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Tendon Force during Occupational Hand Activities

Final Progress Report

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ABSTRACT

The overall goal of this study was to determine the dose-response relationships of fingertip load to tendon load in order to provide guidelines for hand tool design and tool use to minimize tendon loading, and thereby reduce the risk of developing tendon related disorders. The specific goals of this study were to investigate the effects of finger and wrist posture, external force distribution, and fingertip force rate on *in vivo* forces in the flexor digitorum profundus (FDP) and the flexor digitorum superficialis (FDS) tendons of the index finger and to evaluate the ability of a biomechanical model to predict these tendon forces. This information can identify finger positions and motions that are associated with lower tendon forces and may be used to design tools and tasks to decrease risk of tendon overuse injury and improve rehabilitation strategies. Tendon forces were recorded with buckle force transducers and fingertip forces were measured with a load cell in 15 adults during carpal tunnel surgery while hand position was recorded with a video camera.

Flexor tendon to fingertip force ratios depend on positions of the metacarpophalangeal and distal interphalangeal joints as well as the direction of applied external force when subjects isometrically increase their fingertip force from 0 to 15 N. For the same fingertip force, FDP force can be reduced by using a pulp pinch posture and increasing MP joint flexion. FDS force can be reduced in a tip pinch posture and extended MP position.

A static, three-dimensional biomechanical model poorly predicts these *in vivo* tendon forces demonstrating the importance of validating models before they are used to plan prevention strategies.

Applying fingertip force at higher rates (15 vs. 1.5 N/s) does not significantly increase FDP or FDS force ratios during isometric pinch tasks. Thus, reducing fingertip loading rate, within this range, may not reduce risk of injury.

During active, unresisted finger flexion or extension, FDP force is higher when the fingers are in a flexed posture and the wrist is either in a flexed or neutral position while FDS force is higher only when the fingers and wrist are both in a flexed posture. Therefore, tendon forces can be reduced by limiting finger and wrist flexion.
These findings may be used to develop guidelines for reducing tendon forces in order to design strategies for the prevention of tendon disorders of the hand.

SIGNIFICANT FINDINGS

The effects of fingertip loading conditions and hand posture on *in vivo* forces in the flexor digitorum profundus (FDP) tendon and the flexor digitorum superficialis (FDS) tendon of the index finger were investigated during isometric pinching tasks and unresisted finger motion.

The force in both flexor tendons was linearly related to the force simultaneously applied at the fingertip during each isometric task that consisted of increasing the fingertip force from 0 to 15N at a specific rate while the finger was in a natural pinching posture. On average, the force in the FDP tendon was approximately 2.5 times larger than the force at the fingertip while the FDS tendon force was 1.5 times the fingertip force. These ratios of tendon forces to external force were influenced by the position of the finger. Up to 36% of tendon force variation was attributed to MP joint angle, DIP joint angle and the direction of the external force with respect to the fingertip. Force in the FDP tendon per unit fingertip force was higher when the MP joint was extended, the DIP joint was flexed and the external force was applied more parallel to the fingertip, a tip pinch position. In contrast, FDS force ratio was greatest when the MP joint was flexed, the DIP joint was extended and the external force was more perpendicular to the fingertip, a pulp pinch position.

The predictions of a static, three dimensional biomechanical model agreed with these findings regarding the effect of MP joint angle on FDP and FDS tendon force ratios, but differed regarding the influence of the DIP joint. In addition, the model predicted and the measured FDP tendon force ratios for individual trials were not related, even though the MP joint angle, PIP joint angle, DIP joint angle, and external fingertip force direction and position assumed by subjects during each trial were used in the model. Surprisingly, the model predicted and measured FDS force ratios were negatively related. The mean FDP to fingertip force ratio predicted by the model was significantly higher than the mean ratio measured *in vivo* while the mean predicted and measured FDS ratios were not significantly different. The rate of force applied at the fingertip (1.5 N/s to 15 N/s) during the high precision, isometric pinch task did not significantly affect FDP or FDS tendon to fingertip force ratios.

Joint postures also influenced forces in the flexor tendons generated during active, unresisted, composite finger flexion and extension. Mean FDP tendon forces varied with hand posture between 1.3 and 4.0 N while mean FDS forces ranged from 1.3 to 10.7 N. FDP force increased as the fingers were flexed when the wrist was in either a neutral or a flexed position while FDS force increased with finger flexion only when the wrist was also flexed. The similarity in tendon force magnitudes measured during finger flexion and extension was unexpected.

TRANSLATION OF FINDINGS

Since high forces applied cyclically to the flexor tendons may lead to tissue damage, the risk of injury may be reduced by decreasing these internal forces. These forces may be lowered by adjusting external loading conditions in the workplace through improved tool and workstation design. Because flexor tendon forces are directly related to applied fingertip forces, reducing the amount of force required to perform a task will lower tendon forces.

In addition, changing finger joint position and angle of external force application can reduce the amount of tendon force necessary to apply a desired fingertip force. If reduction of FDP tendon to tip force ratio is the objective, then DIP flexion angle should be decreased, the external force should be applied more perpendicularly to the fingertip and MP flexion angle should be increased. If the goal is to decrease FDS tendon to fingertip force ratio and the angle between the fingertip and applied force is greater than 61 degrees, then DIP angle should be increased. If external force angle is less than 61 degrees, then DIP angle should be reduced. These findings apply to finger postures where MP joint angles range between 12° and 39° (mean minimum and maximum angles), DIP joint angles between 14° and 37° and the angle between the external force and fingertip is between 57° and 79°. Since the FDP generates more force than the FDS per unit fingertip force, the FDP may be more susceptible to injury. Thus adjusting posture to reduce FDP force may be more important in tool design. If this is the case, the design should promote a larger MP flexion angle, smaller DIP flexion angle, and a more perpendicular external force application. These adjustments in finger and external force orientation may be achieved by changing the shape of hand tools. The findings of this study suggest that limiting active composite finger flexion to MP angles of 45° or less may be a method of avoiding high tendon forces.

In contrast to position, adjusting fingertip loading rate over the range evaluated here during isometric force application may not be important for reducing tissue forces when designing switches and similar devices.

During tasks with no or low resistance, limiting the range of finger flexion can prevent higher FDP forces while simultaneously limiting finger and wrist flexion angles can reduce FDS forces. The small difference in tendon force between flexion and extension suggests that active motion in either direction leads to similar tendon forces.

LIST OF ABBREVIATIONS

FDP	flexor digitorum profundus
FDS	flexor digitorum superficialis
TE	terminal extensor
RB	radial band
UB	ulnar band
LU	lumbrical
LE	long extensor
UI	ulnar interosseous
RI	radial interosseous
ES	extensor slip
MP	metacarpophalangeal
PIP	proximal interphalangeal
DIP	distal interphalangeal
IP	interphalangeal
EF	external force
T	tendon force or moment
J	passive joint constraint force or moment
R	external force or moment
PCSA	physiological cross-sectional area
MVC	maximum voluntary contraction
EMG	Electromyography
CTR	carpal tunnel release
pFDP	predicted FDP tendon to fingertip force ratio
pFDS	predicted FDS tendon to fingertip force ratio
MP Ext	MP joint angle at the end of extension
MP Flex	MP joint angle at the end of flexion
ANOVA	analysis of variance
RMANOVA	repeated-measures analysis of variance
AIC	Akaike's information criterion

TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

CHAPTER I:

INTRODUCTION

Strain injuries to tendons, nerves, and muscles of the hand and wrist are a well recognized problem in the workplace [1]. In 1998, pain and injury of hands and fingers prompted 6.9 million physician office visits while another 3.1 million visits were related to wrist symptoms according to the National Ambulatory Medical Care Survey [2, 3]. Twenty-three percent of injuries and illnesses that resulted in days away from work in 2002 affected the upper extremities [4]. Based on estimates from the 1988 National Health Interview Survey, 395,000 cases of tendinitis and similar tendon disorders of the hands, wrists, and elbows were medically diagnosed in the working population after the patients experienced prolonged hand discomfort [5].

Work-related musculoskeletal disorders affect a diverse group of workers in many fields including manufacturing, services, and retail trade, but the risk varies depending on the industry and occupation [4]. For example, one study showed that the annual incidence of tenosynovitis and peritendinitis in a meat-processing factory reached 25.3% in an occupation that required strenuous arm movements while it was only 1% among other employees [6]. In 2001, the National Research Council and Institute of Medicine Panel on Musculoskeletal Disorders and the Workplace performed an extensive literature review and concluded that work-related risk factors are associated with the occurance of upper extremity musculoskeletal disorders [1]. The Panel recommended that targeted interventions be introduced in the workplace since they can reduce the risks of disorders and that these preventive strategies should be scientifically designed.

1.1 Importance of Understanding Internal Tendon Forces

1.1a Chronic Tendon Disorders

Risk factors for tendon disorders of the hand and wrist include the amount of force applied by the fingers, exposure duration, sustained extreme hand postures, and the rate of repeated hand and finger motions. Studies have demonstrated an association between hand and wrist disorders and jobs that require high force and/or high repetition [7, 8]. In addition, an increased risk of upper extremity disorders and discomfort has been associated with non-neutral, extreme hand postures [8-11]. A relationship between the velocity of wrist motion during repetitive occupational tasks and a higher rate of upper extremity disorders has also been reported [11, 12]. A better understanding of the role of external loading parameters in the injury mechanism can help us develop targeted preventive measures to reduce the number of injuries.

The postulated mechanism for overuse injury involves the tendon's biological response to external loading conditions. The forces transmitted to tendons during everyday activities cause small strains that can disturb the microscopic fiber structure. Although under normal circumstances tendon cells can repair this microdamage and adapt the tendon to new mechanical stimuli, some repetitive strain levels may cause an amount of damage that exceeds the tenocyte rate of repair and may lead to inflammation or cell death. Eventually, the cumulative damage can result in tendon degeneration and rupture [13].

To better understand injury mechanisms and to develop effective prevention and treatment strategies, it is important to understand how external loads applied during the execution of occupational and other daily activities are related to internal tendon forces, where damage occurs. Knowledge about the effect of loading factors, such as finger posture and movement rate, on the relationship between internal and external forces may be used to improve tool and workstation design and thereby reduce tissue forces and associated injuries. For example, the amount of tendon force required to apply a desired force at the fingertip may be reduced by adjusting the orientation of the hand and external force by changing the shape of hand tools. Similarly, the rate of force application needed to activate switches and similar devices can be controlled to reduce tendon forces if the relationship between internal and external forces depends on rate.

1.1b Acute Tendon Injuries

Information about flexor tendon forces during different motions can also be used to develop more effective rehabilitation treatment protocols for acute tendon injuries. Flexor tendon lacerations are a common clinical problem and the technique selected to manage them will influence the recovery of finger function. Many versions of postoperative rehabilitation procedures that involve applying limited forces to repaired tendons are currently used in the clinic. The principle of rehabilitation has been to allow tendon excursion while limiting tendon forces initially, then gradually increasing both tendon excursion and force with time. Animal and clinical studies have demonstrated that immediate tendon mobilization after repair is beneficial to healing [14-17], but identifying an optimal rehabilitation procedure has proved elusive. Tendon excursions and low forces generated by passive or active finger movement may stimulate healing, prevent adhesion formation and improve repair strength, thus helping the patient regain an increased range of motion faster [14, 15, 17]. However, excessive force during finger motion can cause gap formation, poor healing, and even rupture of repair [18-20].

Currently, no consensus exists concerning the best type of motion or hand posture to use during rehabilitation [21]. A rehabilitation regimen of early passive finger flexion and active extension was introduced 30 years ago [22]. The wrist was positioned in a dorsal extension block splint and a rubber band was attached to the fingertip with a nail suture and tensioned to the volar forearm area. The rubber band provided force for passive finger flexion while the patient actively extended the fingers. Since that time, the original dorsal extension block splint has been modified to increase the range of finger motion [23, 24] and remains one of the most commonly used rehabilitation protocols [25]. The passive flexion – active extension procedure has also been supplemented with assisted controlled passive motion and/or an active flexion component. The other common form of rehabilitation is based on the work of Duran, with passive digital flexion and extension coupled with the use of a dorsal extension block splint [26]. More recently, some clinicians have advocated the use of early controlled active motion to improve finger function [27, 28]. The positions of the wrist and fingers are carefully controlled with a dorsal splint during these rehabilitation procedures to limit the amount of force in the repaired tendons. The wrist and metacarpal-phalangeal (MP) joints are maintained in various degrees of flexion during passive and active proximal interphalangeal (PIP) and distal interphalangeal (DIP) joint movements. In contrast, wrist extension and MP flexion have been advised when the fingers are actively held in cascade contact in a flexed position [21, 29]. A better understanding of forces in the flexor tendons during different finger motions and at different hand postures can aid in improving rehabilitation protocols.

1.2 Previous Investigations of Flexor Tendon Forces

1.2a Biomechanical Models

Complex biomechanical models have been developed to investigate flexor tendon forces during different static and dynamic tasks. Most models predict forces during static or quasi-static situations when the finger acts against an external load, usually representing pinching or grasping activities [30-35]. A dynamic, three-dimensional model to study tendon forces during unresisted finger flexion and extension has also been presented [36]. Static models suggest that both finger joint positions and external force distributions will influence the ratio between the external force applied at the fingertip and the force generated by the finger flexor muscles [30-32]. Dynamic models also predict changes in tendon forces as a function of finger position [36, 37]. Because the finger is an extremely complicated system, models include numerous and differing assumptions regarding finger structure and muscle activity and vary in complexity. They contain various representations of muscles and muscle properties, estimations of muscle moment arms, and force distributions in the extensor mechanism. Additional assumptions are introduced to calculate the unknown muscle and joint forces since the index finger is a redundant system where the number of unknowns exceeds the number of equations. Solutions have been obtained by either assuming a certain distribution of muscle forces

[31-33], by the systematic reduction method, [30, 38] or by using optimization based on hypothesized physiological criteria [33, 37, 39, 40] and the choice of solution method affects predictions [33, 39]. Since these complex models contain multiple assumptions about finger structure and muscle function, their predictive ability should be validated before they can be used with confidence to limit internal tissue forces to reduce injuries and improve treatment protocols.

1.2b Electromyography Studies

In addition to models, fine wire electromyography (EMG) has been used to estimate forces generated by finger muscles during different static and dynamic activities by inserting fine wires into the muscles to measure their electrical activity during force production. During isometric tasks, the flexor muscles as well as the extensors and intrinsic finger muscles all co-contract when either low (1 to 3 N) or high (28 N) forces are applied at the fingertip [35, 41] and muscle activation patterns depend on external force direction and finger position [35, 42]. For example, different muscle excitation levels were observed during palmar than during distal force production even though the finger remained in the same position [35]. Changes in extensor and interossei muscle activities exceeded 25% of reference MVC values. In another study, flexor digitorum superficialis (FDS) activity depended on finger joint position and was significantly lower during circle pinch than Boutonniere pinch when both low and high forces were applied [42]. During dynamic finger flexion, finger flexor and extensor muscles are also co-activated and their activity increases with increasing movement rate and frequency [43-45]. Studies also suggest that agonists and antagonists may act together to control more complex finger movements, such as fast motions, motions limited to one joint, or movements against a resistance [43, 46, 47]. These EMG studies indicate that the coordination of multiple finger muscles is necessary to control finger position and force and to provide the appropriate torques at all finger joints during the execution of both static and dynamic tasks. They also suggest that individual muscle contributions and roles may vary with force, finger posture, force direction, movement rate, and type of motion. However, EMG data only provides information about the relative activity of each muscle during a task. Absolute force contributions of different muscles cannot be compared and passive forces in the system cannot be evaluated. In addition, EMG measurements may be influenced by motion artifacts during dynamic motions, limiting their reliability.

1.2c In Vivo Measurements

Actual flexor tendon forces during different static and dynamic tasks and how they respond to changes in external loading conditions, such as finger position and applied force, can only be obtained with *in vivo* experiments. Since *in vivo* measurements are invasive and difficult to perform, little information about flexor forces is available.

The relationship between force at the fingertip and *in vivo* force in one or both flexor tendons has been measured experimentally during static loading. On average, the force in the flexor digitorum profundus (FDP) tendon was 8 times larger than the force simultaneously applied at the fingertip during a pinching task while the FDS tendon force was almost twice the fingertip force. [48] Another study reported higher FDS to fingertip force ratios that varied between 2 and 6 [33]. It also showed that FDS force ratios increased when the DIP joint was extended or hyperextended. These flexor tendon to fingertip force ratios contain large inter-subject variability, exceed model predictions and disagree with model predictions regarding the effect of DIP joint position on FDS force

[33, 48]. Finger postures corresponding to force measurements were not recorded in the first study, while only the effect of DIP joint angle on force in one flexor tendon was explored in the second study. Neither study controlled the rate of force application.

In vivo forces in the flexor tendons have also been measured during active and passive movements. During a keystroke, maximum FDS tendon force was 3 to 7 times larger than the fingertip force, exceeding model predictions [49]. Active and passive wrist motions resulted in similar, low forces in the FDP and FDS tendons [48]. On the other hand, forces in the two tendons were higher when one of the IP joints was actively flexed [48]. These studies included a limited number of subjects and the effects of finger joint positions on flexor forces were not investigated.

1.3 Dissertation Objectives

Additional information about *in vivo* tendon forces during different loading conditions can improve our understanding of how finger position, external force distribution and loading force rate influence flexor tendon forces, the associated motor control strategies, and distribution of forces among finger muscles. Detailed experimental data is also essential for validation and improvement of existing biomechanical models. In turn, these models can be used to predict tendon forces during a broader set of activities. Internal tendon forces can be reduced in the workplace and during tendon rehabilitation by applying both the empiric data and improved model predictions to specific tasks.

The goal of this dissertation is to explore the effects of external loading conditions on *in vivo* forces in the flexor digitorum profundus (FDP) tendon and the flexor digitorum superficialis (FDS) tendon of the index finger during isometric pinching tasks and unresisted finger motion.

The effects of finger posture on flexor tendon forces during an isometric pinching task will be investigated in Chapter II in 14 subjects. Specifically, the effects of distal interphalangeal (DIP) joint angle, proximal interphalangeal (PIP) joint angle, metacarpophalangeal (MP) joint angle, and the position and direction of external force at the fingertip will be explored.

The ability of a static, three dimensional model to predict flexor tendon to fingertip force ratios will be evaluated in Chapter III. Model force predictions based on finger postures assumed by subjects will be compared with corresponding tendon forces simultaneously measured *in vivo*. The effects of DIP joint angle, PIP joint angle, MP joint angle, and the position and direction of external force at the fingertip on predicted and measured forces will also compared.

The effects of the rate of force applied at the fingertip on flexor tendon forces during an isometric pinching task will be investigated in Chapter IV.

The effects of wrist and finger position on flexor tendon forces during active unresisted finger flexion and extension will be investigated in Chapter V. Force changes caused by increasing finger flexion will be examined and forces in a flexed wrist posture will be compared to forces in a neutral wrist posture. In addition, differences in forces between active finger flexion and active finger extension will be explored.

CHAPTER II:

EFFECT OF FINGER POSTURE ON FORCES GENERATED BY FINGER FLEXOR MUSCLES DURING AN ISOMETRIC TASK: *IN VIVO MEASUREMENTS*

2.1 ABSTRACT

Risk factors for activity related tendon disorders of the hand include applied force, repetition, exposure duration and sustained hand posture. Understanding the relationship between external loading conditions and internal tendon forces can elucidate their role in injury and rehabilitation. The goal of this investigation is to determine how finger posture and external force direction affects *in vivo* forces in the flexor digitorum profundus (FDP) tendon and the flexor digitorum superficialis (FDS) tendon during an isometric pinching task. Tendon forces, recorded with buckle force transducers, and fingertip forces were simultaneously measured during open carpal tunnel surgery as subjects (N = 14) isometrically increased their fingertip force from 0 to 15 N when the index finger was held in two different positions. Finger joint angles as well as the position and direction of the fingertip force were recorded with a video camera. MP joint flexion leads to a decrease in the FDP to fingertip force ratio and an increase in the FDS force ratio. FDP force ratio increases as the DIP joint is flexed and as the external force is applied less perpendicularly to the fingertip. The influence of DIP and external force angles on FDS force ratio is interdependent, but FDS force is greatest when the DIP joint

is extended and the angle between the fingertip and the external force is more perpendicular. These findings can help define finger positions that will reduce forces generated by the flexor tendons to apply a given fingertip force.

2.2 Introduction

Musculoskeletal disorders of the distal upper extremity are a well recognized problem in the workplace[1]. Risk factors for tendon disorders of the hand and wrist include the applied force, repetition, exposure duration and sustained hand posture. These injuries are associated with jobs that require high force and/or high repetition [7, 8]. In addition, several studies have demonstrated an association between non-neutral hand postures and an increased risk of upper extremity disorders and discomfort [8-11].

Because finger posture determines the moment arm of the external force (distance between point of force application on the finger and each finger joint), it will influence the amount of muscle force transmitted to the fingertip. For example, if flexor muscles generate the same force in all postures, maximum fingertip pinch forces should decrease as pinch span increases because the force moment arm increases. In fact, studies have demonstrated that maximum pinch force decreases as pinch width increases during both a five-finger pulp pinch [50] and a two-finger pulp pinch between the index finger and thumb [51]. Grip span also affects the force needed for object manipulation at submaximal force levels [52-54]. The moment equilibrium maintained around each joint during static tasks also suggests that the internal muscle force needed to apply a specific submaximal external force depends on posture. However, the moment arm of the external force is only one factor that will influence the relationship between fingertip and flexor muscle forces since the finger is a complex system with seven muscles acting over three joints. It is important to understand how external loads are related to internal tendon forces and how hand posture and the distribution of the external force on the hand affect the relationship in order to better understand injury mechanisms and to develop effective prevention and rehabilitation strategies.

Biomechanical models have been developed to predict the relationship between external forces applied by the fingers and internal tendon and joint forces during different static tasks. They suggest that both finger posture and external force distribution will influence the ratio between the external force applied at the fingertip and the force generated by the finger flexor muscles. For example, Chao et. al. predicted that FDP to fingertip force ratio ranges from 1.9 to 2.1 during tip pinch (flexed DIP) and from 2.5 to 3.1 during pulp pinch (extended DIP); the corresponding FDS to fingertip force ratios ranges were 1.8 - 2.2 and 0.3 - 1.3 [30]. Other investigators have also used a similar model to examine the effects of finger joint angles and external force position on estimated tendon forces [31, 32]. These models predict that DIP flexion will lead to a decrease in FDP tension and an increase in FDS tension while the influence of other joint positions is less clear. In contrast, a model with a DIP joint constraint moment developed based on experimental data predicted a decrease in FDS force with DIP flexion [33]. Unfortunately, the index finger is a complicated and redundant system where the number of unknown forces exceeds the number of available equilibrium equations. In order to calculate tendon forces, models include assumptions about either the distribution of muscle forces or about the body's optimization function. The choice of solution method affects predictions. In addition, other simplifying assumptions about finger structure are included in models.

Therefore, *in vivo* measurements are needed to determine the actual tendon to fingertip force ratios and to validate models. The relationship between force at the fingertip and *in vivo* force in one or both flexor tendons has been measured experimentally during static loading. The ratio of tendon to fingertip force was 7.9 ± 6.3 for the FDP tendon and 1.7 ± 1.5 for the FDS tendon during tip pinch. [48] However, finger position was not reported. Another study reported FDS to fingertip force ratios ranging from 1.7 to 5.8 [33] and showed that FDS force ratio increased from 2.4 ± 0.6 during tip pinch to 4.4 ± 1.4 during pulp pinch (DIP joint extended or hyperextended). The authors measured force in only one tendon and the effects of other joint angles on tendon force were not explored. The measured force ratios exceed model predictions and disagree with predictions about the effect of DIP joint position on force. Finger position may influence these ratios, the associated motor control strategies, and distribution of forces among the muscles of the finger.

The goal of this investigation was to determine how finger posture affects *in vivo* forces in the flexor digitorum profundus (FDP) tendon and the flexor digitorum superficialis (FDS) tendon during an isometric pinching task. Specifically, the effects of distal interphalangeal (DIP) joint angle, proximal interphalangeal (PIP) joint angle, metacarpophalangeal (MP) joint angle, and the position and direction of external force at the fingertip were explored.

2.3 Materials and Methods

2.3a Data Collection

Data acquisition took place during carpal tunnel surgery. The surgeon instrumented both flexor tendons of the index finger with buckle force transducers and each subject repeatedly pressed his or her finger against a load cell while the finger was in two different positions. Fourteen subjects (9 females and 5 males, average age 41 ± 10 years) who were scheduled for open carpal tunnel release surgery participated in the study after reading and signing a consent form. The Committee on Human Research from the University of California, San Francisco approved the procedures. Subjects had no previous index finger tendon injuries. Several days prior to surgery, the subjects practiced the experimental tasks in a setting that simulated the procedure during surgery. At that time, their hand length was measured and a radiograph of the index finger was taken in the sagittal plane with a calibration bead taped to the finger.

The experiment was conducted during open carpal tunnel release surgery with local anesthesia injected at the incision site. Thus, the subject retained motor control of the forearm muscles and intrinsic hand muscles throughout the procedure. After the flexor retinaculum ligament was released with a longitudinal incision and adequate access to the flexor tendons was gained, the FDP and FDS tendons of the index finger were isolated. Two gas-sterilized buckle force transducers were introduced into the field; one mounted on each flexor tendon of the index finger. The transducers were staggered within and slightly distal to the carpal tunnel so that they would not collide. The transducers were a modified version of a device previously described [55]. The basic design and dimensions were the same, but two additional strain gauges were added to produce a full bridge circuit for temperature compensation. In addition, the edges of the transducer's arches were rounded where they contact the tendon to decrease stress concentrations. The tendon thickness in the transducer was measured using a digital micrometer with a resolution of 0.01 mm (Series 575 Digimatic Indicator, Mitutoyo). The transducers were individually tested and calibrated using a previously described method [55]. An equation was calculated for each transducer that adjusted for tendon thickness and related transducer output to tendon force. The estimated mean errors for the calibration factor ranged from 3.8 to 7.3%. Each transducer's performance was evaluated prior to every surgery by supporting the device at the ends and hanging two weights from the fulcrum. The transducer output differed from the output obtained during initial calibration testing by an average of 0.7%, demonstrating that the calibration factor did not change over the course of the experiments.

After the transducers were inserted, the subject flexed the index finger against a load 20 times to seat the transducers onto the tendons. Then the tendon thickness was measured to enable accurate conversion of transducer output to tendon tension and the tourniquet was released to allow tissue reperfusion (mean tourniquet time was 30 ± 13 min). The subjects were supine with the shoulder abducted to 90 degrees during the procedure. A custom designed apparatus supported the load cell at the end of the index finger in order to achieve the desired hand positions. The hand was placed in the device with the thumb up, the palm facing the feet, and the wrist in 15 extension (Figure 2.1). The device allowed for unrestricted flexion at the wrist and each of the index finger joints and the finger was not restrained by the surgeon. The MP joint of the index finger was positioned by the surgeon in either 45 or 15 flexion using angle brackets while the PIP and DIP finger joints assumed a natural pinching position. The hand positioning apparatus was adjusted so that fingertip force was normal to the surface of a six-axis load cell (ATI Industrial Automation, Apex, North Carolina, USA) while the angle between the fingertip and load cell was determined by the pinching position selected by the subject. Applied force was recorded as the subjects pressed on the load cell's hard surface. Data was simultaneously collected from the tendon transducers and fingertip load cell at 100Hz using a laptop computer with an A/D board. The estimated centers of joint rotation of the index finger and wrist were marked with a surgical pen on the radial side of the hand. A video camera mounted above the operating field recorded finger position in the sagittal plane (30 frames/s).

Data was collected during isometric finger flexion approximately 12 minutes after the tourniquet was deflated to allow time for muscle recovery. The subjects were instructed to steadily increase the force on the load cell for 10 seconds until the fingertip force reached 15N; then decrease the force in the same manner. To help them attain the target force and force rate, subjects observed a computer monitor mounted above their heads that provided real-time feedback of fingertip force. Subjects were able to compare their force-time profile to the desired force profile and adjust their fingertip force to match the desired pattern. Each subject repeated the task several times when the finger was positioned with the MP joint in 45 flexion. Then the hand positioning apparatus was adjusted so the MP joint angle equaled 15 flexion and the procedure was repeated. After the tasks were completed, a final measurement of tendon thickness was taken, the transducers were removed, and the carpal tunnel surgery completed.

2.3b Data Analysis

The voltage output from the buckle force transducers was converted to tendon force by using the calibration equations calculated from *in vitro* testing and the measured tendon thickness. The forces measured with the load cell and buckle transducers were filtered using a sixth-order, low-pass Butterworth filter at a cutoff frequency of 25Hz. The total fingertip force was calculated from the acquired three orthogonal force components. On average, 96% of total fingertip force was directed perpendicular to the load cell surface. The start of each loading task was defined as the time when the fingertip force first exceeded 1 N (2.5N for subjects 3 and 8) and the end occurred when the force reached its maximum value. Tendon and fingertip forces corresponding to these times were extracted. For all the trials, the relationship between force in each tendon and force at the fingertip was defined as the slope of the line of tendon versus tip force, calculated with a linear regression. One to five good trials were selected for each subject at each finger posture for further analysis based on the linearity of the relationship between tendon and fingertip forces ($R^2 > 0.7$) and on how well the fingertip force corresponded to the target force profile.

A video frame, corresponding to the midpoint time of each selected trial, was captured and finger joint angles, as well as the position and angle of the fingertip with respect to the load cell, were measured (Adobe Photoshop). The MP and PIP joint angles were measured by connecting the points that marked centers of joint rotation at the wrist, MP, PIP, and DIP joints (Figure 2.2). A line parallel to the dorsal surface of the distal phalanx and passing through the DIP joint center of rotation was drawn. Another line perpendicular to the surface of the load cell and passing through the center point of contact between the finger and load cell was drawn. These two lines and the PIP joint marker were used to estimate the DIP joint angle and the angle between the finger and the external force applied on the load cell (EF angle). In addition, the distance between the DIP joint center and point of external force application on the load cell was measured in the distal direction (EF distance).

Hand size was represented by hand length and by the length of the middle phalanx of the index finger. Hand length was defined as the distance between the distal wrist crease and the tip of the long finger. Middle phalanx length was estimated from the radiograph as the distance between the center of a circle drawn around the phalanx head and the midpoint of an ellipse drawn inside the proximal concave articular surface (AutoCAD). The 5 mm ball bearing visible in the x-ray was used for calibration. For four subjects without radiographs, middle phalanx length was estimated using the relationship between hand and phalanx length established with a linear regression using data from 10 subjects where both measurements were available ($R^2 = 0.78$). Middle phalanx length was then used to normalize EF distance.

2.3c Statistical Analysis

Multiple trials for each subject at the flexed (MP joint in 45° flexion) and extended (MP joint in 15° flexion) finger positions were used to calculate averages, standard deviations, and coefficients of variation for individual positions and subjects. A two-tailed paired t-test ($p = 0.05$) was used to evaluate differences in posture parameters between the two finger positions. The means and standard deviations of posture parameters and force ratios were determined using individual subject means at the flexed and extended finger postures.

Data from all 91 trials (14 subjects, 2 postures, 1-5 trials per posture) was used to determine whether posture affected the ratios of FDP to finger force and FDS to fingertip force. A separate analysis was performed for each tendon. An univariate mixed procedure analysis (SAS software) was first used to test the individual effects of MP angle, PIP angle, DIP angle, external force angle and external force distance on tendon force ratios. Then, a best fit model was determined by performing the mixed procedure analysis with multiple parameters and their first order interaction terms. The goodness of fit was evaluated using Akaike's information criterion (AIC). [56]

2.4 Results

The average MP, PIP and DIP joint flexion angles at the flexed and extended finger postures as well as the sum of the three angles are reported in Table 2.1. Although the two postures vary among individual subjects, MP, PIP and DIP joint angles are all significantly different at the two positions. In the flexed finger posture, MP and PIP joints are more flexed while the DIP joint is more extended. In the flexed finger position, the external force is directed at a smaller angle with respect to the distal phalanx and is applied farther from the DIP joint than in the extended one (Table 2.1). During multiple trials at each finger position, most subjects reproduced a similar posture. Maximum angle differences between multiple trials at one position average: $3.6° \pm 3.1°$ for the MP joint, $7.3° \pm 5.5°$ for the PIP joint and $7.5° \pm 5.0°$ for the DIP joint. The mean middle phalanx length, defined as the distance between the center of DIP joint rotation and the center of the concave articular surface at the PIP joint, is 19.3 ± 3.3 mm.

The average FDP to fingertip force ratio across finger postures is 2.2 ± 0.6 while the average FDS to fingertip ratio is 1.5 ± 0.8. The mean force ratios for all subjects at two positions are listed in Table 2.2 along with their average variability between trials at each position for individual subjects and between different subjects. The coefficient of variation for all subjects is 0.27 for FDP force ratio and 0.52 for FDS, indicating greater variability in the force generated by the FDS. Between subject variability accounts for a greater percentage of the total variation in FDS force ratio than in FDP force ratio, especially in the flexed finger position.

The relationships between FDP and fingertip force and between FDS and fingertip force are linear as indicated by mean r-square values of 0.97 ± 0.05 and 0.94 ± 0.07. The relationship between FDP and FDS force ratios for each subject is displayed graphically in Figure 2.3 at the two different finger positions. Force in the FDP tendon is greater than in the FDS tendon and forces in the two tendons have a weak, but significant negative relationship ($R^2 = 0.17$ and $p = 0.03$ for all trials). This negative relationship between the force ratios in the two tendons is stronger for the extended finger position ($R^2 = 0.50$ and $p < 0.01$) while it is not significant for the flexed position ($R^2 = 0.04$ and $p = 0.53$).

The ratios between forces generated by the FDP and FDS tendons and the force applied at the fingertip during isometric force production tasks in two different index finger postures depend on DIP angle, MP angle and EF angle. For the univariate analysis of the FDP to fingertip force ratio, only DIP angle and EF angle have significant independent effects ($p = 0.01$ and $p = 0.05$) (Table 2.3). FDP tendon to fingertip force ratio increases as DIP flexion increases and as the angle between the fingertip and loading surface becomes more parallel to the fingertip. However, only 9% and 3% of the variability in FDP force ratio can be attributed to DIP and external force angles. The univariate model that includes only DIP joint angle fits the data better than models with DIP angle plus other parameters. In the best fit model with multiple posture parameters, only the combination of MP angle and EF angle are significant. FDP force ratio increases as external force angle is less perpendicular to fingertip and as the MP joint is extended. The interaction term of these variables is not significant. Although the effects of these posture components on FDP force are significant, they explain only 17% of the variance in the FDP to fingertip force ratio.

For the FDS force ratio, only MP joint angle and the DIP joint angle have significant influence ($p = 0.03$ and $p < 0.01$) when considered independently in the univariate analysis. FDS force ratio increases with MP flexion or DIP extension. However, only 1% and 13% of the variability in the force ratio can be attributed to the MP or DIP angles, respectively. The model that includes only DIP joint angle fits the data better than a model with both DIP and MP angles. The best fit multivariate model indicates that DIP and EF angles jointly influence FDS force ratio and explain 36% of its variance. Because the interaction term of these two variables is significant, the value of each one will influence the effect that the other one has on the FDS force ratio. FDS force ratio is highest when the external force angle is large (force is more perpendicular to fingertip) and when the DIP joint is extended. It is lowest either when the DIP joint is extended and the external force angle is small or when the DIP joint is flexed and the external force angle is large. (Figure 2.4) When the DIP joint angle is less than 34 degrees, FDS force ratio increases as the external force angle increases. However, when the DIP joint is in a more flexed position, FDS force ratio decreases with an increase in the external force angle. Conversely, when the external force angle is less than 61 degrees, an increase in DIP flexion leads to an increase in FDS force ratio while DIP angle has the opposite effect on FDS force when the external force angle is greater than 61 degrees.

2.5 Discussion

Ratios between forces generated by the two flexor tendons of the index finger and the force applied at the fingertip were examined when fingertip force increased linearly from 0 to 15N with the finger in two different natural pinching postures. Force in the FDP tendon is higher than in the FDS tendon for the same externally applied fingertip force. Forces in the two tendons are somewhat inversely related suggesting load sharing between the two flexors.

DIP joint angle, MP joint angle and the direction of the externally applied force with respect to the finger all have significant effects on FDP and FDS tendon to fingertip force ratios. Seventeen percent of FDP force variation is attributed to MP joint and EF angles, while 36% of FDS force variability is due to changes in DIP and EF angles. FDP force decreases as the DIP joint is extended and as the external force is applied more perpendicularly to the fingertip. Thus FDP force is lower in a pulp pinch than a tip pinch posture. On the other hand, the influence of DIP and EF angles on FDS force ratio is interdependent and complex, but FDS force is greatest when the DIP joint is extended and the angle between the fingertip and the external force is large, a position resembling pulp pinch. In addition, MP joint flexion leads to a decrease in FDP force while FDS force decreases with MP joint extension. Changes in these three finger posture parameters have opposite effects on forces in the two flexor tendons. DIP angle is the single variable with the largest influence on FDP and FDS force. In contrast, PIP joint angle and the distance between the external force and DIP joint have negligible effects on force in either tendon. The results indicate that the position of some joints as well as the external force direction must be considered to accurately describe the relationships between internal tendon forces and external fingertip force.

Several limitations of *in vivo* measurements must be considered when interpreting and applying the presented data. First, measurement errors are introduced by the use of buckle force transducers and during the conversion of their output to force. However, these errors are small relative to the forces measured (see methods section) and although they may affect the absolute force values, they should not influence the conclusions regarding the effect of finger posture on tendon to fingertip force ratios. Collection of data during carpal tunnel surgery is another limitation. The muscle forces may not accurately represent control strategies and movements executed during activities of daily living since subjects have lost finger sensory and proprioceptive feedback due to local anesthesia of the median nerve. However, the motor performance during surgery was similar to the practice trials, indicating that muscle activity may not have been altered by the surgery. In addition, 14 subjects is a limited sample size; a larger number of subjects and a larger set of postures would provide more complete characterization of the system.

Ratios between *in vivo* flexor tendon and fingertip forces have been measured previously during pinch tasks. The FDP to fingertip ratio of 7.9 ± 6.3 reported by Schuind et al. [48] during tip pinch is much larger and more variable than the mean value of $2.2 \pm$

0.6 in the present study. The FDS ratio of 1.7 ± 1.5, observed by their group, is similar to our results of 1.5 ± 0.8. Both these FDS to fingertip ratios are smaller than the ratio of 3.3 ± 1.4 measured in another study. [33] That study reported that FDS force ratio increases from 2.4 ± 0.6 during tip pinch to 4.4 ± 1.4 during pulp pinch, supporting our findings regarding the effect of DIP joint angle on FDS force ratio. However, comparisons between studies are difficult because those studies did not report joint angles. Differences in finger positions and experimental techniques may account in part for the difference between the values found in the literature and those from the current study.

In vivo measurements are difficult to perform; hence, biomechanical models have been developed to predict the effect of finger posture on tendon forces and externally applied forces. One model predicts an increase in FDP to fingertip force ratio (from 2.0 to 2.8) and a decrease in FDS to fingertip force ratio (from 2.0 to 0.8) as the DIP joint is extended from a tip pinch to a pulp pinch [30]. Other models offer similar predictions [31, 32]. These predictions disagree with our empiric findings regarding the effect of DIP position on tendon forces. However, another model that included a DIP joint constraint moment when the DIP joint was extended predicted the observed decrease in FDS force with DIP flexion. [33] Since finger postures used by subjects in our experiment were not identical to the postures employed for model predictions, comparisons are difficult. The position of other joints and the applied force may influence how DIP position affects tendon forces. In fact, our findings indicate that the effect of DIP angle on FDS tendon force ratio is simultaneously dependent on the angle between the external force and the fingertip.

Theoretically, as finger joint flexion increases, the flexor tendon should generate less force to apply a constant fingertip force. First, the moment arm of the external force around a joint decreases with joint flexion and second, the flexor tendon moment arm increases as the tendon moves farther from the joint center. Both these effects imply that less tendon force is required to balance the external force. Therefore, FDP force is predicted to decrease as DIP flexion increases and external force angle decreases, since it is the only muscle balancing the moment generated by the external force. In addition, as these changes in posture lead to a fall in FDP force, a rise in FDS force is predicted to maintain adequate total flexor force at the PIP and MP joints. However, our findings indicate that DIP and external force angles have the opposite effects on flexor tendon forces.

Several plausible explanations exist for this apparent contradiction. First, perhaps the external force moment remains constant since the external force is applied farther away from the DIP joint when the external force angle decreases (correlation coefficient equals 0.39). Second, Dennerlein et al. suggested that passive connective tissues may produce a joint constraint moment when the DIP joint is fully extended. [33] This passive moment may balance the externally applied load and reduce the required FDP force; however, additional analysis of our data revealed that an increase in DIP flexion is related to an increase in FDP force throughout the range of joint flexion. Therefore, the observed relationship cannot be explained solely by a joint constraint moment that exists with full DIP extension.

A third explanation is that when the DIP joint is flexed and the external force angle is less perpendicular to the fingertip, higher extensor or intrinsic muscle co-

contraction may be needed to maintain postural stability and provide the appropriate torques at all finger joints, leading to the observed increase in FDP force. Fine-wire electromyography studies have shown activity in all seven index finger muscles during a low force pinch task and during the production of static MVC forces. [35, 41] Different patterns of muscle excitation were observed during palmar and distal force production while the finger remained in the same position. [35] Although activity levels of the FDP and extensor digitorum were not influenced by DIP joint position in another study, the higher FDS activity observed when the DIP joint was extended supports our empiric findings [42]. Thus, the force produced by the other finger muscles involved in isometric pinch functions may influence FDP and FDS forces since their individual contributions are interrelated and vary with finger posture and external force direction.

The findings of this study demonstrate that changes in finger joint position and angle of external force application with respect to the fingertip alter internal flexor tendon forces during a static fingertip loading task. If reduction of FDP tendon to tip force ratio is desired, then DIP angle should be decreased, EF angle should be increased and MP angle should be increased. If the goal is to decrease FDS tendon to fingertip force ratio and the EF is greater than 61 degrees, then DIP angle should be increased. If EF angle is less than 61 degrees, then DIP angle should be reduced. The design of tools and work rehabilitation strategies can be improved based on this knowledge to reduce the risk of tendon disorders. The results of this study demonstrate the importance of *in vivo* measurements in validating existing biomechanical models and improving our understanding of the impact of external factors on internal forces in tendons.

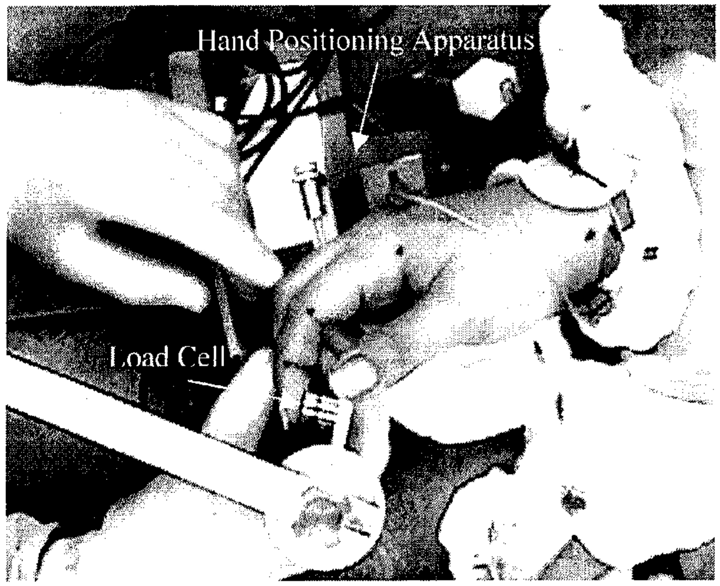


Figure 2.1. View of experimental setup from above. The subject's hand is stabilized in a thumb up posture using the hand positioning apparatus and the joint postures are recorded with a video camera. The fingertip force is measured with a six-axis load cell as the FDP and FDS tendon forces are recorded with two buckle force transducers (not visible). Estimated joint centers of rotation are marked.

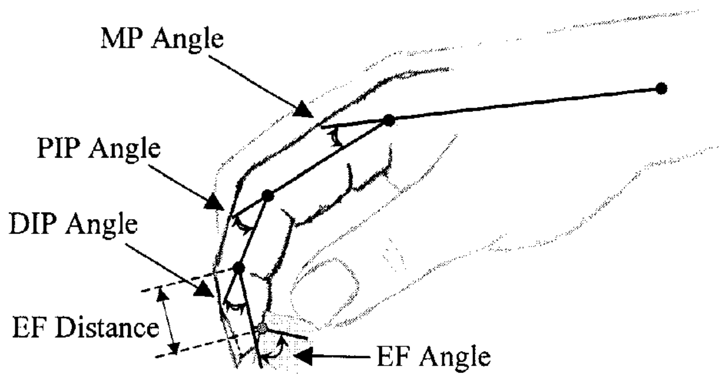


Figure 2.2. Finger posture parameters.
Black circles indicate the marked centers of joint rotation of the wrist, MP, PIP, and DIP joints. The gray circle indicates the center of contact region between the load cell and the finger. These points, the line parallel to the dorsal surface of the distal phalanx, and the line perpendicular to the load cell surface were used to estimate three finger joint angles, the angle between the fingertip and load cell (EF angle) and the distance between the external force point of application and the DIP joint in the distal direction.

	Finger Joint Posture		
	Flexed	Extended	p-values[2]
MP Angle	37° ± 12°	13° ± 11°	< 0.01
PIP Angle	54° ± 23°	33° ± 22°	< 0.01
DIP Angle	19° ± 23°	31° ± 20°	0.03
Sum of Joint Angles	110° ± 19°	78° ± 18°	< 0.01
External Force (EF) Angle	62° ± 10°	75° ± 13°	< 0.01
External Force (EF) Distance [1]	1.1 ± 0.5	0.8 ± 0.4	< 0.01

Table 2.1. Index finger and external force position during loading for two finger postures. Data reported as mean ± standard deviation for 14 subjects when their fingers were in "flexed" or "extended" finger posture.

1 External force distance is normalized to middle phalanx length

2 Paired t-test

	Mean		Standard Deviation		Coefficient of Variation	
	FDP	FDS	FDP	FDS	FDP	FDS
Between Subjects						
Between Trials in Flexed Position	2.07	1.54	0.76	1.01	0.37	0.66
Between Trials in Extended Position	2.33	1.40	0.63	0.76	0.27	0.54
Within Subject						
Between Trials in Flexed Position			0.35	0.22	0.20	0.18
Between Trials in Extended Position			0.32	0.30	0.16	0.28

Table 2.2. Mean FDP and FDS tendon to fingertip force ratios and mean within and between subject variability, expressed as standard deviations and coefficients of variation (N = 14).

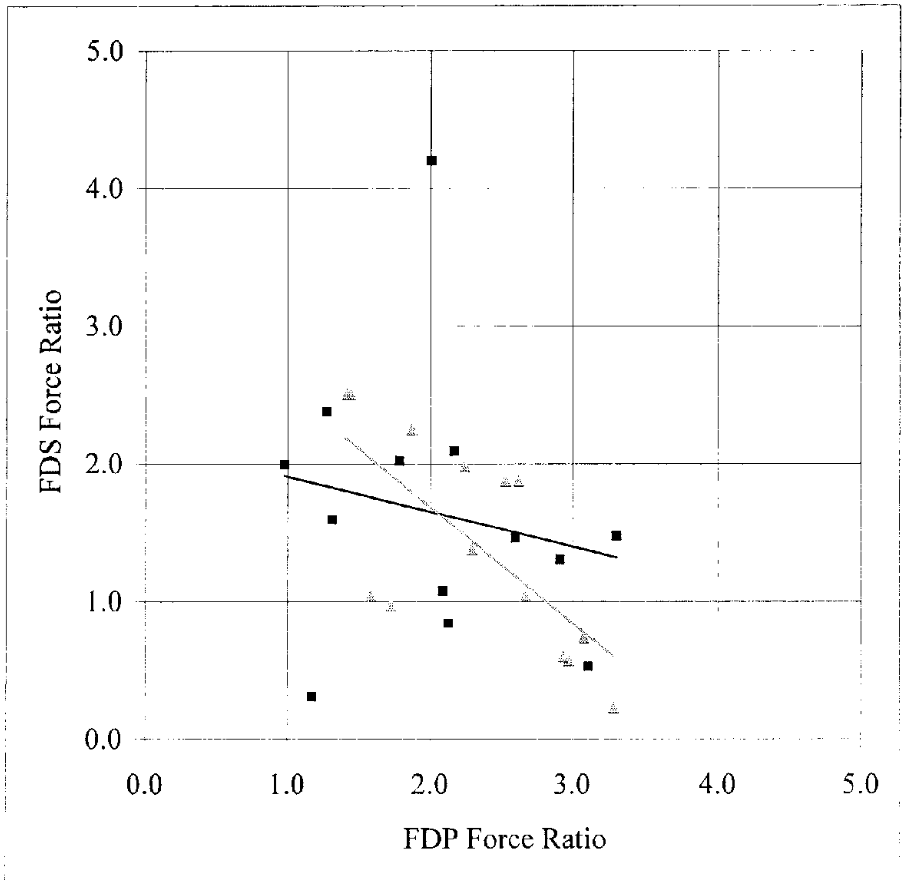


Figure 2.3. Relationship between FDP to fingertip force ratio and FDS to fingertip force ratio for each subject at two finger postures. The best fit lines for each finger position are also displayed.

- ■ Flexed Finger Posture
- △ Extended Finger Posture

FDP to Fingertip Force Ratio			
Univariate Analysis	P-value	AIC*	R^2
FDP = 2.315 - 0.006 * MP Angle	0.20	182	0.08
FDP = 2.335 - 0.004 * PIP Angle	0.36	183	0.00
FDP = 1.899 + 0.011 * DIP Angle	0.01	177	0.09
FDP = 2.847 - 0.010 * EF Angle	0.05	180	0.03
FDP = 2.003 + 0.183 * EF Distance	0.43	175	0.02
Forward Stepwise Model			
FDP = 4.252 - 0.024 * EF Angle - 0.019 * MP Angle	<0.01	177	0.17
FDS to Fingertip Force Ratio			
Univariate Analysis	P-value	AIC	R^2
FDS = 1.289 + 0.009 * MP Angle	0.03	190	0.01
FDS = 1.617 - 0.002 * PIP Angle	0.65	194	0.04
FDS = 1.898 - 0.015 * DIP Angle	0.00	183	0.13
FDS = 1.079 + 0.007 * EF Angle	0.20	192	0.14
FDS = 1.354 + 0.189 * EF Distance	0.42	186	0.04
Forward Stepwise Model			
FDS = -2.399 + 0.111*DIP Angle + 0.061*EF Angle - 0.002*DIP Angle*EF Angle	<0.01	155	0.36

Table 2.3. Effects of posture elements on FDP and FDS tendon to fingertip force ratios. Equations are shown for univariate analyses and for the best-fit forward stepwise model.

* AIC (Akaike's information criterion) indicates the goodness-of-fit of the model (smaller is better fit) and R-square indicates the predictive ability of the model.

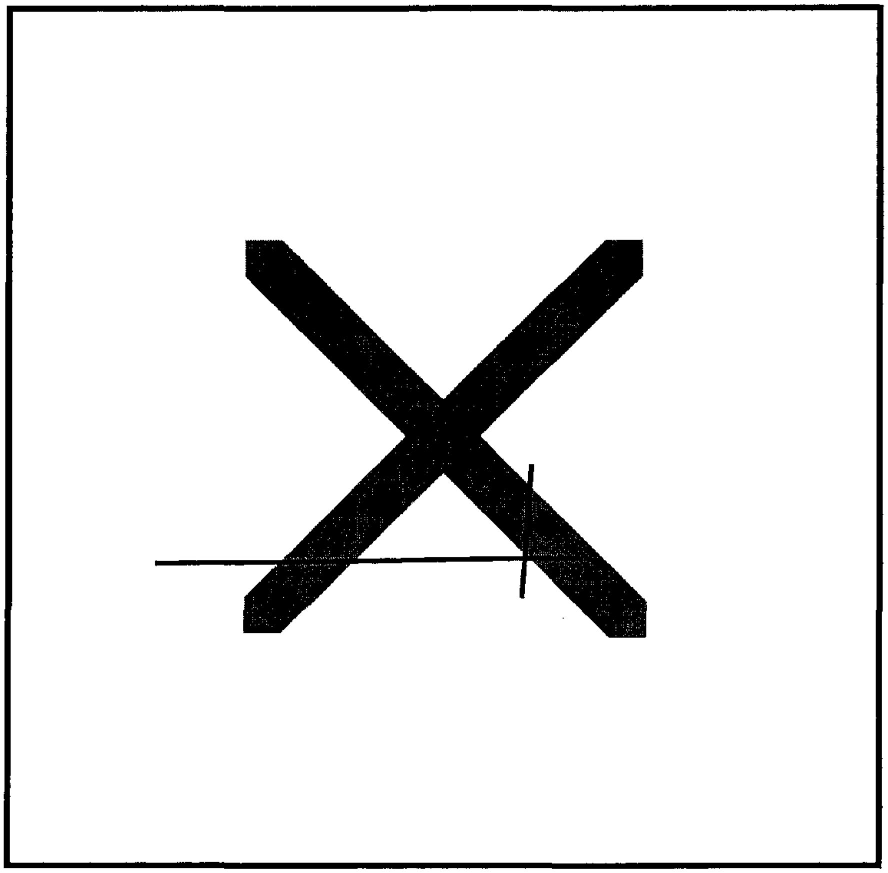


Figure 2.4. Relationship between predicted FDS to fingertip force ratio, DIP joint angle and EF angle, as described by the best-fit multivariate model. The two thick black lines, at DIP = 34° and EF = 61°, indicate the values where the effect of the other variable on FDS ratio changes direction.

CHAPTER III:

Effect of Finger Posture on Forces Generated by Finger Flexor Muscles during an Isometric Task: Biomechanical Model Validation

3.1 Abstract

Understanding the relationship between fingertip loading and internal tendon forces is important for designing tools to reduce the risk of tendon injuries and for improving rehabilitation strategies after tendon repair. Biomechanical models have been developed to predict ratios between internal tendon forces and fingertip pinch force, but due to the complexity of the system these models include many assumptions regarding finger structure and muscle activity. The goal of this investigation is to evaluate the ability of a static, three dimensional model to predict flexor digitorum profundus (FDP) and flexor digitorum superficialis (FDS) tendon to fingertip force ratios. Model predictions based on experimentally observed finger joint angles and external force distributions are compared with corresponding tendon forces measured *in vivo*. The mean predicted FDP to fingertip force ratio of 3.5 ± 1.0 is significantly higher than the mean measured ratio of 2.2 ± 0.8 ($p < 0.01$) while the mean predicted and measured FDS ratios are not significantly different (1.4 ± 0.9 and 1.6 ± 0.8; $p = 0.35$). For individual trials, the measured and predicted FDP ratios are not related ($R^2 \leq 0.02$); while the measured and

predicted FDS force ratios are negatively related ($R^2 = 0.12$). Employing different optimization functions does not substantially improve the predictive ability of the model. The model predicts lower FDP ratios and higher FDS ratios in a tip pinch compared to a pulp pinch finger posture. In contrast, measured FDP forces are lower and measured FDS forces are higher during pulp pinch. Thus, the conventional three-dimensional biomechanical model poorly predicted *in vivo* tendon forces.

3.2 INTRODUCTION

A better understanding of forces in the finger flexor tendons during common tasks can help improve prevention strategies and treatment protocols for tendon injuries. Several studies have demonstrated an association between tendon disorders of the hand and wrist and jobs that require high force and/or high repetition [7, 8, 57]. In addition, an increased risk of upper extremity disorders and discomfort has been associated with non-neutral, extreme hand postures [8, 10, 11, 58]. Forces applied by the fingers are directly related to internal forces in tendons, where injury may occur. The amount of tendon force needed to apply a specific external force is also related to finger joint posture and the orientation and position of the applied force. Knowledge of the effects of finger posture on the relationship between internal and external forces may be used to change tool and workstation design to reduce tissue forces and risk of injury.

Static biomechanical models that predict ratios between external forces applied by the fingers and internal tendon and joint forces have been used to investigate how changes in finger posture [30-34, 39], disease state [40, 59], and treatment procedure [40, 59] affect these ratios. Because the finger is a complicated system, models include numerous and differing assumptions regarding finger structure and muscle activity and vary in complexity. Equilibrium equations at each joint are used to describe the balance between external and internal forces and moments in either two or three dimensions. Muscles are usually represented as force generating elements, but their force-length and activation properties [34, 35, 59, 60] have been included in models that predict maximum external forces. Muscle moment arm lengths at different joint angles are either predicted from models that describe the relationship between tendon excursion and joint angle [34, 37, 40], obtained from the normative model that is based on distances between tendons and bones measured from x-rays [30-32, 61], or approximated by an average value experimentally calculated over the range of motion [35, 59, 60]. In addition to these variations, models also differ regarding their assumptions of force distributions in the extensor mechanism [30, 34, 35, 37, 39]. After the equilibrium and constraint equations are formulated, additional assumptions are necessary to predict the unknown muscle and joint forces since the number of unknowns exceeds the number of equations. Solutions have been obtained by either assuming a certain distribution of muscle forces [31-33], by the systematic reduction method, [30, 38] or by using optimization based on hypothesized physiological criteria [33, 37, 39, 40], such as minimizing the sum of muscle stresses squared [33, 37]. Some models have been used to predict the change in internal force per unit fingertip force due to changes in finger posture and direction and position of the external force [30-33, 39]. Others have been used to predict the muscle force distribution necessary to apply maximum external forces at different finger positions [34, 35].

Since these complex models contain numerous assumptions about finger structure and muscle function, their predictive ability should be validated before they are used with confidence to predict internal tissue forces. The limited *in vivo* tendon force data disagrees with model predictions regarding the effect of DIP joint position on FDS force and demonstrates large inter-subject variability [33, 48]. However, these experimental tendon force measurements are difficult to compare with model predictions because all finger joint angles and the external force distributions were not recorded and used in the models.

The goal of this investigation is to evaluate the ability of a static, three dimensional model to predict flexor digitorum profundus (FDP) and flexor digitorum superficialis (FDS) tendon to fingertip force ratios. Model force predictions based on finger postures assumed by subjects are compared with corresponding tendon forces simultaneously measured *in vivo*. The effects of distal interphalangeal (DIP) joint angle, proximal interphalangeal (PIP) joint angle, metacarpophalangeal (MP) joint angle, and the position and direction of external force at the fingertip on predicted and measured forces are also compared. The initial model is then modified to determine how changes in optimization functions influence its predictive ability. Finally, the importance of various finger structure and posture parameters in the model is evaluated with a sensitivity analysis.

3.3 Materials and Methods

3.3a In Vivo Data Collection and Analysis

The collection and analysis of *in vivo* flexor tendon force, fingertip force, and posture data was described in detail in Chapter II. Briefly, *in vivo* FDP and FDS tendon forces, recorded with buckle force transducers, and fingertip force were simultaneously measured during open carpal tunnel surgery as subjects (N = 14) isometrically increased their fingertip force from 0 to 15N when the index finger was held in two different positions. Finger joint angles (MP, PIP, and DIP) as well as the position and direction of the fingertip force were recorded with a video camera. See Chapter II for range of postures and forces.

3.3b Biomechanical Model

Eighteen equilibrium equations describe the balance between internal and external forces and moments acting at the DIP, PIP, and MP joints in three orthogonal directions. One degree of freedom (flexion-extension) is assumed at each of the IP joints and two degrees of freedom (flexion-extension and adduction-abduction) are allowed at the MP joint. The unknown internal forces acting at the joints consist of muscle forces transmitted by tendons and passive joint constraint forces and moments supplied by soft tissues (Table 3.1) [30]. The external force at the fingertip is treated as a point force located at the center of contact between the fingertip and load cell. Directions and moment arms of tendon forces at each joint are calculated for each trial based on finger posture and size. They are determined by first establishing six coordinate systems, one on the proximal and one on the distal bone of each joint as described by Chao et al. [30]. Finger segment proportions that correspond to distances between these coordinate systems are assumed to be constant and related to the length of the middle phalanx as reported by Chao et al. [30]. The directions of tendon forces and the lengths of tendon moment arms are calculated with the normative model of the hand [30] using experimentally measured finger joint angles and middle phalanx lengths. The normative model is based on the assumption that the distance between a tendon and a bone remains constant at points located on the proximal and distal sides of the joint. Average distances measured in cadaver specimens with x-rays were reported by Chao et al. [30] in three dimensions. The direction of the external force that was applied at the fingertip and its moment arm around each joint is calculated using coordinate transformations and the direction and position of the force with respect to the DIP joint, joint angles and lengths of finger segments. Joint flexion and fingertip force direction are assumed to be limited to the sagittal plane (i.e. MP joint is in 0° abduction, external force is applied on the radio-ulnar centerline of the finger and directed in that plane). The free-body force analysis yields 18 equations (3 dimensions and 3 joints) with 24 unknowns, in the following form:

$$\_\, T_i + J + E = 0$$

T_i = tendon force or moment

J = passive joint constraint force or moment

E = external force or moment

Four additional equalities describe the distribution of forces in the extensor mechanism of the index finger as reported in literature [30]:

$$T_{TE} = T_{RB} + T_{UB}$$

$$T_{RB} = 0.667*T_{LU} + 0.167*T_{LE}$$

$$T_{UB} = 0.333*T_{UI} + 0.167*T_{LE}$$

$$T_{ES} = 0.333*T_{LU} + 0.167*T_{LE} + 0.333*T_{UI} + 0.333*T_{RI}$$

In addition, inequality constraints define upper and lower bounds on muscle and joint forces. Axial joint constraint forces are assumed to be compressive while tendon forces are assumed to be tensional. An upper limit on tendon forces is defined by multiplying each muscle's PCSA [30] by the maximal muscle stress value of 35N/cm^2

[63]. The system of 22 equalities and 23 inequalities is solved by optimization using a quadratic programming routine in Matlab (MathWorks, Natick, Massachusetts). The sum of muscle stresses raised to the second power is minimized to predict all 24 unknown muscle forces and joint constraint forces and moments [33].

3.3c Statistical Analysis

Average predicted and measured FDP and FDS tendon to fingertip force ratios and their standard deviations were calculated using trials for which the model predicted tendon force ratios based on posture data. A paired t-test ($p = 0.05$) was used to evaluate differences between the measured and predicted ratios while r^2 values obtained with a linear regression represent the fit between them.

A mixed procedure analysis (SAS software) was used to determine the effects of MP angle, PIP angle, DIP angle, external force angle (EF angle) and external force distance (EF distance) on predicted tendon force ratios. A univariate analysis followed by a best-fit forward stepwise analysis with multiple variables were used to test how posture affects predicted forces; separate analyses were performed for each tendon. For multivariate analyses, first-order interaction terms were included. The goodness of fit was evaluated using Akaike's information criterion (AIC). [56]

3.3d Model Variations

The initial model was modified to determine if changing certain assumptions would improve the model's ability to predict measured tendon to fingertip force ratios. First, a passive dorsovolar DIP joint constraint moment was added to the equilibrium equations for trials when the DIP joint angle was extended or hyperextended (DIP angle $< 5°$)[32, 33]. For all other trials, the model was not altered. In addition, solutions to the set of equality and inequality constraints were obtained with different optimization functions (Table 3.4). Various combinations of muscle stresses, muscle forces, joint constraint forces and joint constraint moments were minimized.

3.3e Sensitivity Analysis

The influence of different model parameters on mean predicted tendon force ratios was assessed with a sensitivity analysis. Changes in mean force ratios caused by a 10% increase in moment arms of each tendon around each joint in 3 directions were calculated. Individual finger segment lengths (distances between coordinate systems) and muscle PCSA areas were also increased by 10% to determine their influence on tendon force ratios. The effect of errors in measured posture input parameters was also investigated. Finger joint angles and EF angle were increased by 5° while EF distance was increased by 10% of the length of the middle phalanx.

3.4 RESULTS

The mean FDP tendon to fingertip force ratio of 3.5 ± 1.0, predicted by a static three-dimensional model, is significantly higher than the mean ratio of 2.2 ± 0.8 measured *in vivo* ($p < 0.01$) (Table 3.2a). However, the predicted FDS mean of 1.4 ± 0.9 is not significantly different from the measured mean of 1.6 ± 0.8 ($p = 0.35$). The model generated results for 69 of 91 trials. No solutions were obtainable for trials when the DIP joint was in extension and the external force was applied far from the DIP joint. The measured and predicted FDP ratios for 69 trials (Figure 3.1a) and for subject means (Figure 3.1c) are not related ($R^2 \leq 0.02$); even though experimentally observed postures were used to generate corresponding biomechanical model estimates. Surprisingly, the predicted and measured FDS force ratios are negatively related for both individual trials and subject means (Figures 3.1b and 3.1d; $R^2 = 0.12$ and 0.16).

A mixed procedure statistical analysis used to analyze the effects of posture elements on predicted FDP tendon to fingertip force ratio (pFDP) indicates that pFDP increases with MP flexion, DIP extension, and EF distance when each of the variables is considered independently by univariate analysis (Table 3.3). The goodness-of-fit of the statistical model improves when more than one posture parameter is included in the analysis. The best-fit multivariate forward-stepwise analysis indicates that all three joint angles and the distance between the external force and the DIP joint influence predicted FDP tendon to fingertip force ratios. The predicted FDP ratio increases when the MP, PIP, and DIP joints are extended and when the force is applied more distally. The effect of DIP joint angle on FDP force depends on MP joint angle and vise versa since their interaction term (DIP Angle * MP Angle) is significant.

According to the univariate analyses, the predicted FDS to fingertip force ratio (pFDS) increases with increasing MP extension, PIP extension, DIP flexion, a more perpendicular external force application and a decrease in EF distance. The best-fit multivariate analysis indicates that pFDS ratio increases with MP flexion, PIP extension, DIP flexion, and a decrease in EF distance. The external force point of application influences the effect of PIP angle on FDS force and vise versa.

When a passive DIP joint constraint moment is added to the model for trials with DIP extension, the model predicts force ratios for 82 trials (Table 3.2b). The mean predicted FDP force ratio is lower than the one calculated with the original model while the FDS force ratio is higher. However, the mean predicted FDP ratio is still larger than the measured one ($p < 0.01$).

When different optimization functions are employed to solve the indeterminate system of equations, model predictions of flexor tendon forces do not improve substantially (Table 3.4). Mean predicted FDP force ratios, ranging from 3.3 to 20.2, are all significantly greater than measured ratios and the values are not correlated ($R^2 \leq 0.06$). On the other hand, no significant differences between measured and predicted FDS ratios exist when 18 of 32 functions are minimized. However, a negative relationship is always observed between measured and predicted FDS force ratios ($0.01 \leq R^2 \leq 0.25$). Similar forces are predicted by most of the optimization criteria. Mean predicted FDP force ratios range between 3.3 and 3.8 while FDS ratios are between 1.1 and 1.7 for 20 different optimization functions; 12 of these yield identical results. Minimizing the sum of squared muscle stresses and the sum of muscle stresses results in the same mean predicted forces while minimizing the sum of muscle forces yields slightly lower FDP force ratios and slightly higher FDS force ratios. Mean force ratios in both tendons are predicted to increase to minimize radio-ulnar shear joint forces, but change by less than 0.3 to minimize other joint forces. In addition, predicted FDP force ratio increases to minimize all joint constraint moments except the DIP axial joint moment.

A sensitivity analysis demonstrates which model input variables associated with finger structure and posture will have the greatest impact on predicted tendon to fingertip force ratios (Table 3.5). For example, a 10% increase in the FDP tendon moment arm around the DIP joint in the sagittal plane causes a 9.7% decrease in mean predicted FDP force ratio and a 27% increase in the FDS ratio. The lengths of the FDP and FDS tendon moment arm around the PIP joint also influence the predicted FDS force ratio. On the other hand, 10% increases in moment arms of most other tendons and of all tendons in other planes cause less than 1% change in mean force ratios. Similarly, altering the physiological cross-sectional area of muscles has a very limited effect on predicted forces (not shown). Changing finger proportions by increasing the size of individual finger segment lengths by 10% leads to changes in FDP force of less than 1%, but can change the FDS ratio by up to 16.6%. Finger joint angles and the direction and point of application of the external force also influence force ratios, especially the FDS. The factors that have the greatest influence on FDP force ratio are FDP tendon moment arms, DIP angle, EF angle, and the point of application of the external force. The factors that have the greatest influence on FDS force ratio are FDP tendon moment arms, PIP angle, EF angle, the point of application of the external force and the length of the middle phalanx.

3.5 Discussion

Forces in the two flexor tendons of the index finger per unit applied fingertip force are poorly predicted by a static, three dimensional biomechanical model during an isometric pinch. The predicted FDP force ratio is significantly higher than the one measured *in vivo*. The predicted and measured FDP tendon forces for individual trials are not related even though the MP joint angle, PIP joint angle, DIP joint angle and external fingertip force direction and position assumed by subjects during each trial were used in the model. Surprisingly, the predicted and measured FDS force ratios are negatively related. Employing different optimization functions to solve for unknown tendon forces does not improve the predictive ability of the model. Likewise, the fit between predicted and measured forces does not improve when a DIP joint constraint moment is added to those trials where the DIP joint was extended.

Sensitivity analysis indicates that flexor tendon moment arm lengths around the IP joints in the sagittal plane have a large influence on predicted tendon force ratios, especially on the FDS force. Since tendon moment arm lengths depend on both the subject's finger size and selected joint angle, the normative model was chosen to calculate moment arms for each trial because it accounts for joint angle and finger size and is based on experimentally measured tendon to bone distances [30]. The uncertainty in moment arm values estimated with this technique and corresponding changes in tendon forces are difficult to determine because only mean tendon locations were reported and they were only measured at one finger position. In addition to flexor tendon moment arms, finger segment proportions affect the predicted FDS force ratio. Although imprecise values of moment arm and finger segment lengths will influence predicted flexor tendon forces, the direction and magnitude of parameter errors for specific trials cannot be determined based on the available data. Therefore, adjusting these model inputs to improve predictions cannot be justified. Perhaps, better correspondence between measured and predicted forces could be attained with a more accurate representation of each finger for individual subjects and postures. More precise measurement of finger posture may also improve model predictions. Since the FDS tendon force ratio is more sensitive to changes in finger structure and posture variables, they will have a larger impact on FDS force ratio predictions than on predicted FDP forces.

The biomechanical model predicts that FDP force will decrease as MP joint, PIP joint and DIP joint flexion increase and as the external force is applied more proximally. When the joint is flexed, the moment arm of the external force around it decreases and concurrently the moment arm of the flexor tendon increases. As a result, less tendon force is required to balance the external force. Since only the FDP counteracts the external extension moment at the DIP joint, a decrease in FDP force is predicted when DIP joint flexion increases and EF distance decreases. At the same time, FDS force increases with increasing DIP flexion and more proximal force application to maintain the necessary total flexor force at the PIP and MP joints. An increase in FDS force is also predicted with MP flexion and PIP extension. The angles of all finger joints, as well as the position of the external force, simultaneously influence forces in the two flexor tendons and must be considered together to predict these forces. Previous studies have also predicted lower FDP to fingertip force ratios and higher FDS force ratios in a flexed, "tip pinch", finger position than in a more extended, "pulp pinch", posture [30-32].

These predictions can be compared to the effects of finger posture parameters on forces measured *in vivo* (Chapter II). Model predictions agree with *in vivo* findings regarding the effect of MP joint position of FDP and FDS tendon forces, but differ regarding the influence of the DIP joint. In contrast to model predictions, measured FDP force ratio decreases with DIP extension and thus is lower during pulp pinch than tip pinch posture. On the other hand, measured FDS force ratio is highest in a pulp pinch position when the DIP joint is extended and the external force is applied more perpendicularly to the fingertip. Another study also observed a higher *in vivo* FDS force ratio during pulp pinch when the DIP joint was fully or hyperextended [33].

Limitations of *in vivo* data collection during a surgical procedure must be considered. First, measurement errors are introduced by the use of buckle transducers, but these errors are small (4-7%) relative to the force values. Second, because posture data was acquired in two dimensions, the assumptions that joint flexion and fingertip force direction were limited to the sagittal plane and that the camera was parallel to this plane had to be introduced. A custom apparatus that supported the hand and load cell was designed to position the finger in this plane and minimize errors associated with these assumptions. In addition, muscle forces measured during carpal tunnel surgery may not accurately represent motor control strategies used during activities of daily living since subjects have lost sensory and proprioceptive feedback. However, subjects were provided with visual feedback and the task performance during surgery was similar to practice trials, indicating that muscle activity may not have been altered by the surgery. Another limitation is the number of postures tested per subject; a greater range of postures would have improved the model evaluation process.

The biomechanical model may not be able to predict tendon forces accurately because the index finger is a redundant system composed of the FDP and FDS muscles, two extrinsic extensors and three intrinsic muscles acting over three joints. Because the number of muscles exceeds the degrees of freedom, different combinations of muscle forces can generate the same external fingertip force and maintain the same finger posture at submaximal force levels. Flexor tendon forces will be influenced by forces in other muscles, especially the extrinsic extensors, during isometric pinch tasks since all seven finger muscles are involved in force production [35, 41]. One study suggests that variations in muscle activation between trials with identical position and force outputs are due to different motor control strategies [41]. Differences in muscle activation and control mechanisms between subjects may also explain the large percentage of force variability that could not be attributed to changes in finger posture in the present study. Because external force can be produced by different muscle force distributions, one optimization criteria may be inadequate for predicting muscle forces.

Posture may also influence the selected motor control strategy and affect model predictions. For example, additional co-contraction of extensors may be needed to stabilize the finger in a tip pinch posture versus a pulp pinch posture, explaining the increase in FDP force measured with increasing DIP flexion. EMG studies have demonstrated that muscle activation patterns do depend on finger position and external force direction [35, 42]. However, additional information regarding the relationship between finger posture and force produced by different finger muscles is needed to gain a better understanding of motor control strategies, their potential dependence on posture, and to define associated optimization functions.

The findings of this study emphasize the importance of validating biomechanical model predictions with *in vivo* measurements. Knowledge about the effect of external factors, such as finger posture and external force position and orientation, on internal tendon forces can be used to reduce the risk of tendon disorders. The impact of various factors and hand postures of interest on the relationship between tissue and applied forces can be investigated with models. However, their predictive ability must be confirmed and calibrated before they can be effectively used to reduce internal forces and improve the design of tools and work strategies.

Joint	Tendon Forces	Passive Constraints
DIP	terminal extensor (TE)	DIP axial compressive force
	flexor digitorum profundus (FDP)	DIP radio-ulnar shear force
		DIP dorsovolar shear force
		DIP axial moment
		DIP radio-ulnar moment
PIP	extensor slip (ES)	PIP axial compressive force
	flexor digitorum profundus (FDP)	PIP radio-ulnar shear force
	flexor digitorum superficialis (FDS)	PIP dorsovolar shear force
	radial band (RB)	PIP axial moment
	ulnar band (UB)	PIP radio-ulnar moment
MP	flexor digitorum profundus (FDP)	MP axial compressive force
	flexor digitorum superficialis (FDS)	MP radio-ulnar shear force
	radial interosseous (RI)	MP dorsovolar shear force
	long extensor (LE) *	MP axial moment
	lumbrical (LU)	
	ulnar interosseous (UI)	

Table 3.1. Unknown internal forces and moments. List of tendon forces and passive joint constraint forces and moments acting on the three joints of the index finger.

* long extensor tendon includes forces produced by the extensor digitorum communis and the extensor indicis muscles

	Measured	Predicted	Trials	R^2	P-values*
a) Original Model					
FDP	2.23 ± 0.75 (0.57 - 3.89)	3.51 ± 0.96 (1.34 - 6.23)	69	< 0.01	< 0.01
FDS	1.55 ± 0.77 (0.21 - 4.53)	1.40 ± 0.85 (0.02 - 4.17)	69	0.12	0.35
b) Model with DIP Joint Constraint Moment					
FDP	2.20 ± 0.75 (0.57 - 3.89)	3.32 ± 0.99 (1.06 - 6.23)	82	0.03	< 0.01
FDS	1.60 ± 0.90 (0.21 - 4.53)	1.73 ± 1.18 (0.02 - 7.77)	82	0.01	0.44

Table 3.2. FDP and FDS tendon to fingertip force ratios measured *in vivo* and predicted by the original model (a) and by the model with a DIP joint constraint moment (b). Data reported as mean ± standard deviation (range) for trials with model predictions.
* Two-tailed, paired t-test

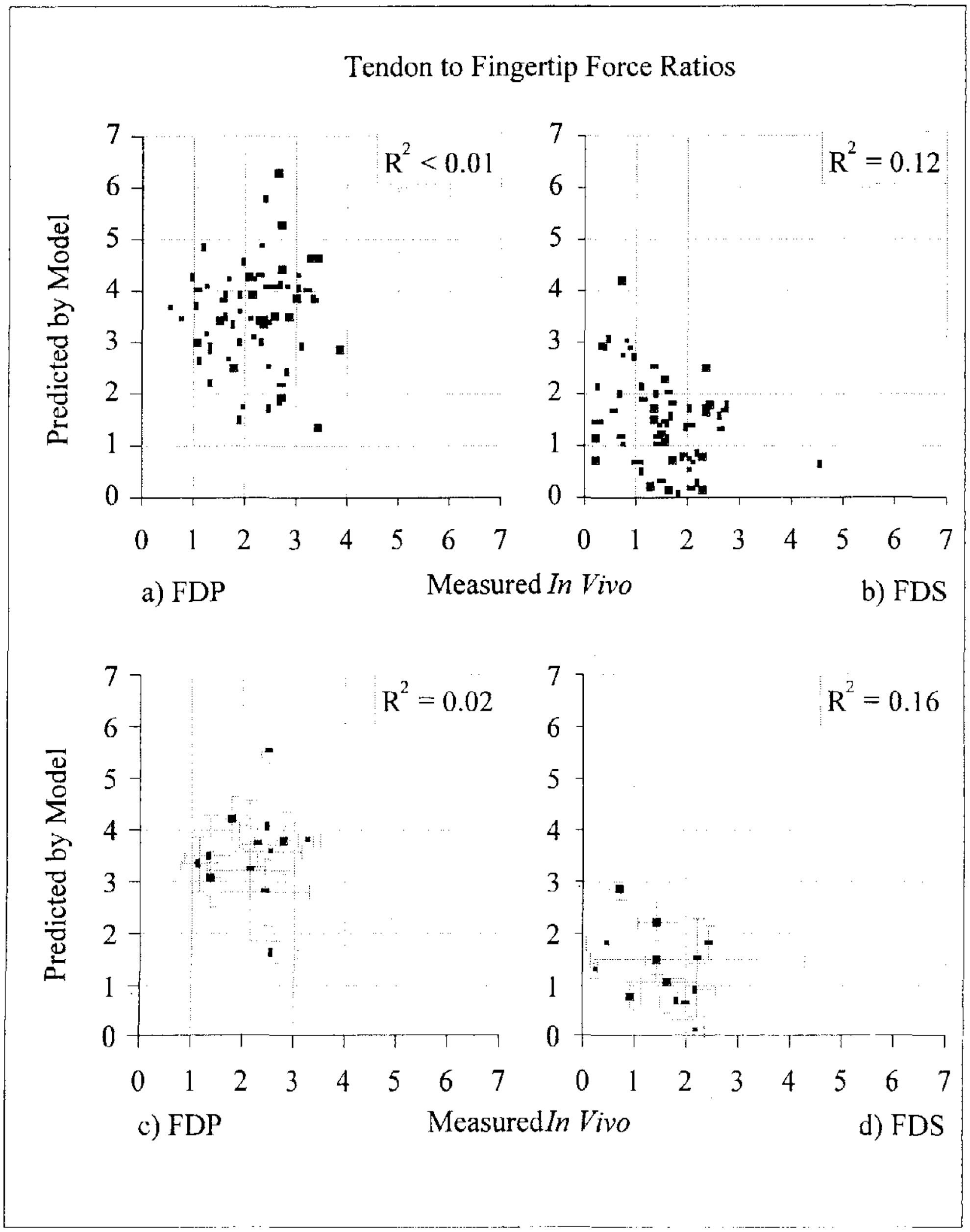


Figure 3.1. Relationship between tendon to fingertip force ratios measured *in vivo* and predicted by a biomechanical model for the FDP (a and c) and FDS (b and d) tendons. Figures a and b show all 69 trials with model predictions while means and standard deviations for 13 subjects are presented in Figures c and d.

Predicted FDP to Fingertip Force Ratio (pFDP)			
Univariate Analysis	P-value(s)	AIC*	R-square
pFDP = 3.278 + 0.013 * MP Angle	0.02	158	0.00
pFDP = 3.169 + 0.011 * PIP Angle	0.08	160	0.09
pFDP = 4.761 - 0.037 * DIP Angle	<0.01	143	0.11
pFDP = 3.726 - 0.002 * EF Angle	0.68	163	0.01
pFDP = 1.736 + 2.237 * EF Distance	<0.01	113	0.62
Forward Stepwise Model			
pFDP = 3.926 - 0.061*MP Angle - 0.026*PIP Angle - 0.031*DIP Angle + 3.791*EF Distance + 0.0004*DIP Angle*MP Angle	<0.01	26	0.64
Predicted FDS to Fingertip Force Ratio (pFDS)			
Univariate Analysis	P-value(s)	AIC	R-square
pFDS = 1.804 - 0.021 * MP Angle	<0.01	137	0.00
pFDS = 2.712 - 0.037 * PIP Angle	<0.01	90	0.63
pFDS = 0.131 + 0.037 * DIP Angle	<0.01	128	0.24
pFDS = 0.220 + 0.017 * EF Angle	<0.01	144	0.06
pFDS = 3.069 - 2.120 * EF Distance	<0.01	100	0.43
Forward Stepwise Model			
pFDS = 2.695 + 0.014*MP Angle - 0.041*PIP Angle + 0.030*DIP Angle - 2.202*EF Distance + 0.020*PIP Angle*EF Distance	<0.01	37	0.90

Table 3.3. Effects of posture elements on predicted FDP and FDS tendon to fingertip force ratios. Equations are shown for univariate mixed procedure analyses and for the best-fit forward stepwise model.

* AIC (Akaike's information criterion) indicates the goodness-of-fit of the model (smaller is better fit) and R-square indicates the predictive ability of the model.

Optimization Criteria: Minimized Function	FDP			FDS		
	Predicted Force Ratio	R^2	p	Predicted Force Ratio	R^{2*}	p
Muscle Forces and Stresses						
_ muscle stress^2	3.51 ± 0.96	0.00	0.00	1.40 ± 0.85	0.12	0.35
_ muscle stress	3.51 ± 0.96	0.00	0.00	1.40 ± 0.85	0.12	0.38
_ muscle force	3.28 ± 1.05	0.01	0.00	1.73 ± 1.18	0.01	0.45
extensor slip force	3.76 ± 0.98	0.00	0.00	1.14 ± 0.90	0.16	0.02
FDP force	3.51 ± 0.96	0.00	0.00	1.40 ± 0.85	0.12	0.38
FDS force	11.60 ± 8.94	0.00	0.00	0.56 ± 0.99	0.25	0.00
long extensor force	3.63 ± 0.97	0.01	0.00	1.20 ± 0.86	0.13	0.05
lumbrical force	15.16 + 8.20	0.05	0.00	4.00 ± 4.95	0.23	0.00
radial band force	3.51 ± 0.96	0.00	0.00	1.40 ± 0.85	0.12	0.38
radial interosseous force	9.33 ± 4.51	0.02	0.00	1.14 ± 2.44	0.18	0.24
terminal extensor force	3.28 ± 1.05	0.01	0.00	1.73 + 1.18	0.01	0.45
ulnar band force	3.76 ± 0.98	0.00	0.00	1.14 ± 0.90	0.16	0.02
ulnar interosseous force	3.79 ± 0.96	0.00	0.00	1.15 ± 0.94	0.15	0.02
Joint Constraint Forces						
_ axial compressive joint force	3.51 ± 0.95	0.00	0.00	1.40 ± 0.85	0.12	0.36
_ shear joint force	3.51 ± 0.96	0.00	0.00	1.40 ± 0.85	0.12	0.38
_ radio-ulnar shear joint force	20.17 ± 11.98	0.06	0.00	3.12 ± 5.18	0.24	0.02
_ dorsovolar shear joint force	3.51 ± 0.96	0.00	0.00	1.40 ± 0.85	0.12	0.38
MP axial compressive force	3.76 ± 0.98	0.00	0.00	1.14 ± 0.90	0.16	0.02
MP radio-ulnar shear force	18.54 ± 10.63	0.06	0.00	4.70 ± 6.08	0.23	0.00
MP dorsovolar shear force	3.76 ± 0.98	0.00	0.00	1.14 ± 0.90	0.16	0.02
PIP axial compressive force	3.51 ± 0.96	0.00	0.00	1.40 ± 0.85	0.12	0.38
PIP radio-ulnar shear force	19.64 ± 12.29	0.06	0.00	2.98 ± 5.00	0.24	0.03
PIP dorsovolar shear force	3.51 ± 0.96	0.00	0.00	1.40 + 0.85	0.12	0.38
DIP axial compressive force	3.51 ± 0.96	0.00	0.00	1.40 ± 0.85	0.12	0.38
DIP radio-ulnar shear force	20.17 ± 11.98	0.06	0.00	3.12 + 5.18	0.24	0.02
DIP dorsovolar shear force	3.51 ± 0.96	0.00	0.00	1.40 ± 0.85	0.12	0.38
Joint Constraint Moments						
_ joint moment	9.91 ± 5.21	0.02	0.00	1.17 ± 3.00	0.14	0.36
MP axial moment	18.36 ± 12.57	0.05	0.00	2.98 ± 5.00	0.24	0.03
PIP axial moment	8.84 ± 4.78	0.06	0.00	1.13 + 1.80	0.18	0.13
PIP radio-ulnar moment	9.58 ± 4.53	0.02	0.00	1.09 ± 2.47	0.18	0.19
DIP axial moment	3.51 ± 0.96	0.00	0.00	1.40 ± 0.85	0.12	0.38
DIP radio-ulnar moment	20.17 ± 11.98	0.06	0.00	3.12 ± 5.18	0.24	0.02

Table 3.4. FDP and FDS tendon to fingertip force ratios predicted by models with different optimization criteria. The muscle stresses, muscle forces, joint constraint forces, and joint constraint moments and combinations of them that were minimized to solve the system of equality and inequality constraints are listed with the resulting predicted force ratios reported as mean ± standard deviation. The predicted ratios were compared to the measured ratios with a paired t-test (p) and the fit between them was evaluated with a linear regression (R^2).
* Negative relationship between measured and predicted FDS force ratios

	Change in Mean Predicted Force Ratios (%)			
	FDP	**(R^2)**	**FDS**	**(R^2)**
Moment Arms in the Z Direction (_ 10%)				
TE tendon around DIP joint *	0.7	(0.00)	-1.9	(0.12)
FDP tendon around DIP joint	-9.7	(0.00)	27.0	(0.11)
FDP tendon around PIP joint *	2.3	(0.01)	-24.0	(0.15)
RB tendon around PIP joint	0.0	(0.00)	0.1	(0.12)
UB tendon around PIP joint	0.0	(0.00)	0.3	(0.12)
FDS tendon around PIP joint	0.6	(0.01)	-8.4	(0.12)
ES tendon around PIP joint	-0.2	(0.00)	2.3	(0.12)
FDP tendon around MP joint	-1.5	(0.00)	-1.9	(0.12)
FDS tendon around MP joint	-0.5	(0.00)	-1.1	(0.12)
RI tendon around MP joint	-0.3	(0.00)	-0.2	(0.12)
LU tendon around MP joint	0.0	(0.00)	-0.1	(0.12)
UI tendon around MP joint	-0.3	(0.00)	-0.2	(0.12)
LE tendon around MP joint	0.0	(0.00)	-0.1	(0.12)
Finger Segment Lengths (_ 10%)				
DIP joint	-0.8	(0.00)	5.3	(0.15)
middle phalanx	-0.5	(0.00)	16.6	(0.11)
PIP joint	0.2	(0.00)	0.8	(0.12)
proximal phalanx	0.9	(0.00)	1.3	(0.12)
MP joint	-0.2	(0.00)	-0.3	(0.12)
Joint and External Force Angles (_ 5°)				
MP Angle	-0.8	(0.00)	-0.7	(0.12)
PIP Angle *	-0.3	(0.00)	-13.7	(0.13)
DIP Angle	-3.6	(0.01)	0.5	(0.08)
EF Angle *	4.0	(0.00)	-13.4	(0.10)
External Force Distance **				
x distance	-10.1	(0.01)	13.4	(0.10)
y distance *	3.8	(0.01)	-4.9	(0.10)

Table 3.5. Sensitivity analysis of model to finger structure and posture variables. Percent change in mean predicted FDP and FDS tendon to fingertip force ratios caused by increasing a model input parameter. The mean force ratios are based on all trials with model predictions.

* indicates that less than 69 trials were predicted by model

** Distance between DIP joint center and center of contact between the fingertip and load cell was increased by 10% of middle phalanx length.

CHAPTER IV:

In Vivo Forces Generated by Finger Flexor Muscles Do Not Depend on the Rate of Fingertip Loading during an Isometric Task

4.1 Abstract

Risk factors for activity related tendon disorders of the hand include applied force, duration, and rate of loading. Understanding the relationship between external loading conditions and internal tendon forces can elucidate their role in injury and rehabilitation. The goal of this investigation is to determine whether the rate of force applied at the fingertip affects *in vivo* forces in the flexor digitorum profundus (FDP) tendon and the flexor digitorum superficialis (FDS) tendon during an isometric task. Tendon forces, recorded with buckle force transducers, and fingertip forces were simultaneously measured during open carpal tunnel surgery as subjects ($N = 15$) increased their fingertip force from 0 to 15N in one, three, and ten seconds. The rates of 1.5, 5, and 15 N/s did not significantly affect FDP or FDS tendon to fingertip force ratios. For the same applied fingertip force, the FDP tendon generated more force than the FDS. The mean FDP to fingertip ratio was 2.4 ± 0.7 while the FDS to tip ratio averaged 1.5 ± 1.0 ($p < 0.01$). The fine motor control needed to generate isometric force ramps at these specific loading rates probably required similar high activation levels of multiple finger

muscles in order to stabilize the finger and control joint torques at the force rates studied. Therefore, for this task, no additional increase in muscle force was observed at higher rates. These findings suggest that for high precision, isometric pinch maneuvers, tendon forces are independent of loading rate.

4.2 Introduction

Tendon disorders of the distal upper extremity are a well recognized problem in the workplace [1]. Risk factors for tendon disorders of the hand and wrist include the applied force, duration and rate of repeated motions, and sustained non-neutral hand posture. These injuries are associated with jobs that require high force and/or high repetition [7, 8]. In addition, several studies have demonstrated a relationship between the velocity of wrist motion during repetitive occupational tasks and a higher rate of upper extremity disorders [11, 12].

External forces applied by the fingers have been measured during different dynamic and static work activities to assess and quantify exposure in order to improve hand tool design. Fingertip forces vary widely between occupational tasks. During typing, peak loads at the fingertips reach 3N [64] while during power tool use the finger forces can be as high as 190 N [52]. To better understand injury mechanisms and to develop effective prevention strategies, it is important to understand how these external loads are related to internal tendon forces and how factors such as movement rate and hand posture affect the relationship. Biomechanical models have been developed to predict internal forces in tendons of the finger, but they include simplifying assumptions or optimization techniques that ignore loading rate, co-contraction and other factors in order to solve an indeterminate problem.

The relationship between force at the fingertip and *in vivo* tendon force in one or both flexor tendons has been measured experimentally during static loading. The ratio of tendon to fingertip force was 7.9 ± 6.3 for the FDP tendon and 1.7 ± 1.5 for the FDS tendon during tip pinch [48]. Another study reported FDS to fingertip force ratios ranging from 1.7 to 5.8 [33]. These ratios exceed model predictions and contain large variability between subjects. Finger position and loading force rate may influence these ratios, the associated motor control strategies, and distribution of forces among the muscles of the finger. The first study did not record finger joint positions, while the second study measured force in only one tendon. Neither study controlled the rate of force application.

Because *in vivo* measurements are difficult to perform, fine wire electromyography has been used to estimate muscle forces and the relative contributions of individual muscles. During isometric tasks, the flexor muscles as well as the intrinsic finger muscles and extrinsic extensors are active when either low (1 to 3 N) or high (28 N) forces are generated at the fingertip [35, 41]. During dynamic finger flexion, finger flexor and extensor muscles are also co-activated and their activity increases with increasing movement rate and frequency. For example, when index finger joints are flexed, the FDP and FDS muscles remain active throughout the motion and their activity levels increased with speed [43]. During fingertip tapping, the activity of both the flexor and extensor muscles increases at higher tapping frequencies [45]. Flexor muscle activity increases with movement rate because additional force is required for higher accelerations of finger segments and attached tissues (e.g. tendons, muscles). The extensors are co-activated to regulate joint stiffness, to provide stability, and to control finger joint positions and torques by decelerating the motion and compensating for errors. During faster motions, antagonist co-contraction increases to provide additional control and stability.

A mechanism of increased agonist and antagonist activity, similar to the one used to control movement rate, may be used to control the rate of force development during isometric contractions. However, the effect of force rate on internal tendon forces has not been investigated. The goal of this investigation is to determine whether the rate of force applied at the fingertip affects *in vivo* forces in the flexor digitorum profundus (FDP) tendon and the flexor digitorum superficialis (FDS) tendon during an isometric task. We hypothesize that the ratio of flexor tendon to fingertip force will increase as the rate of force application increases during static loading. We also hypothesize that the FDP tendon will generate a greater force than the FDS tendon per unit fingertip force in a moderately flexed finger posture.

4.3 Materials and Methods

4.3a Data Collection

In this study, both flexor tendons of the index finger were instrumented and each subject repeatedly pressed his or her finger against a load cell at three different rates. Fifteen subjects (10 females and 5 males, average age 41 ± 10 years) who were scheduled for open carpal tunnel release surgery participated in the study after reading and signing a consent form. The Committee on Human Research from the University of California, San Francisco approved the procedures. Subjects had no previous index finger tendon injuries. Several days prior to surgery, the subjects practiced the experimental tasks in a setting that simulated the procedure during surgery.

The experiment was conducted during open carpal tunnel release surgery with local anesthesia injected at the incision site and a forearm tourniquet. After the flexor retinaculum ligament was released with a longitudinal incision, the FDP and FDS tendons of the index finger were isolated and buckle force transducers were placed around each. The transducers were a modified version of the device previously described by this group [55]. The transducers were tested and calibrated prior to the experiment [55]. A calibration factor was calculated for each transducer to adjust for tendon thickness and relate transducer output to tendon force. The estimated mean errors ranged from 3.8 to 7.3%.

After the transducers were inserted, the subject flexed the index finger against a load 20 times in order to seat the transducers onto the tendons. Then the tendon thickness was measured and the tourniquet was released to allow tissue reperfusion. The subjects were supine with the shoulder abducted to 90 degrees. A custom designed apparatus supported the load cell at the end of the index finger in a predetermined location to achieve the desired hand position. The hand was placed in the device with the thumb up, the palm facing the feet, and the wrist in 15 extension. The MP joint of the index finger was positioned by the surgeon in 45 flexion using an angle bracket while the PIP and DIP finger joints assumed a natural pinching position with the fingertip on the load cell. The centers of rotation of the joints of the index finger and wrist were marked with a surgical pen on the radial side of the hand and this side was recorded with a video camera mounted above the operating field and set perpendicular to the plane of finger flexion.

Data was collected from the tendon transducers and fingertip load cell (ATI Industrial Automation, Apex, North Carolina, USA) simultaneously at 100Hz using a laptop computer with an A/D board.

Subjects were instructed to steadily increase the force on the load cell at three different rates until the fingertip force reached 15N. To help achieve the target force and force rates, subjects observed a computer monitor mounted above their heads that provided real-time feedback of fingertip force. Subjects were able to compare the force that they exerted with the desired force profile and adjust their fingertip force to match the desired pattern. They were instructed to attain the maximum force in one, three, or ten seconds, corresponding to fingertip force rates equal to 15 N/s, 5 N/s and 1.5 N/s (fast, medium, and slow). Three to 33 trials were collected for each subject at each rate. After the tasks were completed, the tendon thickness was measured again, the transducers were removed, and the carpal tunnel surgery completed.

4.3b Data Analysis

The voltage output from the buckle force transducers was converted to tendon force using the calibration factor adjusted for tendon thickness. Due to transducer problems, FDP tendon force for one subject during the trial at the slow force rate was omitted from analysis. The forces measured with the load cell and buckle transducers were filtered using a low-pass Butterworth filter at a cutoff frequency of 25Hz. The total force at the fingertip was calculated from the acquired three orthogonal force components. The start of each loading task was defined as the time when the fingertip force first exceeded 1 N (2.5N for subjects 3 and 8) and the end occurred when the force reached its maximum value. Tendon and fingertip forces corresponding to these times were extracted. For all the trials, the relationship between force in each tendon and force at the fingertip was defined as the slope of the line of tendon versus tip force, calculated with a linear regression. The fingertip force rate at each time point was defined as the slope of a linear fit of tip force and time over 0.1s (10 points). The mean fingertip force rate for each trial was calculated. The best trial was selected for each subject at each rate for further analysis based on how well the fingertip force corresponded to the target force rate. For each trial, the MP, PIP and DIP joint angles were measured from the corresponding video frame with Adobe Photoshop.

A two-factor repeated-measures analysis of variance was used to determine whether the rate of force application (fast, medium, and slow) and tendon (FDP and FDS) affected the ratio of tendon to fingertip force ($p = 0.05$).

4.4 Results

An example of typical tendon and fingertip forces produced at the fast, medium, and slow loading rates is displayed in Figure 4.1. The average maximum tip forces and mean tip force rates that the subjects attained during the trials are summarized in Table 4.1 and should be compared to the target tip force and force rates of 15N, 15N/s, 5N/s, and 1.5N/s. Most people were able to produce the desired maximum force of 15 N at their fingertips, but two subjects were only able to generate maximum forces of 8.9 and 11.6 N. Maintaining the target force rate during each trial was difficult. Although the individual force rates varied from the target force rates, the fast rate was at least five times greater than the slow rate for all subjects except one. One person was unable to press at the fast rate in any of the trials. Subjects maintained a similar finger joint posture throughout all the trials (Table 4.1).

Small differences in tendon versus tip force slopes at different rates are demonstrated by typical force data from one subject (Figure 4.2). Similar data across all subjects (Figure 4.3) shows high linearity for the FDP and tip force and for the FDS and tip force relationships ($R^2 = 0.98$ and 0.95, Table 4.1). The average ratios of the tendon to tip forces for all subjects and rates are listed in Table 4.1 and the individual values are displayed in Figure 4.4. RMANOVA indicates that the interaction term between tendon and rate is not significant. The rate of force application does not have a significant effect on the ratio of tendon to fingertip force for either tendon ($p = 0.29$). The FDP tendon generates more force than the FDS tendon for the same amount of applied fingertip force ($p < 0.0001$). The mean FDP to fingertip force ratio across all the trials was 2.4 ± 0.7 while the FDS to tip ratio averaged 1.5 ± 1.0. Adding peak fingertip force and MP angle as additional factors to the repeated measures ANOVA does not change the conclusion that force rate does not have a significant effect on the tendon to fingertip force ratio ($p = 0.46$).

4.5 Discussion

Ratios between forces generated by the two flexor tendons of the index finger and the force applied at the fingertip were investigated during a task involving increasing fingertip force linearly from 0 to 15N at rates of 1.5 N/s, 5 N/s, and 15 N/s while maintaining a static hand posture. These three different fingertip loading force rates do not significantly affect the FDP to fingertip or the FDS to fingertip force ratios. This finding does not support the hypothesis that increasing fingertip loading rates will increase flexor tendon to fingertip force ratios. Forces in the FDP tendon are significantly higher than forces in the FDS tendon for the same externally applied fingertip force. The force ratios and relative force contributions of each muscle exhibit large inter-subject variability, but the FDP to fingertip force ratio is less variable among subjects than the FDS to fingertip ratio, as indicated by the lower coefficients of variation (standard deviation/ mean).

Ratios between flexor tendon and fingertip force during isometric loading have been measured previously. The FDP to fingertip ratio of 7.9 ± 6.3 observed by Schuind et al. [48] is much larger than the mean value of 2.4 ± 0.7 from our study. The lower variability in our study may be due to a consistent finger posture between subjects. The FDS ratio of 1.7 ± 1.5, from the same study, is similar to our results of 1.5 ± 1.0. Both these FDS to fingertip ratios are smaller than 3.3 ± 1.4, the ratio measured in another study. [33] The smaller values of tendon to fingertip ratios presented in our study are in closer agreement with model predictions. [30-32] Differences in finger positions may account in part for the difference between the values found in the literature and those from the current study.

Surprisingly, the rate of fingertip force application did not affect the amount of force generated by the extrinsic finger flexor muscles per unit fingertip force during the experimental task. It was originally hypothesized that increased antagonist co-activation would be required to provide additional stability and control of finger joint position and torque during faster loading and that flexor forces would also rise to balance this extensor activity. In contrast to the findings of this study, an increase in flexor activity was measured with increasing movement speeds during dynamic finger flexion [43-45] suggesting that muscles may respond differently to static and dynamic conditions. Therefore, the impact of different motion parameters on muscle loads should be evaluated in the context of a particular task. The difference between the results of the two studies may also be explained by the difference in experimental techniques. EMG measurements may be influenced by motion artifacts during dynamic motions, limiting their reliability. On the other hand, a study that investigated different rates of force development (ranging from 5 to 20% MVC/s) during an isometric contraction of the biceps Brachii muscle supports our findings. [65] The overall muscle activity as measured by electromyography did not depend on the rate of isometric force generation even though the motor unit recruitment strategy did change.

This study measured forces in two of seven muscles that control the index finger. The FDP and FDS muscles span multiple joints and act over multiple finger segments to produce the desired flexion force at the fingertip. To control external force output and finger position, other muscles must be activated to maintain postural stability and provide the appropriate torques at all index finger joints. The activity of the three intrinsic muscles and two extrinsic extensors will influence flexor force and must be considered for a complete understanding of the system. Co-contraction of all seven finger muscles has been reported during a low force, precision grip task and during the production of static MVC forces. [35, 41] These studies indicate that all finger muscles are involved in isometric fingertip force generation, but their individual contributions and roles may vary with force, finger posture, and force direction. It is possible, that for this task, similar levels of co-contraction were occurring in the other muscles even at the different loading rates. In our study, it is likely that the fine motor control needed to generate the precise force ramps required high activation levels of intrinsic and extrinsic finger muscles in order to stabilize the finger and control joint torques. Therefore, we observed no additional increase in FDP and FDS forces at the higher rates. Perhaps, less co-contraction at slow rates and lower tendon to tip force ratios would occur during a less constrained loading task. Unfortunately, the activity of the other five muscles and their response to changes in loading force rate cannot be directly estimated. However, the presented results have important implications since the extrinsic flexors are responsible for generating the majority of flexion force and are the most likely candidates for injury or delay in healing.

Several limitations of *in vivo* measurements must be considered when interpreting and applying the presented data. First, measurement errors are introduced by the use of buckle force transducers and during the conversion of their output to force. However, these errors are small relative to the forces measured (see methods section) and although they may affect the absolute force values, they should not influence the conclusions regarding the effect of loading rate on tendon to fingertip force ratios. Collection of data during carpal tunnel surgery is another limitation. The muscle forces may not accurately represent control strategies and movements executed during daily activities since subjects have lost finger sensory and proprioceptive feedback due to local anesthesia of the median nerve. In addition, the upper extremity is constrained for the surgical procedure and subjects do not have visual feedback of finger motion. However, the performance during surgery appeared similar to the practice trials, indicating that the limitations were related to the task itself and not the surgery.

The findings of this study indicate that changes in fingertip loading force rates between 1.5 N/s and 15 N/s during a static fingertip loading task do not alter internal flexor tendon forces. The implications are that the motor control for the isometric loading tasks investigated in this experiment will involve similar recruitment of antagonists. Thus, modifying loading rate does not alter the tendon to tip force ratio. The results of this study demonstrate the importance of *in vivo* measurements in understanding how external factors impact internal forces in tendons and the motor control mechanisms. Design of tools and work rehabilitation strategies can be improved based on this knowledge to reduce the risk of tendon disorders.

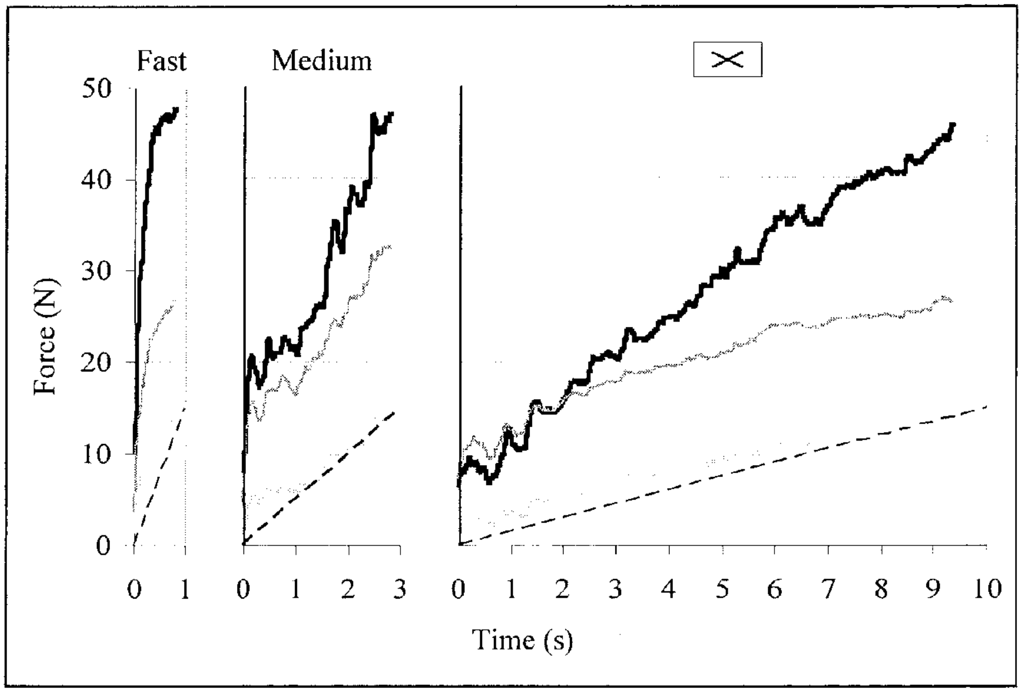


Figure 4.1. Example from one subject of FDP tendon, FDS tendon, and fingertip forces during a typical loading trial at the fast, medium and slow fingertip force rates.

Force in the FDP tendon
Force in the FDS tendon
Force at the fingertip
Target fingertip force ramp

	Fingertip Loading Force Rate			
	FAST	MEDIUM	SLOW	p-values
Maximum Tip Force (N)	16.3 ± 2.7	14.6 ± 2.6	13.9 ± 2.9	0.04
Mean Tip Force Rate (N/s)	15.0 ± 2.9	4.7 ± 1.0	1.4 ± 0.3	<.0001
MP Angle (degrees)	33 ± 12	34 ± 11	37 ± 12	0.06
PIP Angle (degrees)	51 ± 21	54 ± 21	49 ± 24	0.31
DIP Angle (degrees)	24 ± 28	20 ± 23	18 ± 24	0.24
FDP vs. Tip Force R^2	0.99 ± 0.02	0.98 ± 0.03	0.97 ± 0.04	0.29
FDS vs. Tip Force R^2	0.95 ± 0.09	0.96 ± 0.04	0.94 ± 0.08	0.84
FDP vs. Tip Force Slope	2.7 ± 0.6	2.4 ± 0.7	2.2 ± 0.8	0.29
FDS vs. Tip Force Slope	1.6 ± 1.2	1.4 ± 0.8	1.5 ± 1.0	0.29

Table 4.1. Parameters related to tendon and fingertip forces during loading at three different fingertip force rates. Data reported as mean ± standard deviation (N = 15). P-values were calculated with a repeated measures ANOVA.

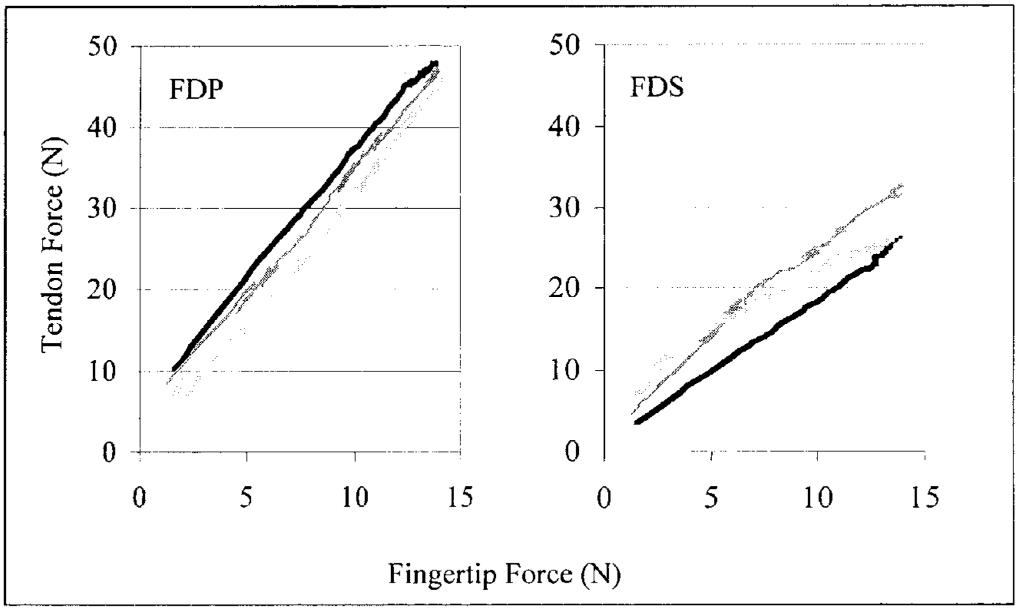


Figure 4.2. Example from the same subject of the relationship between force applied at the fingertip and force in the FDP (left) and FDS (right) tendons during a typical loading trial at the fast, medium, and slow fingertip loading force rates.

Fast
Medium
Slow

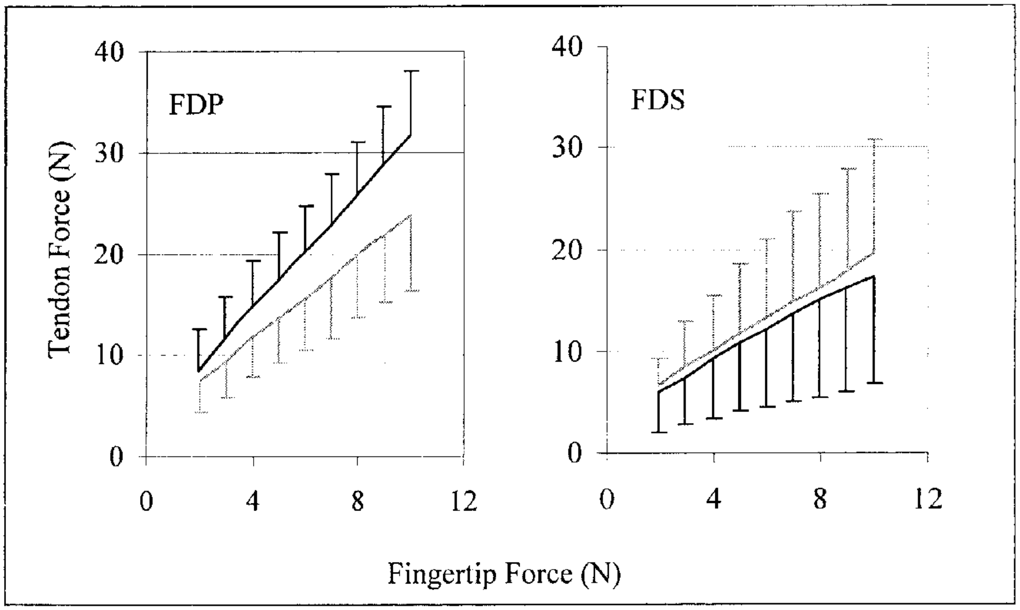
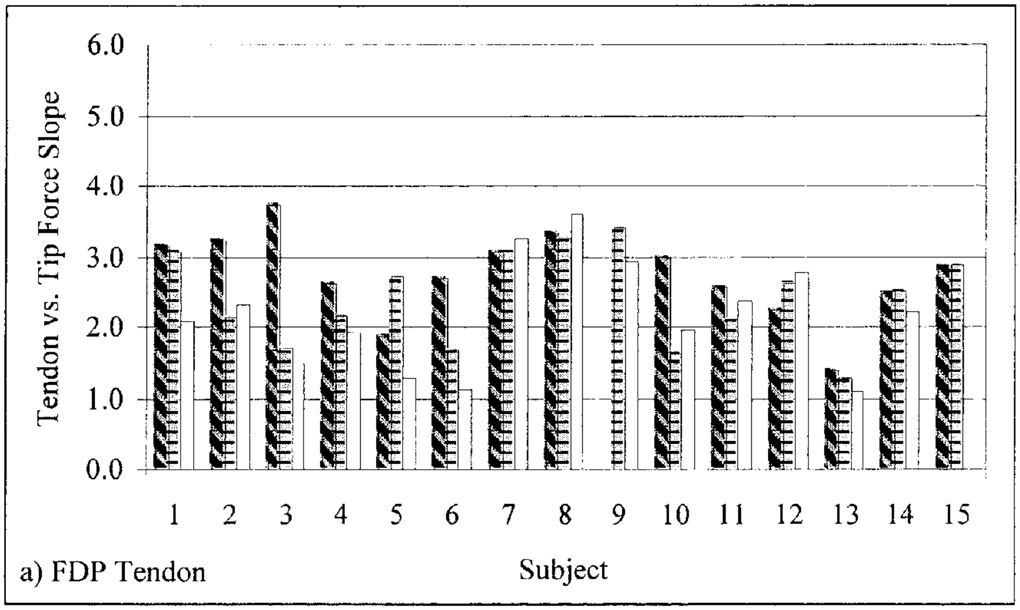


Figure 4.3. Relationship between force applied at the fingertip and force in the FDP and FDS tendons during a loading trial at the fast and slow fingertip force rates. Data is reported as mean ± standard deviation for all subjects who attained fingertip forces of 2 to 10 N during trials at both rates (n = 13 for FDP and n = 14 for FDS).

— Fast

— Slow



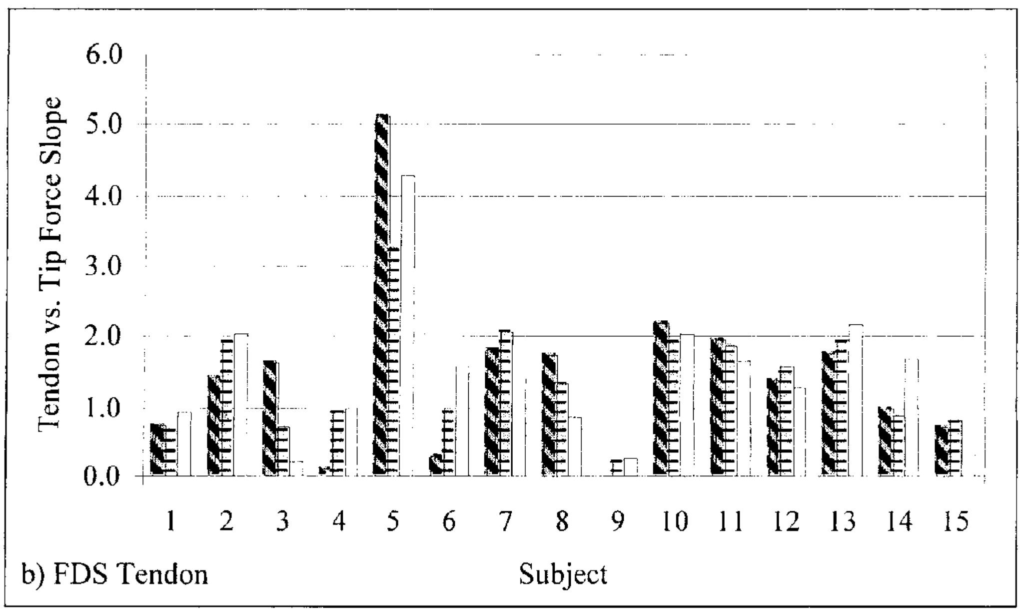


Figure 4.4. Relationship between force in the FDP tendon (a) and at the fingertip and between force in the FDS tendon (b) and at the fingertip during loading at three different fingertip force rates for each subject.

- Fast
- Medium
- Slow

CHAPTER V:

In Vivo Forces Generated by Finger Flexor Muscles Increase With Finger and Wrist Flexion during Active Finger Flexion and Extension

5.1 Abstract

Currently, treatment of flexor tendon lacerations is based on the use of one of many repair techniques and, some sequence of hand motions and positions for early protected motion rehabilitation. However, the effect of these maneuvers on actual forces at the tendon repair site is not well understood. The goal of this study is to determine *in vivo* force histories in human flexor digitorum profundus (FDP) and flexor digitorum superficialis (FDS) tendons of the index finger during active unresisted finger flexion and extension and to examine the effect of wrist and finger position of these forces. During open carpal tunnel surgery, finger positions were recorded on video. Tendon forces were simultaneously acquired with buckle force transducers, devices that measured tendon tension using strain gages [55]. Mean *in vivo* FDP tendon forces vary with hand posture and direction of motion between 1.3 ± 0.9N and 4.0 ± 2.9N while mean FDS tendon forces range from 1.3 ± 0.5N to 10.7 ± 4.4N. FDP force increases during active finger flexion at both wrist angles of 0° or 30° flexion. Force in FDS is unchanged during finger flexion with wrist in neutral. With wrist in flexion, FDS force increases with finger

flexion. Tendon forces are similar regardless of whether the fingers are moving in the flexion or extension direction. The parameters of the wrist and finger position may be adjusted to reduce tendon forces during rehabilitation, while maintaining tendon excursion. Increasing the amount of finger and wrist flexion during active finger motion may increase the tension in the repair and the probability of rupture. Data from this study, combined with information about repair strength and effects of tendon forces on healing, can be used to design more effective rehabilitation protocols that match tendon forces to repair strength during key hand positions and exercises to improve clinical outcomes.

5.2 Introduction

Flexor tendon injuries are a common clinical problem and the technique selected to manage them will influence the recovery of finger function. Different methods of tendon repair and rehabilitation have been studied extensively in the past few decades to improve clinical outcomes. Many versions of postoperative rehabilitation procedures are currently in use. Animal and clinical studies have demonstrated that immediate tendon mobilization after repair is beneficial to healing [14-17], but identifying an optimal rehabilitation procedure has proved elusive. Tendon excursions and low forces generated by passive or active finger movement may stimulate healing, prevent adhesion formation and improve repair strength, thus helping the patient regain an increased range of motion faster [14, 15, 17]. However, excessive force during finger motion can cause gap formation, poor healing, and even rupture of repair [18-20].

The principle of rehabilitation has been to allow tendon excursion while limiting tendon forces initially, then gradually increasing both tendon excursion and force with time. A rehabilitation regimen of early passive finger flexion and active extension was introduced 30 years ago [22]. The wrist was positioned in a dorsal extension block splint and a rubber band was attached to the fingertip with a nail suture and tensioned to the volar forearm area. The rubber band provided force for passive finger flexion while the patient actively extended the fingers. Since that time, the original dorsal extension block splint has been modified to increase the range of finger motion [23, 24] and remains one of the most commonly used rehabilitation protocols [25]. The passive flexion – active extension procedure has also been supplemented with assisted controlled passive motion and/or an active flexion component. The other common form of rehabilitation is based on the work of Duran, with passive digital flexion and extension coupled with the use of a dorsal extension block splint [26]. More recently, some clinicians have advocated the use of early controlled active motion to improve finger function [27, 28]. The positions of the wrist and fingers are carefully controlled with a dorsal splint during these rehabilitation procedures to limit the amount of force in the repaired tendons. The wrist and metacarpal-phalangeal (MP) joints are maintained in various degrees of flexion during passive and active proximal interphalangeal (PIP) and distal interphalangeal (DIP) joint movements. In contrast, wrist extension and MP flexion have been advised when the fingers are actively held in cascade contact in a flexed position [21, 29]. Currently, no consensus exists concerning the best type of motion or hand posture to use during rehabilitation, but there is a trend towards more aggressive protocols that increase the amount of force and excursion applied to the tendons [21]. An understanding of the *in vivo* forces in the flexor digitorum profundus (FDP) and flexor digitorum superficialis

(FDS) tendons during different finger motions and at different wrist postures is needed to improve rehabilitation protocols.

In vivo flexor tendon forces have been measured with buckle force transducers and predicted with models. Schuind et. al. measured FDP and FDS forces during passive wrist motion and during active isolated flexion of the interphalangeal joints [48]. Forces in the two tendons ranged from 0 to 13 N during unresisted active PIP joint flexion and from 0 to 29 N during unresisted active DIP flexion. Passive wrist motion resulted in forces of only 0 to 6 N. However, specific hand postures corresponding to the force measurements were not reported. In addition, isolated movements of one joint do not simulate maneuvers commonly used during postoperative rehabilitation which include either passive or active flexion of multiple finger joints. Because *in vivo* measurements are difficult to perform, models have been developed to predict flexor tendon forces. Most models estimate the relationship between a force applied externally to the finger and the forces in the tendons and are often limited to quasi-static situations [33, 35, 38]. In order to estimate forces during rehabilitation motions, models should include dynamic components and finger movements without external load. One such model predicted no force in the FDS tendon during active finger flexion [36], a result that disagrees with EMG data collected when subjects moved their fingers [43]. Even models that include an accurate representation of the finger structure and motion rely on simplifying assumptions or optimization techniques to solve the indeterminate problem of calculating tendon forces [36] and may not yield correct results.

Additional *in vivo* measurements are necessary for a better understanding of finger flexor tendon forces since previous *in vivo* studies investigated a limited number of motions and the uncertainty associated with model predictions is great. The goal of this study is to determine simultaneous *in vivo* force histories in human FDP and FDS tendons of the index finger during active unresisted finger flexion and extension and to examine the effect of wrist and finger position on these forces in subjects with healthy tendons. Four hypotheses are tested: 1) Flexor tendon forces during active finger flexion will exceed forces during active finger extension; 2) Flexor tendon forces will increase with increasing finger flexion during active finger flexion; 3) Flexor tendon forces will be lower in a flexed wrist posture compared to a neutral wrist posture during active finger flexion; and 4) The patterns of forces generated by the FDP and FDS tendons at different finger and wrist positions will be similar. In this study, *in vivo* tendon forces were measured while subjects actively flexed their fingers with the wrist in two different positions, neutral and 30 degrees of flexion.

5.3 Materials and Methods

5.3a Data Collection

Twelve subjects (8 females and 4 males, average age 42 ± 10 years) who were scheduled for open carpal tunnel release surgery participated in the study after reading and signing a consent form. The study had full IRB approval from the Committee on Human Research . Subjects had no previous index finger tendon injuries. Several days prior to surgery, the subjects had the opportunity to rehearse the experimental protocol.

The experiment was conducted during open carpal tunnel release surgery with local anesthesia injected at the incision site. Thus, the subject retained motor control of the forearm muscles and intrinsic hand muscles throughout the procedure. After the flexor retinaculum ligament was released with a longitudinal incision and adequate access to the flexor tendon was gained, the flexor digitorum profundus and the flexor digitorum superficialis tendons of the index finger were identified. Two gas-sterilized buckle force transducers were mounted on them, one on each flexor tendon of the index finger. The transducers were staggered on the tendons within and slightly distal to the carpal tunnel. They were a modified version of the device previously described [55]. The basic design and dimensions were the same, but two additional strain gauges were used to produce a full bridge circuit for temperature compensation. In addition, the edges of the transducer's arches were rounded where they contact the tendon to decrease stress concentrations at this location. The tendon thickness in the transducer was measured using a digital micrometer with a resolution of 0.01 mm (Series 575 Digimatic Indicator, Mitutoyo). The transducers were individually tested and calibrated using a previously described method [55]. An equation was calculated for each transducer that, adjusting for tendon thickness, related transducer output to tendon force. The estimated mean errors for the calibration factor ranged from 3.8 to 7.3%. Each transducer's performance was evaluated before every surgery by supporting the device at the ends and hanging two weights from the fulcrum. The transducer output differed from the output obtained during initial calibration testing by an average of 0.7%, demonstrating that the calibration factor did not change over the course of the experiments.

After the transducers were inserted, the subject was asked to flex the index finger against a load 20 times to seat the transducers onto the tendons. No bowstringing of tendons was observed when the wrist was in flexion. Then the tendon thickness was measured to enable accurate conversion of transducer output to tendon tension and the tourniquet was released to allow tissue reperfusion (mean tourniquet time for 6 subjects was 30 ± 13 min). The subjects were supine with the shoulder abducted to 90 degrees during the procedure. The hand was positioned with the thumb up and the palm facing the feet. The centers of joint rotation of the index finger and wrist were marked with a surgical pen on the radial side of the hand. A video camera that was mounted above the operating field recorded finger motion in the sagittal plane (30 frames/s). Simultaneously, force data was collected from the tendons at 100Hz using a Power PC Macintosh laptop computer with an A/D board. During data collection, the surgeon tapped his finger rapidly on a force sensor placed in the field of view of the camera. The sensor output was recorded together with the buckle transducer force data and the increases in signal were later synchronized with the times the finger contacted the sensor with an accuracy of 1 frame or 0.03s.

Data was collected during active finger flexion and extension approximately 20 minutes after the tourniquet was deflated to allow time for muscle recovery. The wrist was first positioned by the surgeon in either zero or thirty degrees of flexion using two sterilized angle brackets. The patients were then given specific verbal instructions by the surgeon. During the active finger flexion and extension motion, each subject was asked to bend all the fingers until the fingertips lightly touched the palm as "tip contact" and then to straighten them out. The motion was repeated two times at each wrist position. After the tasks were completed, a final measurement of tendon thickness was taken, the transducers were removed, and the carpal tunnel surgery completed.

5.3b Data Analysis

The voltage output from the buckle force transducers was converted to tendon force by using the calibration equations calculated from *in vitro* testing and the thickness measurements. To extract tendon force at different finger postures, finger position during continuous finger flexion and extension was represented by five MP joint angles: each subject's finger position at the start of the motion, at 15 MP flexion, 45 flexion, 60 flexion and the joint position at the end of the motion and vise versa during extension. Video tapes were viewed to determine times that correspond to these five finger positions. The video frames near each instant of interest were captured and the angle of the MP joint was measured with Adobe Photoshop. The video times were correlated with the acquired force data and the forces in the FDP and FDS tendons at each position were extracted.

Maximum force values for each entire flexion and extension movement were defined as the highest forces reached between the beginning of the flexion motion and end of extension motion. Similarly, the maximum forces when the fingers were in the fixed flexed position were determined using the time between the end of flexion and beginning of extension.

5.3c Statistical Analysis

Mean forces were calculated across trials for each tendon at each MP joint position and wrist position for both active finger flexion and active finger extension. MP joint angles at the extremes of the range of motion were termed "MP Flex" and "MP Ext." The joint angle had to exceed 60 to be classified in the former category and be less than 15 to fit in the latter. Since some subjects did not reach these positions, the sample size for the "MP Flex" and "MP Ext" finger position categories is smaller than for other finger postures. The force data is presented as the average and standard deviation of all the subjects who attained each position during the motion.

The effects of finger and wrist position and the direction of motion on these force values were analyzed with a three-factor repeated measures analysis of variance ($p = 0.05$). A separate analysis was performed for the FDP and FDS tendons. When significant differences were identified with the RMANOVA, Tukey's follow-up test was used for pair-wise comparisons. The effect of wrist position on maximum tendon force values was evaluated with a two-tailed paired t-test ($p = 0.05$).

5.4 Results

FDP and FDS tendon forces during active finger flexion and active finger extension were plotted for all subjects. An example of a typical *in vivo* force history of the two finger flexor tendons during active finger flexion and extension when the wrist is at 0 and 30 flexion is shown in Figure 5.1. The start and finish of both motions are indicated by vertical lines and different finger positions are represented by points that correspond to five MP joint angles. This subject's data demonstrates a commonly observed force pattern. FDP and FDS forces increase as this subject flexes his or her fingers to MP joint angles exceeding 45 (A in Figure 5.1a and 5.1b). Forces in both tendons remain high when the fingers are held in the flexed position (B) and then decrease rapidly when the subject begins to extend his or her fingers. A small increase in force can be seen when the MP joints of the fingers are extended from 45 to 15 when the wrist is in the neutral position (C in Figure 5.1a). The forces in the FDP tendon are higher than in the FDS when the wrist is in the neutral position while the opposite is true when the wrist is in 30 flexion. In addition, forces in both tendons are larger when the wrist is flexed.

The mean FDP and FDS tendon force values and standard deviations are reported in Table 5.1 during active finger flexion and extension at each wrist and finger position. If the MP joint angle is less than 15 during any of the maneuvers, the corresponding force values are shown in the "MP Ext" row. Similarly, if the MP joint exceeded 60 , the forces that correspond to the maximum flexed position are shown in the row labeled "MP Flex". Actual MP flexion angles and the sample sizes at each position are presented in Table 5.2. The actual finger position was similar to the desired joint angles of 15 , 45 , and 60 for all motions. Some subjects were not able to attain 60 of MP flexion and some did not extend their MP joint more than 15 . Therefore, for some postures the sample size is less than 12. The force values when the wrist and fingers are maximally flexed include only three subjects; therefore, they were excluded from the statistical analysis.

Summary data are displayed graphically in Figure 5.2. Mean *in vivo* FDP tendon forces vary with hand posture and type of motion between 1.3 N and 4.0 N while mean FDS tendon forces range from 1.3 N to 10.7 N. Analysis of FDP tendon force using a three-factor repeated measures analysis of variance identified significance ($p = 0.0002$) only for MP angle (Figure 5.2a). Wrist angle, movement direction and all four interaction terms have no significant effect on force. Therefore, FDP force does not depend on wrist flexion or direction of finger motion. However, FDP force increases with increasing MP angle; the force rises with finger flexion and decreases with finger extension. It is significantly higher when the MP joint is positioned at 45 or 60 flexion than when it is at 15 flexion ($p = 0.009$ and $p = 0.0004$ from Tukey's follow-up test). The force is also greater when the MP joint is at 60 flexion than when it is in the maximum extended position ($p = 0.04$).

In contrast, for FDS force, the interaction term of wrist and finger position was significant ($p < 0.0001$), indicating that both wrist and finger position have a significant and mutually dependent effect (Figure 5.2b). Based on follow-up tests, FDS force is greater when the fingers are in a flexed position only when the wrist is at 30 . When the wrist is positioned in flexion, FDS force is significantly higher when the MP joint is at 60 than when it is in the extended starting position, at 15 MP flexion or at 45 MP flexion ($p < 0.0001$). When the MP joint angle equals 60 , wrist flexion leads to a significant increase in FDS force ($p < 0.0001$). However, FDS force does not depend on finger posture when the wrist is in the neutral position. Neither movement direction or interaction terms containing the direction of finger motion were significant; therefore, FDS force does not depend on the direction of finger motion.

In addition to examining tendon force values at specific hand positions, the maximum tendon forces during active finger motion were identified. The maximum forces attained during a complete flexion and extension motion, including the time when the fingers were maintained in a fixed flexed position, are reported in Table 5.3. In addition, the maximum forces for just the fixed flexed finger position are reported. The mean values are similar indicating that forces in the tendons reach their peak when the fingers are held in a flexed position. These maximum force values exceed the forces during motion, reported in Table 5.1. The additional increase in force after MP joint angle attained its maximum value may have been caused by further flexion of the IP joints or by subjects applying additional force against their palms. Maximum force in the FDS tendon when the finger are maintained in a fixed flexed position is significantly higher when the wrist is flexed than when the wrist is in neutral ($p = 0.04$). In contrast, wrist position does not affect maximum FDP forces ($p = 0.4$).

5.5 Discussion

As far as we are aware, this is the first study to report *in vivo* forces in the flexor tendons of the index finger generated during active, unresisted, composite finger motion (simultaneous motion of all index finger joints). The results show that forces in both flexor tendons increase during active finger flexion and decrease during active finger extension. FDP tendon force increases when the MP angle is beyond 45 flexion, at either wrist angle of 0 or 30 flexion. The increase in FDS force with finger flexion is only observed when the wrist is flexed to 30 . In contrast, when the wrist is in a neutral position, flexion of the fingers does not affect FDS force. The magnitude of tendon forces at different wrist and finger positions is similar regardless of whether the fingers are moving in the flexion or extension direction.

During active free finger flexion, flexor tendons must generate sufficient forces to overcome the inertia of finger segments, frictional resistance, passive forces in the finger and any co-contraction of antagonist muscles. Passive forces are produced during movement due to stretching and frictional forces of soft tissues such as muscles, tendons, joint capsules, and skin. Passive forces of extensor muscles increase as their length increases when, for example, the fingers and wrist are flexed [66]. To overcome these increasing forces that oppose finger flexion, the force in the flexor tendons must increase as the fingers or wrist are flexed. This is consistent with the observed increase in flexor tendon forces with wrist and finger flexion. The FDP and FDS muscles work together to move the fingers towards the palm. The timing and direction of force changes in the FDP and FDS tendons are generally consistent between subjects, but the relative magnitude of FDP and FDS force differs as indicated by large standard deviations. The FDP plays the dominant role and generates more force than the FDS in most subjects at most hand postures. However, the force in the FDS increases significantly when the fingers and the wrist are both flexed. Perhaps FDS has to supplement FDP force in order to attain this extreme range of motion. The FDP alone may not be able to generate enough force to overcome the additional passive force from the extensor system.

During active finger extension, FDP and FDS force magnitudes and patterns are similar to ones seen during finger flexion. Since finger extensor muscles generate force to straighten the fingers, the amount of force in the flexor tendons during this movement was unexpectedly high. The data implies that the flexors are actively co-contracting during finger extension, possibly to stabilize the finger. If the forces in the flexors were purely passive, they would be lowest when the wrist and fingers are flexed and would increase with finger extension as the flexor muscles lengthen. As the fingers extended, the expected increase in flexor force was not observed. Active forces may mask any increase in passive muscle forces due to lengthening. Perhaps this increase was not significant at the achieved MP extension angles and would only be visible if the fingers reached fully extended or hyperextended positions. Force patterns in both tendons and the relative amount of force generated by each tendon vary among subjects during extension.

When the flexors act as antagonists to the finger extension motion, their activation levels and patterns are more variable than when they act as agonists during flexion.

Tendon forces during hand motions have been measured directly or estimated indirectly with models. Savage [66] defined "minimal active tension" as the "least active force to produce movement" and stated that it "exceeds the passive tension in the opposite muscle group by small forces…" He concluded that during flexion of the interphalangeal joints the "minimum active tension" in the flexor tendons increased with wrist flexion and was larger when the MP joint angle equaled 90 degrees than when it was 45. Although the actual forces in the tendons were not measured, the trends in force changes with hand position agree with the data presented here. *In vivo* forces in human flexor tendons of the index finger have been reported previously during passive and active finger motion [48, 49]. In one study, FDP and FDS forces were measured during active isolated flexion of one of the IP joints. The FDP forces were higher than the mean peak forces we recorded during simultaneous composite flexion of all finger joints while the FDS forces are similar to the ones we observed when the wrist was flexed. The data is difficult to compare to our study since the motions were different and the hand positions corresponding to the force measurements were not reported in the other study.

In vivo force changes due to changes in wrist position have been reported in canine FDP tendons [67]. When the FDP was stimulated to tetany, tendon force decreased with wrist flexion. In contrast, we observed no difference in FDP forces in the two wrist positions, but did see an increase in FDS tendon force in the more flexed wrist position. The difference in the findings may be explained by the difference in type of muscle activation since the canine muscle was stimulated to tetany when all other muscles were relaxed while our subjects actively exerted the minimum force needed to move the fingers. The maximum force that muscles can actively generate decreases with muscle length, as described by the Blix curve. Although this relationship may account for the results of the canine study, the decrease in length due to wrist flexion is not likely to limit the low tendon forces during unresisted flexion. The primary determinant of required flexor force is probably the amount of resistance they must overcome.

In addition to direct force measurements, muscle activity levels acquired using fine-wire EMGs have been used to estimate muscle forces. One study investigated EMG activity of the seven finger muscles during active flexion and extension of all index finger joints at three different speeds [43]. The FDP and FDS muscles were active during the entire flexion motion and their activity levels were highly correlated. Although that study also showed that the two muscles work together to flex the finger, their relative contribution and coordination at different finger positions cannot be deduced from the reported EMG data. During extension, the extensors and all intrinsic muscles were active throughout the movement. Limited activity was observed in the flexors during fast extensions, but no activity was observed during slow extensions. In contrast, all muscles were co-activated during both isolated flexion and extension of the MP joint. In another study, slow flexion and extension of the MP joint both consisted of a series of sub-movements controlled by cyclic activation of agonist and antagonist muscles [46]. Another study examined FDP and FDS activity during active finger extension executed as part of a rehabilitation procedure with different versions of the Kleinert splint [47]. Flexor co-activity during resisted extension was observed in most subjects, especially in the FDS muscle. The level of activity increased as the amount of resistance increased, but was absent during unresisted extension. Finger muscles span multiple joints and act over multiple segments to produce a desired motion. Coordination of these muscles is necessary to control finger position and to provide the appropriate torques at all index finger joints. Previous studies suggest that antagonist co-activation may be used to control more complex movements of this system, such as fast motions, motions limited to one joint, or movements against a resistance. In contrast to previous studies, our data suggests that flexors were active as antagonists during composite, unresisted finger extension. The co-activation may have been necessary to provide additional control over the motion, perhaps to compensate for lost proprioceptive feedback. The slower rate of motion (mean duration of flexion was 3.7 ± 2.5s) in our study may also lead to increased co-contraction, though the effect of this rate is difficult to determine due to a lack of similar study in the literature.

Other researchers have attempted to predict forces in tendons and joints by developing complex models of the hand. Most models aim to determine forces during static or quasi-static loading of the fingers against an external load to represent pinching or grasping activities. Recently, a dynamic, three-dimensional model to study unresisted finger flexion and extension has been presented [36]. Passive properties of the muscles and joints; muscle forces as a function of length, velocity, activation level and architecture; and inertial effects were considered in the model. The model predicted that FDP force will increase from 0 to 1.8 N during active finger flexion when the wrist is in a neutral position and that it will decrease to 0 N during extension. The effect of changing finger position is similar to the pattern presented in our study, but the force values are about 2 N lower than the mean forces we measured. That model predicted no force in the FDS tendon which contradicts our *in vivo* findings and the EMG measurements [43].

Many factors must be considered to understand and predict forces generated in tendons and the associated muscles at different hand postures. Each muscle must be examined as one component of the complex structure of the finger and the complete biological system must be described accurately. The exact task being performed must also be taken into account since the type and direction of movement, as well as the magnitude of applied external force, are factors that will determine muscle force requirements. For example, during unresisted finger movement, passive forces that depend on position may affect muscle force significantly and should be considered in models that predict tendon forces during low force movements. Even complex models do not predict muscle forces accurately because the finger is a redundant system. Models often include simplifying assumptions and optimization functions to solve the indeterminate problem. The individual muscle forces are difficult to predict since different combinations of muscle forces can result in the same outcome motion. Thus, *in vivo* measurements are critical to helping us understand hand function and the roles of the individual muscles during different tasks and may provide insights that are impossible to gain from model predictions and indirect measurements.

Several limitations of *in vivo* measurements must be considered when interpreting and applying the presented data. First, measurement errors are introduced by the use of buckle force transducers and during the conversion of their output to force. However, these errors are small relative to the forces measured (see methods section) and although they may affect the absolute force values, they will not influence the conclusions regarding the effect of position on force. Collection of data during CTR surgery is another limitation. The recorded motion and associated muscle forces may not accurately represent movements executed during daily activities since subjects have lost sensory feedback from the median nerve due to anesthesia. The lack of proprioception may have also limited the range of motion in some subjects, thus reducing the amount of information we were able to collect. In addition, positions of IP joints could not be monitored for many subjects because their thumbs blocked these joints in more flexed positions. Thus finger position was represented by MP joint angle and the effect of specific IP joint angles on force was not examined.

This study reports force changes in relatively normal, healthy tendons due to changes in finger and hand position during unresisted, low force motion. Forces will be higher in repaired tendons due to additional friction caused by the suture, scarring, and swelling. These additional forces need to be taken into account when applying the data obtained in this study to design of rehabilitation protocols.

Despite these limitations, the results have several clinically important implications. A tendon repaired with a four-strand suture and epitendinous suture should withstand the reported maximum forces generated in the FDP tendon during active flexion and active extension [68]. However, the maximum forces in the FDS tendon may exceed the repair strength, especially if the wrist is maintained in a flexed position. The position of the wrist and fingers may be adjusted to reduce tendon forces during rehabilitation exercises. This study found that flexor tendon forces increase with increasing finger flexion during active finger flexion, supporting hypothesis number two. Hypothesis number three was refuted; the force in the FDS tendon was higher at a flexed wrist posture compared with a neutral posture. Therefore, limiting the amount of finger and wrist flexion during active flexion and extension can decrease the tension in the repair and the probability of rupture. The study found that force in the two tendons was different at the same finger positions, a result that refutes hypothesis four. This differential effect of position on the two tendons should also be considered when designing a rehabilitation procedure. For example, a laceration of the FDP tendon alone may warrant a different protocol than an injury to both tendons since limiting the wrist flexion angle will decrease the amount of force in the FDS, but not the FDP tendon. Another interesting and unexpected finding that is at variance with hypothesis one was the small difference in tendon force between flexion and extension. It suggests that active motion in either direction at the rate performed by the subjects leads to similar tendon forces, and is contrary to the current belief that force can be limited by using active finger extension and passive flexion during the early stages of rehabilitation.

In vivo tendon force measurements can help us understand tendon mechanics during complex hand motions. Data from this study, combined with information about repair strength and the effects of forces on healing at different times following surgery, will lead to rehabilitation protocols scientifically designed to improve clinical outcomes.

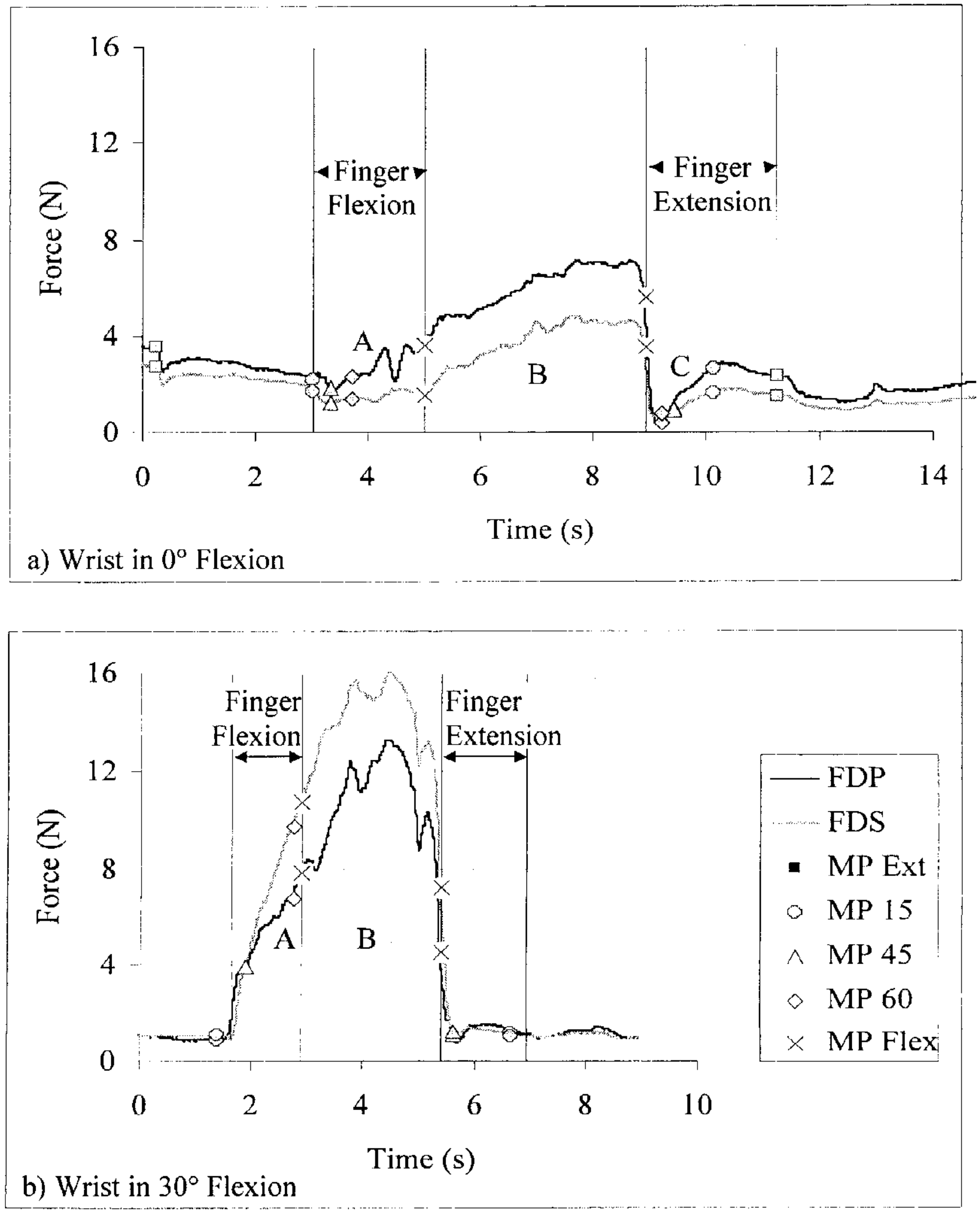


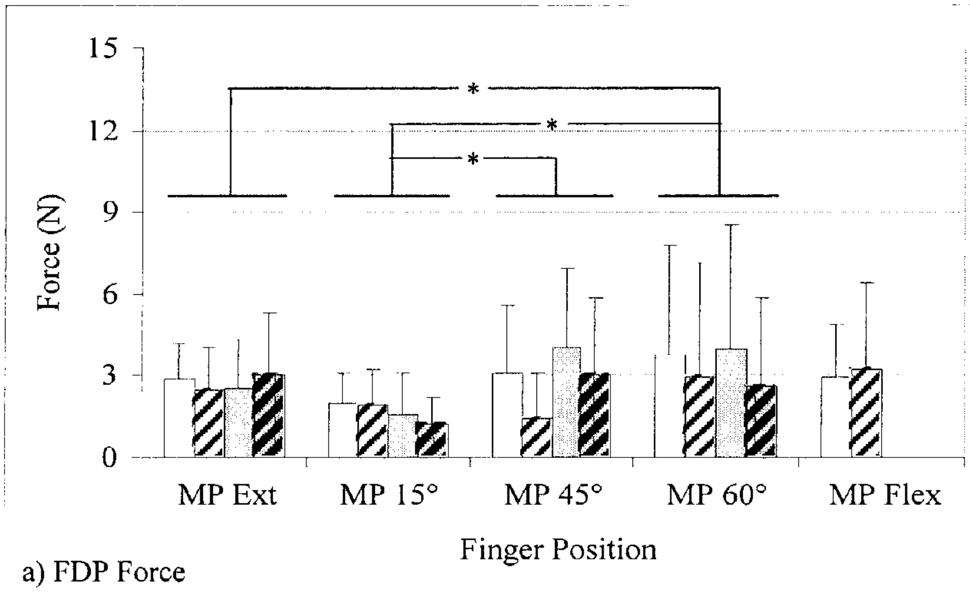
Figure 5.1. Example of forces in the FDP and FDS tendons from one subject during active finger flexion and extension with wrist at 0° flexion (a) and 30° flexion (b). Individual points indicate different finger positions as defined by MP joint angles and vertical lines indicate the start and end of finger motion. Region A corresponds to the time when the fingers are extended from 45° to full flexion; Region B indicates when the fingers are held in a fixed flexed position; and C corresponds to finger extension from 45° to 15° flexion.

Hand Posture	FDP		FDS	
	Active Flexion	Active Extension	Active Flexion	Active Extension
Wrist at 0°				
MP Ext	2.89 ± 1.33	2.47 ± 1.57	2.02 ± 1.16	1.84 ± 0.83
MP 15°	1.99 ± 1.12	1.91 ± 1.33	1.59 ± 0.77	2.04 ± 1.24
MP 45°	3.10 ± 2.46	1.44 ± 1.66	1.55 ± 1.20	1.47 ± 1.10
MP 60°	3.77 ± 4.01	2.96 ± 4.19	1.91 ± 1.98	1.59 ± 1.59
MP Flex	2.96 ± 1.95	3.24 ± 3.13	2.42 ± 4.20	2.74 ± 2.81
Wrist at 30°				
MP Ext	2.58 ± 1.75	3.04 ± 2.24	1.76 ± 1.18	1.93 ± 0.84
MP 15°	1.55 ± 1.58	1.26 ± 0.94	1.26 ± 0.54	1.56 ± 0.73
MP 45°	4.04 ± 2.88	3.02 ± 2.85	3.23 ± 2.49	3.01 ± 2.55
MP 60°	3.97 ± 4.56	2.60 ± 3.26	8.51 ± 10.70	7.16 ± 6.44
MP Flex		2.17 ± 2.35		10.71 ± 4.35

Table 5.1. Forces in the FDP and FDS tendons during active finger flexion and active finger extension at different wrist and finger positions. Data reported as mean ± standard deviation in Newtons.

Hand Posture	Active Flexion	Active Extension
Wrist at 0°		
MP Ext	7 ± 2 (9)	8 ± 4 (5)
MP 15°	16 ± 2 (10)	16 ± 2 (12)
MP 45°	45 ± 4 (11)	45 ± 1 (12)
MP 60°	59 ± 2 (10)	60 ± 1 (10)
MP Flex	69 ± 4 (7)	70 ± 4 (8)
Wrist at 30°		
MP Ext	5 ± 5 (7)	4 ± 4 (5)
MP 15°	17 ± 4 (12)	16 ± 2 (11)
MP 45°	45 ± 3 (12)	45 ± 2 (11)
MP 60°	58 ± 2 (8)	59 ± 2 (6)
MP Flex		65 ± 2 (3)

Table 5.2. Measured MP joint angles during active finger flexion and active finger extension at different wrist and finger positions. Data reported as mean ± standard deviation in degrees (sample size). Sample size is the number of data points (subjects) used to calculate mean force values at each position.



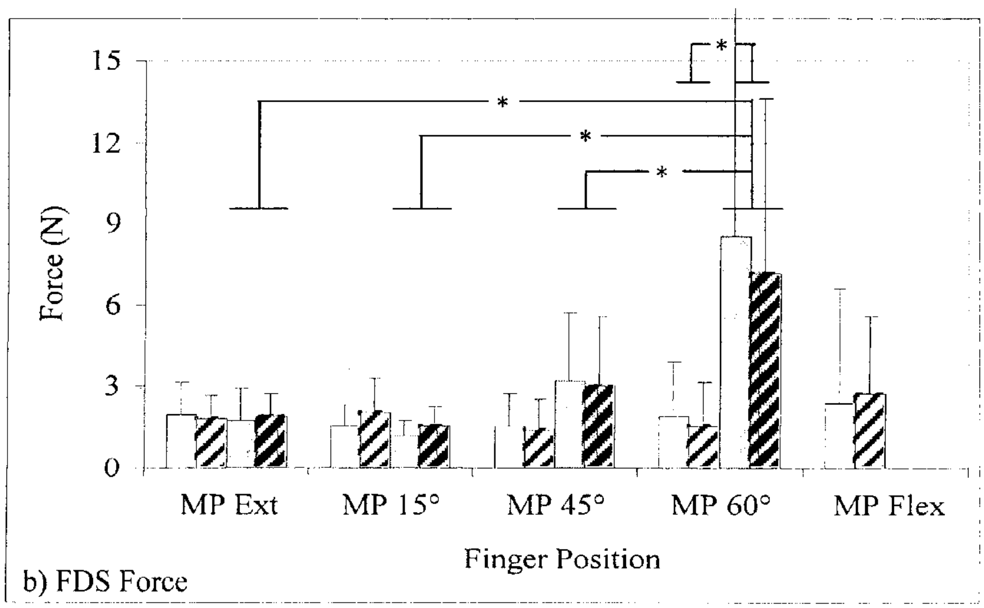


Figure 5.2. Forces in the FDP (a) and FDS (b) tendons during active finger flexion and extension at different wrist and finger positions. Data reported as mean ± standard deviation.

- □ Wrist at 0° Flexion, Finger Flexion Motion
- ▨ Wrist at 0° Flexion, Finger Extension Motion
- ░ Wrist at 30° Flexion, Finger Flexion Motion
- ▨ Wrist at 30° Flexion, Finger Extension Motion

* indicate significant differences in Tukey follow-up tests ($p < 0.05$)

	FDP				FDS			
Entire Motion								
Wrist at 0°	7.55	±	5.11	(0.91 - 17.32)	5.02	±	6.67	(1.20 - 25.58)
Wrist at 30°	6.92	±	4.74	(0.65 - 17.80)	9.03	±	12.87	(1.60 - 47.53)
Fixed Flexed Fingers								
Wrist at 0°	7.01	±	5.18	(0.74 - 17.32)	4.24	±	6.85	(0.50 - 25.58)
Wrist at 30°	5.66	±	4.19	(0.65 - 14.13)	8.57	±	13.09	(1.20 - 47.53)

Table 5.3. Maximum forces in the FDP and FDS tendons during the entire motion (flexion, fixed flexed finger position, extension) and during just the fixed flexed finger position. Data reported as mean ± standard deviation (range) in Newtons (N = 12).

CHAPTER VI:

CONCLUSION

6.1 SUMMARY OF FINDINGS

The effects of fingertip loading conditions and hand posture on *in vivo* forces in the flexor digitorum profundus (FDP) tendon and the flexor digitorum superficialis (FDS) tendon of the index finger were investigated during isometric pinching tasks and unresisted finger motion.

The force in both flexor tendons was linearly related to the force simultaneously applied at the fingertip during each isometric task that consisted of increasing the fingertip force from 0 to 15N at a specific rate while the finger was in a natural pinching posture. On average, the force in the FDP tendon was approximately 2.5 times larger than the force at the fingertip while the FDS tendon force was 1.5 times the fingertip force. These ratios of tendon forces to external force were influenced by the position of the finger. Up to 36% of tendon force variation was attributed to MP joint angle, DIP joint angle and the direction of the external force with respect to the fingertip. Force in the FDP tendon per unit fingertip force was higher when the MP joint was extended, the DIP joint was flexed and the external force was applied more parallel to the fingertip, a tip pinch position. In contrast, FDS force ratio was greatest when the MP joint was flexed, the DIP joint was extended and the external force was more perpendicular to the fingertip, a pulp pinch position. The predictions of a static, three dimensional biomechanical model agreed with these findings regarding the effect of MP joint angle on FDP and FDS tendon force ratios, but differed regarding the influence of the DIP joint. In addition, the model predicted and the measured FDP tendon force ratios for individual trials were not related, even though the MP joint angle, PIP joint angle, DIP joint angle, and external fingertip force direction and position assumed by subjects during each trial were used in the model. Surprisingly, the model predicted and measured FDS force ratios were negatively related. The mean FDP to fingertip force ratio predicted by the model was significantly higher than the mean ratio measured *in vivo* while the mean predicted and measured FDS ratios were not significantly different. The rate of force applied at the fingertip (1.5 N/s to 15 N/s) during the high precision, isometric pinch task did not significantly affect FDP or FDS tendon to fingertip force ratios.

Joint postures also influenced forces in the flexor tendons generated during active, unresisted, composite finger flexion and extension. Mean FDP tendon forces varied with hand posture between 1.3 and 4.0 N while mean FDS forces ranged from 1.3 to 10.7 N. FDP force increased as the fingers were flexed when the wrist was in either a neutral or a flexed position while FDS force increased with finger flexion only when the wrist was also flexed. The similarity in tendon force magnitudes measured during finger flexion and extension was unexpected.

6.2 Implications of Findings for Motor Control and Biomechanical Modeling

These *in vivo* findings can improve our understanding of motor control strategies by elucidating the roles of individual finger muscles during different tasks. Seven muscles act on the four bones and over the three joints of the index finger to produce a desired external force and/or motion. Coordination of these muscles is necessary to maintain postural stability and provide the appropriate torques at all joints in order to control finger position and external force output.

Since the finger is a redundant system where the number of muscles exceeds the degrees of freedom, different combinations of muscle forces can achieve the same goal during submaximal force production. The FDP and FDS muscles function together to produce the desired flexion force at the fingertip and to move the fingers towards the palm. Generally, the FDP generates more force than the FDS during both static and dynamic tasks and the FDP to fingertip force ratio is less variable among subjects than the FDS to fingertip force ratio. These findings imply that the FDP muscle plays the primary role in generation of flexion force while the FDS plays a secondary, less consistent part. However, the force and relative force contribution of each muscle exhibit large inter-subject variability suggesting the use of various control strategies that involve different amounts of load sharing between the two flexors.

The findings that changes in MP angle, DIP angle, and direction of external force all have significant and opposite effects on *in vivo* forces in the two flexor tendons indicate that finger posture influences muscle force distribution during submaximal isometric force generation tasks. Static models that describe how force and moment equilibrium are maintained around each joint also predict changes in muscle forces due to alterations in finger posture. They predict that finger posture will influence the amount of muscle force needed to balance a given external force because posture determines the moment arm length of the external force. However, the lack of agreement between model predictions and *in vivo* measurements implies that balancing internal and external moments is insufficient to describe the effect of finger posture on muscle force distribution. In this redundant system, loading conditions, such as load position and loading rate, may also affect muscle forces by influencing the selected motor control strategy. The fine motor control needed to generate precise force ramps may require high activation levels of the two extrinsic extensors and three intrinsic muscles in order to stabilize the finger and control joint torques. If higher extensor or intrinsic muscle forces are necessary to maintain postural stability at all finger joints during tip pinch than during pulp pinch, FDP force will increase. Such increases in co-contraction may explain the higher FDP forces measured when the DIP joint was flexed and the external force angle was less perpendicular to the fingertip. In contrast, the different loading rates investigated during the selected isometric pinch tasks may have required similar recruitment of antagonists, so no additional increases in FDP or FDS forces were observed at the higher rates.

Antagonist co-contraction may also play a role in motor control of unresisted motions. Although extensor activity during active finger flexion cannot be determined from this study, the similar FDP and FDS force magnitudes during finger flexion and extension suggest that flexors may be actively co-contracting during finger extension. In addition to antagonist muscle activation levels, passive, position-dependent forces in soft tissues (e.g. joint or muscle) may affect tendon forces during tasks with low force requirements, especially at extreme postures. During unresisted active finger flexion, the significant increase in FDS force observed when both the fingers and wrist were flexed may have been necessary to overcome the additional passive force from the extensor system at this position.

The complex structure of the finger and associated tissues and the precise task being performed must be considered to understand and predict forces generated in flexor tendons. Finger and hand posture, the amount of movement, and the magnitude and direction of applied external force, are factors that will determine muscle force distribution and associated motor control strategies.

Appropriate biomechanical models can be useful for investigating the impact of hand postures and other loading conditions on the relationship between tissue and applied forces. However, the complexity of the system necessitates the inclusion of numerous assumptions regarding finger structure and muscle function in models. These assumptions will affect model predictions. The findings that a conventional three-dimensional, biomechanical model poorly predicted *in vivo* tendon forces emphasize the importance of validating the predictive ability of models with *in vivo* measurements. In addition, the need to evaluate model predictions during different loading conditions is demonstrated by the finding that even though the mean measured and predicted FDS force ratios were not significantly different, model predictions regarding the effect of DIP joint position on FDS tendon force disagreed with *in vivo* measurements. The accuracy of certain finger structure and posture variables (e.g. FDP tendon moment arms, DIP angle, PIP angle, and the direction and position of external force) will affect the model's predictions, as demonstrated by the sensitivity analysis. In addition, the predictive ability of the model will depend on the optimization criteria selected to solve for muscle forces. Because external force can be produced by different muscle force distributions, one optimization criteria may be inadequate for predicting muscle forces. The selected motor control strategy that is represented by the optimization criteria may be influenced by loading conditions, including posture.

6.3 Applications of Findings to the Workplace and Clinic

Since high forces applied cyclically to the flexor tendons may lead to tissue damage, the risk of injury may be reduced by decreasing these internal forces. These forces may be lowered by adjusting external loading conditions in the workplace through improved tool and workstation design. Because flexor tendon forces are directly related to applied fingertip forces, reducing the amount of force required to perform a task will lower tendon forces. Changing finger joint position and angle of external force application can also reduce the amount of tendon force necessary to apply a desired fingertip force. If reduction of FDP tendon to tip force ratio is the objective, then DIP flexion angle should be decreased, the external force should be applied more perpendicularly to the fingertip and MP flexion angle should be increased. If the goal is to decrease FDS tendon to fingertip force ratio and the angle between the fingertip and applied force is greater than 61 degrees, then DIP angle should be increased. If external force angle is less than 61 degrees, then DIP angle should be reduced. These findings apply to finger postures where MP joint angles range between 12° and 39° (mean minimum and maximum angles), DIP joint angles between 14° and 37° and the angle between the external force and fingertip is between 57° and 79°. Since the FDP generates more force than the FDS per unit fingertip force, the FDP may be more susceptible to injury. Thus adjusting posture to reduce FDP force may be more important in tool design. If this is the case, the design should promote a larger MP flexion angle, smaller DIP flexion angle, and a more perpendicular external force application. These adjustments in finger and external force orientation may be achieved by changing the shape of hand tools. In contrast to position, adjusting fingertip loading rate over the range evaluated here during isometric force application may not be important for reducing tissue forces when designing switches and similar devices.

During tasks with no or low resistance, limiting the range of finger flexion can prevent higher FDP forces while simultaneously limiting finger and wrist flexion angles can reduce FDS forces.

The position of the wrist and fingers may also be adjusted to limit forces at a tendon repair site during hand movements prescribed for early protected motion rehabilitation. Limiting the amount of finger and wrist flexion during active finger flexion and extension may decrease the tension in the repair and the probability of gap formation and rupture while maintaining sufficient force to stimulate healing, prevent adhesion formation, and improve repair strength. Finger and wrist flexion can be increased over time to increase tendon forces as the repair site becomes stronger. The findings of this study suggest that limiting active composite finger flexion to MP angles of 45° or less may be a method of avoiding high tendon forces.

The small difference in tendon force between flexion and extension suggests that active motion in either direction leads to similar tendon forces. This finding contradicts the current belief that force can be limited by using active finger extension and passive flexion during the early stages of rehabilitation.

The differential effect of posture on force in the two tendons implies that a laceration of one tendon may warrant a different protocol than an injury to both tendons. For example, avoiding wrist flexion will decrease the amount of force in the FDS, but not the FDP tendon. Data from this study, combined with information about repair strength and additional frictional forces due to swelling and suture bulk at different times following surgery, can be used to match tendon forces to repair strength during key hand positions and exercises. Rehabilitation protocols can then be designed scientifically to improve clinical outcomes.

6.4 Limitations of Findings

Several limitations of *in vivo* measurements must be considered when interpreting and applying the presented data. First, measurement errors are introduced by the use of buckle force transducers and during the conversion of their output to force, but these errors are small (4-7%) relative to the measured force values. These errors may affect absolute force values, but they should not influence the conclusions regarding the effects of finger and wrist posture and loading rate on tendon forces. Second, because posture data was acquired in two dimensions, the assumptions that joint flexion and fingertip force direction were limited to the sagittal plane and that the camera was perpendicular to this plane had to be introduced. A custom apparatus that supported the hand and load cell was designed to position the finger in this plane and minimize errors associated with these assumptions during static loading. In addition, the surgeon monitored and adjusted hand position. During finger motion, positions of the DIP and PIP joints could not be determined for some subjects because their thumbs blocked these joints in more flexed postures. Thus finger position was represented by MP joint angle during active finger flexion and extension and the effects of IP joint angles on force were not examined. The small set of postures and loading rates tested per each subject is another limitation.

Collection of data during carpal tunnel surgery may also affect the results. The muscle forces may not accurately represent control strategies and movements executed during activities of daily living since subjects have lost finger sensory and proprioceptive feedback due to local anesthesia of the median nerve. The lack of proprioception may have also limited the range of motion in some subjects, thus reducing the amount of information we were able to collect. However, subjects were provided with real-time visual feedback of their fingertip force during static tasks and were given verbal instructions and feedback by the investigators. The performance during surgery was similar to the practice trials, indicating that some limitations were related to the task itself and not to the surgery. Therefore, muscle activity may not have been altered by the surgery.

6.5 Future Directions

6.5a Better Understanding of Motor Control Strategy and Development of Improved Biomechanical Models

This study addresses several questions regarding the effects of external loading conditions on internal flexor tendon forces and the validity of existing biomechanical models. At the same time, it raises new questions regarding these issues, suggests new hypotheses to examine and new experiments to perform.

The findings suggest that motor control strategy may depend on finger position during submaximal isometric force generation. For example, muscle co-contraction levels may change with joint position and external force direction to provide additional joint stability that may be required at certain finger postures. Such changes in muscle force distribution may explain the reported differences between measured and predicted tendon force ratios. Defining muscle force patterns that are consistent among subjects at submaximal force levels during different loading conditions can increase our understanding of associated control strategies and aid in the development of improved biomechanical models. In order to accurately predict changes in tendon forces due to changes in posture, multiple optimization functions may be needed to represent different motor control strategies and to solve the indeterminate system of equalities and inequalities.

A better understanding of muscle force distributions can be gained with additional *in vivo* measurements. Tendon force measurements for a larger set of postures per subject can provide a more complete characterization of the system and can help to improve biomechanical models. Additional natural pinching positions can help clarify muscle contributions at different postures and define within and between subject differences in motor control strategies. In addition, the effects of individual posture parameters on tendon forces can be determined by altering the position of one variable, such as the angle of one joint, at a time. For example, changes in force due to changes in DIP joint angle can provide information about adjustments in control strategies between tip and pulp pinch positions within subjects. Additional insight can also be gained by simultaneously investigating the activity of other muscles and determining the levels of co-contraction at different postures with EMG techniques.

The information gained by examining muscle forces at different positions can help to define more appropriate optimization functions for a new model and their potential dependence on position. Improvements in data collection techniques can provide more accurate information and improve the predictive ability of the model. The accuracy of finger structure parameters can be improved by measuring each subject's tendon moment arms and finger segment lengths at different positions with *in vivo* imaging techniques such as MRI. Measurements of finger posture and external force position and direction can be improved by acquiring better video images and automating their analysis with appropriate software.

6.5b Tendon Forces during Additional Clinically Relevant Maneuvers

Understanding the magnitude of flexor tendon forces during other rehabilitation motions, commonly used in the clinic, can help to design more effective rehabilitation protocols and improve clinical outcomes. Tendon forces during additional maneuvers can be investigated with *in vivo* measurements. Flexor forces during passive flexion and extension of the fingers can be compared with forces during active flexion and extension at different finger and wrist positions in order to determine the type and extent of motions that should be initiated at different times after the surgery. Since decreasing tendon force variability across subjects can help to generalize the findings and develop recommendations based on them, additional training prior to surgery and more instructions during the procedure can be given to try to obtain more consistent results. To improve our understanding of the effects of different finger and wrist positions on tendon forces, all joint angles can be continuously measured during these motions with either better video data acquisition and processing techniques or with electro-goniometers.

6.5c Tissue Response to Loads

In vivo measurements can help to define flexor tendon forces generated during the execution of occupational and other daily activities. To develop more effective prevention and treatment procedures, the role of these forces in the tendon injury mechanism needs to be better understood. However, information about specific loads that can lead to injury and about tissue changes during early stages of the disease process is difficult to obtain from human studies. *In vitro* organ culture models provide an alternate method to study tissue responses to different, precisely controlled, physiological loading conditions. Changes in the mechanical, structural, and biological properties of tendons in response to different loading regimens can be examined with an *in vitro* organ culture model. Information about the types of tendon load (magnitude, frequency, duration) that lead to damage can be combined with information about the amount of force in a tendon during different external loading conditions to limit exposure to acceptable levels and reduce the risk of tendon disorders. A better understanding of tissue response during early stages of tendon degeneration can also be used to develop and test different treatment options.

Data from different types of studies, including epidemiological investigations, *in vivo* measurements, biomechanical model predictions, and *in vitro* experiments, should be synthesized to gain a better understanding of tendon injury and repair mechanisms in order to design more effective prevention and treatment methods.

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