INFORMATION TO USERS

This manuscript has been reproduced from the microfilm master. UMI films the text directly from the original or copy submitted. Thus, some thesis and dissertation copies are in typewriter face, while others may be from any type of computer printer.

The quality of this reproduction is dependent upon the quality of the copy submitted. Broken or indistinct print, colored or poor quality illustrations and photographs, print bleedthrough, substandard margins, and improper alignment can adversely affect reproduction.

In the unlikely event that the author did not send UMI a complete manuscript and there are missing pages, these will be noted. Also, if unauthorized copyright material had to be removed, a note will indicate the deletion.

Oversize materials (e.g., maps, drawings, charts) are reproduced by sectioning the original, beginning at the upper left-hand corner and continuing from left to right in equal sections with small overlaps. Each original is also photographed in one exposure and is included in reduced form at the back of the book.

Photographs included in the original manuscript have been reproduced xerographically in this copy. Higher quality 6" x 9" black and white photographic prints are available for any photographs or illustrations appearing in this copy for an additional charge. Contact UMI directly to order.

UMI

A Bell & Howell Information Company
300 North Zeeb Road, Ann Arbor, MI 48106-1346 USA
313/761-4700 800/521-0600

**DYNAMIC MODEL OF THE HAND
WITH APPLICATION IN FUNCTIONAL NEUROMUSCULAR
STIMULATION**

by
ALI ESTEKI

Submitted in partial fulfillment of the requirements
for the degree of Doctor of Philosophy

Thesis Advisor: Professor Joseph M. Mansour

Department of Mechanical and Aerospace Engineering
CASE WESTERN RESERVE UNIVERSITY

August, 1995

UMI Number: 9607928

UMI Microform 9607928

Copyright 1995, by UMI Company. All rights reserved.

This microform edition is protected against unauthorized
copying under Title 17, United States Code.

UMI

300 North Zeeb Road
Ann Arbor, MI 48103

CASE WESTERN RESERVE UNIVERSITY

GRADUATE STUDIES

We hereby approve the thesis of

ALI ESTEKI

candidate for the PHD

degree.*

(signed) *Joseph M. Morrison*
(chair) *Patrick E. Crago*
Michael W. Feist
P. H. Bick

date *Apr. 25 1995*

*We also certify that written approval has been obtained for any proprietary material contained therein.

I grant to Case Western Reserve University the right to use this work, irrespective of any copyright, for the University's own purposes without cost to the University or to its students, agents and employees. I further agree that the University may reproduce and provide single copies of the work, in any format other than in or from microforms, to the public for the cost of reproduction.

DYNAMIC MODEL OF THE HAND WITH APPLICATION IN FUNCTIONAL NEUROMUSCULAR STIMULATION

Abstract

by

ALI ESTEKI

A dynamic model of hand was generated and used to study potential hand functions on hand positioning and grasp force and the outcome of surgeries such as tendon transfers and joint fusions, in situations typical of those encountered in FNS. In the model, the hand is being treated as a multi-body system including rigid segments (phalanges and palm) connected by joints. Each joint is subjected to muscle moments, produced by a muscle model, passive joint moment and external force moments. The external forces are the result of interaction forces between the hand segments and a grasped fixed or movable object. In particular, the model was used to study the non-prehensile tip and palmar pinch movements, and to simulate prehensile lateral and palmar grasp functions of the hand.

A quasi-linear viscoelastic model was found to described the joint passive moment. Model simulations showed that, in the absence of external forces, tip pinch posture is not obtained using only the extrinsic muscles. The currently used tendon transfer methods were found useful in normal hands but not significantly effective in

quadriplegic hands. It was also found that activation of multiple thumb muscles (ADP, APB and FPL) without surgical alterations can provide functional lateral pinch with about 70% more grip force than the currently used method with joint fusion. A grasp protocol was introduced and shown successful in palmar grasp and hold of movable cylindrical objects using only extrinsic muscles.

ACKNOWLEDGMENTS

I would like to express my sincere gratitude to my advisor Professor Joseph M. Mansour for his guidance and support throughout the course of this project. I would also like to extend my appreciation to Professors Patrick Crago and Dwight Davy for their useful suggestions in muscle modeling and the usage of optimization techniques.

Last but by no means the least, I would like to thank my wife for her support and encouragement.

This project was supported by a grant from Neural Prosthesis program (N.I.N.D.S.) of the National Institute of Health.

TABLE OF CONTENTS

ABSTRACT	ii
ACKNOWLEDGMENTS	iv
TABLE OF CONTENTS	v
LIST OF FIGURES	vii
LIST OF TABLES	ix
NOMENCLATURE	x
I. INTRODUCTION	1
A. Background	2
<i>Rouvas' model</i>	5
<i>Sarangapani's model</i>	9
<i>Hendrix's model</i>	10
B. Present Work	12
II. JOINT PASSIVE MOMENT	17
A. Introduction	17
B. Methods	21
C. Results	28
D. Discussion	33
III. THE MODEL OF HAND	48
A. Introduction	48
B. ADAMS Review	50
C. Development of The Model	55
<i>Multi-body system</i>	57
<i>Muscle-tendon unit</i>	62
D. Results	80
E. Discussion	82
IV. NON-PREHENSILE PALMAR PINCH	98
A. Introduction	98
B. Method	101
C. Results	110

D. Discussion	112
V. PREHENSILE LATERAL PINCH	126
A. Introduction	126
B. Method	128
C. Results	138
D. Discussion	143
VI. PREHENSILE PALMAR GRASP	158
A. Introduction	158
B. Method	161
Ground-object interaction model	162
Hand-object interaction model	163
C. Results	177
D. Discussion	181
VII. CONCLUSIONS AND FUTURE WORK	196
A. Conclusions	196
B. Future Work	198
REFERENCES	201

LIST OF FIGURES

Figure

- 1.1 Schematic of previous static models of hand.
- 1.2 Schematic of the present dynamic model of hand.

- 2.1 Device used to measure the passive moment at the MCP joint.
- 2.2 Moment relaxation at the MCP joint.
- 2.3 Elastic moment for one of the subjects tested (JM).
- 2.4 Typical passive moment hysteresis loop at the MCP joint.
- 2.5 Measured hysteresis loop thickness as a function of joint angular speed.
- 2.6 Measured damping moment as a function of joint angle.
- 2.7 Measured passive moment and corresponding model results.
- 2.8 Passive moment at MCP joint with the wrist at flexed and extended angles.
- 2.9 Passive moment hysteresis loop at MCP joint of a quadriplegic person.

- 3.1 Schematic of the ADAMS hand model.
- 3.2 Orientation of the local reference coordinate of the phalanges.
- 3.3 MCP flexion/extension joint angle verses muscle activation level.
- 3.4 MCP adduction/abduction joint angle verses muscle activation level.
- 3.5 PIP flexion/extension joint angle verses muscle activation level.
- 3.6 DIP flexion/extension joint angle verses muscle activation level.
- 3.7 CMC flexion/extension joint angle verses muscle activation level.
- 3.8 CMC adduction/abduction joint angle verses muscle activation level.
- 3.9 MP flexion/extension joint angle verses muscle activation level.
- 3.10 IP flexion/extension joint angle verses muscle activation level.
- 3.11 MCP joint angle against time when FDS is activated.
- 3.12 Passive moment hysteresis loop predicted by the model.

- 4.1 Palmar pinch posture of the hand.
- 4.2 Schematic of the interface of the hand model and the optimization routines.
- 4.3 Iteration of index finger muscle activation levels and resultant joint positions.
- 4.4 Iteration of thumb muscle activation levels and resultant joint positions.
- 4.5 Evaluated objective function verses iteration number.
- 4.6 Iteration of abnormal index finger muscle activations and joint positions.
- 4.7 Optimization results with the Zancolli-Lasso tendon transfer, normal hand.
- 4.8 Results with the modified Zancolli-Lasso tendon transfer, normal hand.
- 4.9 Optimization results with the Zancolli-Lasso tendon transfer, abnormal hand.
- 4.10 Results with the modified Zancolli-Lasso tendon transfer on an abnormal hand.

- 4.11 Evaluated objective function verses iteration number in variety of cases.
- 4.12 Evaluated objective function verses variable EI and DII activation levels.

- 5.1 Functional hand posture in lateral prehension.
- 5.2 Schematic of the lateral pinch model.
- 5.3 Plot of the STEP function built in ADAMS.
- 5.4 Schematic of the thumb-object intersection in lateral prehension.
- 5.5 Hysteresis loop of loading and unloading of thumb distal phalanx.
- 5.6 The external force produced by activation of ADP and FPL verses time.
- 5.7 Front view of the thumb when ADP is fully activated.
- 5.8 Front view of the thumb when FPL is fully activated.
- 5.9 Front view of the thumb when ADP is fully activated with IP joint fused.
- 5.10 Front view of the thumb when FPL is fully activated with IP joint fused.
- 5.11 Front view of the thumb when ADP and FPL are activated.
- 5.12 Front view of the thumb when ADP, FPL and APB are activated.
- 5.13 Front view of the thumb when FPL and EPL are activated.

- 6.1 Posture of palmar grasp of a cylindrical object.
- 6.2 Schematic of the palmar grasp model.
- 6.3 Schematic of hand-object intersection in palmar grasp.
- 6.4 Position of the object during optimization in palmar grasp.
- 6.5 Hand posture and object position during palmar grasp, case number one.
- 6.6 Hand posture and object position during palmar grasp, case number two.
- 6.7 Hand posture and object position during palmar grasp, case number three.
- 6.8 Hand posture and object position during palmar grasp, case number four.
- 6.9 Hand posture and object position during palmar grasp, case number five.
- 6.10 Hand posture and object position during palmar grasp, case number six.
- 6.11 Hand posture and object position during palmar grasp, case number seven.
- 6.12 Hand posture and object position during palmar grasp, case number eight.
- 6.13 Same as 6.11 with the Zancolli tendon transfer.
- 6.14 Same as 6.12 with the Zancolli tendon transfer.

LIST OF TABLES

Table

- 2.1 Characteristics of the normal subjects tested.
- 2.2 Constants A and B in the normalized relaxation function.
- 2.3 Passive moment constants at the MCP joint of the subjects tested.
- 2.4 Passive moment constants at the thumb and index finger joints.

- 3.1 Wrist, thumb, and index finger joint characteristics used in the model.
- 3.2 Geometric and inertia parameters of the model element.
- 3.3 Physiological Cross Sectional Area of the selected muscles in hand model.
- 3.4 Parameters and computed forces of the selected muscles in the model.
- 3.5 Tendon moment arms of the thumb, and index finger at different joints.
- 3.6 Excursions of index finger tendons.
- 3.7 The isometric force predicted by the model and measured in the literature.

- 4.1 Summary of the simulations performed.
- 4.2 Summary of results of the optimization in normal and abnormal hands.

- 5.1 Coefficients of regression of force-displ. curves of thumb distal pad.
- 5.2 Summary of the results of variety of trials in lateral pinch.

- 6.1 Summary of static optimizations performed.
- 6.2 Summary of dynamic simulations performed.
- 6.3 Coefficients of regression of force-displacement curves of different pads.
- 6.4 Activation levels required to hold an object resulted from optimization.
- 6.5 Results of dynamic simulations of palmar grasp in variety of cases.

NOMENCLATURE

Joints

CMC	Carpometacarpal (thumb)
MP	Metacarpophalangeal (thumb)
IP	Interphalangeal (thumb)
MCP	Metacarpophalangeal (index finger)
PIP	Proximal Interphalangeal (index finger)
DIP	Distal Interphalangeal (index finger)

Muscles

FDP	Flexor Digitorum Profundus
FDS	Flexor Digitorum Superficialis
EI	Extensor Indicis
EDC	Extensor Digitorum Communis
PII	First Palmar Interosseous
DII _p	First Dorsal Interosseous (parallel head)
DII _t	First Dorsal Interosseous (transverse head)
FPL	Flexor Pollicis longus
FPB	Flexor Pollicis Brevis
EPL	Extensor Pollicis Longus
EPB	Extensor Pollicis Brevis
APL	Abductor Pollicis Longus
APB	Abductor Pollicis Brevis
ADP _o	Adductor Pollicis (oblique head)
ADP _t	Adductor Pollicis (transverse head)
OPP	Opponens Pollicis

Hand Segments

MC1	First Metacarpal (thumb)
TPP	Thumb Proximal Phalanx
TDP	Thumb Distal Phalanx
MC2	Second Metacarpal (palm)
IPP	Index Proximal Phalanx
IMP	Index Middle Phalanx
IDP	Index Distal Phalanx

CHAPTER ONE

INTRODUCTION

Functional Electrical Stimulation (FES) or Functional Neuromuscular Stimulation (FNS) is a developing technology that restores limited voluntary control of skeletal muscles where nervous system damage has rendered them paralyzed. The application of functional electrical stimulation provides a means of permanently activating the function of the muscles involved in performing a desired movement. Clinical applications of functional electrical stimulation have proved that this method can generate controlled contraction of skeletal muscles (Peckham, 1988). Stimulation may be used to gain and maintain control of the daily movements used for manipulation, grasp, locomotion, or posture. With the implementation of this system to the upper extremities, the patient would be able to reach, grasp and pick up an unmodified object and manipulate it for use, independently.

A quantitative simulation of hand function can be quite useful in predicting potential grasp performance, in situations typical of those encountered in FNS. These include operating with a selected set of muscles, fused joints, and surgical alterations such as tendon transfers. In addition, simulation can be used to study the interaction of objects with the hand with both open and closed loop implementations of FNS. Although, the simulation can not fully replace subject testing, the model can be used in place of experimental tests on subjects to make initial evaluations of muscle function and a range of functional improvements that could be obtained in a comprehensive FNS program. It can be utilized to quickly identify rehabilitative procedures that are most likely to result in functional improvement, thereby narrowing the number of procedures tested in clinical trials.

A. Background

Numerous efforts have been made to model the hand function. In general, two distinct approaches have been used. The first approach is to solve the equilibrium equations for the muscle forces as a function of specified hand position (Chao and An, 1977; An *et al.*, 1985). This inverse approach leads to an indeterminate problem since there are more muscles than equilibrium equations. Some optimum criterion has often been introduced to overcome the indeterminacy problem. The second method uses a forward approach, in which the equilibrium equations are solved for joint positions,

driven by muscle forces (Wells *et al.*, 1985; Buchholtz *et al.*, 1987; Buchholts and Armstrong, 1992; Mansour *et al.*, 1994). The inverse approach computes for muscle forces necessary to obtain a certain hand posture, whereas the forward approach computes for the hand position as a function of muscle forces.

Wells *et al.* (1985) introduced a planar model of the index finger with unloaded movement, which allowed no abduction/adduction at the MCP joint. The model recognized the importance of passive moments at the joints in determining the static equilibrium positions of the finger. The muscle model only included force-length property. Buchholtz *et al.* (1987) used this model to predict index finger grip strength as a function of the diameter of cylindrical objects. The phalanges of the finger were modeled as ellipsoidal segments with contact points on the cylinder. In this model, first, a wrap-around protocol was used to determine the contact points with the stationary cylinder. In other words, the joint angles were independently manipulated to cause the proximal phalanx to contact the cylinder first, fixing the PIP joint in space, and then causing the middle and distal phalanges to contact. In this respect, the algorithm was a backward solution to the kinematics problem. Once the joint positions had been determined, the corresponding muscle forces were calculated using force-length relationships. Then the muscle forces were input into the forward kinematics equilibrium routine to determine the corresponding contact forces with the cylinder. Buchholtz and Armstrong (1992) developed a three dimensional model of the thumb and all the fingers with the application in grasp of cylindrical, stationary objects. This

model also solved the reverse kinematic problem by forcing the fingers to wrap around the cylinder. The interaction forces between the object and the hand segments were calculated based on the depths of penetration of the rigid cylinder into the ellipsoidal finger segments using the soft tissue deformation characteristics of the hand.

Several efforts have also been made to model the lower extremity. A computer model of the human lower extremity was developed by Delp *et al.* (1990) to study how surgical changes in musculoskeletal geometry and musculotendon parameters affect muscle force and its moment about the joints. A model for each actuator was formulated to compute its isometric force-length relationship. The kinematics of the lower extremity were included, thus the force and joint moment could be computed for any body position. Tendon transfer and lengthening procedures could be simulated by adjusting the model parameters according to various surgical techniques. Yamaguchi and Zajac (1990) used a computer simulation of bipedal gait to study the performance of FNS as a practical means to restore aesthetic, unsupported gait to paraplegics. The model used a combination of dynamic programming and an open-loop, trial-and-error adjustment process to find a suboptimal set of discretely varying muscle stimulation patterns needed for a three dimensional, eight degree of freedom dynamic model to sustain a step.

A static model of hand function was developed at this institute (Rouvas, 1989; Sarangapani, 1991) to quantitatively investigate the relative roles of the extrinsic and

intrinsic muscles, and to explore the potential for using the intrinsic muscles in hand grasp (Hendrix, 1993). Rouvas initiated the generation of the model, Sarangapani improved it by appending a muscle-tendon unit model, and Hendrix investigated the use of intrinsics in grasp by addition of index finger intrinsic muscles in his version of a grasp model (Figure 1.1). The following is a quick sequential survey of different versions of the static model.

Rouvas' model

The motivation for developing a static model of hand was to answer the question that what position the hand will take if certain muscles are activated. Most of the previous efforts in hand modeling were to predict muscle forces necessary to obtain a certain hand position, which is often called an inverse static problem. However, the task here was to solve a forward static problem. This approach led to a determinant problem that could be solved without the use of additional constraint relations or objective functions.

The model used free body diagrams, with sections at each joint, to formulate the equilibrium equations. Each bone was acted upon by active muscle forces and a set of internal reactions and passive moments at the joints. The muscles crossing the joint were represented by a force vector and all other forces were resolved into an internal

reaction force and a passive moment acting at the center of rotation of the joint. This representation of the joint eliminated the need to identify and measure the sources of internal forces and moments (ligaments, joint capsule, skin and fascia etc.), an extremely difficult if not impossible task. In this model the following quantities had to be known in particular:

1. Tendon moment arms at each joint.
2. Passive moment at each joint.
3. Physical dimensions of the hand.

The mechanical model consisted of the equilibrium equations of the thumb, and the index and middle fingers. The general format of the equations was:

$$\text{Force equilibrium: } \sum \bar{\mathbf{F}} + \bar{\mathbf{R}} + \bar{\mathbf{E}} = \mathbf{0}, \quad (1.1)$$

$$\text{Moment Equilibrium: } \sum (\bar{\mathbf{r}} \times \bar{\mathbf{F}}) + \bar{\mathbf{P}} + (\bar{\mathbf{l}} \times \bar{\mathbf{E}}) = \mathbf{0}, \quad (1.2)$$

where $\mathbf{F}$ is the muscle force, $\mathbf{R}$ is the joint reaction force, $\mathbf{E}$ is the external force, $\mathbf{r}$ is the tendon moment arm, $\mathbf{P}$ is the joint reaction moment, and finally $\mathbf{l}$ is the moment arm of the external force.

a) Rouvas' Model:

b) Sarangapani's Model:

c) Hendrix's Model:

Figure 1.1) Schematic of Rouvas' (a), Sarangapani's (b), and Hendrix's (c) models of hand at a common joint.

In this model only the extrinsic muscles that cross the joints of the digits being investigated were included and the inertia of the fingers was ignored. The solution of the equations was then reduced from a system of differential equations to a system of nonlinear algebraic equations. Also, the position of the wrist was assumed to be fixed in the neutral position.

The tendon moment arms of the index finger used in the model were taken from An *et al.* (1979). For the middle finger, the index finger moment arms were normalized with respect to the length of the middle finger. The thumb moment arms at CMC joint were used from Ou (1980). The values of moment arms at MP and IP joints were taken from the work of Ketchum reported by Brand (1985). Passive moments at index finger joints were experimentally obtained. Physical dimensions were measured directly on the author's hand.

Sarangapani's model

The same muscles were used in this version of the hand model, but instead of the muscles being treated as pure force generators, they were given muscle characteristics. The force-length properties were included in this model of the hand. A static muscle model was developed using the model introduced by Crago *et al.* (1992). Here muscle force was a function of muscle unit length, stimulus period and activation

level. Moreover, the length of the muscle-tendon unit was dependent on the joint angle. An equation representing muscle-tendon length as a function of PIP and MCP joint angles was linear and of the form

$$L_m = L + (d_2 \cdot \Theta^{MCP}) + (d_3 \cdot \Theta^{PIP}), \quad (1.3)$$

where L is the rest length of the muscle, that is, the length of the muscle when all joints crossed by the muscle-tendon unit are at zero degrees of flexion (the finger is straight), d_2 and d_3 are constants based on the slope of the tendon excursion curves from Frankenburg's data (1990), and Θ^{MCP} and Θ^{PIP} are the MCP and PIP joint angles. The excursion due to abduction/adduction was assumed negligible compared to that resulting from flexion/extension, and therefore was ignored, preventing further non-linearity in the model. The muscle force equation was then

$$F = F_1 + (F_2 \cdot \Theta^{MCP}) + (F_3 \cdot \Theta^{PIP}), \quad (1.4)$$

where F_1 is a component of force independent of joint angles, F_2 and F_3 are components of the force that are dependent on the MCP and PIP joint angles, respectively.

Based on the data from Benedict *et al.* (1963), a constant tendon modulus of elasticity was assumed. However, Sarangapani found that the force generated by the model was insensitive to a wide range of change in tendon stiffness. Tendon areas were obtained from Blanton and Biggs (1970), and tendon lengths from Brand (1985)

and Lieber (1990). Ratios of muscle fiber length to muscle length, and pennation angles were also obtained from Lieber (1990).

Hendrix's model:

This version of the hand model sought to achieve the grasp of fixed cylindrical objects using the thumb, index, and middle finger models. An entirely forward solution method to the modeling of grasp, in which muscle stimuli were inputs and the contact forces were outputs, was utilized. This was somewhat similar to a model developed by Wells *et al.* (1985). Wells' model however, was purely planar, allowed no abduction/adduction movement at the MCP joint, was only for one finger and only accepted muscle contraction lengths rather than muscle stimuli parameters.

Previous versions of the hand model by Rouvas and Sarangapani utilized only the extrinsic muscles for the index and middle finger. However, in Hendrix's model, the interosseous muscles were added to the two fingers. Specifically, the muscles added were the first dorsal and first palmar interossei and the second and third dorsal interossei. Moreover, a grasp model was incorporated into the hand model which calculated whether or not one or more phalanges of the finger contacted the cylinder as muscles were stimulated.

Although the geometric model yielded finger pad penetration, it did not directly generate an external force. Consequently, force-displacement relationships for finger pads were incorporated into the model. The contact profile was experimentally obtained using a circular indenter compressing against the different finger pads. The geometric model calculated a displacement, if contact occurred, and the displacement-force equation computed the external forces exerted by the object to the finger.

After numerous grasp model simulations, Hendrix concluded that functional hook, tip-pinch, and wrap-around precision grasps of cylinders could be achieved through the use of extrinsics alone if the object is fixed and placed properly. However, the interossei could indeed overcome the clawing action of a finger controlled by the extrinsics alone and allow the finger to achieve a desired tip-pinch position. He also concluded that the position of the cylinder was a larger factor in determining the type of grasp achieved than the stimulation protocol applied to the muscles.

B. Present Work

All the hand models discussed so far, were static models computing only for the static equilibrium position of the joints, or the steady state grasp force applied to a stationary object. The hand performance in reaching, grasping and holding a movable object had to be investigated. A static model does not provide information about the transition

from one static equilibrium to another static equilibrium position. It cannot predict the prehensile grasp functions of hand in case the object to be grasped is allowed to move, a situation very common in practice. Moreover, the static models do not often include the effect of friction. Therefore, development of a dynamic model of hand, and model of grasp allowing the object to move was necessary.

New applications, such as lateral, palmar and tip pinch functions with or without surgical alterations, also needed to be studied. The static model developed in this institute did not include the intrinsic muscles of the thumb. These muscles, especially the adductor pollicis, have a major role in the movements of the thumb and are commonly used in the prehensile functions involving the thumb.

A dynamic model of hand is quite complex, even when used for limited functions such as lateral pinch and palmar grasp. The wrist, thumb, and index finger alone represent a system with ten degrees of freedom. Any moveable object which is grasped by the model would increase the degrees of freedom and the number of equations. More than fifteen muscles are used to control just the degrees of freedom of the thumb and index finger. Muscle force depends on muscle length as well as velocity which in turn depend on joint angles. The dynamic responses of muscles, joints and finger pads are considerably more complex than their static behavior. The static model can be described by a set of nonlinear algebraic equations, whereas a dynamic model is

described by nonlinear first and/or second order coupled ordinary differential equations.

There are several approaches that could be used to develop and numerically solve the equations of motion for such a system. Since the application is not highly specialized, general purpose codes which can easily handle different applications can be utilized. **ADAMS** (Automatic Dynamic Analysis of Mechanical Systems, Mechanical Dynamics Inc., Ann Arbor, MI) is a software package which incorporates most of the factors needed to study the dynamics of the hand. It generates the equations of motion for a system based on user-specified links (parts) and joints of a multi-body system. User friendly interfaces make defining the model and producing graphic outputs a relatively simple task. The ADAMS program allows us to define our own control and driving forces.

In the present dynamic model (Figure 1.2), the hand is being treated as a multi-body system including rigid segments (phalanges) connected by joints. Each joint is subjected to a total moment which is decomposed into active, passive and external components. The active component of joint moment arises from the reduction of active forces in individual muscles crossing a joint into an equivalent resultant force and moment about some specified point. The passive moment is due to the deformation of passive tissues such as tendons, ligaments, skin, as well as inactive

muscles. The external moment is the result of interaction forces between the hand segments and a grasped object.

Each segment of the multi-body model is specified in a main ADAMS program as a "part" using the geometric and inertial data of the phalanges and the palm. Joints are modeled by their mechanical equivalents in the main program using ADAMS' statements built in for certain kinematic constraints (for instance, "revolute" which allows only one degree of rotational freedom and constrains the other degrees of freedom). The passive moments are directly computed within the main routine and are applied at corresponding joints as force constraints. Moments of muscle and external forces are computed in separate user-written subroutines and are linked to the main ADAMS program. These moments are applied at the joints as driving moments of the multi-body system. The muscle subroutine computes the dynamic muscle forces based on the activation level of the muscles and the state variables of the model, joint angles and joint velocities. The interaction subroutine calculates the external forces according to the intersection kinematics of the phalanges and the grasped object. The external forces include the normal and frictional forces between the object and the finger pads. In the presence of an object, the interaction forces between the object and the ground are also included in the model. The rate dependent terms, such as activation dynamics and force-velocity relationships in muscle model and viscoelastic behavior of the joints and the finger pads, are all considered.

Figure 1.2) Schematic of the present dynamic model of hand at a common joint.

The dynamic model requires several input parameters:

1. Passive moment parameters at joints based on a nonlinear viscoelastic model presented in Chapter two.

2. Muscle model parameters estimated for hand muscles based on a model introduced by Crago (1992) and utilized by Chiel *et al.* (1992). Geometric parameters of the muscle-tendon units are borrowed from the literature.

3. Body segment parameters (mass, moment of inertia, center of mass, and geometry) determined from geometrical measurements on Professor J.M. Mansour's hand. Each phalanx is modeled as a cylinder and the palm as a thin box.

In summary, a dynamic model of hand is generated and used to investigate the effects of muscle selection, including the relative effect of intrinsic and extrinsic

muscles, on hand positioning and grasp force and the outcome of surgeries such as tendon transfers and joint fusions. In particular, the model is used to study the non-prehensile tip and palmar pinch movements, and is complemented by grasp programs to simulate prehensile lateral and palmar grasp functions of the hand with the application of FNS system. In each of these cases, the ability to attain a certain posture and/or grasp a particular object with or without surgical alterations are studied.

CHAPTER TWO

JOINT PASSIVE MOMENT

A. Introduction

In the model of hand, the skeletal system is being treated as a chain of rigid body segments connected by joints. Each joint is subjected to a total moment which can be decomposed into active, passive and external components. The active component of joint moment arises from the reduction of active forces in individual muscles crossing a joint into an equivalent resultant force and moment about some specified point. The external moment about a joint is the result of interaction forces between the hand segments and the grasped object.

The contribution of passive structures to the total force system at a joint is not as easily described as the active component. Motion of a joint causes passive tissues

such as tendons, ligaments, skin, as well as inactive muscles, to be deformed. The deformation of these passive tissues gives rise to a passive resistance or stiffness of the joint. A detailed description of this deformation process would be extremely complex, requiring constitutive properties of each tissue as well as their geometry, and might well involve finite deformations. Alternatively, many investigators have described the passive stiffness at a joint by combining the contributions from individual tissues into a lumped passive moment. Typically, the lumped passive moment is obtained from experimental measurements. This latter approach is generally used to prescribe the passive moment at a joint.

Passive moments have been measured and used to evaluate joint and neuromuscular diseases as well as to provide data for modeling musculoskeletal dynamics (Hayes and Hatze, 1977; Mills and Hull, 1991; Such *et al.*, 1975; Thomas, 1965; Wright and Johns, 1961; Yoon and Mansour, 1982). To ascribe some physical significance to these measurements, attempts have been made to decompose the passive moment into elastic, plastic, viscous, inertial, and frictional components. In a more general sense, the elastic, and frictional responses may be classified as rate independent while the viscous response is rate dependent.

The influence of age, sex, and joint size on the elastic and dissipative components of stiffness in the normal knee joint were studied by Such *et al.* (1975). Although this investigation did not attempt to separate the dissipative component of

the passive moment into viscous and frictional elements, the implication was that the dominant dissipative effects were frictional. Mills and Hull (1991) determined the stiffness of the human knee *in vivo* with a particular emphasis on the effect of foot constraints on passive knee moment. Foot position was shown to have a strong effect on the passive knee moment. In a similar manner, knee position has been shown to influence the passive moment at the hip (Yoon and Mansour, 1982) presumably due to the stretching and shortening of biarticular muscles such as rectus femoris and muscles of the hamstring group. Measurements of the passive moment at the hip were found to be independent of the angular speed, at least in the range of 0.017 to 1.62 rad/sec used in these tests (Yoon and Mansour, 1982). In this study it was also shown that the passive moment at the hip could be fit by exponential functions, but no attempt was made to identify elastic and dissipative components of the passive moment.

There have been several attempts at modeling passive moment. Wright and Johns (1961) used a pendulum type of apparatus (free and forced movement) to measure the passive stiffness at the second metacarpophalangeal joint. Measurements were made at frequencies from 0.003 to 3 cycles per second. In addition, they used their apparatus in a relaxation mode, that is, the joint was moved to a position and held fixed and the moment required to hold this position decayed or relaxed over time. To ascribe some physical significance to these measurements, the total joint stiffness was decomposed into elastic, plastic, viscous, inertial, and frictional components. In a more

general sense, the elastic, plastic and frictional responses may be classified as rate independent while the viscous response is rate dependent. It was found that the contributions to joint stiffness, in decreasing order of importance, were elastic, plastic and viscous effects. Inertial and frictional effects were found to be negligible in normal joints, although some frictional effects were observed in joints with severe rheumatoid arthritis. In contrast, Thomas (1965) believed that there were significant frictional contributions to total joint stiffness but that viscous effects were negligible. He based this conclusion on the failure of a model, with linear viscous damping, to fit experimental data. Studies by both Wright and Johns, and Thomas found the elastic component of the passive moment to be dominant, accounting for about fifty percent of the joint stiffness.

Hayes and Hatze (1977) measured the passive moment at the elbow and, based on these data, developed a viscoelastic model for the passive moment. They modeled the passive moment as the sum of an elastic contribution $F(\Theta)$, which was a nonlinear function of the joint angle Θ , and a viscous contribution $G(\Theta, \omega) = C(\Theta)\omega$, which was a linear function of joint speed. $C(\Theta)$ was a quadratic function of the joint angle. In this model the viscous part of the passive moment was directly proportional to the joint angular speed ω . This suggests a much stronger dependence on velocity than indicated by other studies. Frictional effects were not considered in this model. Although each of these investigations include a nonlinear elastic element, they offer widely differing models for the dissipative part of the passive moment. Some insight into resolving

theses differences might be gained from previous models of peri-articular tissues (Haut and Little, 1972; Fung, 1981; Woo, 1980) which suggest that linear elements can not be used to accurately represent the passive moment.

The objectives of this part of the dissertation are to identify the mechanisms which give rise to the dissipative part of the total passive moment and to develop a complete mathematical model of the passive moment. First, experiments are designed to reveal the qualitative contribution of viscous and frictional effects on the passive moment. From these experimental data a quantitative model is developed. The model parameters are then determined based on the measurements.

B. Methods

To better define the dissipative part of the passive moment, a two part experimental protocol is first developed and implemented. Tests are performed in moment relaxation and at constant velocity. Measurements are made at the MCP joint of the index finger of three normal subjects (Table 2.1). On the basis of these experimental measurements, a quantitative model of the passive moment is developed. Using the experimental results, the model parameters for MCP joint are determined for the three subjects tested. For the subject JM, the model parameters for PIP and DIP joints of the index finger are estimated based on the measurements at index finger joints. The

parameters for the thumb joints are estimated based on the measurements at index finger joints and crude measurements of elastic contribution of the thumb passive moment. In addition, the constant velocity test is also performed on PIP joint of the index finger of subject JM, and on MCP joint of a quadriplegic person. Finally, the effect of wrist position on the passive moment at the MCP joint is evaluated experimentally.

SUBJECT	SEX	AGE	HEIGHT(m)	WEIGHT(Kg)
JM	M	45	1.83	71
MT	F	25	1.60	55
AC	M	26	1.78	78

Table 2.1) Characteristics of the normal subjects tested.

A device (Figure 2.1) is designed to measure the passive moment at metacarpophalangeal (MCP) and proximal interphalangeal joints of index and middle fingers as a function of wrist angle, finger joint angle and finger joint velocity, although the modeling procedure is based on the measurements on the MCP joint of the index finger. All tests are performed with the finger moving in the horizontal plane and the subject's forearm lying on the test rig's table (A). Padded supports at the forearm (B) and palm (C) regions prevent lateral and rotational displacements of these

segments during a test. The subject's finger is strapped to a brace (D) made from aluminum and foam finger splints (Dillon Manufacturing Co., Norcross, GA). The braced finger is suspended at the end of a force measuring beam (E). Palm supports, mounted on plate (F), are used to set the wrist angle.

Figure 2.1) Device used to measure the passive moment at the metacarpophalangeal joint of the index finger. All measurements were made with the finger moving in the horizontal plane to eliminate the effects of gravity. The parts labeled in the figure are: (A) supporting table, (B) forearm support, (C) palm support, (D) finger brace, (E) force transducer, (F) wrist positioning plate, (G) horizontal arm, (H) main shaft, (I) cross slides, (J) center indicator, (K) motor, (L) limit switches and (M) potentiometer.

A vertical cantilever beam, with strain gages mounted near the fixed end, is used to measure the force tangential to the circular path of movement of the finger. This load cell is held by a horizontal arm (G) which was attached to a shaft (H). Two linear slides (I) are used to position the shaft (H) such that it is colinear with the approximate position of the axis of rotation of the metacarpophalangeal joint. This is performed by means of a center indicator (J) lowered from inside the shaft at the adjustment stage.

Joint rotation is driven by a gear-head motor (K, Bodine Electric Co.) directly connected to horizontal arm (G). The speed of rotation is set, and maintained constant, by an electronic controller (Bodine Electric Co.). The range of joint motion is set for each subject using two limit switches (L) mounted on a semi-circular plate. These switches are interfaced with the electronic controller which reverses the direction of rotation of the finger when either switch is closed. Joint angle is measured by a potentiometer (M) mounted on top of the shaft, H.

Two different tests are performed, one in moment relaxation and one where the joint is rotated at a constant velocity over its set range of motion. In the moment relaxation experiment, the joint is started from its zero position (where the finger and the metacarpal bones are aligned) and rotated with a constant angular speed of 1.0 rad/sec. After five cycles, from full flexion to full extension and back to full flexion, the motion is suddenly stopped, the joint is held for approximately one minute at a fixed

flexed or extended position and the relaxation of the moment is measured. Relaxation tests are performed approximately every ten degrees from full extension to full flexion.

In the second experiment, the joint is exercised over its full range of motion at a constant velocity. A complete cycle of data consists of rotation of the finger from full extension to full flexion where its motion is reversed by closing the limit switch, back to full extension. The actual limits of joint flexion and extension are set for each subject by noting the point at which the subject starts to feel uncomfortable. To determine the effect of rate of joint rotation on passive joint moments, measurements are made at angular speeds of 0.05, 0.33, 0.66, 0.85, 0.96 and 1.29 rad/sec. Each measurement is performed with the wrist in a neutral position, that is, with the palm aligned with the forearm. In addition, measurements are made with the wrist fixed at sixty seven degrees of flexion and sixty seven degrees of extension at a single speed of 0.85 rad/sec.

Data from the load cell and the potentiometer are acquired at a rate of 50 samples per second in real time, using a personal computer (80386 processor) in conjunction with a multi-channel analog to digital data acquisition board (DAS-1402, Keithley Data Acquisition, Taunton, MA). Relaxation data are collected for seventy seconds, which is a practical limit beyond which most subjects found it difficult to keep their muscles relaxed. The passive moment is computed from the load cell data and its known distance from the shaft (H).

To account for the possibility of preconditioning, the joint is exercised for five cycles through the test range just prior to data acquisition. Repeatability is also determined by first measuring the passive moment over the full range of joint motion, removing the hand from the apparatus and remounting the hand and making a second set of measurements at the joint.

A preliminary investigation of passive moment and peri-articular tissue models suggests that linear elements cannot be used to accurately represent the passive moment. The applicability of two nonlinear viscoelastic models,

$$M(\Theta, t) = F(\Theta) + G(\Theta)H(t) \quad (2.1)$$

$$M(\Theta, \omega) = F(\Theta) + G(\Theta)D(\omega), \quad (2.2)$$

is investigated, where Θ is the joint angle, ω is the joint angular speed, M is the passive moment at a joint, $F(\Theta)$ is the elastic response of the joint, $G(\Theta)$ is the instantaneous elastic response of the joint modified by either $H(t)$, the normalized relaxation function or $D(\omega)$.

The functions $F(\Theta)$ and $H(t)$ are obtained using data from relaxation tests. The elastic response, $F(\Theta)$, is equivalent to the relaxed joint moment, that is, the moment about seventy seconds after the joint position is fixed. Since the moment relaxes to a non-zero equilibrium value, $H(t)$ is obtained by both shifting the equilibrium value to zero and normalizing by the range of the moment according to

$$H(t) = \frac{M(t) - M_{\min}}{M_{\max} - M_{\min}}, \quad (2.3)$$

where $M(t)$ is the moment at time t , $M_{\min}$ the relaxed value of the moment at $t = 70$ seconds and $M_{\max}$ is the maximum value of the moment, that is, immediately after the joint position is fixed.

The constant velocity test is used to provide data for constructing the functions $G(\Theta)$ and $D(\omega)$. The instantaneous elastic response, $G(\Theta)$, is defined from the differences between the elastic moment ($F(\Theta)$) and the total flexion or extension passive moment over the range of joint angles tested. The function $G(\Theta)$ is found by fitting a curve to these differences as a function of joint angle. For this purpose, the instantaneous elastic response, $G(\Theta)$, is approximated using data from the constant velocity test at the highest velocity used, 1.29 rad/sec. The function $D(\omega)$ is found from the total joint moment M , and computed the functions F and G , using Equation (2.2)

$$D(\omega) = \frac{M(\Theta, \omega) - F(\Theta)}{G(\Theta)}. \quad (2.4)$$

The function $D(\omega)$ is then found by fitting a curve to the values on the right hand side of this equation as a function of joint angular speed ω .

For the subject JM, the parameters of the elastic contribution ($F(\Theta)$) for PIP and DIP joints of the index finger are determined from the constant velocity tests

performed on those joints. Curve fit is employed on several points on the center line of the hysteresis loop instead of the equilibrium joint moments. The parameters of the elastic contribution of the thumb joints are determined based on crude measurements by Hendrix (1993). For the viscoelastic contribution ($G(\Theta)D(\omega)$) of the IP joints of the index finger and the thumb joints, same parameters as determined for the MCP joint are used. It is assumed that the dissipation at thumb joints is slightly greater than finger joints. Therefore, the constant in Equation (2.7) is slightly increased for the thumb joints.

C. Results

Excellent repeatability of the measurements, without removing a subject's hand from the test stand, is observed with maximum differences between any two trials of not more than five percent. If the finger is tested, removed and then retested the difference between two cycles is greater, in the extreme approaching fifteen percent. Tests to evaluate the effects of preconditioning show that after two or three cycles there is no significant change in the passive moment-joint angle relationship.

Moment relaxation is rapid, but not instantaneous, decaying to a non-zero equilibrium value (Figure 2.2). Experimental values of the equilibrium joint moment, as a function of joint angle, are fit with an exponential function of the form

$$F(\Theta) = a(e^{b(\Theta-\Theta_r)} - e^{c(\Theta-\Theta_r)}), \quad (2.5)$$

where a , b and c are constants and Θ_r is the joint angle where the equilibrium joint moment is zero (Figure 2.3). The normalized relaxation function at each joint position is well fit by a logarithmic function of the form

$$H(t) = [\bar{A} \ln(t) + \bar{B}], \quad (2.6)$$

where $\bar{A}$ and $\bar{B}$ are averaged values of constants obtained from relaxation tests at several joint positions (Table 2.2).

JOINT ANGLE (deg)	A	B	r^2
-53.1	-0.149	0.733	0.969
-44.3	-0.134	0.681	0.985
-32.7	-0.127	0.622	0.901
58.4	-0.122	0.568	0.910
69.2	-0.136	0.594	0.974
82.0	-0.143	0.693	0.981
Average	$\bar{A} = -0.135$	$\bar{B} = 0.648$	
Standard Err.	0.010	0.064	

Table 2.2) Constants A and B in Equation (2.6) used to fit the normalized relaxation data at different joint angles. Since the constants A and B show little variation with joint angle, their average values $\bar{A}$ and $\bar{B}$ are used to construct the function $H(t)$. Note that close to the rest position the instantaneous elastic moment is relatively small, therefore the relaxation data are not reliable. Positive angles represent flexion and negative angles are extension.

The values of the constants A and B obtained from each individual test show little variation as a function of the fixed joint angle used in the relaxation tests implying that $H(t)$ is only a function of time. Note that close to the rest position the instantaneous elastic moment is relatively small, approaching the resolution of the measurement system: these relaxation data are not reliable and are not used to obtain $\bar{A}$ and $\bar{B}$. The normalized relaxation function is very close to one when t is close to zero and approaches zero when t goes to seventy seconds.

The instantaneous elastic response $G(\Theta)$, is constructed based on several observations. When the joint is cycled at constant velocity, hysteresis is always seen (Figure 2.4). The total flexion moment, going from full extension to full flexion, and the total extension moment, going from full flexion to full extension, are two distinct nonlinear functions of displacement (Figure 2.4). It is also observed that the hysteresis increases as a function of joint angle measured relative to the rest position (Figure 2.5a). Several different order polynomial expressions are fit to these data, and a second order polynomial of the form

$$G(\Theta) = g_0 + g_1(\Theta - \Theta_r) + g_2(\Theta - \Theta_r)^2 \quad (2.7)$$

is found to give a good fit (Figure 2.6).

Although the hysteresis loop thickness, the differences between the elastic moment and either the total flexion or total extension moment, appears to be almost

independent of the displacement rate, a more careful examination of the data reveals that the hysteresis increases as the rate increases (Figure 2.5a). A power function of the form

$$D(\omega) = \omega^n \quad (2.8)$$

is found to provide a good fit to the normalized hysteresis (Figure 2.5b). The exponent, n , is much less than one for all subjects tested (Table 2.3).

PARAMETER	SUBJECT		
	JM	MT	AC
a (N-Cm)	0.654	0.673	1.263
b	-2.412	-1.745	-1.826
c	4.523	1.585	3.488
g_0' (N-Cm)	-2.779	0.447	-2.234
g_1' (N-Cm)	-3.027	-0.372	-2.089
g_2' (N-cm)	-3.302	1.683	-2.246
g_0'' (N-cm)	2.252	-0.649	1.577
g_1'' (N-cm)	1.744	0.180	2.284
g_2'' (N-cm)	5.053	-0.352	4.744
Θ_r (rad)	0.71	0.44	0.87
n	0.12	0.25	0.16
$\bar{A}$	-0.135	-0.132	-0.139
$\bar{B}$	0.648	0.586	0.602

Table 2.3) Constants in the mathematical models given by Equations (2.1) and (2.2) for the three subjects tested. The superscripts "l" and "u" denote the constants for the lower and upper instantaneous elastic moment functions G .

Measurement of passive moment shows a wide range of normal variation for the three subjects tested. The parameters of the mathematical model for the subjects show a similarly wide range of variation (Table 2.3). Measured passive moment and corresponding model results computed from Equation (2.2), in the region where the velocity was constant, show very good agreement for different subjects and at different joint velocities (Figure 2.7). The applicability of the model is not investigated in the extremes of the motion, where the finger is reversing direction and the velocity is not constant. For one of the subject (JM), model parameters of the passive moment measured for the index finger joints and estimated for the thumb joints are summarized in Table (2.4).

Wrist position is found to have a significant effect on the passive moment at the metacarpophalangeal joint of the index finger (Figure 2.8). Holding the wrist at sixty seven degrees of flexion shifts the moment-displacement curve downward when compared to measurements with the wrist fixed at zero degrees. Sixty-seven degrees of wrist extension has the opposite effect, shifting the curve slightly upward.

The hysteresis loop of the MCP joint passive moment of the quadriplegic person has a similar shape as the normal joint (Figure 2.9). However, the joint range of motion is completely shifted to the flexion side (about fifty degrees).

Parameter	Thumb Joint				Index Finger Joint			
	CMC (flx)	CMC (add)	MP (flx)	IP (flx)	MCP (flx)	MCP (add)	PIP (flx)	DIP (flx)
a (N-Cm)	0.030	0.463	0.629	1.534	0.654	3.662	1.597	0.935
b	-0.010	-0.822	-2.505	-8.231	-2.412	-4.539	-3.275	-5.631
c	29.70	30.11	5.714	2.074	4.523	21.73	2.047	2.610
g_0^I (N-Cm)	3.8	3.8	3.3	2.3	-2.779	2.3	2.3	2.3
g_1^I (N-Cm)	-3.027	-3.027	-3.027	-3.027	-3.027	-3.027	-3.027	-3.027
g_2^I (N-Cm)	-3.302	-3.302	-3.302	-3.302	-3.302	-3.302	-3.302	-3.302
g_0^u (N-Cm)	3.7	3.7	3.2	2.2	2.252	2.2	2.2	2.2
g_1^u (N-Cm)	1.744	1.744	1.744	1.744	1.744	1.744	1.744	1.744
g_2^u (N-Cm)	5.053	5.053	5.053	5.053	5.053	5.053	5.053	5.053
(rad)	0.43	0.39	0.22	0.05	0.71	0.0	0.74	0.38
n	0.12	0.12	0.12	0.12	0.12	0.12	0.12	0.12
$\overline{A}$	-0.135	-0.135	-0.135	-0.135	-0.135	-0.135	-0.135	-0.135
$\overline{B}$	0.648	0.648	0.648	0.648	0.648	0.648	0.648	0.648

Table 2.4) Passive moment parameters for the thumb and index finger joints of subject JM. Index finger joint parameters are measured using the apparatus discussed earlier. Thumb joint parameters are estimated based on the index finger parameters, and crude measurements of elastic contribution of the thumb passive moment.

D. Discussion

The primary motivation for this investigation was the development of a general mathematical model of passive joint moment for use in dynamic simulations of hand

movement. For this application, a simple curve fit of the data, as used in some previous investigations, is not sufficient. A curve fit does not account for variations due to changes in joint angular velocity nor does it account for reversals in motion between flexion and extension at arbitrary joint angles. Of the two models developed in this investigation, that given by Equation (2.2) is more efficient for simulation of hand function. The independent variables in this model, joint position and velocity, are readily available in a simulation. The definition of time in Equation (2.1) complicates the use of this model since it must be initialized at each reversal of joint movement.

Passive moments are an important component of the control of limb movements. To appreciate the contribution of the passive moment to musculoskeletal motion, imagine the movement of a model finger with no passive moment at the joints. If any one muscle is activated, the joint will accelerate unabated, since there is nothing to oppose joint rotation. Of course, in actual joint movements, this does not happen. Either a second muscle opposing the first could be activated or the passive moment could balance the active moment and stop joint rotation. Electromyographic studies have shown that at least in some free movements of the fingers, there is no co-contraction of muscles. The passive moment opposes the active moment and is essential for determining finger position (Long, 1968; Long, 1981; Wells, *et al.*, 1985; Darling and Cole, 1990; Hendrix and Mansour, 1993; Mansour, *et al.*, 1994). For the specific case motivating this study, passive moment is particularly important. In quadriplegic people active muscle forces are sixty to eighty percent less than in normal

which makes the passive contribution highly significant (Yamaguchi, 1990; Lemay, 1994).

Linearity of the viscous response used in other models (Hatze and Hayes, 1977) was not consistent with the experimental results obtained here. A linear dependence of the viscous contribution on joint angular speed suggests a much greater effect of speed than was observed experimentally. Further, the experimental data suggested that for the joint and individuals tested, frictional effects were not significant. Relaxation experiments showed that initially there was a rapid, but not instantaneous drop in the moment. If friction was a dominant mechanism, an instantaneous drop in the moment would have been observed. The models used in this investigation are consistent with the quasi-linear viscoelastic behavior seen in several biological materials (Haut and Little, 1972; Woo *et al.*, 1980; Fung, 1981).

The non-linearity of the viscous behavior, as described by Equation (2.8), may be the cause for the differing findings on the relative contributions of frictional and viscous dissipation. Physically modeling the experimental data showed that the viscous contribution was manifested through joint speed raised to a low fractional power of velocity. This type of behavior has been observed in other joint systems (Wu *et al.*, 1990). A power of zero would indicate that the damping was independent of velocity and therefore frictional in nature, whereas a power of one would be an indication of linear viscous dissipation. Although the dissipation was found to be rate dependent, it

was a weak function of velocity and approximated a rate independent behavior. As can be seen in Figure (2.5b), an implication of this type of damping is the existence of a region where the damping is almost independent of velocity. Tests performed in this region could lead to the conclusion that friction was the dominant dissipative mechanism while tests performed only at lower velocities might suggest a linear viscous behavior.

The quasi-linear model for the dissipative part of the passive moment, $G(\Theta)D(\omega)$, provides a continuous switch between the two parts of the hysteresis loop, that is the flexion/extension and extension/flexion curves (Figure 2.4). When the direction of motion changes, first joint velocity decreases to zero, and then increases in the opposite direction. The decrease in velocity, decreases the value of the function $D(\omega)$ (Figure 2.5), and therefore reduces the hysteresis thickness. When the velocity is zero, the function $D(\omega)$ is zero, the hysteresis thickness is zero, and the total passive moment is equal to the relaxed elastic moment, $F(\Theta)$. The direction of velocity then changes and its magnitude increases, which in turn increases the function $D(\omega)$, and therefore increases the magnitude of the function $G(\Theta)D(\omega)$. However, this time, the function $G(\Theta)$ is negative and is subtracted from the relaxed elastic moment, $F(\Theta)$, resulting in the lower curve of the hysteresis loop. Same process, with a different sign change, occurs in switching from the lower to the upper curve.

There are some experimental limitations inherent in using human subjects. While making experimental measurements, muscle activity was not monitored. This could potentially corrupt the data and subsequent analysis. There were several reasons for believing that muscle activity was not an issue in these measurements. The joint speeds used were within the range used by both Thomas (1965) and Wright and Johns (1961). In these previous studies no electrical muscle activity was found. To assess the magnitude of muscle contraction on the measured moment, we also asked subjects to voluntarily resist the movement of their joint. Even small resistance was readily apparent, causing noticeable changes in the measured joint moment. It was found that some subjects were not able to fully relax their muscles while experimental measurements were being made: they tended to want to drive the device rather than let the device drive their joint. In these subjects the passive moment showed considerable variation over the joint range of motion or over time in the relaxation experiments. Data from all such subjects were eliminated from this investigation. The possibility of eliciting a stretch reflex at higher velocities leads to a necessary approximation for determining the instantaneous elastic response. This approximation is inherent in viscoelastic material testing (Fung 1981). We used the highest joint speed in our test protocol to approximate the instantaneous elastic response. Experience showed that approximately seventy seconds was a practical limit on the duration of the relaxation measurements. If the test was extended beyond this time, subjects would begin to show signs of active muscle contraction. Finally, some variability in repeated

50
measurements on the same subject are to be expected. These variations were found to be much less than those between subjects.

The effects of wrist position on the passive moment at the MCP joint would appear to be due to stretching of extrinsic finger flexor and extensor muscles. When the wrist is in a flexed position, the extrinsic extensor tendons are stretched and the flexors are slackened. Thus, with respect to the neutral position, the torque is larger in flexion and is lower in extension. This shifts the torque-displacement curve downward indicating that the finger rest position shifts to a less flexed angle. In contrast, wrist extension stretches the extrinsic flexors and slackens the extensors which in turn decreases the torque in flexion and increases the torque in extension. The result is shifting the torque-displacement curve upward, which predicts a more flexed rest position for the finger. Overall, changing wrist position produced a vertical shift in the passive moment curve but with little change in the shape of the curve. Similar shifts in passive moment due to wrist position have been observed at the proximal interphalangeal joint (Brand, 1985). The fact that the passive moment curve is shifted, but does not change shape, suggests that the models developed here can be applied more generally to include the influence of wrist position.

In summary, models of the passive moment were developed based on experimental measurements. These models are consistent with the quasi-linear viscoelastic behavior seen in several biological materials (Haut and Little, 1972; Woo

et al., 1980; Fung 1981). One of the models, given by Equation (2.2), is well suited for use in dynamic simulations of hand function since the independent variables, joint position and velocity, are readily available, while the other is better suited for describing the passive moment over time. Questions raised by previous studies regarding the influence of rate dependent and rate independent effects on the passive moment were addressed, and it was shown that models which employ linear viscous damping elements can not accurately describe the dissipation in passive moments. Although the dissipative part of the passive moment might appear to be dominated by rate independent effects, it was found that a nonlinear rate dependent model described this behavior. The manifestation of the rate dependency, as a low fractional power of velocity, shows that the dissipative part of the passive moment resembles, but is not, a rate independent process.

Figure 2.2) Moment relaxation at the metacarpophalangeal joint at a fixed joint angle of 42 degrees of extension (subject JM). Note that the moment does not relax to zero. In most subjects the moment was at or near equilibrium after seventy seconds, the practical upper limit after which active muscle contraction began to corrupt the data.

Figure 2.3) Elastic moment for one subject (JM) as function of the fixed joint angles used in the relaxation tests. The elastic moment was equivalent to the relaxed or equilibrium joint moment. The experimental data is represented by the (+) signs and the model, $F(\Theta) = a(e^{b(\Theta - \Theta_r)} - e^{c(\Theta - \Theta_r)})$ is shown by a solid line. Values of the constants a , b and c are given in Table (2.3).

Figure 2.4) Typical passive moment at the metacarpophalangeal joint measured at a constant angular speed of 0.85 rad/sec. The speed was constant between forty degrees of extension and seventy degrees of flexion. Data outside this region were not used to develop the model of passive moment since the motor was either stopping or starting. Positive angles represent flexion and negative angles are extension. The upper curve represents joint movement from extension to flexion, the lower from flexion to extension.

Figure 2.5) a) Measured hysteresis loop thickness as a function of joint angular speed and at four joint angles: 40 degrees (□), 30 degrees (Δ), 20 degrees (+) and 10 degrees (×) of extension. b) Hysteresis loop thickness normalized by the maximum value at each joint angle for several joint positions. The normalized loop thickness is essentially independent of joint angle. As the velocity increases the hysteresis loop thickness approaches a constant value, that is, it becomes independent of angular speed. The solid line represents the model of the form $D(\omega) = \omega^n$, $n = 0.12$.

Figure 2.6) Measured instantaneous elastic moment as a function of joint angle at 1.29 rad/s, the highest speed used in these tests. The measurement at this speed is used as an estimate of the instantaneous elastic moment. Measurements from flexion to extension are represented by (Δ), and those from extension to flexion by ($\square$). The solid line represents the modeled quadratic estimation of the data given by Equation (7).

Figure 2.7) Measured passive moment (---) and corresponding model results (solid line) computed from Equation (2.2). The data and model are shown for the region where the velocity was constant. a) Velocity 0.85 rad/sec, subject JM. b) Velocity 0.85 rad/sec, subject MT. c) Velocity 0.05 rad/sec, subject JM.

Figure 2.8) Passive moment as a function of joint angle with the wrist fixed at sixty seven degrees of flexion (upper curve) and sixty seven degrees of extension (lower curve). The shift in the passive moment curve is consistent with the stretching of extrinsic flexor and extensor muscles. With the wrist flexed, the extrinsic extensor muscles are lengthened and tensioned resulting in a higher passive moment in flexion at the metacarpophalangeal joint. The opposite is true when the wrist is extended.

Figure 2.9) Passive moment hysteresis loop at MCP joint of a normal subject (JM) and a quadriplegic person (KT). The horizontal arm of the apparatus could not be rotated more than 105 degrees of flexion, therefore, for subject (KT), the data is not recorded beyond that point.

CHAPTER THREE

THE HAND MODEL

A. Introduction

A biomechanical model of hand function is developed to assist the implementation of Functional Neuromuscular Stimulation (FNS) systems. This model can be directly utilized to simulate non-prehensile movements of the hand, or can be complemented by models of interaction of the hand with different objects to simulate prehensile grasp functions. A dynamic quantitative simulation of the hand function may be quite useful in predicting potential grasp performance, in situations typical of those encountered in FNS. In addition, simulation can be used to study the interaction of objects with the hand with both open and closed loop implementations of FNS. Although the simulation can not fully replace subject testing, the dynamic model can be used in place of experimental tests on subjects to make initial evaluations of muscle function and a

range of functional improvements that could be obtained in a comprehensive FNS program. It can be utilized to quickly identify rehabilitative procedures that are most likely to result in functional improvement, thereby narrowing the number of procedures tested in clinical trials.

A static model of hand function was developed by Mansour *et al.* (1994) and Hendrix and Mansour (1993) to explore the potential for using the intrinsic muscles in hand grasp, and to quantitatively investigate the relative roles of the intrinsic and extrinsic muscles. The model included the index and long fingers and the thumb. The metacarpophalangeal (MCP) and carpometacarpal (CMC) articulations were modeled as universal joints and each of the more distal articulations was modeled as a hinge joint. Position of the finger and thumb was actively controlled by fifteen muscles. The intrinsic muscles were included in the model of hand function since they were used in FNS systems, and have the potential for stimulation and possible improvement of hand function. Finger position was predicted from the equilibrium of moments of muscle and external forces and passive moments at joints. The last version of the static model completed by Hendrix (1993) showed convergence problems in the program's numerical solution routine. Addition of any more complexity to the model could easily worsen the convergence problem and disable the simulation.

Studies on kinetics of hand movements have revealed that dynamic torques associated with finger movements are large in relation to joint torques produced by

muscles, especially in faster movements (Cole and Abbs, 1986; Darling and Cole, 1990). Electromyographic measurements (EMG) and torque patterns associated with finger movements at different speeds indicate that muscle activity is necessary not only for producing motion at the joints but also to counteract dynamic inertial torques (Darling and Cole, 1990). In a dynamic model of hand movements, appropriate velocity dependent terms and inertial terms have to be taken under consideration.

In this part of the work, the model of the hand generated using ADAMS is presented. An overall output/input sensitivity analysis (joint position to individual muscle activation pattern) is then performed to be used in later chapters to select appropriate muscles for specific tasks. In subsequent chapters, the hand model is complemented by optimization and interaction routines to simulate some of important daily functions of the hand.

B. ADAMS Review

ADAMS (Automatic Dynamic Analysis of Mechanical Systems, Mechanical Dynamics Inc., Ann Arbor, MI) is a computer software package used in the field of Multi-body System Analysis to simulate motion of such mechanical systems. Multi-body system analysis is characterized by large displacements and gross motion of the system; i.e., the extent of the relative motion of the components can be on the order of

the overall dimensions of the system. This mechanical system consists of rigid and/or flexible parts interconnected by various types of joints and other possible constraining forces. The net reaction and applied forces drive the relative motion of the parts.

In an ADAMS simulation, the locations and orientations of the parts in a mechanical system are fully described in terms of six coordinates; three translational and three angular. The time-dependent translational and angular displacements, velocities, and accelerations are stored in the state vector that also contains the current values of the reaction and applied forces at all of the constrained and inertial locations on each of the parts. Thus, the state vector provides a complete description of the mechanical system for a multi-body system analysis (An Overview of How to Use ADAMS, Mechanical Dynamics Inc., Ann Arbor, MI, 1991).

The field of multi-body analysis is based on the laws of physics. Newton's laws of motion, or an equivalent formulation of the physical principles, are implemented in a multi-body system analysis tool. The formulation chosen for the ADAMS code is the system of Euler-Lagrange equations of motion. This system is composed of algebraic and second order differential equations that require sophisticated numerical methods for their solution.

The system of equations governing the behavior of any mechanism in ADAMS consists of:

- ◆ Six first-order dynamic equations for each part relating forces and moments to translational and rotational accelerations.
- ◆ Six first-order kinematic equations for each part relating positions to velocities.
- ◆ A single algebraic constraint for each scalar, holonomic, motion constraint.
- ◆ A single differential equation for each scalar, non-holonomic, motion constraint.
- ◆ A single algebraic equation for each scalar force component.
- ◆ Any number of user-defined algebraic or first-order differential equations.

ADAMS formulates all of the equations on the above list and assembles them into a single system to be solved simultaneously for the time-dependent values of the entries in the state vector. Equations in the system can be written in general form of three blocks of equations:

$$\mathbf{M}(\bar{\mathbf{q}}, \bar{\mathbf{u}}, \bar{\ddot{\mathbf{u}}}, \bar{\mathbf{f}}, t) = \mathbf{0} \quad (3.1)$$

$$\bar{\mathbf{u}} - \dot{\bar{\mathbf{q}}} = \mathbf{0} \quad (3.2)$$

$$\mathbf{N}(\bar{\mathbf{q}}, \bar{\mathbf{f}}, t) = \mathbf{0} \quad (3.3)$$

where the elements of the column matrix $\mathbf{q}$ are the generalized coordinates containing the translational and rotational variables. The column matrix $\mathbf{u}$ contains the time derivatives of the components of $\mathbf{q}$ as indicated by the second block of equations

which are the kinematic differential equations. The dot denotes the time derivative. The constraint forces and applied forces comprise the column matrix $\mathbf{f}$, and t is time. The column matrix $\mathbf{M}$ represents all of the differential equations governing the dynamics of the model plus any user-defined differential equations. The column matrix $\mathbf{N}$ includes all of the algebraic equations that make up the constraints for the model.

The ADAMS code can perform six different analyses depending on the characteristics of the problem and the interests of the user:

1. Initial condition analysis
2. Kinematic analysis
3. Static equilibrium analysis
4. Quasi-static analysis
5. Dynamic analysis
6. Linear analysis

"Initial condition analysis" (the assembly process) checks for the consistency of the initial set up of the system before a simulation can begin. "Kinematic analysis" is used when the overall system has no independent degree of freedom (ADAMS' manual defines the total number of degrees of freedom of a system as "the number of independent motions that characterize the system"). "Static analysis" looks for the steady state equilibrium solution by equating all rate dependent terms to zero. In this case, differential equations of motion are altered into a set of algebraic equations. For

"dynamic analysis" of multiple degree of freedom systems, ADAMS uses numerical integrations. The differential equations representing the system are numerically integrated to provide positions, velocities, accelerations, and forces acting on the system at successively later times. The integration algorithms invoked by the three ADAMS optional integrators are all modified implementations of the Backward Differentiation Formula method for the solution of systems of differential and algebraic equations.

The integrator GSTIFF which is used in dynamic analysis of the present study uses the predictor corrector integration method (Gear, 1971). Each integration step consists of two phases, predictor and corrector. During the predictor phase, previous values of each variable are used to predict the value of that variable at a new time, by approximating the solution with polynomials in time (extrapolation). The predicted values for the variables are then used to solve the governing equations, by a modified Newton-Raphson algorithm. This iterative solution, the corrector, uses the Jacobian matrix which is a sparse matrix of partial derivatives. After the corrector converges to a solution, the integrator estimates the amount of error in the solution based on the difference between the predicted and the corrected values (ADAMS reference manual). If the estimated error is greater than that specified in the program by the user, the integrator rejects the solution and takes a smaller time step. If the estimated error is less, the integrator accepts the solution and takes a new time step. The integrator repeats this process until it reaches the simulation end time specified in the simulation

command. The integrator controls both the order of integration, the order of the polynomials used in the solution, and the size of the time step to keep the error in each time step below that specified in the program.

C. Development of The Model

In general, the hand is modeled as a multi-body system acted upon by moments at the joints. Each segment of the multi-body system (a phalanx or palm) is specified in an ADAMS main routine as a rigid "part" using the geometric and inertial data of the phalanges and the palm (figure 3.1). The joints are modeled as kinematic constraints between the adjacent segments using ADAMS' "joint" statements (for instance, "revolute" which allows only one degree of rotational freedom and constrains the other degrees of freedom). The passive moments are directly computed within the main routine and are applied at corresponding joints as moment constraints. Muscle and external forces can be reduced to forces at and moments about the joints. The resultant forces are actually action-reaction workless constraint forces which have no effect on our computations, and are not included in the model. Therefore, only the resultant moments are applied about the corresponding joints.

Moments of muscle and external forces, which are applied at the joints, are computed in separate user-written subroutines and are linked to the ADAMS main routine. The muscle subroutine computes dynamic muscle moments based on the

activation level of the muscles coming from the input file, the state variables of the model (joint angles and joint velocities) coming from the main routine, and individual muscle characteristics given in the subroutine. In prehensile applications, the interaction subroutines calculate the moments of the external forces according to the intersection kinematics of the phalanges and the grasped object, and the state variables. The moments of the muscle and external forces are returned to the main routine and are applied about the corresponding joints.

Figure 3.1) Schematic of the ADAMS hand model. The input to the model is a set of muscle activation levels, and the output is as requested (angular positions, velocities, accelerations, and moments of any predetermined "marker"). The interaction subroutine is active in prehensile applications.

The model of hand includes four major partitions: joint passive moments, multi-body system, muscle-tendon units, and interaction model. Due to the importance of the joint passive moments, a separate chapter (Chapter two) was assigned to the development of a mathematical model and measurements of the joint passive moments. The multi-body system and muscle-tendon unit models will be individually discussed, thereafter. In chapters 5 and 6, two distinct interaction models for lateral pinch and palmar grasp will be presented.

Multi-body System

In the hand model, each phalanx is modeled as a cylindrical rigid body connected to the adjacent segment(s). The palm is modeled as a thin box, connected proximal to the forearm (in our model, the ground), and distally to the index finger. The local reference coordinate of each segment is located at the center of rotation of the proximal joint, and is oriented such that the X direction is along the center line of the cylinder, and Z and Y directions are defined as the flexion-extension and adduction-abduction rotational axes, respectively (Figure 3.2). Flexion and adduction are defined as right handed rotation about Z and Y axes. Only the thumb and the index finger are included in the present model. The other fingers can easily be incorporated into the model since the general structure of the fingers is very much the same.

In summary, the hand which is modeled as a multi-body system includes eight rigid segments (including the ground) connected by seven joints with ten rotational degrees of freedom. The geometric dimensions are measured from Prof. Mansour's hand. The masses are computed based on the volume of the cylindrical model of each segment assuming specific mass of 1.1 Kg/cm^3 (Seireg and Arvikar, 1989). Principal moments of inertia are calculated from the geometry and mass of each segment. The geometric and inertial values of the segments used in the model are summarized in Table (3.1).

Figure 3.2) The local reference coordinate of each part is located at the center of rotation of the proximal joint, and is oriented such that the X direction is along the center line of the cylinder, and Z and Y directions are defined as the flexion-extension and adduction-abduction rotational axes, respectively. Flexion and adduction are defined as right handed rotation about Z and Y axes, respectively.

Model Element	Length (cm)	Radius (cm)	Mass (Kg)	Moment of Inertia (Kg.cm ²)	
				I_{xx}	$I_{yy} = I_{zz}$
MC1	7.2	1.5	0.05	0.05625	0.892125
TPP	3.9	1.1	0.015	0.009075	0.080588
TDP	2.0	1.1	0.009	0.005445	0.011723
Palm	box: 8×6×3		0.36	2.67	4.75 , 2.56
IPP	5.0	1.3	0.015	0.012675	0.131338
IMP	2.7	1.1	0.009	0.005445	0.024593
IDP	2.4	1.0	0.006	0.003	0.01302

Table 3.1) Geometric and inertia parameters of the model element. MC1 denotes the first metacarpal bone. Thumb consists of two phalanges, proximal (TPP) and distal (TDP), whereas the index finger includes three phalanges, proximal (IPP), middle (IMP), and distal (IDP).

The skeleton of the hand consists of three major segments: the carpus or the wrist bones; the metacarpus or palmar bones; and the phalanges or bones of the fingers. There are eight carpal bones which are arranged in two rows. There are five metacarpal bones in the palm which articulate at their proximal extremities with the carpal bones and at their distal ends with the proximal phalanges. Thumb consists of two phalanges, proximal and distal, whereas the index finger includes three phalanges, proximal, middle, and distal.

The structure of the wrist articulation is quite complex; however, the wrist altogether acts as a two degree of freedom universal joint with flexion-extension and radio-ulnar deviation movements (Table 3.2). The two perpendicular axes of rotation are believed to intersect (Brumbaugh *et al.*, 1982) or to be very close (Youm *et al.*, 1978; Andrews and Youm, 1979).

The three joints of the thumb provide five degrees of freedom and a wide range of motion for the thumb. The distal joint, often called interphalangeal (IP), has flexion-extension movement providing one degree of freedom. The proximal joint, metacarpophalangeal (MP), has flexion-extension and abduction-adduction movements, or two degrees of freedom. The basal joint, carpometacarpal (CMC) or trapeziometacarpal, has flexion-extension and abduction-adduction motion (two degree of freedom) as well as a small, limited amount of axial rotation. Based on a study by Cooney *et al.* (1981), axial rotation of the carpometacarpal (CMC) joint can occur passively when the ligaments are lax and the joints' articular surfaces are distracted, but during active motion and function, such as pinch and grasp, muscle forces and the compressive force across this joint constrain rotation such that axial rotation is not considered as an independent degree of freedom of the thumb. Motions of the carpometacarpal (CMC) joint can be described based on the saddle-shaped articular surfaces of the trapezium and the first metacarpal bone. Flexion/extension, therefore, is defined as the movement in the plane perpendicular to the convex surface of the central ridge of the articular surface of the trapezium. Abduction/adduction is defined

as the movement perpendicular to the flexion-extension plane and parallel to the concave-convex articular surface of the central ridge.

In ADAMS, it is not possible to implement one degree of freedom of a part as a function of the other degrees of freedom of the same part. Due to this technical difficulty in implementation of the dependent rotation of the thumb as a function of its flexion, and the fact that this rotation is very small in the practical range of motion of the thumb during pinch and grasp, the carpometacarpal (CMC) articulation is modeled as a two degree of freedom universal joint (Table 3.2). Likewise, a one degree of freedom revolute joint is chosen to represent the metacarpophalangeal (MP) as well as the interphalangeal (IP) articulations (Table 3.2).

Model Element	Joint	Mechanical Equivalent	Degrees of Freedom
Wrist	Wrist	Universal Joint	2
Thumb	CMC	Universal Joint	2
	MP	Revolute Joint	1
	IP	Revolute Joint	1
Index Finger	MCP	Universal Joint	2
	PIP	Revolute Joint	1
	DIP	Revolute Joint	1

Table 3.2) Wrist, thumb, and index finger joints characteristics used in the multi-body system model.

The movements of the finger joints are much simpler than those of the thumb. The metacarpophalangeal (MCP) articulation of the index finger is represented by a two degree of freedom universal joint, whereas, the two interphalangeal articulations, proximal (PIP) and distal (DIP), are modeled as one degree of freedom revolute joints (Table 3.2).

Muscle-tendon unit

Muscles in the hand are like the actuators in robotics arm. They produce the force required to position different parts of the hand and to generate force to accomplish a specific task. The literature on muscle mechanics is so vast that a detailed survey is far beyond the scope of the present study. In this section we will review some of the major developments that lead to present day models, discuss some of the selected models, and finally present the muscle-tendon unit model utilized in the present model of hand.

One of the early attempts at the quantitative description of muscle mechanics was made by Hill (1938). He observed that the velocity of shortening of active muscle is a function of the force it generates. He also observed that after a single stimulus the muscle becomes fully active for a more or less fixed period of time. Based on these observations, he proposed a model of the mechanical activities of muscle consisting of

two components; one active component which he called the contractile element and one passive elastic component in series with the active component. Later Hill (1953) proposed a relationship now famous as force-velocity relationship. This relationship is used mainly to characterize the dynamic behavior of the contractile element. The necessity for a series elastic element is inferred from data such as the Levin and Wyman experiment (1927), which shows that there is an instantaneous length change in the muscle for sudden changes in the applied load.

The active state, the muscle force when it is stimulated, in Hill's force-velocity relationship was assumed to be a constant function of time, and is identified with the maximum isometric force of the muscle. This quantity was obtained by fully stimulating the muscle while it is at its optimum length, which is close to the rest length. Thus, at a different level of stimulation and at a different length, the muscle is expected to have a different active state. This idea was realized by Ritchie and Wilkie (1958) who extended Hill's model by modulating the active state with a function of the length of the muscle and a function of time. The force-length relationship was obtained experimentally for the contraction phase of activity and depicted graphically.

A modeling procedure that was based on mammalian rather than amphibian muscle was that of Bahler (1968). He proposed a model of the contractile element in which the output of the element is a function of the length, velocity of contraction and duration of stimulation. He based his model on Hill's two element model. In addition,

he modeled the series elastic element by a cubic function of the force and the length-tension relationship by a quadratic function of the length of the contractile element. A step function of time and stimulus frequency was used to model the stimulation-time relationship. Bahler's model comes close to describing the excitation dynamics of the muscle in addition to the contraction dynamics as defined by the force-velocity and the force-length relationship. However, this excitation dynamics has no physiological basis.

In all the models discussed thus far, the force-velocity relationships are only useful for the shortening phase of contraction. For the lengthening phase, the models break down. This is because the hyperbola used to describe Hill's force-velocity relationship increases continuously for positive (lengthening) contractions, whereas experimental evidence has shown that muscle exhibits yielding for such velocities (Sugi, 1972). By muscle yielding, it is meant that at positive contraction velocities the length of the muscle increases for no change in the force produced.

The question of excitation dynamics requires an understanding of the underlying mechanism of the excitation-contraction coupling. The fact that presence of calcium in muscle tissue is intimately connected with the excitation-contraction coupling was used by Taylor (1969) in modeling the excitation dynamics of muscle. He proposed that the magnitude of the active force is proportional to the concentration of calcium. He went further to account for relaxation in his model by

postulating the presence of a carrier pump mechanism for removing calcium from the sarcoplasm of the muscle fiber. Based on the chemical kinetics of the bound calcium and that of the carrier pump, he derived a set of three differential equations; and along with Hill's force-velocity relationship, he defined the complete dynamics of the contractile element using four first order differential equations. Computer simulation of the model was found to predict muscle behavior (quick release, re-development of tension etc.) with a good degree of precision.

Hatze (1976), in his study of kicking motions, proposed a model whose excitation dynamics closely follows that of Taylor. He also defined the active state in his model as the amount of calcium bound to troponin (a muscle molecule which is associated with the phenomenon of contraction and force production). In addition, the model was that of a muscle fiber, and that the recruitment of fibers in a given muscle constitutes part of the excitation dynamics of the muscle. In his model, he also modified Hill's force-velocity relationship to include the effect of muscle yielding during lengthening. This was done by using the hyperbolic tangent function rather than the hyperbola of Hill's equation. Hatze's muscle model is made up of three first order differential equations that define the excitation and contraction dynamics. The earlier model was then modified by Hatze (1980) to include the phenomenon of recruitment in further detail. This resulted in a muscle model defined by five first order differential equations. Contrary to the others who validated their models on the intrinsic properties

alone, he validated his model on a gross musculo-skeletal motion (a minimum time kicking problem).

The design of functional neuro-muscular stimulation systems requires a computer model that has the ability to activate many muscles to effect functional control of paralyzed limbs. Critical to this task is a model of each muscle and tendon that will be stimulated. Zajac *et al.* (1986) developed a dimensionless, first order model of musculo-tendon (actuator) contraction dynamics that is based on fundamental architecture and properties of muscle and tendon. Only three parameters: muscle strength, optimal muscle fiber length and tendon slack length are needed to scale the generic representation to a specific actuator. For the lumped-parameter mechanical model for tendon and muscle, he assumed that tendon is a conservative spring with variable stiffness, and that passive muscle develops force as would a spring with variable stiffness, and that muscle, once activated, develops active force from its contractile properties. He further assumed that the properties of active muscle are given by two elements: a contractile element described by a force-velocity relationship dependent on activation and muscle length, and a series elastic element to account for muscle's short-range stiffness. The contractile mechanism of muscle is assumed, in steady-state, to develop isometric force that depends on its length and its activation. Actually, it is through isometric force that the force-velocity relationship is assumed to depend on activation and length.

Simple second order linear models (single input single output) fit experimental data quite well if the experimental conditions are constrained (Bawa *et al.*, 1976). The underlying non-linear behavior of muscle is revealed only when the conditions are changed (isometric length, stimulus rate, etc.). Thus, linear second order models do not have sufficient generality to capture muscle behavior under a wide range of operating conditions. Hill-based muscle models attempt to describe muscle behavior under general conditions by incorporating separate activation, length-tension, force-velocity, and series elasticity terms. Each term is assumed to be independent of the others and thus they are measured separately. In spite of their popularity and wide spread use, Hill-type models have not been tested quantitatively as extensively as the simple second order models during simultaneously changing length, velocity and activation.

Two types of input/output models are introduced by Crago (1992) to capture muscle behavior under a wide range of conditions. These two types of models are evaluated for their ability to fit experimental data and their sensitivity to operating conditions. In one, muscle output is calculated as the product of three independent factors: activation dynamics, torque-angle, and torque-velocity properties, as is typically done in Hill-based models. In the second model, activation is coupled to velocity. Both models can predict muscle output (joint torque) well, but parameters for the uncoupled model are found to be quite sensitive to operating conditions. The Hill-type uncoupled model lacks generality. Introducing the activation-velocity

coupling substantially improves the generality. The parameters of these two models must be directly measured for the human muscles, prior to their application in a model.

The simple linear Hill-type model implemented by Chiel *et al.* (1992) in a model of lapping by a reptilian tongue is selected to model the muscle behavior. In this model, active muscle force is a product of activation dynamics, length-tension and force-velocity properties. The passive force properties of the muscles are included in the measurements of the joint passive moments. Parameters of the selected muscles (activation time constant, length-tension-firing period, and force-velocity properties) are estimated based on the parameters measured on cat's soleus (Crago 1992).

The muscle activation dynamics is described by a linear first order differential equation,

$$\dot{A} = p(\mu - A), \quad (3.4)$$

which relates the neural input μ and activation A . Here t is the time and $1/p$ is the time constant of isometric force development. A is equal to zero at t equal to zero, and is equal to μ as t approaches infinity. μ determines the level of activation to the stimulated muscle. The solution to the differential equation (Equation 3.4) for a ramp input, $\mu = at + b$, is

$$A(t) = at + \left(b + \frac{a}{p}\right)(1 - e^{-pt}). \quad (3.5)$$

The activation time constant is estimated to be 33 milliseconds that is typical for fast muscle (Bawa, 1976).

The isometric length-tension property of muscle is modeled as a linear spring. The isometric force is linearly increasing as a function of the difference between the muscle length L_m and the slack length L_{n_0} (i.e., the length at which muscle generates no force), with the normalized stiffness K_m at the maximal recruitment level:

$$F_{iso} = A(t)K_m(L_m - L_{n_0}). \quad (3.6)$$

Increasing the firing period (activation period), increases the stiffness and the slack length according to the following relationships:

$$K_m = bF_0T_F \quad (3.7)$$

$$L_{n_0} = L_{\max} - 1/(bT_F) - a/b. \quad (3.8)$$

Inserting the above two equations into Equation (3.6) we have

$$F_{iso} = A(t)F_0(1 + (a + b(L_m - L_{\max}))T_F) \quad (3.9)$$

where AF_0 is the estimated maximum force when the muscle is stimulated at an infinitely high frequency, $L_{\max}$ is the maximum muscle length *in situ*, a and b are constants that depend on muscle properties, and T_F is the firing period.

Weights of the muscles are neglected, because their effect is small compared to the maximal isometric strength of the muscles. The maximal isometric strength F_0 is estimated from the cross-sectional areas of the muscles, assuming a constant proportionality coefficient of 20 N/cm². Measured cross-sectional areas from two sources, An *et al.* (1983) and Lieber *et al.* (1991), are given for comparison although, due to availability of the data for all the muscles used in the model, measurements from An *et al.* are used (Table 3.3).

Parameters for the normalized length-tension-firing period relationship; i.e. α and b , are computed based on the measurements for cat soleus (Crago, 1992), but are adjusted for the correct physiological length range for each muscle in the model. These parameters and muscle geometric parameters (Sarangapani, 1989) are presented in Table (3.4).

Muscle	PCSA (cm ²)	
	An	Lieber
FDP	4.1 ± 2.4	1.77 ± 0.16
FDS	3.6 ± 2.1	1.71 ± 0.28
EI	N/A	0.56 ± 0.11
EDC	1.1 ± 0.3	0.52 ± 0.08
PII	1.4 ± 0.8	N/A
DII	1.4 ± 1.0	N/A
FPL	5.1 ± 2.6	2.08 ± 0.22
FPB	2.3 ± 0.6	N/A
EPL	1.9 ± 0.8	0.98 ± 0.13
EPB	1.3 ± 0.7	N/A
APL	3.9 ± 2.0	N/A
APB	1.5 ± 0.6	N/A
ADP_o	2.4 ± 1.2	N/A
ADP_t	0.9 ± 0.4	N/A
OPP	2.8 ± 1.3	N/A

Table 3.3) Physiological Cross Sectional Area (PCSA) of the selected muscles in hand model from An *et al.* (1983) and Lieber *et al.* (1991). An's measurements are used in the model. N/A denotes that the data is not available.

Muscle	Geometric Parameters		Length-Tension Parameters		Computed Forces	
	$L_{\max}$ (cm)	$L_{\min}$ (cm)	$a \times 10^{-4}$ (1/sec)	$b \times 10^{-3}$ (1/cm.sec)	F_0 (N)	F_{iso} (N)
FDP	46.2	37.8	1.0	0.895	82	57.9
FDS	47.0	39.9	"	1.054	72	60.9
EI	33.0	27.4	"	1.342	22	13.7
EDC	48.5	40.9	"	0.984	22	13.8
PII	7.7	5.7	"	3.75	28	19.2
DI1p	10.0	9.0	"	7.5	28	19.5
FPL	40.1	34.8	"	1.41	102	81.9
FPB	6.8	6.5	"	24.19	46	27.7
EPL	29.7	23.9	"	1.28	38	22.7
EPB	19.6	15.5	"	1.84	26	19.9
APL	18.0	16.0	"	3.750	78	65.5
APB	8.1	7.8	"	25.0	30	17.3
ADPo	8.0	5.0	"	3.0	48	45.2
ADPt	7.5	4.0	"	3.0	18	17.1
DI1t	8.5	5.0	"	2.14	42	40.2
OPP	3.0	2.4	"	15.0	56	29.0

Table 3.4) Parameters and computed forces of the selected muscles in the model with all joints at zero degrees.

The firing period of the motor neurons is assumed constant ($T_f = 80$ ms), but activation, $A(t)$, is varied as a function of time (i.e., recruitment modulation).

The force-velocity relationship is modeled as a simple linear damping element scaling the isometric force;

$$F_m = F_{iso}(1 + C\dot{L}_m), \quad (3.10)$$

where C (sec/cm) is the linear damping coefficient, and is estimated from the following relationship

$$C = \frac{1}{10(L_{\max} - L_{\min})}, \quad (3.11)$$

for all the muscles used in the model (Chiel *et al.*, 1992).

The compliance of a tendon alters the muscle length for a given set of joint positions. This would therefore change the force experienced by the muscle (Equation 3.10). The stiffness of tendon K_t is computed from the tendon area A (Blanton and Biggs, 1970), tendon length L_t (Brand, 1985; Lieber, 1990), and constant modulus of elasticity E_t (Benedict *et al.*, 1968) as following:

$$K_t = \frac{A_t E_t}{L_t}. \quad (3.12)$$

The force developed by the tendon F_t is

$$F_t = K_t(L_t - L_{t_0}), \quad (3.13)$$

where L_{t_0} is the slack length of the tendon. The tendon length is

$$L_t = L - L_m, \quad (3.14)$$

where L is the muscle-tendon unit length. L is in turn related to the joint angles Θ_i , which the tendon crosses:

$$L = L_0 + \sum_{i=1}^n d_i \cdot \Theta_i . \quad (3.15)$$

d_i 's are the constants based on the slopes of the tendon excursion curves, L_0 is the muscle-tendon unit length when all joints are at zero degrees of flexion, and n is the number of joints the tendon crosses. Inserting Equations (3.14) and (3.15) into Equation (3.13), the tendon force is

$$F_t = K_t [L_0 + (\sum_{i=1}^n d_i \cdot \Theta_i) - L_m - L_{t_0}] . \quad (3.16)$$

The muscle force and the tendon force must be equal, by equating Equations (3.6) and (3.13) we have

$$AK_m(L_m - L_{m_0})(1 + C\dot{L}_m) = K_t [L_0 + (\sum_{i=1}^n d_i \cdot \Theta_i) - L_m - L_{t_0}] . \quad (3.17)$$

Thus, the muscle length is

$$L_m = \frac{K_t [L_0 + (\sum_{i=1}^n d_i \cdot \Theta_i) - L_{t_0}] + AK_m L_{m_0} (1 + C\dot{L}_m)}{AK_m (1 + C\dot{L}_m) + K_t} . \quad (3.18)$$

By inserting Equation (3.18) into Equation (3.6) and simple mathematical manipulations, the total force exerted by the muscle-tendon unit, F_m , is determined by the following equation:

$$F_m = K(l + \sum_{i=1}^n d_i \cdot \Theta_i), \quad (3.19)$$

where

$$K = \frac{AK_m K_t}{AK_m(1 + CL_m) + K_t}, \quad (3.19a)$$

and

$$l = L_0 - (L_{m_0} + L_{t_0}). \quad (3.19b)$$

The force produced by the muscles apply moment about the joints by means of tendons. The distance from the center of the tendon to the center of the joint is called "moment arm" of the tendon. In most of the extrinsic and some of the intrinsic hand muscles, the moment arm is almost constant in the practical range of movement of the hand (Landsmeer, 1961; Armstrong and Chaffin, 1978; An *et al.*, 1983). In such cases the moment M produced by a muscle about a joint is

$$M = F_m \times r \quad (3.20)$$

where F_m is the muscle force and r is the constant moment arm.

The assumption that the moment arm is almost constant is not valid for some of the intrinsic muscles, specially the intrinsic thumb muscles, where the moment arms change drastically as the thumb moves. For such muscles, the "line-of-action" of the muscle is specified instead of its moment arm. Using this approach, the muscle force acts along a straight line between the origin and the insertion points of the muscle.

The following muscles are included in the model using constant moment arm approach: flexor pollicis longus (FPL), flexor pollicis brevis (FPB), extensor pollicis longus (EPL), extensor pollicis brevis (EPB), abductor pollicis longus (APL), abductor pollicis brevis (APB), flexor digitorum profundus (FDP), flexor digitorum superficialis (FDS), extensor indicis (EI), extensor digitorum communis (EDC), first palmar interosseous (PII), and the parallel head of dorsal interosseous ($DI1_p$). The line-of-action approach is used for the rest of the muscles: oblique head of adductor pollicis (ADP_o), transverse head of adductor pollicis (ADP_t), transverse head of dorsal interosseous ($DI1_t$) and opponens (OPP). The constant moment arms are obtained from An *et al.* (1983, Table 3.5) and the insertion and origin points from Seireg and Arvikar (1989).

The model for some of the above muscles may need more clarification. For the adductor pollicis, two distinct heads are modeled: an oblique head (ADP_o), which arises from the capitate bone and the bases of the second and third metacarpal bones, and a transverse head (ADP_t), which originates from the palmar surface of the third

Muscle	Thumb Moment Arms (cm)			
	CMC (flx)	CMC (add)	MP (flx)	IP (flx)
FPL	0.83	0.85	1.0	0.98
FPB	1.87	0.78	1.0	0
EPL	-1.37	0.25	-1.07	-0.65
EPB	-0.84	-0.95	-1.03	0
APL	0.47	-1.11	0	0
APB	1.46	-0.50	1.32	0

Muscle	Index Finger Moment Arms (cm)			
	MCP (flx)	MCP (add)	PIP (flx)	DIP (flx)
FDP	1.11	0.11	0.79	0.41
FDS	1.19	0.17	0.62	0
EI	-0.90	0.13	-0.27	-0.21
EDC	-0.86	-0.02	-0.27	-0.21
PII	0.66	0.58	-0.26	-0.16
DII_p	0.37	-0.61	0	0

Table 3.5) Tendon moment arms of the thumb, and index finger at different joints from An *et al.* (1983) and Ou (1980). Flexion and adduction are the positive directions.

metacarpal bone. Both heads converge toward the thumb to insert to the ulnar side of the base of the proximal phalanx. Two heads of the first dorsal interosseous, which insert into the base of the index proximal phalanx and the radial band of the extensor mechanism of the index finger, are also modeled. The transverse head (DII_t)

originates from the first metacarpal bone and acts on both the index finger and the thumb, and the parallel head (DII_p) arises from the second metacarpal bone. The two heads of the adductor pollicis, ADP_o and ADP_t , are stimulated independently, but the first dorsal interosseous (DII) is activated as one muscle. The opponens pollicis (OPP) is a small, triangular muscle placed beneath the abductor pollicis brevis (APB). It arises from the trapezium and from the flexor retinaculum and inserts into the radial side of the metacarpal bone of the thumb.

It was previously shown that the isometric force a muscle produces is a function of the angle of the joints the tendon crosses, using the slopes of the tendon excursion curves (Equation 3.19). Excursions of index finger muscles about finger joints are taken from measurements by An *et al.* (1983, Table 3.6).

Muscle	Excursions of index finger tendons (cm) per one radian of joint rotation				
	Wrist (flx)	MCP (flx)	MCP (add)	PIP (flx)	DIP (flx)
FDP	-3.015	-1.088	-0.167	-0.806	-0.418
FDS	-3.497	-1.206	-0.229	-0.620	0
EI	N/A	0.864	-0.199	0.308	0.169
EDC	N/A	0.800	0.192	0.320	0.174
PII	N/A	-0.451	-0.583	0.271	0.137
DII	N/A	-0.337	0.657	0	0

Table 3.6) Excursions of index finger tendons from Armstrong and Chaffin (1978) over wrist joint, and An *et al.* (1983) over index finger joints. N/A denotes that the data is not available.

Excursions of thumb tendons are not available in the literature and therefore, are estimated as the product of thumb moment arm and the joint angle. The total excursion of a tendon is the sum of the excursions of that tendon at the crossing joints. All of the hand extrinsic muscles cross the wrist, thus the wrist angle also influences the forces in them. Although our hand model does not include the wrist muscles, the wrist joint is incorporated in the model, in case the hand performance is to be studied at some specific wrist angle. Values for the slopes of the tendon excursion curves (d) for muscles crossing wrist joint are available only for FDP and FDS muscles (Armstrong and Chaffin, 1978; Table 3.6).

The hand model is used to perform a joint-position/muscle-activation sensitivity analysis. Individual muscles in the hand model implemented in ADAMS are activated in ten steps from zero to a hundred percent of activation level and the resulting joint positions are determined (static analysis). Then, the joint angle due to the activation of each muscle is plotted against activation level of that muscle. In the static simulations, the maximum relative displacement error is determined as 0.01 radian and the maximum moment imbalance is specified as 0.01 N-cm.

To show the importance of the effect of dynamic torques on the hand movement, and therefore on the muscle stimulation pattern, a dynamic simulation is also performed. The index finger joints are positioned in zero angle, FDS activation is ramped up to twenty percent in one tenth of a second and is held at that level (a

protocol similar to the study performed by Darling and Cole, 1990 and 1994). The MCP joint angle is then plotted versus time. In the dynamic simulation, the maximum integration error in each step is set to 0.001.

The index finger model is used to mimic the experimental protocol used in the previous chapter on passive moment measurements at the MCP joint. The IP joints in the model are fused, to simulate joint immobilization in the test protocol. The MCP joint is then rotated using measured data as the input. A cubic spline interpolation function is used to approximate the data points measured during the experiment. The passive moment at the MCP joint computed by the model, presented in the previous chapter, is then plotted against joint position to compare results of the model with the experimental data, especially at the extremes where velocity of the joint is not constant.

D. Results

The output/input sensitivity analysis shows that FDS and FDP are the most powerful flexors at MCP joint among the four potential flexors: FDS, FDP, PII and DI1 (Figure 3.3). Between the two intrinsic muscles, PII and DI1, the former reveals more flexion effect than the latter. The two extensor muscles (EI and EDC) have almost the same extension effect. Among all the finger muscles implemented in the hand model, only

DI1 produces significant abduction at MCP joint (Figure 3.4). The rest of the muscles adduct the joint for only a couple of degrees due to the resistance of the middle finger.

At the PIP joint, only FDP and FDS produce flexion (Figure 3.5). DI1 has no effect, while EI, EDC and PI1 show almost similar linear but only small extension effect. FDS and DI1 do not cross the DIP joint, and thus have no effect at that joint (Figure 3.6). FDP flexes and the rest of the muscles extend this joint.

Among the six extrinsic thumb muscles, the two flexors (FPL and FPB), and the two abductors (APL and APB) flex CMC joint, FPL being the most and APB being the less effective muscles (Figure 3.7a). Between the two extensors, EPL is the most effective muscle, capable of extending the joint to about 17 degrees. None of the intrinsic thumb muscles, except ADP_O which flexes CMC joint about 10 degrees, seems to have significant flexion/extension effect at CMC joint (Figure 3.7b). Sensitivity analysis of adduction/abduction at CMC shows that APL, EPB, and APB have respectively the most abduction effects; whereas, EPL, FPB, FPL, ADP, and PI1 act as adductors (Figure 3.8). All extrinsic muscles, except APL, and one intrinsic muscle, ADP, cross the MP joint. Sensitivity analysis reveals that FPL, ADP_O, FPB and APB are the strongest flexors of the MP joint (Figure 3.9). Extension of the joint is fulfilled by APL, and EPB. At the IP joint, only FPL and APL act, the former being a strong flexor, and the latter being an extensor (Figure 3.10).

Two different Dynamic simulations were performed to compare the results of the model with the experimental measurements available in the literature. In the first dynamic simulation, where all the index finger joints were initially at zero degrees and only FDS was activated, the MCP joint was first extended for about ten degrees, and then flexed to its final flexion angle after an overshoot of about ten degrees (Figure 3.11). The result of the second dynamic simulation, in which the IP joints of the index finger were fused and MCP joint was moved with a motion similar to the constant velocity test in Chapter Two, is in good agreement with the experimental results (Figure 3.12).

E. Discussion

The variation of some of the model parameters among subjects is relatively high. This variation of the parameters is more significant between the normal and abnormal joints to the extent that totally different muscle activation patterns may be required to achieve a certain posture. For instance, the rest position of the index finger joints of the quadriplegic person tested (KT) is much more flexed than that of the normal hand (JM). Therefore, to get to a certain posture, the flexor muscles of the normal hand might be activated while the extensor mechanism of the quadriplegic person with overly flexed joints is to be stimulated. However, some of the general results of simulation of the model of a specific hand may be used to draw general conclusions

applicable to all subjects. To better predict one's hand function, it is necessary to measure and/or estimate the parameters which have significant effect on the simulation results on that specific hand; especially the elastic component of joint passive moments, maximum strength of muscles, and hand geometry.

The maximum isometric forces generated by the FDP, FDS and PII muscles are in good agreement with the experimental measurements by Ketchum *et al.* (1978, Table 3.7).

Muscle	Max. Muscle Isometric Force (N)	
	Our Model	Ketchum <i>et al.</i>
FDP	57.9	60.6±25.9
FDS	60.9	67.8±23.2
Extensor	27.5	58.7±28.1
PII	19.2	27.2±11.5
DII	19.5	53.4±24.7

Table 3.7) Maximum isometric force of the index finger muscles predicted by our model and measured by Ketchum *et al.* (1978) on live subjects.

They use moment balance of external force and muscle forces about the joints of the index finger of live subjects to measure the individual and aggregate maximum muscle forces. The FDP and FDS maximum muscle forces are measured at DIP and PIP joints, respectively. The only flexor tendon crossing the DIP joint is FDP, and the only

flexor tendon crossing PIP, other than FDP, is FDS. Thus, the two muscles may be assumed isolated from the other muscles of the hand at these joints. The extensor mechanism and DII muscles in our model predict much less force than the measured values from Ketchum *et al.* Using the same method as described in the previous paragraph, Ketchum *et al.* also determines the maximum extensor muscle force at the DIP joint. Since PII and first lumbrical are connected to the extensor mechanism distal to the DIP joint, we cannot assume that the only muscle producing extension moment at DIP joint is the extensor mechanism. Therefore, the measured force may be an overestimate of the actual muscle force. Specific muscle forces are measured at MCP joint, where the intrinsic and extrinsic combine their influences. The isolation of muscles at this joint is performed by sequential denervation (nerve block) of the intrinsic muscles. Since no electromyograms are performed, it is not clear if the denervated muscles are fully inactive during the measurements. Thus, the maximum intrinsic muscle forces may also be overestimated. The physiological cross sectional area of the muscles presented in Table (3.3) supports the fact that Ketchum's measurements overestimate the maximum forces in extensor mechanism and DII.

A detailed analysis of sensitivity of the model output (joint positions in non-prehensile grasp and grasp forces in prehensile simulations) to the parameters of the model requires a great effort. More than two hundred parameters are used in the hand model. The model consists of seven joints with eight degrees of freedom. Thirteen parameters are required to model the passive moment at each degree of freedom of a

joint. For each one of the fifteen muscles implemented in the model, nine parameters are needed. The geometry and inertia of each part is determined by three parameters. However, from the formulation of the model, it is clear that some of the parameters have significant effect on the simulations while the others do not have noticeable influence on the results.

In later chapters, the hand model described in the previous sections will be directly used to simulate non-prehensile movements of hand, and will be complemented by models of interaction of hand with different objects to simulate prehensile grasp functions. Before then, the effectiveness of each muscle in producing a desired posture of hand is required. The results of the overall sensitivity analysis will be frequently used in selection of appropriate muscles in performing different tasks.

The results of output/input sensitivity analysis at index finger joints (Figures 3.2 to 3.4) reveals that even thirty percent activation of either of the extensor mechanism muscles can move the MCP joint into zero position. However, none of the extensors can solely neutralize PIP and DIP joints. This fact may be the reason for the little effect of FNS system in opening a quadriplegic patient's hand with claw posture where the interphalangeal joints are in extreme flexion angles. One possible approach that can resolve this problem is to increase the passive and/or active extension moment at the interphalangeal joints by means of tendon transfer.

The experimental measurement on dynamic motion of the index finger by Darling and Cole (1990) is in very good agreement with the result of our simulation following a similar protocol. It is shown that dynamic inertial torques have significant effect in hand movements and muscle activation pattern. When FDS is activated to flex MCP joint from its zero angle position, the joint initially extends (Figure 3.11). At the zero angle position, both passive moment and muscle moment produce joint flexion. However, the extension moment resulting from angular acceleration of the distal segments, which is at first larger than the flexion moment, initially extends the joint. Later, when the inertial moment reduces, the flexion moment overcomes the extension moment and flexes the joint. The inertial moment is also responsible for the overshoot of the joint flexion angle when the finger reaches its final position. Because the inertial torques are large in relation to the joint torque, the finger movements would be substantially altered if the inertial torques were not included.

The hysteresis loop of the passive moment at the MCP joint of the index finger predicted by the model is also in good agreement with the experimental measurements presented in the previous chapter (Figure 3.12). However, the curved corner of the ends of the hysteresis loop are not consistent with the sharp corners in our simulation. It could be due to the fact that, in the simulations, the velocity at the ends are estimated from the displacement-time data recorded during experiment. Or, it could be due to the fact that the model used in the simulation (Equation 2.2), which is a function of joint angle and joint speed, does not include relaxation over time and

predicts a much quicker transition from loading to the unloading curve of the passive moment.

Figure 3.3) MCP flexion/extension joint angle verses muscle activation level for the six index finger muscles used in the hand model. Positive angle is flexion and negative angle is extension.

Figure 3.4) MCP adduction/abduction joint angle verses muscle activation level for the six index finger muscles used in the hand model. Positive angle is adduction and negative angle is abduction.

Figure 3.5) PIP flexion/extension joint angle verses muscle activation level for the six index finger muscles used in the hand model. Positive angle is flexion and negative angle is extension.

Figure 3.6) DIP flexion/extension joint angle verses muscle activation level for the six index finger muscles used in the hand model. Positive angle is flexion and negative angle is extension.

Figure 3.7) CMC flexion/extension joint angle verses muscle activation level for a) the six extrinsic and b) the four intrinsic thumb muscles used in the hand model. Positive angle is flexion and negative angle is extension.

Figure 3.8) CMC adduction/abduction joint angle verses muscle activation level for a) the six extrinsic and b) the four intrinsic thumb muscles used in the hand model. Positive angle is adduction and negative angle is abduction.

Figure 3.9) MP flexion/extension joint angle verses muscle activation level for a) the six extrinsic and b) the four intrinsic thumb muscles used in the hand model. Positive angle is flexion and negative angle is extension.

Figure 3.10) IP flexion/extension joint angle verses muscle activation level for the six extrinsic thumb muscles used in the hand model. None of the intrinsic thumb muscles cross this joint. Positive angle is flexion and negative angle is extension.

Figure 3.11) MCP joint extends initially when FDS is activated to flex the joint from its zero angle position. The joint angle reaches its final value after a ten-degree-overshoot. FDS is activated up to twenty percent in one tenth of a second and the activation is held at that level.

Figure 3.12) The passive moment at the MCP joint of the index finger, measured ($\square$, Fig. 2.4) and predicted (solid line) by the hand model. The motion applied to the joint is approximated by a cubic spline function using the data points measured during the test. The data set includes joint angle as the dependent and the time as the independent variable. The speed is constant (0.85 rad/sec) between forty degrees of extension and seventy degrees of flexion. Positive angles represent flexion and negative angles are extension. The upper curve represents joint movement from extension to flexion, the lower from flexion to extension.

CHAPTER FOUR

NON-PREHENSILE PALMAR PINCH

A. Introduction

The major functions of the hand can be categorized as non-prehensile and prehensile movements (Napier, 1956). Non-prehensile movements are defined as functions in which no seizing is involved. The significant element in such a movement is the posture of hand or, in other words, the joint angles. In prehensile functions, on the other hand, both the posture of hand and the grasp force produced by the hand are to be taken under consideration. The prehensile grasp is in turn categorized as power and precision grip (Napier, 1956). In precision grip, where the objects are often light and the grasp forces are low, the first priority is to obtain an appropriate posture.

Palmar pinch is one of the most common and most useful types of precision grip. In palmar pinch, the interphalangeal joints of the index finger are extended and

the MCP joint is flexed with respect to the rest position of the joints (Figure 4.1). The thumb is often stabilized in a position against the index finger, and the object is pinched between the flexor aspects of the distal phalanges of the index finger and the opposing thumb.

Figure 4.1) Palmar pinch posture of the hand.

Electromyographic measurements have shown that both extrinsic and intrinsic muscles of the index finger are involved in a palmar pinch posture (Long, 1981). In functional neuromuscular stimulation (FNS) applications, however, implementation of the intrinsic muscles is technically difficult and increases the complexity of grasp

synthesis (Keith *et al.*, 1987). The palmar pinch posture might be achieved by stimulation of the extrinsic muscles alone, without or with the help of surgical alterations such as tendon transfer. The hand model, with normal or altered characteristics, can be used to investigate the possible patterns of muscle activation, including or excluding intrinsic muscles, throughout which the palmar pinch posture is obtained.

The hand model alone, introduced in the previous chapter, cannot directly predict the muscle activation pattern required to achieve a desired posture. It is indeed an open-loop model of the hand movements. Activation of selected muscles at certain levels (input) determines the joint positions (output), while the muscle activation patterns required to reach a desired posture is not known. In an implementation of a static model (Hendrix, 1993), tasks were performed by a manual trial and error approach for muscle selection and activation. This relatively inefficient use of the model had grown more cumbersome as complexity of the model increased. The number of possible combinations of muscle activation levels is very large when multiple muscles are used in a simulation. The hand model can be complemented by optimization techniques as a means for selecting the activation level of multiple muscles for a prescribed position of the hand. By controlled trial and error, optimization may automatically find the best activation pattern which provides the desired posture.

Tendon transfer techniques are utilized to restore the grasp abilities of the hand in the absence of intrinsic muscles. One common type of tendon transfer is known as the Zancolli-Lasso technique. In this approach, the FDS tendon is detached from the middle phalanx of the index finger and is attached to the palmar aspect of the proximal phalanx. In another technique, which we call the modified Zancolli, the FDS tendon, detached from its original insertion point, is wrapped around the proximal phalanx and is sutured to the extensor tendon in the dorsal aspect of the PIP joint (Keith, 1994). The former approach makes the FDS muscle a flexor of only the MCP joint, while the latter technique resembles the function of intrinsic muscles, flexion at MCP and extension at interphalangeal joints.

The hand model introduced in chapter three along with optimization techniques is used to investigate the possibility of achieving palmar pinch posture with activation of extrinsic muscles alone. Possible surgical alterations are also simulated to study the ability of normal and quadriplegic hands to reach a satisfactory palmar pinch posture by activation of the extrinsic muscles only.

B. Method

To search for the muscle activation pattern required to provide a desired posture, the hand model is first interfaced with the commercially available optimization software

packages, DOT and DOC (Design Optimization Tool and Design Optimization Control, VMA Engineering, CA). The interfaced program is then utilized to find the muscle activation pattern which minimizes the differences between the current and the desired joint positions in various conditions: normal and quadriplegic hands, including and excluding intrinsic muscles, and with and without tendon transfers. The optimization problem to be solved is then,

$$\text{Minimize: } Obj = f(\Theta_1(\mu_1, \dots, \mu_n), \dots, \Theta_m(\mu_1, \dots, \mu_n))$$

$$\text{Subject to: } 0 \leq \mu_1 \leq 1, \quad 0 \leq \mu_2 \leq 1, \quad \dots, \quad 0 \leq \mu_n \leq 1$$

where Θ_1 to Θ_m are the joint angles and μ_1 to μ_n are the muscle activation levels.

The DOT program uses numerical search methods to seek a minimum value of an objective function subject to constraints (DOT reference manual). The search algorithms are in two steps. The first is to determine a "search direction" which defines how the design variables will be changed. The key idea is that all design variables will be changed simultaneously in a fashion such that objective function moves toward a minimum. The second step is to determine how far to move in that direction, which is called a "one-dimensional search." This process of finding a search direction and then searching is repeated until it converges to the optimum. This is the basic approach used by the Modified Method of Feasible Directions contained in DOT for constrained minimization. If no constraints are imposed, the minimization is called unconstrained

and DOT uses the Broydon-Fletcher-Goldfarb-Shanno or Fletcher-Reeves algorithms. These algorithms also proceed in the two step process of finding a search direction and performing a one-dimensional search.

The DOC program assists with interfacing the DOT program to a computational model with a minimum of effort through a user-written FORTRAN routine, named ANALIZ (DOC reference manual). Several other options such as multi-objective optimization, parametric studies and approximate optimization methods are also available in DOC. In particular, DOC extends the nonlinear capabilities of DOT by using approximated responses. The approximation methods contained in DOC may be the most efficient optimization method if there are relatively few design variables (about five, DOC reference manual). With this method, at each iteration step, objective function(s) is(are) approximated by fitting a curve to the data points evaluated up to the current step. During the optimization process, the function values will come from the approximation functions that have been created. The approximation that is used in DOC is a least squares fit to a series of function values. The order of the approximation may be up to a full second-order Taylor series function. The order can be chosen to be one (linear approximation), two (linear terms plus the diagonal terms of the Hessian matrix), or three (linear terms associated with the entire Hessian matrix).

An algorithm to search for the required muscle activation patterns using approximate optimization method is designed. The hand model implemented in ADAMS is interfaced with DOT and DOC through a FORTRAN user-written routine, ANALIZ. The algorithm works as the following (Figure 4.2): In the first step, the hand model in ADAMS is called to compute the initial joint positions (design responses) related to the initial arbitrary muscle activation levels (design variables). In the next step, since the number of data points are not sufficient for approximation, DOC is called to perturb the initial muscle activations, once or twice based on the order of approximation, to provide sufficient data points for the initial approximation of the objective function. The objective function is then evaluated at the computed joint angles as

$$OBJ_i = \sum_{j=1}^m [T_j - \Theta_j(\mu_1, \dots, \mu_n)]^2, \quad (4.1)$$

where the T_j 's are the desired joint positions, the Θ_j 's are the current joint angles, m is the number of joints, n is the number of muscles activated, and i is the iteration number. Then, DOC approximates the objective function by fitting a curve to the evaluated points, and calls DOT to minimize the approximate objective function. At this point, ADAMS is called again to compute the next set of joint positions using the new set of muscle activations resulting from minimization. In the last step, ANALIZ evaluates the proposed optimum (Equation 4.1), and the results are added to a "basis",

a list of the values of the objective function. This process is repeated until it converges to a final optimum where the change in objective function is less than the acceptable tolerance determined by the user. The most recent activations are therefore the required muscle activation patterns.

Figure 4.2) Schematic of the interface between the hand model implemented in ADAMS and the optimization routines. Optimization routines control muscle activation levels to move the joints into the desired positions by minimizing the differences between the initial and the desired joint angles.

Initially, the hand model with all muscles was interfaced to the optimization routines. However, the optimization would not converge, apparently due to low sensitivity of the objective function to some of the muscles. The results of joint-position/muscle-activation sensitivity analysis performed in the previous chapter are used to select the muscles which produce the greatest joint movements toward the desired posture of the hand. Using a reduced set of muscles, chosen for their effectiveness in producing the desired motion, may result in a convergent optimal solution for finger position.

Moreover, the thumb and index finger models are uncoupled and separately interfaced with the optimization routines to break down the problem into two simpler ones with fewer number of inputs (design variables) and outputs (design responses). In the hand model, the thumb and index finger are coupled only through the act of the transverse head of the first dorsal interosseous ($DI1_t$). $DI1_t$ originates from the first metacarpal bone and inserts into the distal phalanx of the index finger. To uncouple and simulate the two models separately, the activation level of this muscle is used as a design variable in the index finger positioning algorithm, but is set to the same level in the thumb positioning scheme.

To study the effect of FDS tendon transfers, first the moment produced by the FDS at the PIP joint is removed from the hand model to simulate the Zancolli-Lasso procedure. To simulate the modified Zancolli method, the moment arms of the

extensor mechanism at PIP and DIP joints are used in conjunction with FDS force to produce extension moments at the interphalangeal joints when the FDS muscle is activated. The function of the FDS muscle at the MCP joint is left intact in both tendon transfer models.

The desired positions of the joints in a successful palmar pinch posture of the hand are prescribed as: MCP 90 degrees flexion, PIP and DIP neutral (zero degrees flexion) for the index finger, and CMC 30 degrees flexion and 5 degrees abduction, and MP 5 degrees flexion for the thumb. These values are not the exact joint angles that are to be reached in a successful palmar pinch, but rather are the extreme values. Any set of joint angles close to the determined values are also acceptable, providing that a convenient palmar pinch posture is obtained. The thumb IP joint is not included in this analysis due to the fact that IP joint position is not affected by the selected muscles.

Based on the results of the sensitivity analysis, presented previously, the following muscles: FDS, EI, DII, EPB, and APL are selected. Simulations are performed with joint passive moments of both normal and quadriplegic hands. Passive moment parameters for the quadriplegic hand are estimated from the measurements performed at the MCP joint of a quadriplegic patient at the MetroHealth Hospital, Cleveland, OH (Chapter 2).

As the simulation protocol, first all the muscles are initially turned off (zero activation level) and then, the following simulations are performed (Table 4.1).

1. Execution of the optimization program interfaced with the model of a normal index finger, using FDS, EI and DI1 muscles to search for the best activation pattern which positions MCP at 90 degrees flexion, and PIP and DIP at zero degrees flexion.
2. Execution of the optimization program interfaced with the model of a normal thumb, using EPB and APL muscles to search for the best activation pattern which positions CMC at 30 degrees flexion and 5 degrees abduction, and MP at 5 degrees flexion.
3. Same as number one, but using only the muscles EI and DI1, and implementing the passive moment parameters of the abnormal joint for the quadriplegic hand described in Chapter two.
4. Same as number one, using only the extrinsic muscles (FDS and EI) with Zancolli-Lasso tendon transfer.
5. Same as number four, but with modified Zancolli tendon transfer.
6. Same as number four, but with implementing the passive moment parameters of the abnormal joint for the quadriplegic hand and limiting the activation levels to 40%.

7. Same as number five, but with implementing the passive moment parameters of the abnormal joint for the quadriplegic hand and limiting the activation levels to 40%.

Case No.	Model	Joint	Muscles	Transfer
1	index	normal	FDS, EI, DII	none
2	thumb	normal	EPB, APL	none
3	index	abnormal	EI, DII	none
4	index	normal	FDS, EI	Zancolli-Lasso
5	index	normal	FDS, EI	Mod. Zancolli
6	index	abnormal	FDS, EI	Zancolli-Lasso
7	index	abnormal	FDS, EI	Mod. Zancolli

Table 4.1) Summary of the simulations performed using the index or thumb model along with the optimization routine.

The first two simulations are mostly performed to evaluate the interfaced program. The rest of the simulations study the effect of joint passive moments and tendon transfers on positioning the index finger. The muscles, in each case, are selected based on the results of the sensitivity analysis performed in the previous chapter as well as the results of the previous cases. In all the simulations, the maximum acceptable convergence error of the change in objective function is set to 0.01 radian. The convergence errors of static equilibrium joint moment and joint position are set to 0.001 Newton-centimeter and radian, respectively.

C. Results

In a normal hand, the closest joint positions to the desired joint angles are achieved by full activation of DII and about ten percent stimulation of EI, EPB, and APL (Figures 4.3 and 4.4, case numbers 1 and 2 in Table 4.1). Joint positions show large oscillations at initial steps of the optimization, but the overall trend is toward convergence to the desired values. The evaluated objective function in the thumb positioning routine converges to about zero, that is, the desired position of the thumb is reached (Figure 4.5). However, for the index finger, converge to the desired joint angles not as good. The joint positions converge to values much larger than the desired angles resulting in convergence of the objective function to a large value. Nevertheless, the posture of hand is significantly altered toward the desired posture after the optimization is employed.

In the case where the estimated parameters of joint passive moments of the quadriplegic index finger are employed in the model (case number 3 in Table 4.1, an abnormal hand), even with full activation of DII and about seventy percent activation of EI, the final joint positions are still far from the desired values (Figure 4.6). In all case numbers 1, 2 and 3, joint positions are closer to the desired angles after optimization. However, joint positions are closer to the desired angles after optimization in normal hand than abnormal hand (Table 4.2).

The Zancolli-Lasso tendon transfer, however, brings the normal index finger to a relatively convenient palmar pinch posture (Figure 4.7). The modified Zancolli tendon transfer, in which the FDS acts similar to an intrinsic muscle, results in even closer joint positions to the desired values in the normal hand (Figure 4.8). In the quadriplegic hand used for these simulations, however, the tendon transfers seem not to be effective (Figures 4.9 and 4.10). Activation of the other extrinsic extensor (EDC), too, does not significantly improve the joint angles since the joints have already reached their extreme extension angles where the resistant passive moments are relatively high.

JOINT	JOINT ANGLE (deg)				
	Desired	Normal		Abnormal	
		Initial	Final	Initial	Final
CMC (flx)	30	31	30	----	----
CMC (add)	-5	25	-7	----	----
MP	5	17	4	----	----
MCP (flx)	90	39	73	79	81
PIP	0	42	21	83	75
DIP	0	22	9	42	30

Table 4.2) Summary of results of the optimization in normal and abnormal hands. The optimization routine alters stimulation level of FDS, EI, DII, EPB, and APL muscles to move the joints toward the desired positions. Joint angles before and after optimization show large improvement in normal hand, but little improvement in abnormal hand.

A comparison among the evaluated objective functions of the above cases (Cases number 4 to 7 in Table 4.1) shows that the introduced tendon transfer methods are effective in the normal hand but not in the quadriplegic hand (Figure 4.11). The lowest value of the objective function, that is, the closest posture to the desired posture, is achieved by the modified Zancolli tendon transfer in the normal hand with the activation of both FDS and EI muscles. The largest value, that is, the farthest posture to the desired posture, is achieved by the Zancolli tendon transfer in an abnormal hand.

D. Discussion

The muscle activation patterns required to reach a desired posture were determined using the hand model interfaced with optimization packages. The joint positions in a palmar pinch posture were selected as the desired angles and the interfaced program was used to search for the muscle activation pattern which provides the palmar pinch posture. It was shown that this posture is not reached without activation of the intrinsic muscles. In a hand with normal passive moments, tendon transfer techniques provided a satisfactory posture. However, due to the hyper-flexion of the joints and weakness of the muscles, tendon transfer alone was not very effective in restoring normal hand functions in this quadriplegic subject.

The thumb and the index finger models were uncoupled and a reduced set of muscles were used in optimal finger positioning routines to reduce the number of design variables and responses. This was believed to decrease the non-linearity of the objective function, resulting in a convergent optimal solution. An approximation method was used in the optimization procedure and was found effective. However, the approximation may reduce the robustness of the optimization in highly non-linear problems. The approximated curves are changed as the optimization process proceeds and the number of points in the basis increases. Therefore, in each optimization trial, a different approximated function is minimized. This fact is more significant when the system is highly nonlinear and the actual objective function changes rapidly.

In some cases, the objective function converges while some of the muscle activation levels are still changing. When a joint is at its extreme position, the passive moment curve is very steep and a small change in joint angle results in a large change in passive moment. At the extremes, even a large increase in muscle moment about the joint is easily resisted by the passive moment. Therefore, the joint angle, and as a result, the object function does not change significantly. This reveals that the objective function at the extreme joint angles is not sensitive to the activation of the muscles.

The muscle activation pattern selected by the optimization method in a palmar pinch posture of the index finger is in agreement with the literature (Long, 1968 and 1981). Based on Electromyographic studies, Long has found that to reach or hold the

MCP joint at flexed and the interphalangeal joints at extended positions, contraction of the intrinsic muscles is required, and to a lesser extent contraction of the extrinsic extensor mechanism. In our simulations, the optimization method chose to turn off the extrinsic flexor (FDS), fully stimulate the intrinsic muscle (DI1) and partially activate the extensor mechanism (EI).

In the normal hand, the joint angles converge close to the desired angles after optimization, provided that both the intrinsic and extrinsic muscles are used. However, there are still significant differences between the desired and the final values of index finger joint angles (Table 4.2). This seems to be due not to the failing of the optimization scheme but rather, due to the fact that the selected muscles cannot do better, and the fact that the subject (JM) cannot flex his MCP joint more than eighty degrees. A three dimensional plot of the objective function versus variable combinations of the activation levels of the selected muscles (EI and DI1 for index finger and APL and EPB for the thumb) reveals that optimization has converged to the global minimum in the feasible region (Figures 4.12a, 4.12b). Since the first dorsal interosseous (DI1) flexes MCP joint while extending the PIP and DIP joints, which is exactly what is needed here, a closer posture to the desired posture might be found if this muscle was more powerful (i.e., a minimum outside the possible region plotted). When the modified Zancolli tendon transfer is employed, DI1 is in fact replaced by a stronger muscle (transferred FDS). It was shown that, in such a case, even activation of the transferred FDS alone can provide the desired posture in the normal hand.

A palmar pinch posture is not obtained when the joint passive moment parameters of the quadriplegic person (KT) are implemented in the hand model with or without tendon transfers. This could be due to the fact that passive moments of the quadriplegic person force his joints into a more flexed position than the normal person (JM) which requires much more extension strength at PIP and DIP joints to achieve a palmar pinch posture. Moreover, as it was discussed in the previous chapter, in the normal hand with less flexed joints, the activation of either of the extrinsic extensor muscles can move the MCP joint into zero position, while none of them can solely neutralize PIP and DIP joints. Thus, the extensor muscles certainly cannot position the abnormal hand, with hyper-flexed joints, as the normal hand, resulting in larger differences between the final and the desired joint angles. Tendon transfer alone is not found to be very effective in restoring normal hand functions in the quadriplegic person, probably due to the high passive moments holding the finger in flexion and weakness of the muscles, especially the extensors. If the normal passive moments could be retrieved by somehow shifting the abnormal passive moments from a hyper-flexed angle to a less flexed angle, the modified Zancolli tendon transfer was effective.

Figure 4.3) Results of simulation case number one (Table 4.1). Iteration of index finger muscle activation levels (a), and resultant joint positions (b) by the optimization routine in order to achieve the desired posture (90 degrees flexion at MCP, zero degree flexion at PIP and DIP).

Figure 4.4) Results of simulation case number two (Table 4.1). Iteration of thumb muscle activation levels (a), and resultant joint positions (b) by the optimization routine in order to achieve the desired posture (30 degrees flexion and 5 degrees abduction at CMC, and 5 degrees flexion at MP joints).

Figure 4.5) The evaluated objective function is almost converged to zero in the thumb positioning routine, whereas it converges to a non-zero value in the index finger routine.

Figure 4.6) Results of simulation case number three (Table 4.1). The desired posture is not achieved when the passive moments of a quadriplegic patient are implemented. Index finger muscle activation levels (a), and joint positions (b) are plotted verses iteration number.

Figure 4.7) Results of simulation case number four (Table 4.1). The Zancolli-Lasso tendon transfer technique performs convergence of the joint positions of a normal hand toward the desired joint angles using FDS and EI muscles. Index finger muscle activation levels (a), and joint positions (b) are plotted verses iteration number.

Figure 4.8) Results of simulation case number five (Table 4.1). The modified Zancolli tendon transfer technique performs even a better convergence of the joint positions of a normal hand toward the desired joint angles using FDS and EI muscles. Index finger muscle activation levels (a), and joint positions (b) are plotted verses iteration number.

Figure 4.9) Results of simulation case number six (Table 4.1). The Zancolli-Lasso tendon transfer is not effective on a quadriplegic hand using FDS and EI muscles. Index finger muscle activation levels (a), and joint positions (b) are plotted verses iteration number.

Figure 4.10) Results of simulation case number seven (Table 4.1). In a quadriplegic hand (passive moments of a quadriplegic patient and muscle activations limited to 40%), the modified Zancolli tendon transfer technique is not efficient to perform the desired posture using FDS and EI muscles. Index finger muscle activation levels (a), and joint positions (b) are plotted verses iteration number.

Figure 4.11) The evaluated objective function verses iteration number in variety of cases: Zancolli tendon transfer on a normal hand (■), modified Zancolli on a normal hand (+), Zancolli on an abnormal hand (*) and modified Zancolli on an abnormal hand (□). Lower objective function (i.e., closer joint positions to the desired angles) is achieved by the modified Zancolli tendon transfer compared to the Zancolli-Lasso technique. In quadriplegic hand, the objective function converges to a relatively high number showing that the tendon transfer is not effective.

Figure 4.12) Results of simulation case number one (Table 4.1). The evaluated objective function (root mean square of the differences of the current and the desired joint positions: 90 degrees flexion at MCP, zero degree flexion at PIP and DIP) versus variable combinations of activation level of EI and DII (a), and APL and EPB. The objective function is minimum when EI is stimulated by about 10% and DII is fully activated (for the index), and when APL and EPB are activated by about 8%. This is exactly what was found by the optimization routine.

CHAPTER FIVE

PREHENSILE LATERAL PINCH

A. Introduction

One of the most important functions of the hand is prehensile lateral pinch. In lateral pinch, the object is placed in between the palmar aspect of the distal phalanx of the thumb and the radial or lateral side of the index finger (Figure 5.1). The index finger along with the other fingers are flexed, providing a support for the object. The thumb is extended and possibly adducted from its rest position so that it opposes the index finger more proximal (refer to Chapter three for definition of the thumb movements). Activation of the thumb flexors and/or adductors provide the grip force, while the adduction/abduction passive stiffness of the finger joints support the object and oppose the grip force.

When the thumb is in contact with an object, movement of the thumb is constrained by interaction forces. During contact, the object deforms compliant finger pads, which produces forces normal and tangential to the surface of the object. Moreover, possible slippage of the pads on the surface of the object produces tangential friction forces. The total interaction force is the vector sum of the normal and the tangential forces.

Figure 5.1) Functional hand posture in lateral prehension. Thumb is extended and possibly adducted to oppose the lateral aspect of the index finger more proximal.

A successful lateral pinch, which is grouped as a type of power grip (Napier, 1956), requires relatively large grip force which can be difficult to achieve. The moment arm of the grip force is generally much larger than moment arm of tendons

about the thumb joints. Therefore, strong muscles are needed to generate enough force to balance the moment of the grip force. Muscles in quadriplegic people using functional neuromuscular stimulation (FNS) are much weaker than normal muscles. Therefore, to produce sufficient grip force, they may need to use a larger number of muscles than normal people.

In applications of FNS, the distal joint of the thumb (IP) is sometimes surgically fused at zero degrees (in a straight line along the proximal phalanx; Kilgore *et al.*, 1988). This prevents significant potential hyper-extension as well as possible extreme flexion of the IP joint in quadriplegic patients, and provides appropriate posture. However, IP joint fusion may alter the potential grip force. It also removes one degree of freedom from the thumb which may be needed in other functions of the hand. An alternative approach which provides the appropriate posture and adequate grip force, without surgical alterations, has to be found.

B. Method

In simulation of a prehensile function of hand, the ADAMS multi-body model is acted upon by muscle moments as well as moments of external contact force produced by the interaction of the thumb and the object (Figure 5.2). The moments of the external force at the thumb joints are computed in an interaction user-written subroutine which

is linked to the ADAMS main routine. The interaction subroutine calculates the external force according to the intersection kinematics of the phalanges and the grasped object. The external force includes the normal (pinch) and frictional forces between the object and the finger pads. External force to the thumb is reduced to resultant forces at, and moments about, all the joints proximal to the point of application of the force. Only resultant moments of the external forces are returned to the main routine and applied about the corresponding joints. The resultant forces are actually action-reaction workless constraint forces which have no effect on our computations and are not included in the model.

Figure 5.2) Schematic of the model utilized to simulate lateral pinch. The input to the model is a set of muscle activation levels, and the output is the pinch force.

To model the interaction of the thumb and an object in lateral pinch, it is assumed that the object has a rectangular shape and is fixed to the fingers. It is also assumed that the fingers can be independently positioned, holding the object horizontally. Hence, the surface of the object fixed to the fingers, is always fixed and parallel to the ground. The geometry of the contact surface of the object is therefore modeled as a horizontal flat plane fixed to the ground.

It is also assumed that only the distal phalanx of the thumb is in contact with the object. The geometry of the palmar surface of the distal phalanx is simply modeled as a sector of a sphere. Available x-rays of a thumb from three views (perpendicular to the palmar surface, and about twenty and forty degrees inclined from the perpendicular view) are used to find several points on the surface of the palmar aspect of the distal phalanx with respect to the center of the distal joint. In each view, a circle is drawn through the points using a compass. The average position of the center of the three circles, with respect to the center of the IP joint, is selected as an estimate of the center of the imaginary sphere ($X=1.3\pm0.2$, $Y=-1.55\pm0.2$ and $Z=0\pm0.1$ cm). Likewise, the average of the radii of the three circles is chosen as the radius of the sphere (2.8 ± 0.3 cm).

When the distal phalanx of the thumb makes contact, it is assumed that its surface penetrates into the surface of the object. The magnitude of normal interaction force, if the thumb and the object are in contact, is computed from the magnitude and

velocity of relative penetration of the contact surfaces. A quasi-linear viscoelastic model, similar to the joint passive moment model introduced in Chapter two,

$$F_n = f_1(d) + f_2(d) \cdot (\dot{d})^r, \quad (5.1)$$

was used to represent this relationship. Here, F_n is the magnitude of the normal force applied to the finger pad at the contact point as a function of the penetration depth, d , and velocity of penetration, $\dot{d}$. f_1 and f_2 are respectively elastic and instantaneous elastic functions, which are determined experimentally. The penetration depth is determined as

$$d = D - r, \quad (5.2)$$

where D is the perpendicular distance from the center of the imaginary sphere to the surface of the object and r is the radius of the sphere. The penetration velocity is the perpendicular velocity of the center of the imaginary sphere.

Since the force perpendicular to the contact surface is always perpendicular to the horizontal plane, thus,

$$\bar{\mathbf{F}}_n = F_n \hat{\mathbf{k}}, \quad (5.3)$$

. In the following formulations all vectors are given in the global reference frame with the unit vectors $\hat{\mathbf{i}}$, $\hat{\mathbf{j}}$ and $\hat{\mathbf{k}}$.

The finger pad may slide while it is in contact with the object. As a result, besides the normal force, a tangential friction force is exerted at the contact point. Assuming a constant coefficient of friction, the magnitude of the friction force is a proportion of the magnitude of the normal force,

$$F_f = \mu F_n, \quad (5.4)$$

where μ is the coefficient of friction. μ is set equal to 0.5, representing the coefficient of friction between finger pads and polished surfaces (Westling and Johansson, 1984). The friction force is in the direction opposite to the tangential velocity of the contact point with respect to the object, which is determined by the unit vector

$$\bar{\mathbf{v}} = \frac{V_x}{V} \hat{\mathbf{i}} + \frac{V_y}{V} \hat{\mathbf{j}}, \quad (5.5)$$

where V is the magnitude of the tangential velocity of the contact point, and V_x and V_y are the X and Y components. Thus,

$$\bar{\mathbf{F}}_f = -\mu F_n \left(\frac{V_x}{V} \hat{\mathbf{i}} + \frac{V_y}{V} \hat{\mathbf{j}} \right), \quad (5.6)$$

The friction force in Equation (5.6) is discontinuous when the direction of motion changes (Figure 5.3). This discontinuity, which is especially significant when the normal force is high, causes convergence problems in the numerical solution. Moreover, this equation is not defined when the tangential velocity of the contact

point is zero. Therefore, the "STEP" subroutine already built in ADAMS, is used to calculate the coefficient of friction as an approximated function of the magnitude of the tangential velocity of the contact point. The STEP subroutine approximates a step function with a cubic polynomial of the form:

$$\begin{aligned} \mu(V) &= \mu_0 & V &\leq V_0 \\ \mu(V) &= \mu_0 + (\mu_1 - \mu_0) \left(\frac{V - V_0}{V_1 - V_0} \right)^2 \left[3 - 2 \left(\frac{V - V_0}{V_1 - V_0} \right) \right] & V_0 < V < V_1, \\ \mu(V) &= \mu_1 & V &\geq V_1 \end{aligned} \quad (5.7)$$

where V_0 and V_1 are the velocities at which the step approximation begins and ends, and μ_0 and μ_1 are the initial and the final values of the friction coefficient (Figure 5.3).

Figure 5.3) The friction force is discontinuous if μ is assumed constant (solid line). The STEP function built in ADAMS which approximates a step function with a cubic polynomial of the form presented in Equation (5.7) is used (dashed line).

Based on our approximation, in which the coefficient of friction is zero when the velocity is zero, μ_0 and μ_0 are set to zero. Different values of μ_1 were tested to find the steepest slope, as close as possible to a step function, where the program still converges. Values less than 0.01 cm/sec were found to cause convergence problems.

The total force exerted at the contact point on the phalanx can therefore be computed from Equations (5.3) and (5.6) as,

$$\bar{\mathbf{F}} = F_n \left[-\frac{\mu}{V_r} (V_x' \hat{\mathbf{i}} + V_y' \hat{\mathbf{j}}) + \hat{\mathbf{k}} \right]. \quad (5.8)$$

The reaction force applied to the object has the same magnitude but is in the opposite direction.

The total force at the contact point is reduced into an equivalent resultant force and moment about the center of the imaginary sphere representing the thumb pad (Figure 5.4). In ADAMS, a force or moment must be applied at a predetermined point, called a "Marker". The location of the contact point is not known before the contact between the phalanx and the object occurs, while the center of the imaginary sphere is a known, predetermined point. The normal force always passes through the center of the sphere, thus, has no moment about it.

The tangential force remains always perpendicular to the radius of the sphere producing an equivalent moment of

Figure 5.4) Schematic of the thumb-object intersection. XYZ is the ground reference frame. S is a slice of the imaginary sphere, representing the surface of the phalanx pad. F_f is the friction force and F_n is the normal force applied to the contact point P. The equivalent force F and moment M are applied at C, the center of the sphere.

$$\vec{M} = \vec{r} \times \vec{F} = F_n \left[\frac{r\mu}{V} (V_y \hat{i} - V_x \hat{j}) + 0\hat{k} \right] \quad (5.9)$$

where $\vec{r}$ is the position vector of the contact point with respect to the center of the sphere. Thus, as the mechanical equivalent, the force computed from Equation (5.8) and the moment from Equation (5.9) are applied to the center of the imaginary sphere.

To find the force-displacement relationship for a finger pad, Hendrix (1993) used an already available testing rig in the biomechanics laboratory to simulate the contact profile when grasping a cylinder. A four centimeter cylindrical indenter attached to a load cell was compressed against each finger pad. The average of the maximum and the minimum force on the hysteresis loop at a given displacement was

plotted against the amount of penetration of the indenter into each finger pad. A polynomial of order three was then fit to the experimental data resulting in four constant coefficients for each finger pad relating the force to the magnitude of penetration.

The same apparatus was used in this study to experimentally determine the parameters of the force-displacement-velocity relationship introduced by Equation (5.1). Two rectangular aluminum blocks ($8 \times 5 \times 2$ cm, with the widest side horizontal) were mounted on the apparatus. The upper block was attached to a load cell, which was driven vertically at a constant velocity by means of a gear-motor system. The load cell measured the force exerted vertically to the block, and a LVDT measured the distance it traveled. The lower block was mounted on a stand which was driven in the vertical direction by means of a manually driven screw.

Two tests were performed. In the first experiment, the thumb was secured to the lower block by means of two Velcro straps, so that its palmar aspect was facing up. The upper block was then driven downward at a constant velocity by the gear-motor system, and was pressed against the palmar aspect of the distal phalanx of thumb until the subject (author) started to feel uncomfortable (about 25 N). At this point, the direction of motion of the block was reversed until it was completely separated from the thumb. Due to the experimental set up, the finger pad was only deformed vertically, not laterally. This test was only performed at the highest constant

velocity that could be achieved by the gear-motor system (about 0.03 cm/sec). This velocity was still much lower than the velocity at which lateral pinch actually occurs. In the second experiment, the relaxation test, the thumb was again secured to the lower block and the upper block was placed just above the thumb. The lower block was then quickly raised by the manual gear system, in about a second, and was held at variety of penetration depths for about sixty seconds. Finally, the parameters of the model of the phalanx pad (Equation 5.1) were determined and implemented in the intersection subroutine, which is linked to the thumb model generated by ADAMS.

As the first step of the simulation protocol of lateral pinch, EPB is partially activated (about thirty percent) to position the thumb in the functional lateral pinch posture just above the object (static simulation, convergence moment error = 0.01 N-cm). Then, ADP and FPL, which are the primary muscles activated in lateral pinch, are independently stimulated from zero to full activation level in one second (dynamic simulation, integration error = 0.001). The full level of activation is held, and the dynamic simulation is continued up to five seconds, where the velocity of the contact point falls below 0.01 cm/sec. This is where the friction effect starts to vanish due to the approximation in Equation (5.7). The resultant normal forces exerted on the thumb are finally plotted against time. Moreover, the range of motion of the MP joint is shifted toward flexion for 20 and 40 degrees, and ADP is stimulated. To study the effect of IP joint fusion, the joint is fixed at zero, 15, and 30 degrees of flexion, and ADP and FPL muscles are independently activated as just described.

The lateral pinch model was used to investigate current strategies as well as possible alternatives for achieving a lateral pinch. The goal is to provide enough grip force, while a convenient posture is maintained, that is extreme flexion and/or hyper-extension of IP and MP joints are prevented. In this view, three combinations of muscles are activated and the posture of the thumb as well as the relaxed normal force at five seconds are recorded: 1) full activation of ADP along with partial stimulation of FPL; 2) activation of FPL and EPL together; 3) full activation of ADP and APB along with partial activation of FPL. Muscle selection is guided by the results of the sensitivity analysis presented in Chapter three.

C. Results

Loading and unloading a finger pad clearly shows hysteresis (Figure 5.5). The loading and unloading curves are far apart, showing a large damping effect. Finger pad relaxation tests, however, yielded little useful data. At relatively high penetration depths, the force does not completely relax after sixty seconds, and holding the finger more than a minute fatigues the subject, resulting in useless data. At lower penetration depths, the amount of force is not enough to overcome the noise due to hand vibrations and the finger pulsation. Hence, the elastic function, f_1 , is estimated as the average between the loading and unloading curves,

$$f_1 = \frac{1}{2}(F_l + F_u), \quad (5.10)$$

where F_l is the loading and F_u is the unloading force. The instantaneous elastic function, f_2 , is then estimated as half of the difference between the loading and unloading curves,

$$f_2 = \frac{1}{2}(F_l - F_u). \quad (5.11)$$

By substituting Equations (5.10) and (5.11) into Equation (5.1), we have

$$F_n = \frac{1}{2}[F_l(1 + d^r) + F_u(1 - d^r)]. \quad (5.12)$$

r is estimated as 0.1, representing very little dependence of the force on the penetration velocity.

The loading and unloading curves of the hysteresis loop are well fit (Figure 5.5) by exponential functions of the form

$$F = ae^{bd} \quad (5.13)$$

where F is either the loading, F_l , or unloading force, F_u , and a and b are constants of regression (Table 5.1).

Coefficients of Regression $F = ae^{bd}$					
a_l	b_l	r^2	a_u	b_u	r^2
0.0568	8.395	0.985	0.0204	8.803	0.989

Table 5.1) Coefficients of regression of the loading and unloading force-displacement curves of the thumb distal pad.

Dynamic simulations of lateral pinch reveal that the instantaneous grip force, after the muscle(s) is(are) activated, may be significantly larger than the static or relaxed grip force (Figure 5.6). Both the instantaneous and the relaxed normal forces generated by FPL are greater than the forces produced by ADP. In addition, the relaxed force, holding the activation level constant, is higher for FPL than ADP. When the IP joint is fixed at zero degrees of flexion, the normal force produced by activation of ADP increases about twenty percent with respect to the case where the joint was free (Figures 5.6a). As the joint fusion angle increases to 15 and 30 degrees of flexion, the normal force decreases, however, it is still about fifteen percent larger than when IP joint is not fused. On the other hand, the force generated by stimulation of FPL decreases about twenty five to thirty percent when IP joint fusion angle increases (Figure 5.6b).

When ADP is stimulated without IP joint fusion, CMC joint is flexed but, MP and IP joints are hyper-extended (Figure 5.7a). Hyper-extension of MP and IP joints leads to an inconvenient posture for lateral pinch. However, when the range of motion of the MP joint is shifted toward flexion, due to the passive moment shift, hyper-extension of the MP joint is prevented and a convenient posture is achieved (Figure 5.7b, c). Activation of FPL without joint fusion results in extreme flexion of IP joint as well as flexion at MP joint (Figure 5.8). Extreme flexion of IP joint, which flexes the thumb toward the palm, also leads to an inconvenient lateral pinch position. In cases where the IP joint is fused, no matter what the fusion angle is, activation of ADP leads to hyper-extension of the MP (Figure 5.9 and Table 5.2). On the other hand, when FPL is activated, no matter what the fusion angle is, the achieved posture is convenient (Figure 5.10 and Table 5.2).

No convenient lateral pinch posture is achieved when ADP is fully activated along with partial stimulation of FPL (Table 5.2). Low level activation of FPL results in MP joint hyper-extension and high level stimulation (about fifty percent of full activation) causes IP extreme flexion (Figure 5.11). On the other hand, full activation of ADP and APB along with partial stimulation of FPL (about 30 percent) results in a convenient posture similar to the case where FPL is activated with IP joint fusion (Figure 5.12). Relatively, significant normal force, 16.6 Newton, is produced in this case (Table 5.2). Full activation of EPL does not overcome the extreme flexion of IP joint caused by full activation of FPL (Figure 5.13). Reducing the stimulation of FPL

to about fifty percent may result in a more convenient posture, however, this will reduce the normal force (Table 5.2).

In summary, among all the combinations studied, the FPL muscle with IP joint fusion, and the combination of ADP, APB and FPL muscles without IP fusion are found to provide more convenient posture for lateral pinch function of the hand (Table 5.2). The accumulated integration steps was often about three hundred steps, and the CPU time was about ten minutes. In one case tested, reducing the integration tolerance by one order of magnitude had no significant effect on the results, but increased the CPU time drastically.

MUSCLE COMBINATION	IP JOINT FUSED	CONVENIENT POSTURE	NORMAL FORCE (N)
ADP	No	No	-----
FPL	No	No	-----
ADP	@ 0 deg	No	-----
ADP	@ 15 deg	No	-----
ADP	@ 30 deg	No	-----
FPL	@ 0 deg	Yes	10.1
FPL	@ 15 deg	Yes	10.7
FPL	@ 30 deg	Almost	11.2
ADP+FPL	No	No	-----
ADP+APB+FPL	No	Yes	16.6
FPL+EPL	No	No	-----

Table 5.2) Summary of the results of activation of a variety of muscle combinations with and without IP joint fusion. If the IP joint is fused, the angle at which it is fused is presented. The final relaxed force is shown in the last column if a convenient posture is achieved. The combination of ADP, APB, and FPL without IP joint fusion is an alternative for FPL stimulation with joint fusion.

D. Discussion

Modeling and simulations of lateral prehension of a fixed object revealed that an alternative approach, other than the strategy currently used in FNS (IP joint fusion), may also provide convenient lateral pinch. In the alternative approach, moment ballance of the three activated muscles (ADP, APB, FPL) prevent hyper-extension and extreme flexion of the IP joint, without any need for joint fusion. Fusion of the IP joint removes one degree of freedom of the thumb which may be needed for other functions of the hand, and decreases the normal force when either ADP or FPL are activated (Figures 5.6a and 5.6b). In addition, if the muscle moments were to be ballanced by the passive moments, joint behavior would change over long period of time. The grip force produced by the alternative approach was shown to be about seventy percent larger than the force produced in the current approach. In quadriplegic people, whose muscles are much weaker, as much muscle as possible should be involved to provide adequate grip force although, there is a limitation on the total number of muscles that can be activated in practice. Simulations also showed that activation of other combinations of the muscles, such as ADP+FPL and FPL+EPL cannot provide a convenient posture for a lateral pinch function.

As it was mentioned earlier, the variation of some of the model parameters among subjects is relatively high. This variation of the parameters, which is more significant between the normal and abnormal subjects, may simply alter the results of

the simulations. For instance, when the ADP muscle with or without IP joint fusion was activated in our simulations, the resultant posture was found inconvenient due to the hyper-extension of the MP joint. This is because the passive moment of the hand modeled here allows about 30 degrees hyper-extension of the MP joint. However, when the range of motion of the MP joint was shifted toward flexion, the hyper-extension of the MP joint was prevented by large passive moment, and a more convenient lateral pinch posture was achieved (Figure 5 7). To better predict one's hand function, it is necessary to measure and/or estimate the parameters which have significant effect on the simulation results on that specific hand; especially the joint passive moments, maximum strength of muscles, and hand geometry.

Several assumptions were made in modeling the interaction of the thumb and object in lateral pinch. It was assumed that the object is placed in between the thumb and the lateral aspect of the index finger, and the distal phalanx of the thumb applies the grip force while the lateral passive stiffness of the fingers support the object. When an object is relatively thin, it is placed in the gap between the index finger, and both the distal and proximal phalanx of the thumb positioned in a straight line and parallel to the object. Both distal and proximal phalanges may, therefore, be in contact. However, only the distal phalanx which is positioned against the index finger applies the pinch force. If the object is relatively thick, the proximal phalanx is used to elevate the distal phalanx providing a larger gap between the index finger and the thumb. In this posture, only the distal phalanx of the thumb is in contact and applies the pinch force. Another

assumption was that only the palmar side of the distal phalanx is in contact with the object during lateral pinch. Therefore, a model of the geometry of the whole phalanx did not seem essential. It was also not essential to precisely model the geometry of the palmar surface of the distal phalanx, since it does not have a significant effect on the posture and computation of the pinch force. The pad of the phalanx was modeled as a sector of the surface of a sphere. This, keeping sufficient accuracy, drastically simplified the formulation of the intersection model.

The sliding friction, between the thumb and the object, is believed to have a significant effect on both the posture of the thumb and the grip force. If the friction force is neglected, the thumb may move freely, tangential to the object, to an altered position which results in a different posture as well as a different moment arm of the external force about the joints. The moment of the total external force, that is, friction and normal force, about a joint is balanced by the joint passive moment and muscle moments. Thus, alteration of the moment arm about a joint, as well as the alteration of the total external force due to neglecting the friction force, may change the grip force, significantly.

Dynamic simulations were terminated when the tangential velocity of the contact point was about 0.01 cm/sec. This often occurred after about four seconds from the time muscle activation was held constant. This is because of the implementation of the STEP function (Equation 5.7) which was used to overcome the

numerical problems discussed earlier. Based on this approximation, when the magnitude of the tangential velocity falls below 0.01 cm/sec, the coefficient of friction does not remain constant and the friction force reduces with the reduction of the tangential velocity. At the time the system reaches its static equilibrium, where the tangential velocity is zero, the coefficient of friction and therefore the friction force reduce to zero. This static equilibrium position is not what we are looking for, since the effect of the friction force has vanished. The dynamic equilibrium configuration of the system however, when the velocity equals 0.01 cm/sec, is a quite close approximation of the static equilibrium configuration of the system including the friction effect.

Normal force was reduced by fusion of the IP joint when either ADP or FPL are activated (Figures 5.6a and 5.6b). It was observed that increasing the joint fusion angle decreases the normal force when ADP is activated (Figure 5.6a) whereas, the normal force is increased when FPL is stimulated (Figure 5.6b). This may be explained by the position of the contact point with respect to the joint centers, that is, the moment arm of the external force. Assume that resultant muscle and passive moments at a joint are constant. The moment of the external force about the joint is to be balanced by the resultant moment. If the moment arm of the external force increases, that is the contact point moves away from the joint, the external force at that joint must decrease. On the other hand, a larger external force has to be produced if the contact point is closer to the joint. When the IP joint is not fused, stimulation of ADP

hyper-extends the MP and IP joints resulting in a proximal contact point (Figure 5.7). Fusion of the joint flexes it with respect to its hyper-extended position (Figure 5.9), moving the contact point away from the joint, resulting in less grip force. On the other hand, activation of FPL flexes the IP joint toward the palm (Figure 5.8). When IP joint is fused, the joint is extended with respect to its flexed position (Figure 5.10), increasing the moment arm of the external force and therefore decreasing the external force. As the fusion angle increases, the joint is more flexed resulting in smaller moment arm and larger external force.

The summary of the results in Table (5.2) reveals that, among all the cases tested, both full activation of FPL with the IP joint fused, either at zero or at 15 degrees of flexion, and activation of a combination of ADP, FPL and APB may lead to a convenient lateral pinch posture. The normal forces produced in both cases are relatively significant. However, activation of FPL with joint fusion generates less grip force. Moreover, activation of FPL with IP joint fusion requires fusion of a joint which reduces one degree of freedom from the thumb. All the three muscles, in the alternative approach, generate flexion moments at CMC and MP joints, which counterbalance the extension moments of the external force. ADP and FPL adduction moments at CMC joint are balanced by APB abduction moment and the external force moment mostly due to the tangential friction force. The moment of the external force at IP joint is balanced by the FPL flexion moment at this joint.

The average strength of a normal hand in lateral pinch is about 80 Newton (An *et al.*, 1985). In our simulations, only selected muscles are stimulated. Thus, it is not surprising that the predicted normal forces are much less than 80 Newtons. This value may only be an estimate of the upper bound of the pinch force predicted by our model. In a study by Kilgore *et al.* (1988), selected muscles were electrically activated and the pinch force was measured. The average pinch force due to stimulation of the adductor group (ADP, DII, PII, and part of FPB) had a magnitude of 9.7 ± 8.8 Newton with an average angle of 32 ± 23 degrees from the normal direction. Assuming that in the adductor group most of the normal pinch force is generated by ADP, this measurement is quite close to what our model predicts (7.5 N). In the same study, the average force was 2.8 ± 1.7 Newton with an average angle of 14 ± 14 when only FPL was activated. This is much less than about 15 Newton which is predicted by our model. A comparison between the physiological cross sectional area (PCSA) of ADP and FPL (Table 3.3), which is a good estimate of muscle strength in producing force, shows that FPL is about two fold stronger than ADP. Besides, the moment arms of the FPL tendon crossing all the three joints of the thumb (Table 3.5a) are respectively large. Therefore, it may not be surprising that FPL generates more pinch force than ADP. High variation of hand parameters among subjects may well be the source of the difference.

Figure 5.5) Hysteresis loop for loading and unloading of the palmar aspect of thumb distal phalanx (□), and the modeled exponential curves (solid lines). The parameters of the exponential curves are given in Table (5.1).

Figure 5.6) The normal force produced by activation of ADP (a), and FPL (b) verses time. The muscle activation is linearly increased up to full level in one second and is held at that level for four seconds. IP joint is either free to move (solid line) or fused at zero (dash line), 15 (long dash-dot line) and 30 (short dash-dot line) degrees.

Figure 5.7) Front view (looking at the tip of the index finger when it is in zero position) of the thumb when ADP is fully activated in one second, and held at that level for four more seconds. The resultant posture is inconvenient due to the hyper-extension of the MP joint (a), but to a convenient posture when range of motion of the MP joint is shifted toward flexion by 20 (b), and 40 degrees (c).

Figure 5.8) Front view (looking at the tip of the index finger when it is in zero position) of the thumb when FPL is fully activated in one second, and held at that level for four more seconds. Extreme flexion of IP joint leads to a non-functional posture.

Figure 5.9) Front view (looking at the tip of the index finger when it is in zero position) of the thumb when ADP is fully activated in one second, and held at that level for four more seconds. The IP joint is fused at zero (a), 15 (b) and 30 (c) degrees. No functional posture is achieved.

Figure 5.10) Front view (looking at the tip of the index finger when it is in zero position) of the thumb when FPL is fully activated in one second, and held at that level for four more seconds. The IP joint is fused at zero (a), 15 (b) and 30 (c) degrees. Functional lateral pinch posture is achieved.

Figure 5.11) Front view (looking at the tip of the index finger when it is in zero position) of the thumb when ADP is fully activated along with partial activation of FPL. Final activation levels are reached in one second, and is held for four more seconds at 10% (a), 30% (b) and 50% (c). No functional lateral pinch posture is achieved.

Figure 5.12) Front view (looking at the tip of the index finger when it is in zero position) of the thumb when ADP and APB are fully and FPL is partially (30%) activated. Final activation levels are reached in one second, and is held for four more seconds. A functional lateral pinch posture is achieved. The normal force produced is 16.6 Newton.

Figure 5.13) Front view (looking at the tip of the index finger when it is in zero position) of the thumb when FPL and EPL are fully activated. Final activation levels are reached in one second, and is held for four more seconds. Full activation of EPL does not overcome the extreme flexion of IP joint caused by activation of FPL. No functional lateral pinch posture is achieved.

CHAPTER SIX

PREHENSILE PALMAR GRASP

A. Introduction

A spatial model of palmar grasp of movable cylindrical objects is presented in this chapter. An open-loop model of the hand was developed in Chapter three and was linked to an optimization program to simulate non-prehensile palmar pinch (Chapter four). Then, it was complemented by an interaction subroutine to simulate prehensile lateral pinch (Chapter five). Prehensile lateral grasp alone is not sufficient in picking-up and holding activities of daily living (Peckham *et al.*, 1988). Indeed, most grasping functions may be achieved through palmar grasp or a combination of palmar and lateral prehension (Figure 6.1).

Figure 6.1) Palmar grasp of a cylindrical object.

Figure 6.2) Schematic of the model utilized to simulate palmar grasp of cylindrical objects. The input to the model is a set of activation levels, and the output is the object position and the grasp force.

In simulation of a prehensile function of the hand, the ADAMS multi-body model is driven by muscle moments as well as moments of external contact forces produced by the interaction of the hand and the object (Figure 6.2). In the model of palmar grasp of a cylindrical object, the interaction subroutine, linked to the main routine, calculates the external forces based on the intersection kinematics of the phalanges and palm, and the grasped object.

In simulations of non-prehensile functions of the hand, where there was no object and therefore no external forces, the input was the activation level to the selected muscles and the overall output was joint positions. In the present model, however, the output is either the grasp force (external force) applied to the object, if the object is fixed, or the grasp force as well as the position of the object, if it is free to move.

Several attempts have been made to measure the isometric strength of fingers in palmar pinch (An *et al.*, 1978; An *et al.*, 1985; Radwin and Oh, 1992) and palmar grasp of cylinders (Amis, 1987). Static models of cylindrical grasp have also been generated to predict grip strength for various sized cylinders (Buchholz *et al.*, 1988), and the final grasp configuration of the hand and contact forces (Hendrix, 1993). In all these attempts, the object to be grasped is fixed. In FNS applications, however, the patient should be able to grasp an unmodified object and manipulate it for use, independently. In practice, the object is often not fixed but is free to move. A dynamic

palmar grasp model is generated by linking a kinematic model of the interaction of the hand and a cylindrical object, and the dynamic model of hand presented in Chapter three. The grasp model is utilized to study the performance of the hand in grasping moveable cylindrical objects by stimulating selected muscles.

B. Method

In addition to the hand segments, a rigid part representing the cylindrical object, is added to the ADAMS main routine. The object interacts with the ground, palm, index finger, and thumb. The ground-object interaction forces include kinematic constraints and frictional forces between the ground and object. The hand-object interaction forces include kinetic constraint forces produced by the normal compliance of the finger pads, and frictional forces between the object and the phalanges. These interaction forces are treated as action-reaction forces in ADAMS model, that is the force is applied to one part and its reaction is applied to the other. The action part of the hand-object interaction force is applied to the hand and constrains hand movements. The reaction part of the hand-object and the ground-object interaction forces are applied to the object and constrain object movements.

Ground-object interaction model

Two different cases of ground-object interaction are modeled. In the first case, the object is fixed to the ground (zero degrees of freedom). In the second case, the object is free to slide on the ground, modeled as a horizontal flat plane. In both cases, the wrist is in neutral position and the palm is fixed to the ground. When the object is fixed to the ground, the constraint forces and moments are automatically applied by ADAMS to balance the applied forces and moments to the object and maintain the relative position of the object with respect to the ground. When the object is free to move, the "PLANAR" joint, already built in ADAMS, is used. Due to the kinematic constraints defined in a PLANAR joint, the object is allowed to slide on a horizontal flat plane (ground X-Y plane), and rotate about the axis perpendicular to the horizontal plane (three degrees of freedom). The kinematic constraint forces and moments are automatically applied by ADAMS, however, action-reaction frictional forces and moments have to be incorporated into the equations of motion by the user.

Friction between the object and the ground, in the case where it is free to slide, is computed as a proportion of the normal force as in the previous chapter. The coefficient of friction is approximated by the STEP function (Equation 5.6). The object may both rotate and translate with respect to the ground. The translational friction force is applied to the center of the circular base of the object in a direction opposite to the tangential velocity of the object (on the X-Y plane). The friction force

due to the rotation of the object is assumed to act on the edge of the circular base of the object. Therefore, the rotational friction applies a moment, with a moment arm equal to the radius of the object, about the axis perpendicular to the ground horizontal plane (Z axis) in the opposite direction of rotation of the object. The coefficient of friction, between the object and the ground, is set to 0.3, representing friction between polished surfaces.

Hand-object interaction model

The hand-object interaction is modeled similar to the interaction model presented in the previous chapter, with the exception that the object is cylindrical in shape and may not be fixed to the ground. In summary, when a phalanx makes contact with the object, normal and frictional forces at the contact point are produced. The normal force is a function of the magnitude and velocity of penetration of the contact surfaces. The friction force, which is perpendicular to the normal force and in the direction opposite to the relative velocity of the contact point, is a proportion of the normal force. The total external force is the vector sum of the normal and the friction forces.

Total external force on each phalanx is reduced to an equivalent force at and moment about all the joints proximal to the point of application of the force. Only

resultant moments of the external forces are applied about the corresponding joints, because the constraint forces do not have any effect on the simulations. The external forces acting on the object are reduced to an equivalent moment and force applied at the center of mass of the object.

To determine the parameters of the normal compliance of the finger pads in Equation (5.11), the same apparatus and same method discussed in the previous chapter, are used. This time, however, a cylindrical indenter with a four centimeter diameter is attached to the load cell. The index finger or the thumb is secured to the lower block and each finger pad is loaded and unloaded by the cylindrical indenter. Exponential functions are fitted to the loading and unloading curves, and the coefficients of the force-displacement relationship for each finger pad are obtained. The velocity in Equation (5.11) is raised to power n , which is estimated as 0.1, representing very little dependence of the force on the penetration velocity.

The grasp model is broken down into four steps. In the first step, the position and velocity of local coordinates of all point of interests, such as joints and center of mass of the object, are obtained with respect to the global reference frame using the system information routines in ADAMS. In the second step, if a phalanx actually contacts the cylinder, the position and velocity vector of the contact point are computed. If the distance between the proximal joint center and the point of contact, that is the moment arm of the external force, is greater than zero and less than the

length of the contacted phalanx, the amount of penetration of the phalanx into the object and the velocity of penetration are computed. In the third step, the normal force, using the relation between the force and the magnitude and velocity of penetration (Equation 5.11), and the friction force, as a proportion of the normal force, are computed. In the fourth step, the resultant moments about the proximal joints to the points of contact, and the resultant forces and moments at the center of mass of the object are determined and returned to the ADAMS routine. The same steps are followed at each iteration.

The magnitude of the normal force, as mentioned in the previous chapter (Equation 5.1), is a function of magnitude and velocity of penetration of the contact surfaces. The magnitude of penetration is determined as

$$d = p - (R_c + R_p), \quad (6.1)$$

where p is the magnitude of the common perpendicular to the center line of the object and the center line of the contacted phalanx, p (Figure 6.3), R_c is the radius of the object and R_p is the radius of the cylinder representing the contacting phalanx. The magnitude of the velocity of penetration is the rate of change of the magnitude of the common perpendicular vector, p .

The normal force is in the direction of the common perpendicular (Figure 6.3).

In Figure 6.3, XYZ is the object local reference frame at the object center of mass (C) defined such that the Z axis corresponds to the center line of the cylindrical object.

Figure 6.3) Schematic of hand-object intersection. XYZ is the object local reference frame at the object center of mass (C), with Z axis coinciding with the center line of the cylindrical object. $x_i y_i z_i$ represents the local reference frame of the contacted phalanx located at its proximal joint center (O_i). x axis coincides with the center line of the cylindrical phalanx. O_j 's are the center of the joints proximal to the contacted phalanx.

$x_i y_i z_i$ represents the local reference frame of the contacted phalanx located with its origin (O_i) at the center of the joint proximal to the contact and its x axis coinciding with the center line of the phalanx. The O_i 's are the centers of the joints proximal to the contacted phalanges ($i = 1, 2, \dots, m$, where m is the number of phalanges in contact), and O_j 's are the joint center of the non-contacting proximal phalanges ($j = 1, 2, \dots, n$, where n is the number of joints proximal to O_i). The center line of the cylindrical object is assumed to be always parallel to the ground Z axis. Therefore, the common perpendicular vector, $\mathbf{p}$, always remains parallel to the ground X-Y plane. In other words,

$$P_z = 0. \quad (6.2)$$

Since $\mathbf{p} \perp \mathbf{m}$, and $\mathbf{p} \cdot \mathbf{m} = 0$, we have

$$p_x m_x + p_y m_y = 0, \quad (6.3)$$

where $\mathbf{m}$ is the moment arm vector of the external contact force. We know that

$$\bar{\mathbf{o}} + \bar{\mathbf{m}} = \bar{\mathbf{h}} + \bar{\mathbf{p}}, \quad (6.4)$$

where $\mathbf{o}$ is the position vector of the proximal joint center with respect to the center of mass of the object. We also know that $\mathbf{h}$ is perpendicular to the X-Y plane, thus from Equation (6.4) we have

$$o_x + m_x = p_x, \quad (6.5)$$

$$o_Y + m_Y = p_Y, \quad (6.6)$$

$$o_Z + m_Z = r_Z, \quad (6.7)$$

where $\mathbf{r}$ is the position vector of the contact point with respect to the center of mass of the object. By substituting Equations (6.5) and (6.6) in Equation (6.3), we conclude that

$$o_X m_X + m_X^2 + o_Y m_Y + m_Y^2 = 0. \quad (6.8)$$

The moment arm vector can be represented as

$$\bar{\mathbf{m}} = m_X \hat{\mathbf{i}} + m_Y \hat{\mathbf{j}} + m_Z \hat{\mathbf{k}} = m(l_1 \hat{\mathbf{i}} + l_2 \hat{\mathbf{j}} + l_3 \hat{\mathbf{k}}), \quad (6.9)$$

where l_1 , l_2 and l_3 are the cosine directions of the moment arm vector. Assuming $m \neq 0$, Equations (6.7) and (6.8) yield

$$m = -\frac{l_1 o_X + l_2 o_Y}{l_1^2 + l_2^2}, \quad (6.10)$$

where m is the magnitude of the moment arm vector. Substituting Equation (6.10) in Equations (6.5) and (6.6), the components of the common perpendicular vector, $\mathbf{p}$, are

$$p_X = o_X - \frac{l_1(l_1 o_X + l_2 o_Y)}{l_1^2 + l_2^2}, \quad (6.11)$$

$$p_Y = o_Y - \frac{l_2(l_1 o_X + l_2 o_Y)}{l_1^2 + l_2^2}. \quad (6.12)$$

The normal force to the finger pad is in the direction of vector $\mathbf{p}$, i.e.,

$$\bar{\mathbf{F}}_n = F_n \left(\frac{p_x}{p} \hat{\mathbf{i}} + \frac{p_y}{p} \hat{\mathbf{j}} + 0\hat{\mathbf{k}} \right), \quad (6.13)$$

where p is the magnitude of the common perpendicular vector, and F_n is the magnitude of the normal force.

The friction force is in the opposite direction of the relative velocity of the contact point on the phalanx and the contact point on the object, both with respect to the ground reference frame. The tangential velocity vector of the contact point is found as

$$\begin{aligned} \bar{\mathbf{v}} &= \bar{\mathbf{v}}_{P_1 O} - \bar{\mathbf{v}}_{P_2 O} + \bar{\mathbf{v}}_{O C} \\ &= [(\bar{\boldsymbol{\omega}}_O \times \bar{\mathbf{s}}_1) + (\bar{\boldsymbol{\omega}}_C \times \bar{\mathbf{r}})] + \bar{\mathbf{v}}_{O C} \end{aligned} \quad (6.14)$$

where p_1 and p_2 are located at the contact point fixed to the finger pad and the object, respectively. $\bar{\boldsymbol{\omega}}_O$ and $\bar{\boldsymbol{\omega}}_C$ are the angular velocities of the phalanx and the object, respectively. Assuming a constant friction coefficient, μ , the friction force vector is

$$\bar{\mathbf{F}}_t = -\mu F_n \left(\frac{v_x}{v} \hat{\mathbf{i}} + \frac{v_y}{v} \hat{\mathbf{j}} + \frac{v_z}{v} \hat{\mathbf{k}} \right), \quad (6.15)$$

where v is the magnitude of the tangential velocity of the contact point. Combining Equations (6.13) and (6.15), the total force at the finger pad is

$$\bar{\mathbf{F}}_p = F_n \left[\left(\frac{p_x}{\rho} - \mu \frac{v_x}{v} \right) \hat{\mathbf{i}} + \left(\frac{p_y}{\rho} - \mu \frac{v_y}{v} \right) \hat{\mathbf{j}} - \left(\mu \frac{v_z}{v} \right) \hat{\mathbf{k}} \right]. \quad (6.16)$$

The position vector of the contact point with respect to the next most proximal joint center is defined as

$$\bar{\mathbf{s}}_i = \bar{\mathbf{m}} - (R_p - D) \frac{\bar{\mathbf{p}}}{\rho}. \quad (6.17)$$

where R_p is the phalanx radius and D is the penetration depth.

The moment about the next most proximal joint, due to the total force is

$$\bar{\mathbf{M}}_{\alpha_i} = \bar{\mathbf{s}}_i \times \bar{\mathbf{F}}_p. \quad (6.18)$$

The contact force on a phalanx produces moments about all the other proximal joints, too. Thus

$$\bar{\mathbf{M}}_{\alpha_j} = \bar{\mathbf{s}}_j \times \bar{\mathbf{F}}_p, \quad (6.19)$$

where

$$\bar{\mathbf{s}}_j = \bar{\mathbf{s}}_i + \bar{\mathbf{d}}_{ij}. \quad (6.20)$$

The contact force on the finger produces an opposite, but equivalent force on the object,

$$\bar{\mathbf{F}}_c = -\bar{\mathbf{F}}_p. \quad (6.21)$$

Therefore, the moment of the contact force about the center of mass of the object is

$$\bar{\mathbf{M}}_C = -\bar{\mathbf{r}} \times \bar{\mathbf{F}}_p, \quad (6.22)$$

where

$$\bar{\mathbf{r}} = R_C \frac{p_x}{p} \hat{\mathbf{i}} + R_C \frac{p_y}{p} \hat{\mathbf{j}} + h \hat{\mathbf{k}}, \quad (6.23)$$

and R_C is the radius of the object.

The grip force required to hold an object is influenced by the weight of the object and the friction between the skin and the object. A study by Westling and Johansson (1984) reveals that the grip force employed by human subjects to hold an object is always more than the minimal grip force necessary to prevent slipping. The difference between the minimal force and the employed force is called the "safety margin". The relative safety margin defined as the safety margin in percent of the minimum required grip force value is found to be almost constant (same reference).

Assuming that the weight of the object is balanced by the friction force between the finger pads and the object, the minimal normal grip force required to hold an object without slipping is

$$F_n^{\min} = \frac{W}{\mu}, \quad (6.24)$$

where W is the weight of the object and μ is the coefficient of friction. According to the results of the measurements by Westling and Johansson (1984), the employed grip force to hold an object without slipping is about 1.4 times the minimal grip force, thus the required normal grip force is

$$F_n^{req} = 1.4 \frac{W}{\mu}. \quad (6.25)$$

It is assumed that the weight of the object is equally supported by the thumb and the fingers. Thus, half of the weight of the object is supported by the fingers, which is in turn assumed to be equally distributed between the index finger and the middle finger (Amis, 1987; Radwin *et al.*, 1992). Thus,

$$F_n^{index} = \frac{1.4}{4\mu} W. \quad (6.26)$$

In our simulations the coefficient of friction was set to 0.5 representing friction between the finger pads and polished surfaces (Westling and Johansson, 1984). Therefore, the required normal force applied to the object by the index finger to hold the object is found from the following equation:

$$F_n^{index} = 0.7W \quad (6.27)$$

To search for the muscle activation pattern which provides the required normal force applied to the object by the index finger to hold the object, the hand model is

interfaced with the optimization packages **DOT** and **DOC**. The interface is described in Chapter four (Figure 4.2), with the exception that, the design response is the magnitude of the total force applied to the object by the index finger. The interfaced program is utilized to find the muscle activation pattern which minimizes the differences between the current and the desired normal forces determined by equation (6.27). The optimization problem to be solved is then to minimize

$$Obj = (F_n^{index} - 0.7W)^2, \quad (6.28)$$

subject to

$$0 \leq \mu_1 \leq 1, \quad 0 \leq \mu_2 \leq 1, \quad \dots, \quad 0 \leq \mu_n \leq 1$$

where F_n is the magnitude of the total normal force applied by contacting phalanges of the index finger, and μ_1 to μ_n are the muscle activation levels.

Prior to executing the optimization, the hand and the object are positioned as the following: The object is placed in a fixed position, corresponding to the approximate final position of the object after a successful grasp. This approximate position was found to be where the object is supported by the proximal phalanx of the thumb and the tissue between the first and the second metacarpal bones (Figure 6.4). The thumb is held stationary in an extended and abducted position by stimulation of the extensor pollicis brevis (EPB) and the abductor pollicis brevis (APB). The extrinsic flexor muscles (FDP and FDS) are selected to provide the grasp force.

Figure 6.4) The object is placed in a fixed position where it is supported by the proximal phalanx of the thumb and the palm during optimization. The thumb is abducted and extended by activation of APB and EPB.

The initial activation level is selected such that the index finger is in contact with the object. If the finger loses contact, the optimization program is lost and is not able to search for the minimum. Finally, the optimization program is utilized to search for the required activation pattern. The optimization is repeated for a variety of cylindrical object diameters (4, 6 and 8 centimeters), and object masses (0.3, 0.9 and 1.8 kilograms, Table 6.1).

Diameter (cm)	Mass (Kg)	Fixed Position (cm)	
		X	Y
4.0	0.3	6.9	5.6
	0.9	6.9	5.6
	1.8	6.9	5.6
6.0	0.3	8.4	5.7
	0.9	8.4	5.7
	1.8	8.4	5.7
8.0	0.3	9.6	5.9
	0.9	9.6	5.9
	1.8	9.6	5.9

Table 6.1) The optimization program is utilized to search for the required activation pattern for a variety of cylindrical object diameters and object masses at fixed positions corresponding to the approximate final position of the object after a successful grasp.

Optimizations are repeated for the cases presented in Table (6.1) with the upper limit of the side constraints (muscle activation levels) equal to 0.4, instead of 1, as if only forty percent of the muscle strength is used. This models the lower strength of the muscles when they are electrically activated using FNS (Yamaguchi, 1990; Kilgore, 1994).

Based on the results of optimization, and the practical palmar grasp procedure often used in FNS applications (Kilgore, 1994), a series of dynamic simulations of palmar grasp are also performed (Table 6.2) with the upper limit of the activation

levels limited to a maximum of one hundred percent (cases 1 to 6) and forty percent (cases 7 and 8). The last two cases, 7 and 8, are repeated with a Zancolli tendon transfer.

Case No.	Diameter (cm)	Mass (Kg)	Initial Position (cm)		Req. Force (N)	EI activation (%)
			X	Y		
1	4.0	0.3	9.5	3.8	2.1	8.0
2	4.0	0.3	13.5	3.8	2.1	25.0
3	4.0	0.3	16.5	3.8	2.1	48.0
4	6.0	0.9	10.5	4.8	6.3	18.0
5	6.0	0.9	14.5	4.8	6.3	35.0
6	8.0	1.8	11.5	5.8	12.6	18.0
7	6.0	0.9	10.5	4.8	6.3	18.0
8	6.0	0.9	14.5	4.8	6.3	35.0

Table 6.2) Dynamic simulations are performed using a variety of object sizes, masses and initial positions. Position of center of mass of the object is given relative to the center of rotation of the wrist. The required normal force is computed from Equation (6.27). Prior to dynamic simulations, the index finger is positioned close to the object, by activation of EI. In case numbers 7 and 8, the upper limit of the activation levels were limited to 40% during optimization.

In each simulation, first the thumb is abducted and extended by activation of APB and EPB so that the thumb opposes the index finger. The index finger is extended by activation of EI such that when the object is placed in the predetermined position (Table 6.2), the index finger does not touch the object, but is very close to it.

This is performed only to prevent slapping of the object which may result in convergence problems in the numerical solution. The object is then placed at the predetermined position, and FDS is activated. The activation is brought up to the maximum (Table 6.4) in one second and is held at the maximum level to the end of the simulation. As a result, the index finger may slide the object and push it against the thumb and/or the palm. A *successful capture* is defined by the object being captured between the thumb (and/or palm) and the index finger. When the FDS activation level reaches its maximum value, FDP is stimulated up to the maximum (Table 6.4) in one second and is held at the maximum level to the end of the simulation. The object is *successfully held* when the normal force applied by the index finger is greater or equal to the required force computed from equation (6.27).

C. Results

When the finger pads are loaded and unloaded, hysteresis is clearly seen. As in the previous chapter, the loading and unloading curves of the hysteresis loop for each finger pad are fitted by exponential functions of the form $F = ae^{bf}$, where F is the loading, F_l , or unloading force, F_u , and a and b are constants of regression (Table 6.3). The coefficients for the part of the palm which is mostly in contact with the object, that is the tissue between the first and second metacarpal bones, are estimated from the

coefficients of the finger pads and based on the fact that this tissue is more compliant than the finger pads.

Finger Pad	Coefficients of Regression					
	$F = ae^{bd}$					
	a_l	b_l	r^2	a_u	b_u	r^2
PALM	0.06	7.0	----	0.02	8.0	----
TPP	0.078	9.225	0.953	0.0231	10.003	0.961
TDP	0.0599	8.395	0.993	0.0218	8.803	0.980
IPP	0.0337	10.247	0.980	0.0030	12.414	0.980
IMP	0.0689	12.217	0.956	0.0134	14.480	0.978
IDP	0.0503	15.029	0.924	0.0094	17.330	0.965

Table 6.3) Coefficients of regression of the loading and unloading force-displacement curves of the palm, thumb and index finger pads. The coefficients for the palm are estimated from the coefficients of the finger pads.

The results of optimization for normal and electrically activated muscles show that as long as FDS can provide the required normal force, FDP is not activated (Table 6.4). When activation of FDS reaches its highest level (100% in normal and 40% in electrically activated muscles), FDP is stimulated to produce adequate grasp force. When the object is relatively heavy (1.8 Kilograms), the electrically activated muscles cannot provide a successful hold where normal muscles can. It is also observed that as the weight of the object increases, the required activation level increases. In contrast,

the activation level decreases when the size of the object increases. In other words, thick light cylinders are easier to grasp than slender heavy objects.

Object's Diameter (cm)	Object Mass (Kg)					
	0.3		0.9		1.8	
	Muscle Activation Level (%)					
	FDS	FDP	FDS	FDP	FDS	FDP
4.0	32.4	0	77.9	0	100	45.3
6.0	21.3	0	61.2	0	100	25.2
8.0	17.3	0	49.5	0	100	17.3
4.0	32.4	0	40	36.9	Not Sufficient Normal Force	
6.0	21.3	0	40	22.2		
8.0	17.3	0	40	18.6		

Table 6.4) Muscle activation levels of FDP and FDS required to produce adequate normal force to hold a cylindrical object in palmar grasp with variety of object sizes and masses. The first three rows are activation levels found for normal muscles, and the last three are activations determined for electrically stimulated muscles which are assumed to produce only 40% of the strength of a normal muscle.

The results of the dynamic simulations of palmar grasp using selected cases from Table (6.4) at different initial positions of the object's center of mass reveal that, in almost all cases, the cylindrical object is successfully grasped and adequate normal force to the object is obtained (Table 6.5). The total residual moment about the center of mass of the object was never equal to zero. The final posture of hand and the final position of the object, in the cases simulated, show that by activation of FDS the

object is dragged into the palm and palmar grasp prehension, instead of palmar pinch prehension, is achieved (Figures 6.5 to 6.12, except Figure 6.7). In the case where a relatively small object ($D = 4$ cm) is placed close to the tip of the index finger, the tip of the distal phalanx, which is modeled as a cylinder penetrates the object resulting in an unsuccessful palmar grasp, but probably a successful palmar pinch (Figure 6.7).

Case No.	Muscle Activation (%)		Final Position (cm)		Grip Force (N)	Residual Moment (N-cm)
	FDS	FDP	X	Y		
1	32.4	0	8.0	4.5	3.6	1.0
2	32.4	0	8.0	4.4	3.7	0.8
3	32.4	0	Unsuccessful Palmar Grasp			
4	61.2	0	8.4	5.4	7.2	3.8
5	61.2	0	8.2	5.8	6.5	0.4
6	100	17.3	9.3	6.1	13.0	3.9
7	40	22.2	8.5	5.3	7.4	2.2
8	40	22.2	8.2	5.8	6.7	1.6

Table 6.5) Results of dynamic simulations of palmar grasp with different object masses, sizes and initial positions (Table 6.1). Muscle activation in each case is determined by the results of optimization (Table 6.4). Successful grasp and hold is achieved for all the cases except case number three where a small object is placed close to the tip of the index finger. In each case, the magnitude of the total residual moment produced about the center of mass of the object is recorded. In case numbers 7 and 8, the upper limit of the activation levels was limited to 40% during optimization.

Simulations of cases 7 and 8 with the Zancolli tendon transfer reveals that if the object is placed relatively close to the palm (Figure 6.13), a more convenient grasp will be achieved.

D. Discussion

Modeling and simulations of palmar grasp of a cylindrical movable object revealed that, for the object sizes used in this study, extrinsic muscles are sufficient to grasp and hold cylindrical objects using the palm for support. The protocol which has to be followed is to abduct and extend the thumb by activation of APB and EPB, extend the index finger, move the hand toward the object to place it as proximal as possible to the palm, activate FDS to grasp the object and then, if necessary, activate FDP to provide adequate grip force. The residual moment about the center of mass of the object may be balanced by flexion of the other fingers, and/or by the moment due to the friction forces between the phalanges and the object through the abduction/adduction movements of the thumb and/or the index finger by activation of the intrinsic muscles.

To simplify modeling of the ground-object interaction, three degrees of freedom, rotations about X and Y axes and translation in the Z direction, are constrained. However, in practice, a moment due to the friction force on the bottom of the object and the force from a contacting digit may tilt the object about a vector in the

sliding plane. If the contact point between the index finger and object is above the center of mass of the object, the moment due to the finger force is added to the moment due to the ground-object friction force. The scenario is worse if the thumb contacts the object below its center of mass. The simplified model does not include the effect of such a possible rotation of the object. However, if all the fingers other than the index finger are flexed, the ones which are below the center of mass of the object may counterbalance the residual moment.

The residual moment may also cause trouble after the object is successfully grasped, that is captured between the thumb, index finger and/or the palm. A necessary condition to lift an object, without tilting, is that it is in static equilibrium, that is the sum of forces and moments about the object center of mass should be equal to zero. At first, the multi-objective optimization method (available in the employed optimization packages, DOT and DOC) was used to search for the muscle activation pattern which provides this condition. The goal was to minimize the difference between the magnitude of the applied normal force to the object by the index finger and the required normal force from Equation (6.27). At the same time, the magnitude of the total moment about the center of mass of the object has to be minimized, too. The objective function was then

$$Obj = [a(F_n^{index} - 0.7W)^2 + b(T_{\zeta}^{TSF} - T_{\zeta}^{index})^2]^{1/2},$$

where a and b are the weighting coefficients, and T_c^{thumb} and T_c^{index} are the magnitude of the moments about the center of mass of the object due to the opposing thumb and palm and the index finger forces, respectively. Unfortunately, the program rarely converged to a satisfactory solution. The reason is apparently due to the fact that multi-objective optimization method works based on "compromise programming" (DOC reference manual), where the minimum of the composite objective function is somewhere in between the minimum of each term of the objective function. Therefore, it may provide the most reasonable compromised result, but the result does not satisfy the static equilibrium requirements. Weighting the terms of the objective function could move the solution toward the minimum of one term, but would move it away from the minimum of the other term.

The muscle activation level required to hold an object was found using optimization programs which work with the static model of the index finger. One reason for using the static model is the fact that the grip force relaxes from the instantaneous value to its elastic value as time increases. Relatively large differences between the loading and unloading curves in finger pad loading experiments reveals that the difference between the instantaneous and the elastic response is significant. The grip forces in our dynamic simulations, right after the object is grasped, are larger than the required forces to hold the object, but they relax to the required values as time goes by. Activation levels found based on the instantaneous value of the grip force underestimate the required force, and may not be sufficient to hold an object for

relatively long times. The grip force relaxation is also reported by Johansson and Westling (1981). Their measurements show that most of the relaxation happens in first twenty seconds after the grasp force is applied.

It was shown that the required muscle activation decreases with increasing cylinder diameter. When the object diameter increases, the index finger makes contact with the object in a less flexed posture, increasing length of the flexor muscles. Due to the length-tension property, increasing the muscle length increases the isometric force the muscle can produce, which requires a lower muscle activation level. In addition, when the object diameter increases, the index finger makes contact with the object in a less flexed angle (with respect to the rest position), where the flexion passive moment is larger. Thus, less muscle activation level is needed to grasp larger objects. This observation is valid only when the position of the object is such that increasing the size decreases the joint flexion angles with respect to the rest positions of the joints.

The final grip forces presented in Table (6.5) are larger than the required grip forces in Table (6.2), resulting in an increased safety margin. The reason for this might be the fact that the required activation levels are determined by a different protocol using the optimization method. In that protocol, the object position is fixed and the static model is utilized to compute for the external forces. The final position of the object (Table 6.5), after it is grasped, is often different from the fixed position of the object (Table 6.1).

Our major intention in this part of the study was to predict performance of hand in palmar grasp not palmar pinch prehension. In Chapter four, it was shown that in palmar pinch prehension, before grasping the object, the intrinsic muscles are required to provide an appropriate posture if the object is free to move. It was shown previously by Hendrix (1993) that if the object is fixed to the ground, tip pinch or palmar pinch can be achieved through the use of extrinsics alone in the size range of cylinders tested. If the object is allowed to move, a more precise model of the distal phalanges of the thumb and the index finger are to be used.

At last but by no means least, it should be noticed that the results of this chapter are only applicable to the normal hand with normal passive moment characteristics. If the abnormal passive moments were implemented in the model, the extensor muscles could not open the hand to place most of the objects between the fingers, thumb and the palm. In fact, prior to any successful prehensile grasp, the hand should obtain a proper posture. In Chapter four, it was also observed that because of the hyper flexion of the abnormal joints (with shifted passive moments to the flexion side of the subject KT), even with appropriate tendon transfers, the muscles could hardly provide the proper posture. These facts reveal the importance of the passive moment in the posture of the hand, and suggest that to restore normal functions in a quadriplegic person with hyper flexed joints, the normal passive moments must first be restored.

Figure 6.5) Case number one (Tables 6.2 and 6.5). The cylinder (diameter = 4 Cm. and mass = 0.3 Kg) is successfully grasped between the index finger and the palm. Initial, final and two intermediate pictures of the hand motion are presented.

Figure 6.6) Case number two (Tables 6.2 and 6.5). The cylinder (diameter = 4 Cm, and mass = 0.3 Kg) is successfully grasped between the index finger and the palm. Initial, final and two intermediate pictures of the hand motion are presented.

Figure 6.7) Case number three (Tables 6.2 and 6.5) The cylinder (diameter = 4 Cm, and mass = 0.3 Kg) is not successfully grasped. The tip of the index finger is penetrated into the object. Initial, final and two intermediate pictures of the hand motion are presented.

Figure 6.8) Case number four (Tables 6.2 and 6.5). The cylinder (diameter = 6 Cm, and mass = 0.9 Kg) is successfully grasped between the index finger and the palm. Initial, final and two intermediate pictures of the hand motion are presented.

Figure 6.9) Case number five (Tables 6.2 and 6.5). The cylinder (diameter = 6 Cm, and mass = 0.9 Kg) is successfully grasped between the index finger and the palm. Initial, final and two intermediate pictures of the hand motion are presented.

Figure 6.10) Case number six (Tables 6.2 and 6.5). The cylinder (diameter = 8 Cm, and mass = 1.8 Kg) is successfully grasped between the index finger and the palm. Initial, final and two intermediate pictures of the hand motion are presented.

Figure 6.11) Case number seven (Tables 6.2 and 6.5). The cylinder (diameter = 6 Cm, and mass = 0.9 Kg) is successfully grasped between the index finger and the palm. Initial, final and two intermediate pictures of the hand motion are presented.

Figure 6.12) Case number eight (Tables 6.2 and 6.5). The cylinder (diameter = 6 Cm, and mass = 0.9 Kg) is successfully grasped between the index finger and the palm. Initial, final and two intermediate pictures of the hand motion are presented.

Figure 6.13) Case number seven (Tables 6.2 and 6.5) repeated with Zancolli tendon transfer. The cylinder (diameter = 6 Cm, and mass = 0.9 Kg) is successfully grasped between the index finger and the palm. Initial, final and two intermediate pictures of the hand motion are presented.

Figure 6.14) Case number eight (Tables 6.2 and 6.5) repeated with Zancolli tendon transfer. The cylinder (diameter = 6 Cm, and mass = 0.9 Kg) is grasped between the index finger and the thumb. Initial, final and two intermediate pictures of the hand motion are presented.

CHAPTER SEVEN

CONCLUSIONS AND FUTURE WORK

A. Conclusions

Several main conclusions may be drawn from the results of this study. The direct conclusions from modeling itself are

1. There is a large variation of joint passive moment among normal subjects and between normal and quadriplegic joints tested.
2. The dissipative part of joint passive moment is relatively significant compared to the elastic component, and has significant effect on dynamic movements of the hand. The dissipative part is a non-linear function of both joint position and joint rotational speed.

3. The dynamic inertial torques have significant effect in hand movements and muscle activation patterns.
4. Optimization is an effective tool for solving an inverse static problem using the forward static model of the hand.

From the application of the hand model in prehensile and non-prehensile hand functions the following conclusions can be drawn:

1. The extrinsic muscles of the thumb are adequate to position the thumb into a tip or palmar pinch posture, however intrinsic muscles of the index finger are required to provide a convenient tip or especially palmar pinch posture.
2. For the tip and palmar pinch postures, the modified Zancolli tendon transfer is more effective than Zancolli transfer.
3. Neither Zancolli nor modified Zancolli tendon transfers are very effective on patients with hyper-flexed joints.
4. An inconvenient lateral pinch posture will be obtained if the MP joint of the thumb is free to hyper-extend and the ADP is activated, or the IP joint is free to hyper-flex and the FPL is simulated.

5. Activation of ADP, FPL, and APB all together may provide a convenient lateral pinch, even though the IP joint is free to hyper-flex and the MP joint is free to hyper-extend.

6. In palmar grasp of movable cylindrical objects, larger diameter, light cylinders require less muscle force to grasp than slender, heavy objects.

7. A hand with normal joints but weak muscles can perform palmar grasp of moveable, cylindrical objects of various sizes and weights. Following the proposed protocol (Chapter six) provides more convenient palmar grasp.

Overall, it was confirmed that joint passive moment is a dominant factor in the hand's potential to reach required postures. This fact was strongly supported as more hand functions were simulated.

B. Future work

Chapters Two, Three, and part of Chapter Four were concentrated on the generation of the hand model and hand positioning program. Three applications of the model were investigated; non-prehensile palmar pinch, prehensile lateral pinch, and prehensile palmar grasp. In addition to the applications presented in here, for the future applications, the prehensile tip and palmar pinch of circular and flat walled objects may need be modeled. This requires a more precise geometric model for the distal phalanx

of the index finger, instead of a simple cylinder. The interaction model may then need to be modified to include the interaction of the distal phalanx of the index finger with different types of objects. As was performed in Chapters Five and Six for lateral pinch and palmar grasp, new interaction models can easily be linked to the hand model.

Future evaluations of the model are needed. A set of experimental tests can be performed to measure the position and velocity of the joints during a non-prehensile movement of the hand, and the grip force during a prehensile function, when specific muscles are stimulated. The results may be used to better evaluate the function of selected muscles in dynamic functions of the hand. The model can be further evaluated by comparison between the experimental and the simulated results.

A serious post-surgical problem in tendon transfer is tendon-to-tendon and/or tendon-to-bone adhesion. Models of these two types of adhesions can predict possible post-surgical complexities and their effect on the hand movements. Moreover, the hand model lacks the presence of Lumbrical muscles. These tiny but important intrinsic muscles should be added to the already available muscles in the model. The index finger lumbricals originate from the FDP tendon. Therefore, despite the present approach in muscle modeling in which all the muscles are uncoupled, the index finger lumbricals should be coupled to the FDP muscle model.

To better predict one's hand function, it is necessary to implement parameters of that specific individual in the model. Simulations of the model can then be used to

predict the potential functions of the individual's hand. These include operating with a selected set of muscles, fused joints, and tendon transfers. In addition, simulation can be used to study the interaction of objects with the individual's hand with both open and closed loop implementations of FNS. Although, the simulation can not fully replace subject testing, the model can be used in place of experimental tests on subjects to make initial evaluations of muscle function and a range of functional improvements that could be obtained in a comprehensive FNS program. It can be utilized to quickly identify rehabilitative procedures that are most likely to result in functional improvement, thereby narrowing the number of procedures tested in clinical trials.

REFERENCES

- Amis, A.A. (1987) Variation of finger forces in maximal isometric grasp tests on a range of cylinder diameters. *J. Biomed. Eng.*, **9**, 313-320.
- An, K.N., Ueba, Y., Chao, E.Y., Cooney, W.P. and Linscheid, R.L. (1983) Tendon excursion and moment arm of index finger muscles. *J. of Biomech.* **16**, 419-425.
- An, K.N., Cooney, W.P., Chao, E.Y. and Linscheid, R.L. (1978) Functional strength measurement of normal fingers. In: *Advances in bioengineering*, N.Y., ASME, 89-90.
- An, K.N., Chao, Y.S., Cooney, W.P. and Linscheid R.L. (1983) Biomechanics of the hand: A basic research study, Submitted to: The American Academy of Orthopaedic Surgeons for Consideration of the Kappa Delta Award.
- An, K.N., Chao, Y.S., Cooney, W.P. and Linscheid R.L. (1985) Forces in the normal and abnormal hand, *J. Orthop. Res.*, **3**, 202-211.
- Andrews, J.G. and Youm, Y. (1979) A biomechanical investigation of wrist kinematics. *J. of Biomech.* **12**, 83-93.
- Armstrong, T.J. and Chaffin, D.B. (1978) An investigation of the relationship between displacements of the finger and wrist joints and the extrinsic finger flexor tendons. *J. of Biomech.* **11**, 119-128.
- Bahler, A.S. (1968) Modeling of mammalian skeletal muscle. *IEEE Trans. on Biomed. Engr.* **15**, 249-257.
- Bawa, P., Mannard, A. and Stein R.B. (1976) Predictions and experimental tests of a visco-elastic muscle model using elastic and inertial loads. *Biol. Cybern.* **22**, 139-149.
- Benedict, J.V., Walker, L.B. and Harris, E.H. (1968) Stress-strain characteristics and tensile strength of unembalmed human tendons. *J. of Biomech.* **1**, 53-63.
- Blanton, P.L. and Biggs, N.L. (1970) Ultimate tensile strength of fetal and adult human tendons. *J. of Biomech.* **3**, 181-189.
- Brand, P.W. (1985) *Clinical Mechanics of the Hand*. The C.V. Mosby Co., St Louis.

- Brumbaugh, R.B., Crowninshield, R.D. and Blair, W.F. (1982) An *in vivo* study of normal wrist kinematics. *J. of Biomech. Engin.* **104**, 176-181.
- Buchholz, B., Wells, R. and Armstrong, T.J. (1987) The influence of object size on grasp strength: Results of a computer simulation of cylindrical grasp. 12th Mtg. of ASB, Univ. of Illinois, Sept. 28-30.
- Buchholz, B. and Armstrong T.J. (1992) A kinematic model of the human hand to evaluate its prehensile capabilities. *J. Biomech.* **25** N2, 149-162.
- Chao, E.Y. and An, K.N. (1977) Graphical interpretation of the solution to the redundant problem in biomechanics, ASME, Paper No. 77-Bio-1, 1.
- Chiel, H.J., Crago, P., Mansour, J.M. and Hathi, K. (1992) Biomechanics of a muscular hydrostat: a model of lapping by a reptilian tongue. *Biol. Cybern.* **67**, 403-415.
- Cooney, W.P., Lucca, M.J., Chao, E.Y.S. and Linscheid, R.L. (1981) The kinesiology of the thumb trapeziometacarpal joint. *The J. of Bone and Joint Surgery* **63-A**, 1371-1381.
- Crago, P.E. (1992) Muscle input/output model: the static dependence of force on length, recruitment and firing period. *IEEE Trans. on Biomed. Eng.* **39**, 871-874.
- Darling, W.G. and Cole, K.J. (1990) Muscle activation patterns and kinetics of human index finger movements. *J. Neurophys.*, **63**, 5, 1098-1108.
- Delp, S.L., Loan, J.P., Hoy, M.G. and Zajac, F.E. (1990) An interactive graphics-based model of the lower extremity to study orthopaedic surgical procedures. *IEEE Trans. Biomed. Engr.*, **37**(8), 757-767.
- Frankenberg, E.P. (1990) Experimental evaluation of tendon moment arms at the MCP and PIP joints of the hand. M.S. Dissertation, Case Western Reserve University.
- Fung, Y. C. (1981) *Biomechanics Mechanical Properties of Living Tissues*. Springer-Verlag, NY.
- Gear, C.W. (1971) Simultaneous Numerical Solution of Differential and Algebraic Equations. *IEEE Transactions on Circuit Theory* **18**, No. 1, 89-95.
- Gielen, C.C.A.M. and Houk, J.C. (1984a) Nonlinear viscosity of human wrist. *J. Neurophys.* **52**, 3, 553-569.

- Gielen, C.C.A.M. and Houk, J.C., Marcus, S.L. and Miller, L.E. (1984b) Viscoelastic properties of the wrist motor servo in man. *Ann. Biomed. Engr.*, **12**, 599-620.
- Hatze, H. (1976) The complete optimization of the human motion. *Math. Biosc.* **28**, 99-135.
- Hatze, H. (1980) *Myocybernetic control models of skeletal muscle*. Univ. of South Africa Press, Pretoria.
- Haut, R. C. and Little, R. W. (1972) A constitutive equation for collage fibers. *J. Biomech.* **5**, 423-430.
- Hayes, K.C. and Hatze, H. (1977) Passive visco-elastic properties of the structures spanning the human elbow joint. *J. Appl. Physiol.* **37**, 265-274.
- Hendrix, L. A. and Mansour, J. M. (1993) Functional grasp potential of the intrinsic minus hand. *IEEE Trans. on Rehab. Engr.*, **1**, 3.
- Hendrix, L.A. (1993) M.S. Thesis, Dept. of Mechanical and Aerospace Engin., Case Western Reserve University.
- Hill, A.V. (1938) The heat of shortening and the dynamic constants of muscle. *Proc. of the Royal Soc.* **126**, 136-195.
- Hill, A.V. (1953) A reinvestigation of two critical points in the energetics of muscular contraction. *Proc. of the Royal Soc.* **141**, 503-510.
- Ketchum, L.D. and Thompson, D.E. (1985) An experimental investigation into the forces internal to the human hand. In *Clinical Mechanics of the Hand*. (Edited by Brand, P.W.), The C.V. Mosby Co., St Louis.
- Kieth, M.W., Peckham, P.h., Thrope, G.B., Stroh, K.C., Smith, B., Buckett, J.R., Kilgore, K.L. and Jatich, J.W. (1988) Implantable functional neuromuscular stimulation in the tetraplegic hand. *J. Hand Surg.*, **14A**, 524-530.
- Kieth, M.W. (1994) Personal communications.
- Kilgore, K.L., Peckham, P.H., Keith, M.W. and Thrope, G.B. (1988) Electrode characterization for functional application to upper extremity FNS. *IEEE Trans. in Biom. Engin.* **37**, 12-21.
- Kilgore, K.L. (1994) Personal communications.
- Landsmeer, J.M.F. (1961) Study in the anatomy of articulation 1. the equilibrium of the 'intercalated' bone. *Acta Morph. neerl. scand.* **3**, 287-303.

- Lemay, M. (1994) Personal communications.
- Levin, A. and Wyman, J. (1927) The viscous elastic properties of muscle. *Proc. of the Royal Soc.* **101**, 218-243.
- Lieber, R.L., Fazeli B.M. and Botte, M.J. (1992) Architecture of selected muscles of the arm and forearm. *J. Hand Surg.*, **17**(5), 804-809.
- Long, C. (1968) Intrinsic-extrinsic muscle control of the fingers. *J. Bone Jt. Surg.*, **50-A**, 973-984.
- Long, C. (1981) Electromyographic studies of hand function. *The Hand, Volume I* (Edited by Tubiana, R.), W. B. Saunders, Philadelphia.
- Mansour, J. M., Rouvas, C., Sarangapani, J., Hendrix, L. and Crago, P. E. (1994) Quantitative functional anatomy of finger muscles: application to controlled grasp. In *Advances in the Biomechanics of the hand and Wrist* (Edited by Schuind, F., An, K., Cooney, W.P., and Garcia-Elias, M.), Plenum Press, New York.
- Mills, O.S. and Hull, M.L. (1991) Apparatus to obtain rotational flexibility of the human knee under moment loads *in vivo*. *J. Biomech.* **24**, 6, 351-369.
- Napier, J.R. (1956) The prehensile movements of human hand. *J. of Bone and Joint Surg.* **38B**, 902-913.
- Ou, C.A. (1980) The biomechanics of carpometacarpal joint of the thumb. PhD Dissertation, Louisiana State University.
- Peckham, P.H., Keith, M.W. and Freehafer, A.A. (1988) Restoration of functional control by electrical stimulation in the upper extremity of the quadriplegic patient. *J. of Bone and Joint Surg.* **70-A**, 144-148.
- Radwin, R.G., Oh, S., Jensen, T.R. and Webster, J.G. (1992) External finger forces in submaximal five-finger static pinch prehension. *Ergonomics* Vol. **35**, No. **3**, 275-288.
- Ritchie, J.M. and Wilkie, D.R. (1958) The dynamics of muscular contraction. *J. of Physiol.* **143**, 104-113.
- Rouvas, C. (1989) The development of a mechanical model of the hand. M.S. Thesis, Case Western Reserve University.
- Seireg, A. and Arvikar, R. (1989) *Biomechanical analysis of the musculoskeletal structure for medicine and sports*. Hemisphere Publ. Co., N.Y.

- Such, C.H., Unsworth, A., Wright, V. and Dowson, D. (1975) Quantitative study of stiffness in the knee joint. *Ann. rheum. Dis.* **34**, 286-291.
- Sugi, H. (1972) Tension changes during and after stretch in frog muscle fibers. *J. of Physiol.* **255**, 237-253.
- Taylor, C.P.S. (1969) Isometric muscle contraction and the active state: An analog computer study. *Bioph. J.* **9**, 759-789.
- Thomas, D.H. (1965) The physical properties of the human finger. PhD thesis, Case Institute of Technology.
- Wells, R.P., Ranney, D.A. and Keller, A. (1985) The interaction of muscular and passive elastic forces during unloaded finger movements: A computer graphics model. *Biomech. Current and Interdisciplinary Research* (Edited by Perren, S. M. and Schneider, E.), 743-748, Martinus Nighoff.
- Westling, G. and Johansson, R.S. (1981) Coordination between grip force and vertical lifting force when lifting objects between index finger and thumb. *Soc. Neurosci. Abstr.* **7**, 247.
- Westling, G. and Johansson, R.S. (1984) Factors influencing the force control during precision grip. *Exp. Brain Res.* **53**, 277-284.
- Winter, D. A. (1980) Overall principle of lower limb support during stance phase of gait. *J. Biomech.* **13**, 923.
- Woo, S. L-Y., Simon, B. R., Kuei, S. C. and Akeson, W. H. (1980) Quasi-linear viscoelastic properties of normal articular cartilage. *J. Biomech. Engr.*, **102**, 85-90.
- Wright, V. and Johns, R.J. (1961) Quantitative and qualitative analysis of joint stiffness in normal subjects and in patients with connective tissue diseases. *Ann. rheum. Dis.* **20**, 36-45.
- Wu, C-h., Houk, J.C., Young, K-Y. and Miller, L.E. (1990) Nonlinear damping of limb motion. *Multiple Muscle Systems: Biomechanics and Movement Organization* (Edited by Winters, J. M. and Woo, S. L-Y.), 214-235, Springer-Verlag, New York.
- Yamaguchi, G.T. and Zajac, F.E. (1990) Restoring unassisted natural gait to paraplegics via functional neuromuscular stimulation: A computer simulation study. *IEEE Trans. Biomed. Engr.*, **37**, 886-902.

- Yoon, Y.S. and Mansour, J.M. (1982) The passive elastic moment at the hip. *J. Biomech.* 15, 12, 905-910.
- Youm, Y., McMurty, R.Y. and Flatt, A.E. (1978) Kinematics of the wrist. I. An experimental study of radial-ulnar deviation and flexion-extension. *The J. of Bone and Joint Surgery* 60A, 423-431.
- Zajac, F.E., Topp E.L. and Stevenson, P.J. (1986) Musculotendon actuator models for use in computer studies and design of neuromuscular stimulation systems. *RESNA 9th Annual Conf.*, Minneapolis, MN, 442-444.