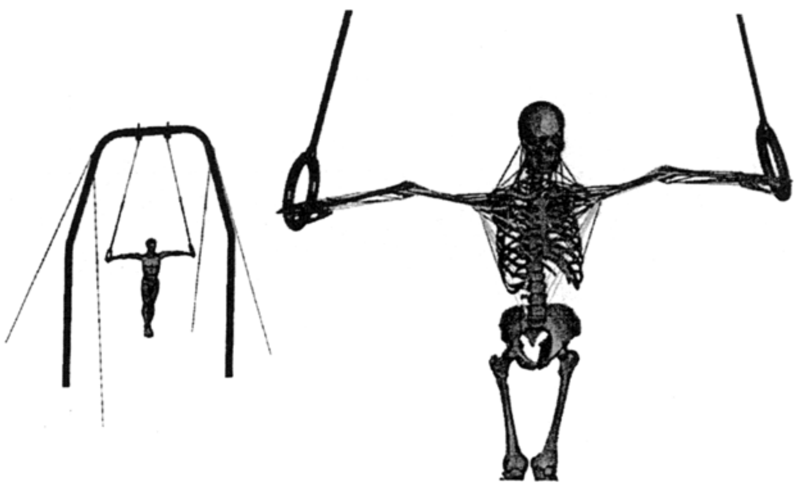


(b) Intermediate positions with static muscle forces (colors)

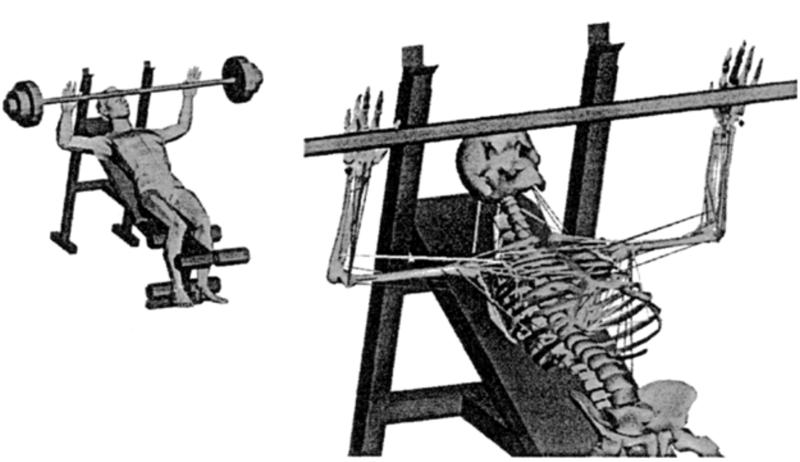
Fig. 11 Robby stowing a bike



(a) Exercise number 1: working the swing arm



(b) Exercise number 2: working the latissimus dorsi



(c) Exercise number 3: working the pectorals

Fig. 12 Robby the sports champion

in the gripping action, say, of a heavy tool or object. The numerical tool can evaluate the muscle activation levels needed to hold the object in a static position The program can also simulate the dynamic action of gripping the object under imposed final voluntary muscle force levels of the contributing set of muscles, as expressed in the animation snapshots of Fig. 10(b). This example demonstrates the great scope of situations that can be simulated and analyzed by the prototype compute models and software.

5.2 Stowing of a bike

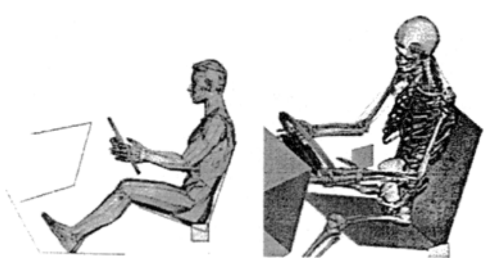
Figure 11 illustrates the action of stowing a bike on the rear bike transportation rack of a passenger car. The animation snapshots, Fig. 11 (a), show the imposed trajectory. The muscle forces involved in holding the bike in two intermediate positions is shown in Fig. 11(b). This example demonstrates how the numerical tool can be used to evaluate the feasibility and the comfort for the human body to carry out potentially difficult physical tasks.

5.3 Robby the champion

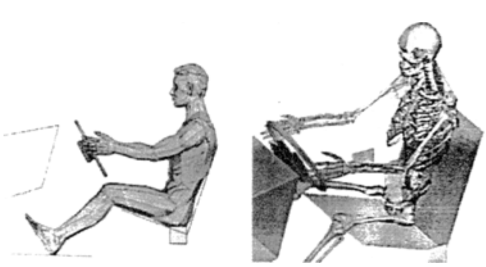
Figures 12(a) to (c) surprise the Robby H-ARB™ model doing various sports exercises. These pictures demonstrate that for each of the exercises the distributions of the muscle forces correspond well to the expected results, i. e., that the right muscles come into action. It is therefore clear that the methodology works in the sense that when the skeleton is equipped with the contributing muscles, then the methodology will find the good solution tendency. Not all muscle groups are presently implemented, however.

5.4 Different driver positions

Figure 13 shows the driver of a passenger car in two different positions. The forces applied by the hands on the steering wheel are (a) 15 N and (b) 21 N, respectively. The calculated arm and neck muscle forces are different in both positions and they can be correlated with muscle fatigue for long term action.

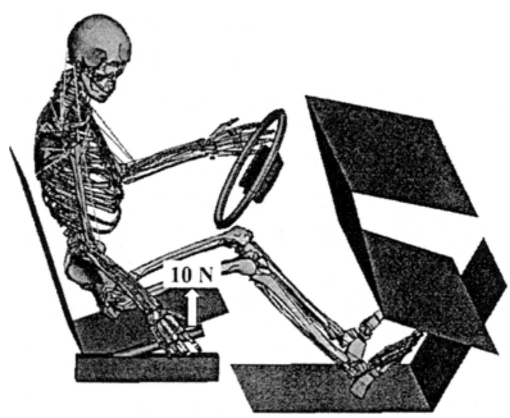


(a) Position 1



(b) Position 2

Fig. 13 Different driver positions



(a) High position

(b) Low position

Fig. 14 Driver activating the hand brakes

5.5 Manipulating the hand brakes

Figure 14, finally, shows the driver of a passenger car activating the lever of the hand brakes with a vertical force of 10 N, using the arm and shoulder muscles (the muscles of the back are not yet implemented in the model). For this short term action the relative muscle forces are much greater, and the question can be whether or not the strength of an individual driver is sufficient to set and unset the hand brakes at activation forces higher than 10 N. In the higher brake lever position, Fig. 14(a), the driver must lean backward, which leads to less favorable lever arms for the muscles of the shoulder. The muscles are therefore activated to a somewhat higher level. In the lower brake lever position, Fig. 14(b), the driver must lean forward, which seems to somewhat release the shoulder muscles. In this forward position the muscles of the back (not modeled) will surely have a more important role. The arm muscle activation levels are comparable in both cases, and coherent (main players: biceps, brachialis). In both cases, the left arm and shoulder muscles contribute to the overall stability by small but non-zero activations when the driver holds on to the steering wheel.

6. Conclusions

This document presents a short overview on the emerging ESI Group comfort and ergonomics models of the human body, that are developed to study the activation levels of the skeletal muscles, needed to sustain various load conditions. The shown examples indicate the wide spectrum of potential fields of application. The numerical methodology used to calculate the skeletal muscle forces proofs to be remarkably efficient and leads in all studied cases to remarkably intuitive results. More validation studies must be performed, including electro-myographic measurements on volunteers. The models of the muscled skeleton must be completed for the still missing muscles, and scaling and morphing technologies must be used to produce models of different sizes. Extensions to dynamic forces and moving subjects are possible. The models are part of an emerging library of compute models in computational biomechanics ("BioLib": H-Model™, Robby™, etc.), which contains models of the human body that are conceived and validated mainly for studies of occupant safety of transport vehicles, comfort, ergonomics and biomedical applications. All models benefit from the synergy created from their different fields of application.

Acknowledgements

The Felin project was carried out by ESI SA with the French military organization DGA. Jean -Pierre Verriest, INRETS-LBMC, acted as a consultant on ergonomic matters. The material in this document was collected by Eberhard Haug and the ongoing PAM-Comfort™ code developments are carried out by Alain Trame on (atr @esi. fr), (both ESI Software, 20 rue Saarinen, Silic 303, 94588 Rungis Cedex, eh@esi.fr). The modeling of the muscled skeleton and most examples were performed by Jean-Christophe Allain (jca@esi.fr) (ESI SA, 20 rue Saarinen, Silic 270, 94578 Rungis Cedex). Muriel Beaugonin (Leader of the Biomechanics Safety Group, ESI Software, mb@esi.fr) and Hyung-Yun Choi (Associate Professor, Hongik University ME Dept., http://safety.hongik.ac.kr/, Seoul Korea) supervised the modeling work of the underlying H-ARB models.

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