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Tensions of the flexor digitorum superficialis are higher than a current model predicts

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Abstract

Existing isometric force models can be used to predict tension in the finger flexor tendon, however, they assume a specific distribution of forces across the tendons of the fingers. These assumptions have not been validated or explored by experimental methods. To determine if the force distributions repeatably follow one pattern the *in vivo* tension of the flexor digitorum superficialis (FDS) tendon of the long finger was measured in nine patients undergoing open carpal tunnel release surgery. Following the release, a tendon force transducer (Dennerlein et al. 1997 *J. Biomechanics* 30(4), 395–397) was mounted onto the FDS of the long finger. Tension in the tendon, contact force at the fingertip, and finger posture were recorded while the patient gradually increased the force applied by the fingertip from 0 to 10 N and then monotonically reduced it to 0 N. The average ratio of the tendon tension to the fingertip contact force ranged from 1.7 to 5.8 (mean = 3.3, s.d. = 1.4) for the nine subjects. These ratios are larger than ratios predicted by current isometric tendon force models (mean = 1.2, s. d. = 0.4). Subjects who used a pulp pinch posture (hyper-extended distal interphalangeal joint (DIP)) showed a significantly ($p = 0.02$) larger ratio (mean = 4.4, s.d. = 1.5) than the five subjects who flexed the DIP joint in a tip pinch posture (mean = 2.4, s.d. = 0.6). A new DIP constraint model, which selects different force distribution based on DIP joint posture, predicts force ratios that correlate well with the measured ratios ($r^2 = 0.85$). © 1998 Elsevier Science Ltd. All rights reserved.

Keywords: Finger; Tendon; *In vivo* force; Force model; Pinch

1. Introduction

Although injuries to the tendons at the wrist and adjacent tissues associated with repetitive work are reported to be high (55% of all work related repetitive motion disorders: Bureau of Labor Statistics, 1995), the injury mechanisms are not well understood. Along with motion, posture and vibration, the force exerted during a task is a risk factor for tendon related disorders (Armstrong et al., 1987; Moore and Garg, 1994; Silverstein et al., 1986). Understanding how force is transmitted from the site of external application (fingertip) to the internal tissue (finger flexor tendons) is critical to understanding the mechanics of repetitive motion disorders. This understanding determines how an externally applied force is supported by the internal biological tissues and such knowledge identifies tendons which are exposed to higher forces. The knowledge of *in vivo* tendon tension also guides techniques of tendon repair, procedures for rehabilitation (Schuind et al., 1992; Soejima et al., 1995; Komi, 1990) and the design of joint replacements (Weightman and Amis, 1982).

Currently clinicians and researchers use models predicting tendon force (e.g. Harding et al., 1993; Chao et al., 1989) based on the equations of static equilibrium at each joint of the finger and a set of tendon tension constraints to evaluate the supporting loads. Because there are more unknown tendon tensions than equilibrium equations, solution of this indeterminate problem requires assumption of additional constraint relationships, usually constraints on how the forces are supported by the muscles

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and tendons (see the methods and An et al. (1984) for description and evaluation of different constraint techniques).

Validating the aforementioned models by experimental measurement was undertaken in one study by Schuind et al. (1992); however, variations in force distributions were not explored or discussed. Schuind et al. (1992) reported the maximum finger flexor tendon-to-tip force ratios for pinch task but did not report joint posture thus precluding comparison of theoretical predictions with the measured tendon forces.

The goals of this study were to measure the *in vivo* tendon tension in the flexor digitorum superficialis (FDS) of the long finger during isometric pinch and to model the measured forces in the tendon based on the subjects' joint thickness, hand length, finger joint posture and the contact force at the fingertip. The specific research question was: are the forces, as reflected in the force of the FDS, distributed across the tendons of the finger consistent with the patterns predicted in the literature? This question impacts the treatment, the prediction, and the research of repetitive motion disorders associated with the tissues of the hand and wrist. The results suggest that the current models do not predict all possible force distributions for the finger. We, therefore, propose a new model in the discussion which better predicts the wider range of force distribution based on observed finger postures.

2. Methods

Nine subjects (eight females and one male, mean age 48 ± 21 years), undergoing open carpal tunnel release surgery at the University of California, San Francisco (UCSF) participated in the study. The procedures were approved by the University of California San Francisco Committee on Human Research and U.C. Berkeley Committee for the Protection of Human Subjects. Prior to surgery, the hand length and joint thicknesses of the metacarpal phalangeal joint (MP), the proximal interphalangeal joint (PIP), and the distal interphalangeal joint (DIP) were measured for each subject. Hand lengths, measured from the distal wrist crease to the tip of the long finger, ranged from 5 to 57 percentile of the population and the average joint thicknesses ranged from 60 to 93 percentile (Garrett, 1970 a,b). The subjects practiced the pinch tasks to be performed during the experiments prior to their surgery.

Surgery was performed, as usual, under local anesthesia at the incision site, and each subject retained motor control of the forearm musculature throughout the procedure. The subjects were prone with the shoulder abducted to 90° and the palm rotated upwards resting on the operating table. The carpal tunnel contents were exposed through a 5 cm longitudinal carpal tunnel incision. The flexor digitorum superficialis (FDS) tendon of the long finger was identified and the synovium removed. A gas-sterilized tendon force transducer (Dennerlein et al., 1997) was then mounted onto the FDS tendon. The subject flexed the long finger 10 times to precondition and seat the transducer onto the tendon. The tendon thickness was then measured *in situ* for use in the transducer calibration factor (An et al., 1990; Dennerlein et al., 1997). The subject's forearm was rotated 90° toward a neutral forearm posture with the thumb upward and with the palm towards the feet. The tourniquet was released prior to data collection and the forearm and wrist were manually stabilized during the pinch tasks with the wrist straight.

The subject gradually (>10 s) increased from 0 to 10 N the force applied by the fingertip to a single axis load cell (GreenLeaf Medical Pinch Meter, Palo Alto, CA) and then decreased it monotonically to 0 N while observing a measure of the force on a visual monitor. These ramps were repeated two to three times. The force range is typical of that applied in keyboard and other occupational tasks (Rempel et al., 1994; Johnson et al., 1993). Data from the load cell and the tendon force transducer were recorded on a computer data acquisition system at 50 samples per second.

The gradual pinch tasks were conducted at three different finger postures which ranged from an extended to a flexed pinch posture. Preset goniometers guided the surgeon to set the angle of the MP joint and the load cell was aligned such that the fingertip would not slip off the load cell. During pinch subjects self-selected either a tip pinch posture (DIP joint flexed) or a pulp pinch posture (DIP joint fully extended or hyper-extended, DIP angle $\leqslant 0°$). A video camera mounted above the surgical field recorded the sagittal view of the finger posture and load cell alignment during the isometric pinch tasks for all but the first subject. For the first subject the video viewed the hand from the side and thus specific posture data from this subject was not acquired and ratios not predicted. The angle of intersection of lines aligned with the dorsal surface of adjacent finger segments defined the approximate joint angle which is later used in the models (Long et al., 1960).

The surgeon removed all transducers once the tasks were completed, and surgery continued. The procedure added approximately 20 minutes to surgery.

A model, incorporating the three observed postures and subjects' anthropometry, predicts the tendon tension to tip force ratio for the eight subjects with posture measurements. The three predictions of the force ratio were averaged for each subject because the variations of the tendon tensions and the ratio predictions within subjects were small. The tendon tension model represents the finger as three movable rigid bodies, (the proximal, middle and distal phalanxes) linked by hinges at the phalangeal joints and moving in the sagittal plane. The tendons of the six muscles that move the long finger span these joints (Table 1 and Fig. 1). The tendons facilitating extension through the contraction of the extensor and intrinsic (Ulnar Interosseous (UI), Radial Interosseous (RI), Lumbrical (LU)) muscles of the long finger are: extensor slip (ES), radial band (RB), ulnar band (UB), and terminal extensor (TE) (Chao et al., 1989).

$$TE = RB + UB,$$

$$RB = 0.133\ RI + 0.167\ EDC + 0.667\ LU,$$

$$UB = 0.313\ UI + 0.167\ EDC,$$

$$ES = 0.133\ RI + 0.313\ UI + 0.167\ EDC + 0.333\ LU. \quad (1)$$

Tension in the tendon acting at a distance from the axis of rotation (the moment arm) and a load applied to the fingertip create moments at the joints. Hence, the static equilibrium balance for the jth joint is:

$$\sum_i F_i r_{ij} + F_{\text{tip}} r_{j,\text{tip}} = 0, \quad (2)$$

Table 1
Tendons of the long finger that cross each joint and their contribution to either flexion or extension

Joint	Flexion	Extension
MP	Flexor digitorum superficialis (FDS), Flexor digitorum (FDP), Ulnar interosseous (UI), Radial interosseous (RI), Lumbrical (LU)	Extensor digitorum Communis (EDC)
PIP	FDS, FDP	Extensor slip (ES), Radial band (RB), Ulnar band (UB)
DIP	FDP	Terminal extensor (TE)

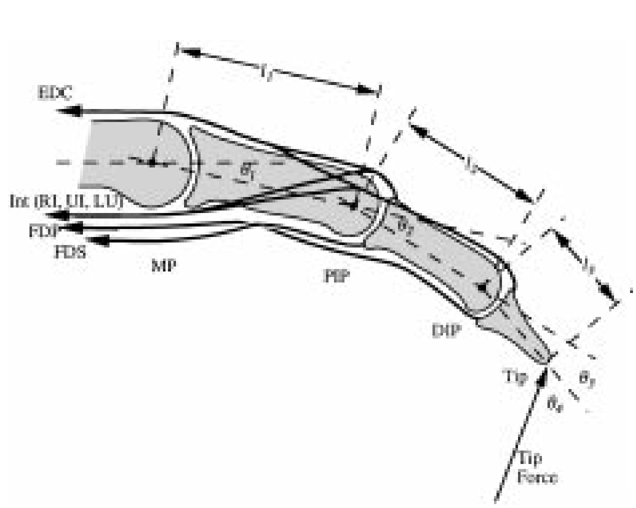

Fig. 1. The two dimensional model of the finger. Three segments are connected by hinges at the MP, PIP, and DIP joints.The relative joint angles (θ_1, θ_2, θ_3), the angle of the applied fingertip load θ_4, and segment lengths (l_1, l_2, l_3) are shown.

Table 2
Three solution methods for the indeterminate problem

	Method	Reference	DIP constraint
1	Set tension in RB, UB, TE, and EDC = 0;	Johnson et al. (1995)	Set tension in FDP = 0 when DIP $\leqslant 0°$.
2	Lump RI, UI and LU into INT and EDC = 0 INT = RI + UI + LU	Harding et al. (1993), Weightman and Amis (1982), Smith et al. (1964)	Set tension in FDP = 0 when DIP $\leqslant 0°$.
3	Optimization methods; $J = \sum_{\text{tend}} F_{\text{tend}}^2/\text{PCSA}^2$ muscle stress	Chao et al. (1989)	Add constraint moment at the DIP joint when DIP $\leqslant 0°$

where F_i is the tension in the ith tendon, r_{ij} is the moment arm for the ith tendon at the jth joint, and $F_{\text{tip}}\ r_{j,\text{tip}}$ is the moment at the jth joint of the force at the fingertip. The joint thickness and the posture measurements are used to predict the moment arms (Armstrong and Chaffin, 1978; An et al., 1983). Hand length measurements determined the distances between the axes of rotation of adjacent joints (Buchholz and Armstrong, 1992). Eqs. (1) and (2) plus the equilibrium moment balances at the MP joint under abduction and adduction axis yield four constraint balances involving the six unknown muscle-tendon tensions. Additional constraints maintain the muscle-tendon forces as tensile only (force > 0).

Three methods of solution were applied to the indeterminate problem (Table 2). Method 1 prescribes that only the extrinsic flexor tendon forces act about the PIP and DIP joints, providing two equations and two unknowns (FDS and FDP; Johnson et al., 1995). Method 2 assumes the three intrinsic muscles act in synchronization and assumes the tension in the extensor (EDC) is zero, giving three unknowns and three balances (Harding et al., 1993; Weightman and Amis, 1982; Smith et al., 1964). Method 3 assumes the muscle contraction balances the tension to minimize the sum of the squares of the muscle stress (An et al., 1984; Pedotti et al., 1978):

$$J = \sum_i (F_i^2/\text{PCSA}_i^2), \quad (3)$$

where PCSA_i is the physiological cross sectional area of the ith muscle from Chao et al. (1989). Quadratic programming techniques provide a solution for the six muscle-tendon forces.

3. Results

The ratios of FDS tendon-to-tip force during isometric pinch varied substantially between subjects. For each subject the tension in the FDS tendon was proportional to the force applied at the fingertip (Fig. 2). For each posture, two or three force histories were collected. The slopes of the linear regression of the tendon and tip force data were equal for the loading and unloading portion of the curves and the variation (standard deviation) of the slopes between the repeated force histories ranged from 0.1 to 0.7. Therefore the slopes of the loading and unloading regions were averaged to represent the tendon-to-tip force ratio for each posture. Correlation of the linear regression ranged from $r^2 = 0.85$ to 1.00 across all subjects. Subject tendon-to-fingertip force ratios averaged across the three postures ranged from 1.7 to 5.8 (mean = 3.3, s.d. = 1.4). Within each subject, the variation (standard deviation) of the ratios across the three postures ranged from 0.1 to 1.9.

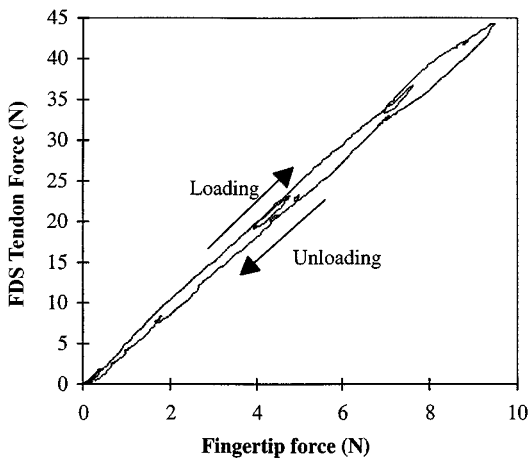

Fig. 2. The FDS tendon and tip force for one subject. The tendon force is proportional to the tip force with correlation coefficients ranging from 0.91 to 1.0 across all subjects.The tendon-to-fingertip force ratio is the slope of the linear regression of the tendon and fingertip force data.

While the postures varied widely between subjects and covered a large range of pinch scenarios, the predicted ratios from the implemented tendon force models varied little and were smaller than the measured values. The most extended posture involved joint angles of 32°, 11°, and −15° for the MP, PIP, and DIP and 72° for the angle between the distal phalanx and the applied tip force, while the most flexed posture involved joint angles of 30°, 95°, 0°, and 60°. Joint angles ranged from 0° to 47° for the MP, from 11° to 95° for the PIP, and from −15° to 44° for the DIP, and the angle between the distal phalanx and the applied tip force ranged from 24° to 86°. The predicted (solution method 3) tendon-to-tip force averaged across the three postures ranged from 0.7 to 1.9 (mean = 1.2, s.d. = 0.4) across all the subjects. The within subject variation ranged from 0.0 to 0.3. Table 3C presents the results of the two other solution methods. The three different solution methods for the indeterminate problem did not yield significantly different predictions for the tendon tension (Table 3C), similar to the results of Pedotti et al. (1978) for the lower extremity during human locomotion. The mean predicted ratio of 1.2 is less than half the value of the measured ratio and the largest average predicted ratio of 1.9 is only slightly larger than the smallest measured ratio of 1.7. Furthermore, the subject averaged predicted ratios correlate poorly ($r^2 = 0.04$) with the measured ratios (Fig. 3).

The ratio of FDS tendon-to-tip force depended upon the DIP posture selected by the subjects during the

Table 3
In vivo and predicted finger tendon force ratios during fingertip loading. (A) The measured FDS tendon force from this study. (B) *In vivo* tendon forces reported by Schuind et al. (1992). (C) Predicted forces using measured joint postures and the three different solutions methods. (D) Predicted forces using the three different solution methods with DIP constraint added to the model. (E) Predicted tendon forces reported in the literature

	Source	FDS[a]	FDP[a]	Intrinsic[a]	EDC	r^2 [b]	RSME[c]
A	*In vivo*	3.3 (1.4)					
B	Schuind et al. 1992 (*In vivo*)	1.7 (1.5)	7.9 (6.3)				
C	Method 1	1.1 (0.4)	3.1 (0.7)			0.05	1.3
	Method 2	1.2 (0.4)	3.9 (0.7)	2.8 (1.0)		0.02	1.3
	Method 3	1.2 (0.4)	3.8 (0.8)	2.8 (0.6)	0	0.04	1.3
D	Method 1 (DIP Constraint)	3.1 (1.8)	1.5 (1.4)			0.62	0.8
	Method 2 (DIP Constraint)	3.7 (2.0)	1.6 (1.7)	1.7 (1.1)		0.71	0.7
	Method 3 (DIP Constraint)	2.9 (1.7)	2.2 (1.4)	2.3 (0.6)	0	0.85	0.5
E	Chao et al. (1989)	0.3–2.1	1.9–3.1	2.4–3.9	0		
	Harding et al. (1993)	0.8–2.7	1.2–3.2	0.5–3.2			
	Weightman and Amis (1982)	1.6–2.8	2.1–2.6	1.3–2.6			

[a]Mean (standard deviation) tendon force in units of applied tip force.
[b]Correlation (r^2) between the predicted and the measured force.
[c]RSME is the root square mean error.

isometric pinch task. Five subjects applied a tip pinch posture (DIP joint flexed) and four subjects applied a pulp pinch posture (DIP joint fully to hyper extended, DIP angle $\leqslant 0$; Chao et al., 1989). The tendon-to-tip force ratios measured for the four pulp pinch subjects (mean = 4.4, s.d. = 1.4) were significantly higher (two-sample t-test performed and $p = 0.022$) than the ratios observed during tip pinch postures (mean = 2.4, s.d. = 0.6). This trend contradicts those predicted by the implemented model (Table 4) and the predictions of Chao et al. (1989).

4. Discussion

Direct measurement of tension in the tendon provides a measure of forces within the human musculoskeletal system that have been earlier predicted by isometric force models. These models make assumptions about the distribution of forces in the tendons. By directly measuring the tension in the FDS we observed that the FDS tendon force varies widely across subjects, more so than models predict. We propose that now the force is supported by different tendons varied across subjects and can be predicted based on the observation of extreme joint postures.

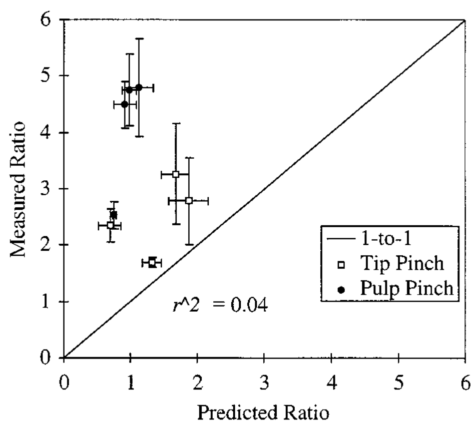

Fig. 3. Mean measured FDS tendon-to-tip force ratios and the predicted isometric force ratios using solution method 3. The error bars represent one standard deviation of the three measurements (vertical) and of the three predictions (horizontal) for the three postures. Current models underpredict the large range of measured ratios.

Schuind et al. (1992) conjectures that the large variation and the differences between the predicted and observed values (s.d. = 1.4) were due to a difference between actual and theoretical postures. The model here used the large range of observed postures; yet, it did not predict the large variations (s.d. = 0.4) seen between subjects (Table 3C and Fig. 3). The predicted values were within the range of predicted values reported by others in the literature (Table 3E). The FDS tendon-to-tip force ratios measured here were larger than the ratios reported by Schuind et al. (1992) but they limited the DIP posture to tip pinch with flexion of the DIP. The mean ratio of the five tip pinch subjects in this study are slightly larger (mean = 2.4, s.d. = 0.6) than that reported by Schuind (mean ratio of 1.7).

The error in the ratios predicted by the model from errors in estimation of tendon moment arms, segment lengths and measured postures is smaller than the observed variation of the ratios between subjects. The sensitivity of the model prediction to errors in the anatomical input parameters and joint posture angles was examined by perturbing the subject average data. Each flexor moment arm was altered ± 1 mm and the force ratio was calculated. Segment length (± 2 mm) and the relative load angle and joint angles ($\pm 10°$) were tested independently. The selected change in moment arm and segment length values were equivalent to the standard errors of these measures as reported by Armstrong and Chaffin (1978) and Buchholz and Armstrong (1992), respectively. The values of the joint angles are the estimated errors of recording the joint angles with the video based system. The model was most sensitive to varying the FDP moment arm at the DIP joint; this caused relative changes of between 0.25 and 1.0 in the predicted FDS tendon-to-tip force ratio. Varying the moment arm for the other joints led to median variation of the predicted ratios of between 0.1 and 0.6. Variation of the segment lengths and joint angles changed the predicted force ratio less than 0.4 and 0.3, respectively.

The measured difference between the FDS force ratios of the pulp and tip pinch postures results from the mechanics of the DIP joint in those postures. In the theory of the models the DIP acts as an unrestrained hinge joint and predicts FDS force ratio decreases in the pulp pinch posture. As the DIP extends, the angle of the applied load and hence the mechanical advantage of the applied

Table 4
Ratios of FDS tendon to tip force, subject mean (standard deviations)

	In vivo	Schuind et al. (1992)	Predicted[a]	Chao et al. (1989)	New Model[a]
Tip pinch	2.4 (0.6)	1.7 (1.5)	1.4 (0.5)	1.7–2.1	1.4 (0.5)
Pulp pinch	4.4 (1.4)	N/A	0.9 (0.2)	0.3–1.3	4.2 (1.0)

[a]Predicted using solution method 3.

fingertip load about the DIP joint increases. The tension of the FDP, the only flexor tendon at the DIP, increases to maintain equilibrium at the DIP joint. The increased FDP tension allows a decreased FDS tension required for equilibrium at the PIP joint. The results of Chao et al. (1989) and the model implemented here predict this balance of FDP and FDS forces at equilibrium. Yet, the predicted decrease of FDS tension was not observed between the tip and pulp pinch subjects.

In actuality the FDS tension of the pulp pinch subjects was significantly higher than the tension measured in the tip pinch subjects (Table 4). We propose that a passive joint constraint moment, produced by soft, connective tissues of the DIP joint, contributes to static equilibrium at the DIP joint near full extension. A pulp posture when the DIP joint is fully or hyperextended suggests a joint constraint moment is required to maintain equilibrium. Assumption of an unrestrained hinge joint is unrealistic in this circumstance.

A new DIP constraint model which selects one of two model structures based upon observed posture was developed to explain the higher FDS tendon force. In the tip pinch model the DIP is an unrestrained hinge joint where tendons alone maintain equilibrium. In the pulp pinch posture the new model includes a constraint moment at the DIP joint that assists the FDP by supporting the moment of the load at the fingertip. The implementation of the constraint moment for the three solution methods is presented in Table 2. For methods 1 and 2 the new model assumes the FDP force vanishes and the DIP joint is rigid (distal and middle phalanxes are a single rigid body). Method 3 adds a joint constraint moment at the DIP joint to the set of variables in Eq. (3). The moment required is predicted by optimization.

The predicted tendon-to-tip force ratios using this new DIP joint constraint model (Table 3D) correlated well ($r^2 > 0.62$) with the ratio of the measured tendon-to-tip forces (Fig. 4). The variance of the subject data accounted for by this new model ($r^2 = 85\%$) and the root mean square error (0.51) were best with Method 3. The new model predicts a large increase in FDS tendon force during pulp pinch (Table 4) whereas the old model did not.

During the isometric pinch the tendon force was proportional to force applied by the fingertip as expected because the mechanical advantage of the tendon to the tip remains constant. The linearity also suggests that recruitment of synergist muscles, like the flexor digitorum profundus, follows the flexor digitorum superficialis.

Only one of the six tendons of the long finger was measured in this study, providing a limited view of all the internal forces. Simultaneously measuring FDP force would further support the DIP constraint model, however the behavior of the FDS follows well the predictions of the model. Other limitations include the invasiveness

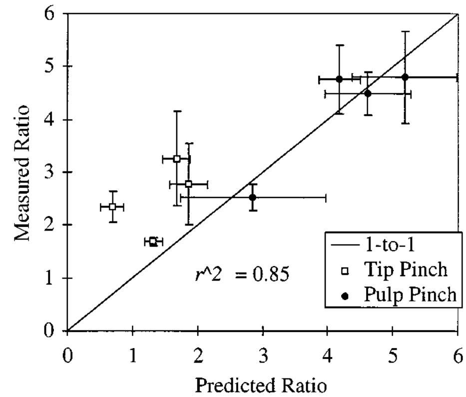

Fig. 4. Mean measured FDS tendon-to-tip force ratios and the predicted isometric force ratios from the new DIP constraint model using solution method 3. The error bars represent one standard deviation of the three measurements (vertical) and of the three predictions (horizontal) for the three postures. This new model better explains the large variation of force ratios by selecting different force distribution assumptions for different observed postures.

of the *in vivo* procedure. The awkward posture of the upper extremity, the slight sedation of the subjects, and the use of local anesthesia during surgery may alter the process of pinching by a subject. However, hyperextension of the DIP joint in the pulp pinch posture does occur during work tasks such as touch typing (Dennerlein et al., 1993) and writing. Moreover, the anesthesia was applied locally at the carpal tunnel and does not alter the mechanics linking tendon force to tip force.

A limitation of the models is that they do not include the effects of co-contraction of antagonist muscles and therefore underpredict tendon force ratios. The mean difference between the measured and predicted force ratios for the pulp pinch postures is 0.07, yet, in the tip pinch postures the model under-predicts the measured ratio by approximately 1 (Fig. 4). While model sensitivity may explain some of this difference, antagonist co-contraction may exist. The tip pinch posture requires active stabilization of the DIP joint by co-contraction whereas the pulp pinch posture can rely on the passive joint constraint to stabilize the joint. Therefore, increased co-contraction is expected in the tip pinch postures.

In conclusion, the isometric, *in vivo*, tendon force measurements indicate a large variation of FDS tendon-to-tip force ratios across individuals for a pinch task. The measured range of FDS tensions was larger than the range of ratios predicted by isometric tendon force models because the models select only one distribution of forces across the finger tendons. The new DIP constraint model accounts for the different finger kinematics between the pulp and tip pinch postures and predicts the wide range of measured tendon-to-tip force ratios with high correlation. The findings once again support the need to validate models predicting tendon force.

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