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Simulation of Cyclic Forearm Rotations by Means of a Biomechanical Model

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Cognition and biomechanics seem to be linked by the fact that the system responsible for selecting movements must know about the body's biomechanical properties. The aim of the present study is to analyse effects in human hand-arm movements by means of a biomechanical model. Especially the so called middle-is-faster effect in cyclic rotations of the forearm is investigated. The biomechanical model consists of 11 rigid bodies, representing the skeletal of the hand-arm system and several force elements representing 38 selected arm muscles. By means of the model cyclic pronating and supinating movements are simulated in the medial, the middle, and the lateral motion range. The movements result by tuned stimulation patterns of the model muscles. It can be shown that in two examples the pronating movements are faster than the supinating movements and that the middle-is-faster effect exists. However, it is difficult to simulate repeated movements because the environmental demands alter with each motion cycle. In reality the fit of the movement outcome to the altered environmental demands is probably attained by varying the parameters in a closed-loop mechanism.

Keywords: *biomechanics, hand-arm system, model, motor control, simulation*

INTRODUCTION

Whereas in former times the movements of the human legs or the trunk were the object of a great number of biomechanical studies, the movements of the hand-arm system were seldom analysed. Concomitant with the progress in the development of biomechanical methods more investigations about the hand-arm movements are carried out. So Pennestri *et al.* (2007) developed a musculo-skeletal model of the human upper limb, consisting of four rigid bodies and 24 muscle elements. The aim of their study was to assess the muscular activity and the joint reactions during driving and to test virtual changes in the cockpit design. Another actual investigation was carried out by Dennerlein *et al.* (2007). They assessed the joint torques for the metacarpal-phalangeal, wrist, elbow, and shoulder joints during single-finger tapping. In both investigations the movements of the hand-arm system were analysed by means of inverse dynamics methods.

Due to the complex structure of the hand-arm system and its large motion range in the prevailing number of studies, the methods of the forwards dynamics were applied to simulate only small movements of the systems in selected directions. This means that, for example, the movements of the hand-arm system were simulated under vibration stress (Dong *et al.* 2007, Fritz 1991, Rakheja *et al.* 2002 or Reynolds and Falkenberg 1982). The models in these studies consisted of several discrete masses connected by linear springs and dampers. A more realistic model was developed by Lemay and Crago (1996). The aim of their study was to simulate the functional neuromuscular stimulation procedures for the control of upper limb movement in tetraplegics.

In experimental psychology the human hand-arm system is normally treated as a black box and its responses to different experimental conditions are investigated. So Rosenbaum *et al.* (1990) studied the simple task of reaching for a bar and placing it precisely on a target. To achieve the task with the right hand, the subjects had to rotate the bar about 90° to the left or to the right side. In dependence upon the direction of the rotation the subjects picked up the bar with an overhand grip or with an underhand grip. The subjects seemed to select that kind of grip which afforded a comfortable final posture of the arm rather than a comfortable initial posture. Rosenbaum *et al.* (1990) called this the 'end-state comfort effect'. They assumed that the source of the end-state comfort effect was the subjects' expectation that back-and-forth movements would be quicker, and hence more efficient for positioning, in the middle of the range of motion than at either extreme (Rosenbaum *et al.* 1996).

In order to pinpoint the source of the 'end-state comfort effect' Rosenbaum *et al.* (1996) performed further experiments concerning object-manipulation. By their experiments Rosenbaum *et al.* (1996) showed that subjects could in fact oscillate the forearm more quickly in the middle of the range of motion than near the extremes of the range. They called this the 'middle-is-faster effect' and asked why, from a biomechanical perspective, this effect exists.

Although cognition and biomechanics pertain to motor control, they have rarely been considered together.

Cognition has usually been concerned with the decision making. On the other hand, biomechanics has usually been concerned with musculoskeletal motion and stability, but with little regard for the way, patterns of performance are selected. The link between both disciplines is given by the fact that the system, responsible for selecting movements, must 'know about' the body's biomechanical properties. Regarding this Bernstein (1967) argued that as actors become more proficient in a task, they exploit biomechanical properties of the body in its interaction with the external environment. According to the hierarchical neural network model of Kawato *et al.* (1987) the biomechanical properties are exploited in the internal dynamic model which is obviously acquired by the *spinocerebellum – magnocellular red nucleus system*. By means of this model a possible, error of the movement can be predicted and the motor commands can be updated. It is assumed that this feedforward control is faster than that of the long-loop sensory feedback.

In the present study a biomechanical model of the human hand-arm system is described which enables the simulation of flexion and extension of the system, of abduction and adduction and of rotations around the longitudinal axis of the forearm. Especially the realisation of these rotations is difficult because some arm muscle, for example the biceps, wrap around the radius during pronation and supination. The simulated movements are used to answer from a biomechanical perspective the question expressed by Rosenbaum *et al.* (1996): 'why can movements be made more quickly in the middle of the range of motion than in the extremes' and to check their statement that the pronating movement can be performed faster than the supinating movement.

METHOD

Structure of the Biomechanical Model

The biomechanical model imitates the right upper extremity, when the upper arm, the forearm, the hand, and the fingers are held in the middle of the motion range (Fig. 1). The model consists of 11 rigid bodies, articulated by joints. The first rigid body represents the right part of the upper trunk, including the *clavicle* and the shoulder blade. This body is fixed in the coordinate system. The body, representing the upper arm, is connected by a ball-and-socket joint to the first body. Restricted by this construction, the upper arm can only be abducted up to the horizontal posture.

The forearm is represented by two rigid bodies, namely one body for the *radius* and the surrounding muscles and the other body for the *ulna* and the muscles (Fig. 1). The model body '*ulna*' is connected by a pin joint to the upper arm (*elbow joint*), whereas the body '*radius*' is attached by a universal joint (*radiohumeral joint*). At the distal end the body '*radius*' is formed like an 'L'. The short side of this 'L' is connected with the '*ulna*' by a cylindrical joint whose axis of rotation is parallel to the longitudinal *ulna* axis. This configuration assures that the body '*radius*' slides and rotates on the fixed '*ulna*' similar to the forearm.

The right hand and the fingers are represented by seven rigid bodies (Fig. 1), namely

- *Carpal bones* and the *metacarpal bones* of finger II to V
- *Proximal bones* of finger II to V
- *Middle bones* of finger II to V
- *Distal bones* of finger II to V
- *Metacarpal bone* of the *thumb*
- *Proximal bone* of the *thumb*
- *Distal bone* of the *thumb*

The *wrist joint* is modelled as a universal joint, which is attached solely to the body '*radius*'. The two perpendicular axes of rotation enable extension, flexion, abduction, and adduction of the hand. The joints between the finger bones are technically expressed simple pin joints which enable extension and flexion of the fingers II to V (*index finger* to *little finger*). The joint between the *lateral carpal bone* and the *metacarpal bone* of the *thumb* is modelled as a universal joint whose axes of rotation are parallel to the axes of the *wrist joint*. Finally, the two joints between the *metacarpa*l, the *proximal* and the *distal bone* of the *thumb* are pin joints whose axes of rotation are perpendicular to the axes of the finger joints. By this construction it is possible to move the *thumb* independent of the movement of the other four fingers.

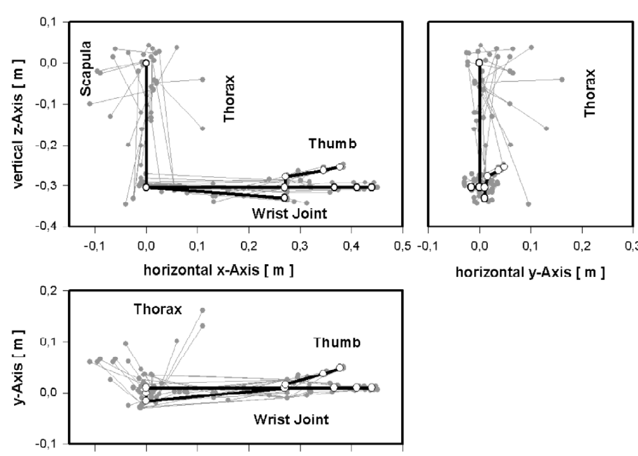

Figure 1: Schematic of the Model Skeleton (Black Bold Lines) and Muscles (Grey Thin Lines) of the Hand-arm System in the sagittal view (xz- plane) frontal view (yz-plane) and transversal view (xy-plane)

The motion range of the joints in the human body is restricted in both rotation directions, for example in extension and flexion. In order to reduce the motion ranges of the rigid bodies, so-called internal torques are implemented in the joints. The torques depend upon the joint angle and rotation velocity according to the equation, presented by Winters and Stark (1985) and Lemay and Crago (1996). By the equation it is attained that the torques are nearly zero in the common motion range of the joints and increase exponentially with joint angles near the extremes of the motion ranges. This means that very high muscles forces are used to move the arm segments in the border ranges of motion. However, Lemay and Crago (1996) computed the torques only for the elbow, forearm, and wrist joint. Therefore new parameters must be assessed for the torques in the other joints of the model. These parameters are derived from the parameters, given by Lemay and Crago (1996), and are fitted to the different motion ranges in the remaining joints.

Muscles

38 muscles were selected to be included in the model of the hand-arm system. The muscles are imitated by so called force elements. In contrast to the muscles the force elements have only a straight line of action between the attachment points. To attain a realistic imitation of the muscle lines of action, some muscles are modelled by one to five force elements forming an open chain. For example, one force element represents the *brachialis* whereas five elements represent the *extensor digitorum*. This means that the first force element models the muscle belly of the *extensor digitorum* and the four other elements model the long tendon of this muscle so that the muscle force is transmitted from the muscle belly to the mobile bone, namely in this case the finger tips. The force elements, representing the muscle-tendon unit, are linked and supported by small bodies without mass. These bodies can shift parallel to the longitudinal axes of the contiguous bodies of the hand-arm system and can rotate around the transverse axes. By this it is attained that the force elements stay nearly parallel to the bodies of the hand-arm system during the movements of the system.

Similar constructions are used to imitate the functions of the *pronator teres* and the *supinator* which insert at the lateral side of the *radius* and for the tendon of the *biceps brachii* which is fixed at the *radial tuberosity*. During supination or pronation these muscles and the tendon wrap around the *radius*. Supporting bodies must be included in the model in order to attain nearly realistic moment arms of these muscle during the rotation of the *radius* around the *ulna*.

Relatively to the adjacent joints, the lever arms of the model muscles are not constant. The corresponding force elements have a straight line of action between the origin and the insertion. Hereby the lever arms of the muscles decrease or increase with the movements in the joints. By the supporting bodies it is avoided that the lever arms can be smaller than zero (e.g. *pronator teres* and *supinator*) or increase to great values as it would be given by a straight line between the origin at the upper arm and the insertion at the finger tip (e.g. *flexors* and *extensors* of the hand and fingers).

Muscles exert forces during active contraction and passive stretching. This mechanical behaviour has to be imitated by the force elements, representing the muscle bellies. The force-strain properties of the force elements correspond to the three-element Hill model consisting of an contractile element, a parallel elastic element and a series elastic element. The relation between the force f_{PE} and the mechanical strain e of the parallel elastic element is simulated by the linear-hyperbolic equation of Luo and Goldsmith (1991)

$$f_{PE} = \frac{k\ \varepsilon}{1-\varepsilon/0.7} A_{muscle} \quad \text{for } 0 < \varepsilon \le 0.65$$

In this equation the value of spring constant k equals 33400 N / m^2 and A_{muscle} is the cross-sectional area of the corresponding muscle.

Corresponding to the study of Hatze (1981) the force output of the contractile elements can be described by the equation

$$f_{CE} = q(\xi, v)\ k(\xi)\ g(\xi) - f_0(\xi)$$

which is a combination of the following functions

–$q(\xi, v)$ as the active-state function

–$k(\xi)$ as the length-tension relation

–$g(\xi)$ as the velocity-dependence function

and $f_0(\xi)$ as the internal resistance tending to extend the muscle fibres during contraction. The independent variables of the four functions are the normalized length of the muscle fibres $\xi (0.58 \le \xi \le 1.8)$ and the relative stimulation rate v(t) ($0 < v < 1$) describing the frequency of the stimulus which leads to the contraction of the muscle. To attain a controlled movement of the biomechanical model the time courses of the stimulation rates of the 39 model muscles must be given.

According to the arrangement of the three components in the Hill muscle model the force of the series elastic element must be the same as the force of the contractile element. If the interplay between the muscle forces and the movements of the model sufficiently simulates reality, the forces of the series elastic elements must coincide with the strain of these elements. Furthermore, regarding the force equilibrium of the muscle model, the force of the tendon is the sum of the force of the parallel elastic element and of the contractile element.

Model Parameters

The proportions between the masses, the inertias, and the centre of mass locations of the 11model bodies are derived as far as possible from Dempster (1955) and the values of these quantities are fitted to a human body having a height of 1.74 m and a body mass of 75 kg (Table 1). The mass of the forearm is divided between the *ulna* and *radius* such that the resulting centre of gravity is located at that point of the forearm which corresponds, according to the literature, to the centre of gravity of the forearm.

The data of the origin and insertion of the muscles at the different bones are derived from the corresponding coordinates listed by Seireg and Arvikar (1989). Seireg and Arvikar (1989) assessed the coordinates for their different models whose properties also based on the data of Dempster (1955). Concerning the deflection of the muscle tendons by the supporting bodies some modifications of the data have to be done. The cross sectional area of the different muscles are drawn from Schumacher and Wolff (1966) (Table 2). Finally, the resting length of the muscles are computed by the model itself. It is assumed that the posture shown in Fig. 1 simulates the neutral position of the joints of the hand-arm system. In this position the muscles are neither

Table 1
The Mass and the Inertia about the Axes through the Centre of Gravity of the 11 Model Bodies

Model Body	*Mass [kg]*	*Inertias [kg m^2]*		
		x-axis	*y-axis*	*z-axis*
Upper trunk	The body is fixed in the coordinate system			
Upper arm	2.31	0.0151	0.0151	0.0022
Radius	0.536	0.0003	0.00425	0.00425
Ulna	0.804	0.00035	0.00425	0.00425
Carpal and metacarpal bones of finger II to IV	0.237	0.000155	0.000329	0.000198
Proximal bones of finger II to IV	0.086	$0.55 * 10^{-4}$	$0.66 * 10^{-4}$	$0.17 * 10^{-4}$
Middle bones of finger II to IV	0.057	$0.36 * 10^{-4}$	$0.38 * 10^{-4}$	$0.6 * 10^{-5}$
Distal bones of finger II to IV	0.025	$0.16 * 10^{-4}$	$0.15 * 10^{-4}$	$0.2 * 10^{-5}$
Metacarpal bone of the thumb	0.021	$0.11 * 10^{-5}$	$0.35 * 10^{-4}$	$0.35 * 10^{-4}$
Proximal bone of the thumb	0.015	$0.8 * 10^{-6}$	$0.2 * 10^{-5}$	$0.2 * 10^{-5}$
Distal bone of the thumb	0.01	$0.4 * 10^{-6}$	$0.7 * 10^{-6}$	$0.7 * 10^{-6}$

Table 2
Origin and Cross Sectional Areas of the 38 Model Muscles (++ second origin at the forearm)

Origin at the shoulder / trunk		*Origin at the upper arm*		*Origin at the forearm*	
Muscle	*Area [m^2]*	*Muscle*	*Area [m^2]*	*Muscle*	*Area [m^2]*
deltoid, posterior	0.00109	triceps brachii, medial head	0.00055		
supraspinatus	0.00033	triceps brachii, lateral head	0.00055		
infraspinatus	0.00057	brachialis	0.00046		
subscapularis	0.00094	brachioradialis	0.00016		
teres minor	0.00016	supinator	0.00006	++	0.00011
teres major	0.00050	pronator teres	0.00008	++	0.00008
latissimus dorsi	0.00104	extensor digitorum	0.00024	extensor indicis	0.00004
coracobrachialis	0.00016	extensor carpi radialis longus	0.00019	extensor pollicis longus	0.00005
pectoralis major	0.00071	extensor carpi radialis brevis	0.00015	extensor pollicis brevis	0.00004
triceps brachii, long head	0.00055	extensor carpi ulnaris	0.00012	abductor pollicis longus	0.00008
biceps brachii, long head	0.00017	flexor digitorum sublimis	0.00020	++	0.00019
biceps brachii, short head	0.00017	flexor carpi radialis	0.00014		
		flexor carpi ulnaris	0.00005	++	0.00011
		palmaris longus	0.00004	flexor digitorum profundus	0.00050
		flexor pollicis longus	0.00010	pronator quadratus	0.00012
Origin at the carpal bones				opponens pollicis	0.00006

contracted nor stretched which correspond to the resting length. Thus the distances between the attachment points of the force elements are equated with the resting length of the muscles and their tendons.

Movements

The movements of the model are started from the neutral position. At first an alternative pronation-supination movement is simulated. It is tried to carry out this movement over more than one cycle. In the next simulation the movements are carried out in the revised turn. The third simulated movement begins with a maximal pronation and then a cyclic supination-pronation follows. The fourth movement begins with a maximal supination followed by a cyclic pronation-supination. Regarding Rosenbaum's studies the motion range of the cyclic movements should correspond to rotations around the longitudinal axis of the forearm of nearly 30°.

In order to simulate the four movements with the model adequate time courses of the relative stimulation rates must be assessed. At first those muscles are extracted which are dominantly connected with the selected movements in the literature. For these selected muscles stimulation time course are derived with cyclic increase and decrease of the relative stimulation rate. Starting with these a-priori sets the time courses are modified by trial and error till a sufficient conformity is attained between the simulated and realistic movements. For each of the four movements additional muscles have to be activated in order to fixate the upper arm or to reduce uncontrolled movements of the hand-arm system.

RESULTS

In Fig. 2 the time courses of rotations in the shoulder joint, the elbow joint, and the wrist joint are shown. The rotation about the longitudinal axis of the forearm corresponds to the supination-pronation movement and the increase of the angle corresponds to the supination. The supination-pronation induces small rotations on the transverse axes in the hand-arm joints as it is shown by the three other time courses. The simulated movement starts from the neutral position with a pronation over -15 ° which corresponds to a counter-clockwise rotation. Then the forearm is supinated over 30°. This movement takes 0.254 s. It follows the first complete pronation over -30° taking 0.248 s. The second supination takes 0.262 s. The following pronation could not be stopped at an angle of –15° and transferred into a further supination by the given activation pattern of the selected muscles. Thus it was not possible to simulate more than one complete supination-pronation cycles over the selected motion range without changing of the pattern or activating further muscles.

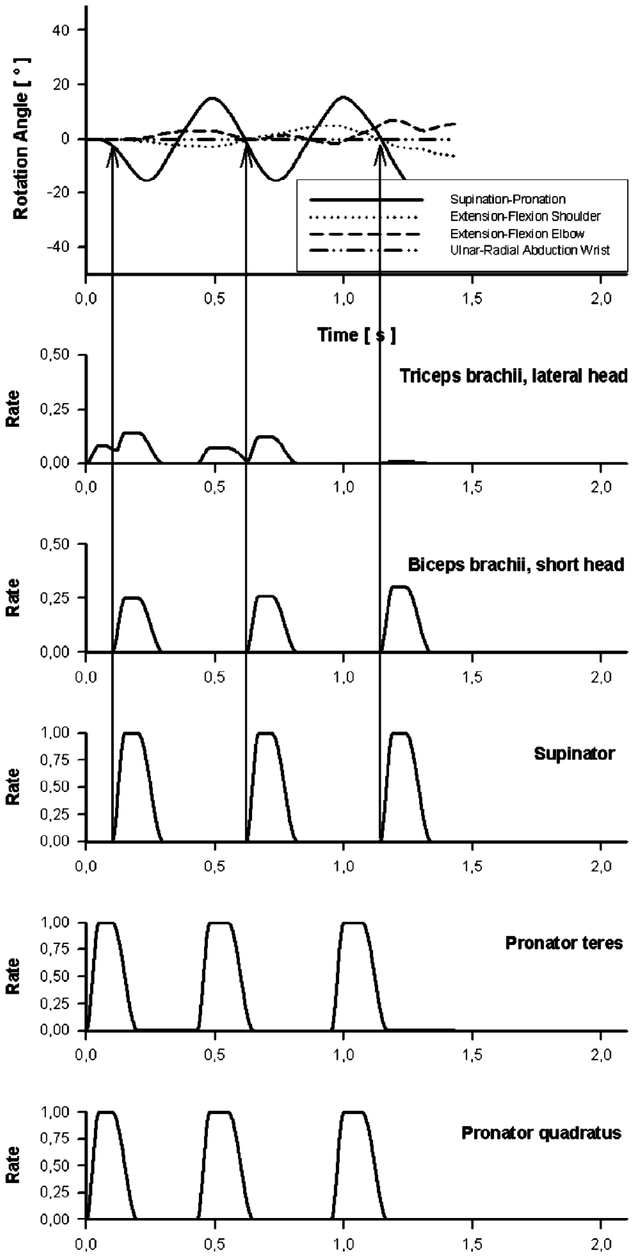

Figure 2: Cyclic Supinating Pronating Movement, Starting with a Pronation in the Central Motion Range. The Upper Plot Shows the Time Courses of the Supination-pronation, the Extension-flexion in the Shoulder and the Elbow Joint, and the Radial-ulnar Abduction in the Wrist Joint. The Five Lower Plots Show the Time Courses of the Relative Stimulation Rates of the Muscle which Produce the Movement

The five lower plots of Fig. 2 show the time courses of the relative stimulation rates of the muscle which produce the movement. The small activity of the *triceps brachii, lateral head,* reduces the flexion in the wrist joint which is induced by the supination-pronation. The activity of the *supinator* increases and decreases in turn with the activity of the *pronator* teres and the *pronator quadratus*. Performing the supination movement the *supinator* is supported by small activities of the *biceps brachii, short* *head*. Further muscles, especially the different parts of the *deltoideus*, have to be activated in order to reduce the rotations in the shoulder joint. Comparing the different time courses it can be seen that the activity of a muscle group starts before the forearm rotates in the corresponding direction. For example the stimulation rate of the *supinator* increases, before the supination of the forearm begins (see arrows in Fig. 2). In this phase the *supinator* acts eccentrically and decelerates the pronating movement.

Fig. 3 shows the time courses of a movement where the supination - pronation are perform in an inverse turn to the former movement. The movements start with a supination over 16°. Then a pronation over 31° follows and lasts 0.288 s. The following supination rotates the forearm over 30° within 0.205 s. During the second cycle the pronation rotates the forearm over 30° and takes 0.216 s. Again it was not possible to simulate the complete second rotation cycle with the concluding supination over 30°. According to the starting supination movement, the *supinator* is active before the *pronator teres* and the *pronator quadratus*.

The third simulated movement starts with a pronation over -50° (Fig. 4). It is followed by a supination over 30° taking 0.331 s and a further pronation of about -30° with a duration of 0.385 s. In contrast to the two former movements, it was possible to simulate a second supination-pronation cycle. The durations of the

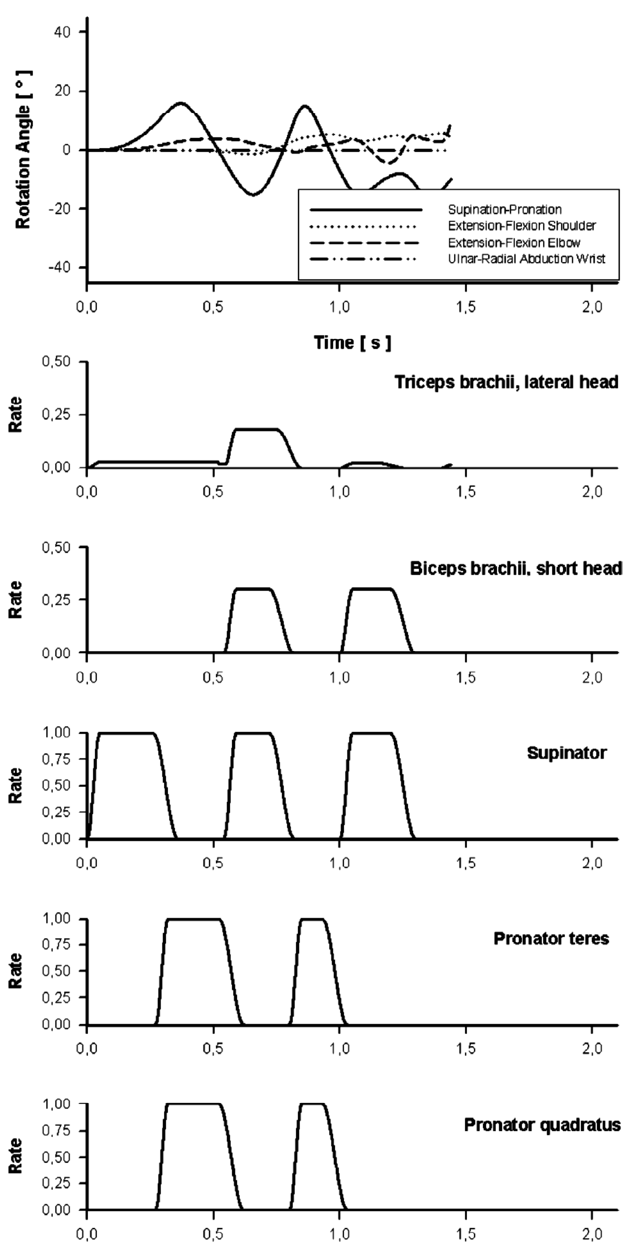

Figure 3: Cyclic Supinating Pronating Movement, Starting with a Supination in the Central Motion Range (Line Assignment as in Fig. 2)

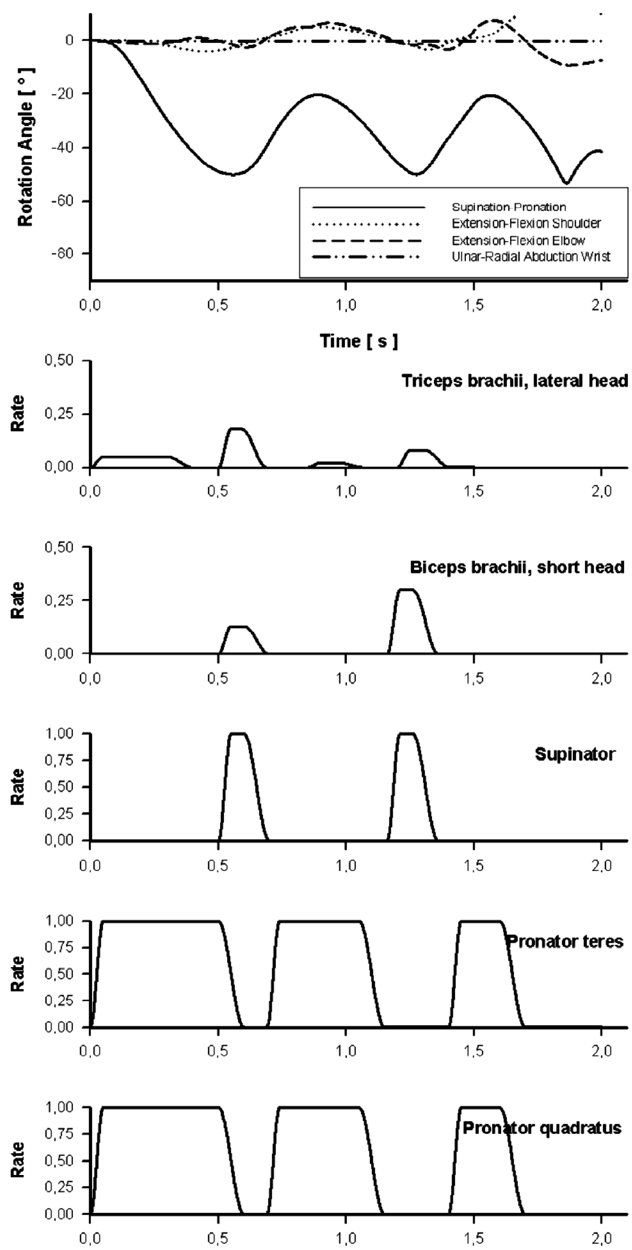

Figure 4: Cyclic Supinating Pronating Movement, Starting with a Pronation to the Medial Position of the Motion Range (Line Assignment as in Fig. 2)

corresponding motion parts are 0.287 s and 0.300 s and thus they differ distinctly from those of the first motion cycle. Compared with the first movement (Fig. 2) the greater pronation angle is attained by a longer activation of the *pronator teres* and the *pronator quadratus*.

The last movement starts with a supination over 74° (Fig. 5). It is followed by a pronation over -39°, duration 0.245 s, and a supination over 28°, duration 0.281 s. Likewise the cyclic movement, starting from the extreme medial posture, it was possible to simulate a complete second pronation-supination cycle. The corresponding rotation angles are –28° and +28° and the durations are 0.318 s and 0.303 s respectively.

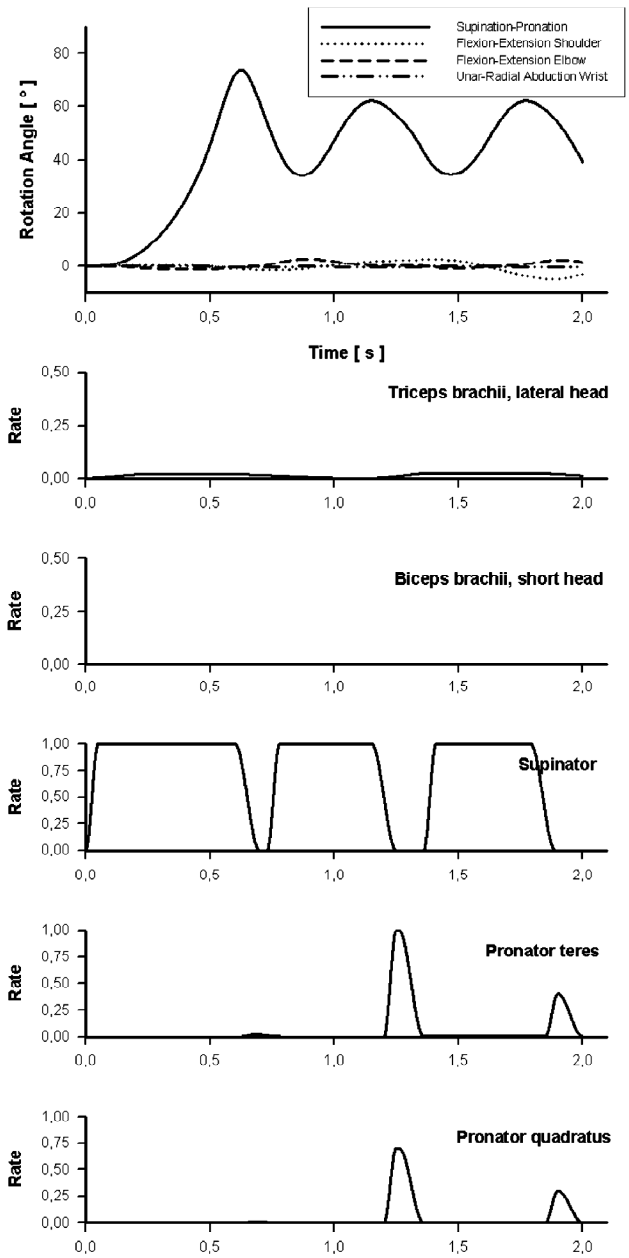

Figure 5: Cyclic Supinating Pronating Movement, Starting with a Supination to the Lateral Position of the Central Motion Range (Line Assignment as in Fig. 2)

DISCUSSION

Limitations of the Model

According to Nigg (1994) a biomechanical model is an attempt to present reality. This means that more or less differences exist between the properties and movements of the human body and those of the model.

The differences between the present model and the human hand-arm system are given by:

- the idealized geometry of the model bodies, which differs from the real shape of the hand-arm segments, the masses, and the moments of inertia.
- the idealized motion range and resistance in the model joints.
- the muscle model with its relation between the activation level and the exerted force.
- the simplified line of action of the imitated arm muscles.

The equations of motion of the model were derived by the methods of the dynamics of multibody systems. This implies that the imitated bodies, e.g. the segments of the hand-arm system, were rigid bodies. However, in reality the segments are not riged bodies. During movements they are deformed near the joints. Furthermore, the model bodies, representing the fingers, are a very rough imitation of reality.

In the model joint the axis of rotation are parallel or perpendicular to another and cross at the centre point of the joint. This is not the case in the joints of the hand-arm system. Additionally the axis can shift and change their orientation during movement. The motion range is limited by the shape of the articular surfaces, the joint capsule, and by the ligaments. In the model joints these motion resistances are imitated by internal torques which increase with increasing rotation angles.

The relationship between the relative stimulation rate and the contraction of the contractile elements of the muscles was calculated by the equations given by Hatze (1981). The contraction of the contractile elements results in a stretching of the series elastic elements which is proportional to the active muscle force. Thus the active shortening of the whole muscle fibres is the sum of the shortening of the contractile and the stretching of the series elastic elements. In the human body a permanent conformity is given between shortening of the muscles fibers and the shortening of centroid muscle lines spread between the origin and insertion points. In the model the body fixed attachment points, representing the origin and insertion of the muscles, approach or withdraw from another by the motion of the model bodies. The motions result from the simulated muscle forces. However, it must be assumed that during the motions the change of the distances between the attachment points not always coincide with the force related shortening of the muscles. In order to avoid this effect a permanent equalization between the simulated lengths of the muscles and the forces has to be carried out.

In the model the curved centroid line of the muscle-tendon system is approximated by one (e.g. *brachialis*) to four straight lines (e.g. *extensor digitorum*) which represent the directions of the force elements. The approximation probably results in differences between the lever arms of the arm muscles and the model muscles. The effect of these differences can be that the torques, produced by the simulated muscles, are lower or greater than in reality or have a divergent direction.

Comparison of the Movements

Rosenbaum et al. (1996) could demonstrate by their experiments two effects, namely those:

- the pronating movement can be performed (somewhat) faster than the supinating movement (Table 3).
- the oscillation of the forearm can perform faster in the middle position of the motion range than in end positions (Table 3).

The described oscillations of the forearm were simulated by stimulating the supinator, the pronator teres, the *pronator quadratus* and to a minor extent the *biceps brachii* (Fig. 2–5). The low activation of the *triceps brachii* was necessary in order to hold the forearm in a nearly horizontal position. However, Table 3 shows that the simulated durations of the pronation and supination are distinctly longer than the corresponding mean values of Rosenbaum et al. (1996). One explanation may be that even the maximal stimulation rate resulted in muscles forces which were smaller than the real muscles forces. This effect may result by the different parameters in the equation of Hatze (1981). Another reason can be that the simulated muscle forces are smaller than the real muscle forces in relation to the mechanical properties of the arm segments. The anthropometric and anatomical data based on different studies with partly small numbers of subjects or donators.

As it can be depicted from the figures or Table 3 that the durations of the pronating movements were only shorter in two from four simulated examples than those of the corresponding supination. Rosenbaum *et al.* (1996) could show by their experiments that pronating movements tended to be quicker than supinating movements. They explained this effect as a result ‘of differences in either the acceleration phase (pronating muscles acting concentrically) or the deceleration phase (supinating muscles acting eccentrically)’. During the simulations the forearm was supinated by the *supinator* and by the *biceps brachii*. Variations of the *biceps brachii* activity showed that the duration of the supination decreases with increasing activity of this muscle. Thus the discrepancies between the simulated movements and those in the experiments obviously are the result of relatively high activities of the simulated *biceps brachii*. It must be assumed that during real supinating movements without additional loads the *biceps brachii* activity is low in order to economize the energy consumption as discussed in a next paragraph.

Although the simulated durations of pronation and supination are longer than the corresponding mean values of Rosenbaum *et al.* (1996), the middle-is-faster effect can also be derived from the simulated motion cyclic in the medial, the central, and the lateral position of the motion range (Table 3). Regarding only the results of Winter and Kleweno (1993) this effect is not easily to explain. Winter and Kleweno (1993) demonstrated that the isometric torque patterns for supination and pronation are mirror images of one another. This means that the supinators are strongest in the medial position (extreme pronation) and the pronators are strongest in the lateral position (extreme supination), whereas in the middle position the torques of the supinators and the pronators have nearly the same level. From a mechanical point of view the best strategy to perform cyclic movements is to satisfy two requirements, namely that the torques are high as possible resulting in high acceleration and thus in short durations of the movements, and secondly that the torques in both rotation directions (supination and pronation) are approximately equal resulting in harmonic movements. According to Winter and Kleweno (1993) in the middle position the torques in both rotation directions are relatively high and equal which enables fast movements in this position. In contrast to this in the two extreme positions high torques would be possible only in one rotation direction and low in the opposite direction. In order to satisfy the second requirement of equal torques in the medial position, the activity of the supinators should be submaximal and corresponding in the lateral position the activity of the pronators should be submaximal which results in relatively slow movements in both extreme positions. During the simulations this relationship was attained by the low stimulation rate of the biceps brachii (Fig. 4) and the short activity of the *pronator terres* and the *pronator quadratus* (Fig. 5).

The simulated supination was performed with small contraction of the *biceps brachii, short head*. The *biceps brachii* supports supination by acceleration of the movement. The main function of the *biceps brachii* is the flexion of the arm in the elbow joint or in the case of holding an object in the hand the *biceps brachii* has to balance the torque of this load. If only a supination should

Table 3
Durations [ms] for Supinating and Pronating Movements in the Medial, Central, and Lateral Positions of the Motion Range. Mean Values from Two Experiments of Rosenbaum *et al.* (1996) and the Shortest Simulated Values Depicted from Fig. 2 to 5

Grey areas ⇒ values are drawn from the movements shown in Fig. 3 and 5, starting with a supination

	Movement positions					
	Medial		*Central*		*Lateral*	
	Supination	*Pronation*	*Supination*	*Pronation*	*Supination*	*Pronation*
Experiment 2	138	137	120	114	149	135
Experiment 3	188	169	149	140	191	171
Simulation	287	300	254205	248216	281	245

Note: In their first experiment Rosenbaum *et al.* (1996) tested the hypothesis that the so called end-state comfort effect which stemmed from an expectation that movements can be made more quickly in the middle of the pronation-supination range than at either extreme.

be performed the *biceps brachii* induced flexion must be compensated by the *triceps brachii*. However, the activation of an additional muscle increases the energy consumption which is necessary for the arm movement. Therefore it can be assumed, that the activity of the *biceps brachii* is increased only so far that the supination and the pronation movement has nearly the same duration (see Table 3). Hereby a harmonic rotation of the forearm is attained which can be performed with relatively low energy consumption.

By means of the model the forearm rotation can be simulated for only one or two cycles in contrast to reality where the cycles can be repeated till muscle fatigue occurs. At the end of the first or second cycle the values of the motion related variables differ from the values at the beginning of the cycle. It follows from this that the pattern of the muscle stimulation rates has to be changed in order to simulate further cycles. Regarding the studies of Schmidt (1975, 1991), it can be assumed that a generalized motor program exists for such a relatively simple movement as the rotation of the forearm. By varying the parameters which determine the way, movements are constructed, the actual movement outcome can be fitted to the altered environmental demands and the desired movement can be carried out. The information about the variation of the parameter can be received by a feedback loop. Concerning the feedback loop duration Kawato *et al.* (1987) mentioned that there are substantial delays. Citing the study of Evarts (1981) they wrote that, for example, the transcortical loop for the control of short duration movements requires 0.04 to 0.06 s. According to the results of Rosenbaum et al. (1996) one cycle of the forearm rotation lasted longer than 0.25 s on average. Thus there should be enough time for a closed-loop mechanism which obviously results in a variation of the motion parameters as described by Roth and Willimczik (1999). Hereby the errors in response execution can be reduced (Schmidt 1976).

CONCLUSIONS

By varying the pattern of the muscular stimulation rates selected, movements can be simulated by means of the biomechanical model which implies many details of the mechanical properties of the human hand-arm system. In the case of realistic movements the system, responsible for selecting the movements and activating the muscles, does probably not know these details. Rather, during an initial phase, the movements are carried out slowly and corrected by a long-loop sensory feedback. Hereby general motor programs in the sense Schmidt (1975) are learned and saved. The programs seems to be the link between cognition and biomechanics.

From a mechanical point of view a harmonic time course of alternative rotations can only be attained, if the muscle torques have nearly the same value in both rotation directions. For the cyclic pronating and supinating movements this relationship is given in the central position of the forearm and the muscle torques in both rotation directions are relatively great. This results in the effect that the forearm can be cyclic rotated more quickly in the middle of the motion range than near the extremes.

It is difficult to simulate repeated movements because the environmental demands can alter with each motion cycle. This requires the change of the stimulation pattern for each cycle. In reality the fit of the movement outcome to the altered environmental demands is probably attained by varying the parameters which determine the way movements are constructed. Hereby the errors in response execution can be reduced.

REFERENCES

[1] Bernstein, N. (1967), *The Coordination and Regulation of Movements*. London: Pergamom.

[2] Dempster, W. T. (1955), *Space Requirements of the Seated Operator.* Wright Air Development Center (WADC), Tech. Rep. No. 55-159.

[3] Dennerlein, J. T., Kingma, I., Visser, B. & van Dieën, J. H. (2007), The Contribution of Wrist, Elbow and Shoulder Joints to Single-finger Tapping. *Journal of Biomechanics* 40: 3013 – 3022.

[4] Dong, R. E., Dong, J. H., Wu, J. Z. and Rakheja, S. (2007), Modeling of Biodynamic Responses Distributed at the Fingers and the Palm of the Human Hand-arm System. *Journal of Biomechanics* 40: 2335–2340.

[5] Evarts, E.W. (1981), Role of Motor Cortex in Voluntary Movements in Primates. In: Brooks, V. B. (ed) Handbook of Physiology, sect 1: vol 11, part 2. American Physiological Society, Bethesda, pp. 1083–1120.

[6] Fritz, M. (1991), An Improved Biomechanical Model for Simulating the Strain of the Hand-arm System under Vibration Stress. *Journal of Biomechanics* 24: 1165-1171.

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

[8] Kawato, M., Furukawa, K. and Susuki, R. (1987), A Hierarchical Neural-network Model for Control and Learning of Voluntary Movement. *Biological Cybernetics* 57: 169–185.

[9] Lemay, M. A. and Crago, P. E. (1996), A Dynamic Model for Simulating Movements of the Elbow, Forearm, and Wrist. *Journal of Biomechanics* 29: 1319–1330.

[10] Luo, Z. and Goldsmith, W. (1991), Reaction of Human Head/ Neck / torso System to Shock. *Journal of Biomechanics* 24: 499–510.

[11] Nigg, B. M. (1994), General Comments about Modelling. In: Nigg, B. M. & Herzog (Eds.), *Biomechanics* of the Musculo-Skeletal System (pp. 367–379), Chichester, New York: John Wiley & Sons.

[12] Rakheja, S., Wu, J. Z., Dong, R. G. and Schopper, A. W. (2002), A Comparison of Biodynamic Models of the Human Hand-arn System for Applications to Hand-held Power Tools. *Journal of Sound and Vibration* 249: 55–82.

[13] Reynolds, D. D. and Falkenberg, R. J. (1982), Three- and Four-degrees of Freedom Models of the Vibration Responses of the Human Hand. In: Vibration Effects on the Hand and Arm in Industry (Brammer, A. J. and Taylor, W., editors), 117–132, John Wiley & Sons, New York.

[14] Rosenbaum, D. A., Vaughan, J., Barnes, H. J., Marchak, F. & Slotta, J. (1990), Constraints on Action Selection: Overhand Versus Underhand Grips. In: Jeannerod M. (Ed.), *Attation and performance XIII* (pp. 321 – 342). Hillsdale, NJ: Lawrence Erlbaum Associates.

[15] Rosenbaum, D. A., Heugten, C. M. van & Caldwell, G. E. (1996), From Cognition to Biomechanics and Back: The End-state Comfort Effect and the Middle-is-faster Effect. *Acta Psychologica*, 94: 59–85.

[16] Roth, K. & Willimczik, K. (1999), *Bewegungswissenschaft*. Reinbeck: Rowohlt.

[17] Pennestri, E., Stefanelli, R., Valentini, P. P. & Vita, L. (2007), Virtual Musculo-skeletal Model for the Biomechanical Analysis of the Upper Limb. *Journal of Biomechanics* 40: 1350 – 1361.

[18] Schmidt, R. A . (1975), A Schema Theory of Discrete Motor Skill Learning. *Psychological Review*, 82: 225–260.

[19] Schmidt, R. A. (1976), Control Processes in Motor Skills. *Exercise and Sport Sciences, Review*, 4: 229–261.

[20] Schmidt, R. A. (1991), *Motor Learning & Performance*. Champaign, IL: Human Kinetics Books.

[21] Schumacher, R. H. & Wolff, E. (1966), *Trockengewicht und Physiologischer Querschnitt der Menschlichen Skelettmuskulatur - II. Physiologische Querschnitte*. Anatomischer Anzeiger 119, 259–269.

[22] Seireg, A. & Arvikar, R. (1989), *Biomechanical Analysis of the Musculoskeletal Structure for Medicine and Sport*. New York: Hemisphere Publishing Corporation.

[23] Winter, J. M. & Kleweno, D. G. (1993), Effect of Initial Upper-limb Alignment on Muscle Contributions to Isometric Strength Curves. *Journal of Biomechanics* 26: 143–153.

[24] Winters, J. M. & Stark, L. (1985), Analysis of Fundamental Human Movements Patterns through the Use of in-depth Antagonistic Muscle Models. *IEEE Trans. Biomed. Engng* 32: 826–839.