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Effect of Finger Posture on the Tendon Force Distribution Within the Finger Extensor Mechanism

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Abstract

Understanding the transformation of tendon forces into joint torques would greatly aid in the investigation of the complex temporal and spatial coordination of multiple muscles in finger movements. In this study, the effects of the finger posture on the tendon force transmission within the finger extensor apparatus were investigated. In five cadaver specimens, a constant force was applied sequentially to the two extrinsic extensor tendons in the index finger, extensor digitorum communis and extensor indicis proprius. The responses to this loading, i.e. fingertip force/moment and regional strains of the extensor apparatus, were measured and analyzed to estimate the tendon force transmission into the terminal and central slips of the extensor hood. Repeated measures analysis of variance revealed that the amount of tendon force transmitted to each tendon slip was significantly affected by finger posture, specifically by the interphalangeal (IP) joint angles ($p < 0.01$). Tendon force transmitted to each of the tendon slips was found to decrease with the IP flexion. The main effect of the metacarpophalangeal joint angle was not as consistent as the IP angle, but there was a strong interaction effect for which MCP flexion led to large decreases in the slip forces ($> 30\%$) when the IP joints were extended. The ratio of terminal slip force: central slip force remained relatively constant across postures at approximately 1.7:1. Force dissipation into surrounding structures was found to be largely responsible for the observed force-posture relationship. Due to the significance of posture in the force transmission to the tendon slips, the impact of finger posture should be carefully considered when studying finger motor control or examining injury mechanisms in the extensor apparatus.

Keywords

Finger; Tendon slip force; Extensor hood; Finger posture

1. Introduction

The forces generated by skeletal muscles are converted into joint torques, which subsequently generate movements of the corresponding body segments or exert forces on external objects. The transformation of muscle forces into a proper sequence of net joint torques to achieve task-specific goals requires elaborate coordination of different groups of muscles. Muscle activation patterns in various human tasks have been examined in order to better understand the principles underlying such complex motor control. Furthermore, muscle-driven simulations of various motor tasks, generally adopting optimization techniques that minimize task-specific objective

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functions, have been performed. These computer simulations allow for precise identification of the role of each muscle in conducting given tasks (for review, see [1]).

Performing similar analysis for the finger kinetics in manual tasks is difficult, mostly because of the anatomical complexity of the musculoskeletal structure of the fingers. Although the loading characteristics of finger flexor tendons in manual tasks were examined in several studies [2,3], quantifying distribution and integration of tendon forces within the finger extensor apparatus, a complex tendon network that transmits multiple tendon forces to different finger segments [4,5], remains challenging. The functional/kinetic outcomes of each muscle within the extensor mechanism should be precisely understood in order to comprehend the sophisticated temporal and spatial coordination patterns of multiple muscles involved in finger movements [6–8]. In addition, as many finger injuries involve rupture of the terminal or central slip of the extensor apparatus (e.g., Boutonniere deformity or mallet finger [9]) information about tendon force transmission to these slips could aid in understanding of the injury mechanisms or in prevention of these injuries.

Force distribution characteristics within the extensor mechanism can be deduced from its mechanical response to the applied tendon force. Strains and geometric changes in selected regions of the finger extensor mechanism have been measured under multiple tendon loadings that generated various finger postures [10–12] in order to examine the interaction of multiple tendon forces within the extensor mechanism and to better understand the functions of the extensor muscle-tendon units. In a recent study [5], the tensions at the proximal (central) and terminal slips of the extensor apparatus were directly measured, using force transducers, during different combinations of applied tendon forces. Note that in these studies, however, multiple tendon forces were applied simultaneously and the resultant behavior of the extensor mechanism was observed. Although concurrent activations of multiple muscles are generally observed in human manual activities [6,13], the propagation characteristics of individual tendon forces within the extensor apparatus become difficult to identify under the simultaneous tendon loading conditions. More importantly, the effects of the finger posture on the tendon force transmission characteristics have not been investigated in previous studies. Finger posture was found to have a significant effect on the joint moment generation in a recent study [14].

In this study, therefore, we aimed to elucidate the effects of finger posture on the transmission characteristics of the individual tendon forces within the extensor apparatus. It was hypothesized that the force transmitted to distal and terminal slips from the extrinsic extensor tendons will be significantly affected by the change in finger posture. Specifically, two extrinsic extensor tendons, extensor digitorum communis (EDC) and extensor indicis proprius (EIP), of the index finger were examined due to their functional importance in finger extension [15]. Two responses of the extensor apparatus to the tendon loading, i.e. fingertip force/moment and specific regional strains of the extensor hood, were measured and analyzed. These data were used to elucidate the tendon force transmission into the terminal and central slips that insert into the distal and middle phalanges of the finger, respectively.

2. Methods

2.1 Experimental protocol

Five fresh-frozen cadaveric hand specimens (3 female, 2 male; 3 right hands, 2 left hands), transected midway between the wrist and elbow, were used for the experiment (index finger length mean $\pm$ SD: 101.8 ± 5.8 mm). Anatomical parameters of each specimen, i.e. segment length (i.e. distal phalanx, middle phalanx, and proximal phalanx) and the thickness of each joint, i.e. distal interphalangeal (DIP) joint, proximal interphalangeal (PIP) joint, and metacarpophalangeal (MCP) joint, were measured. Two main extensor tendons, which terminate within the extensor apparatus of the index finger, extensor digitorum communis

(EDC) and extensor indicis proprius (EIP), were examined in this study. These tendons were exposed at a location roughly 4 cm proximal to the wrist. Silk sutures (3-0) were sewn directly to the tendons, and secured to low-friction monofilament strings (Spiderwire, Pure fishing USA, Spirit Lake, IA).

All skin, fascia, and other soft tissue surrounding the extensor hood were carefully removed. Black plastic markers (diameter = 1.6 mm) were then secured (using cyanoacrylate) to the hood surface (Fig. 1) to measure its deformation under tendon loadings. Distances between markers were maintained at approximately 4 mm. The number of markers used for the strain measurement varied across specimens depending on the size of the specimens (i.e. 42 for the smallest, and 58 for the largest specimen). The marker movements resulting from tendon loading were recorded with a three-dimensional motion analysis system (DMAS, Spica Technology Corp., Maui, HI).

Each hand was mounted on a fixation device (Agee-Wristjack, Hand Biomechanics Lab, Sacramento, CA). The fingertip was secured to a 6 degree-of-freedom (DOF) load cell (JR3, Inc., Woodland, CA) to measure the fingertip forces and moments (Fig. 2). A set of 9 finger postures, consisting of three pairs of proximal and distal interphalangeal (PIP, DIP) joint angles $\{(0^\circ, 0^\circ), (30^\circ, 20^\circ), \text{ and } (45^\circ, 30^\circ)\}$ explored at each of three metacarpophalangeal (MCP) angles $(0^\circ, 30^\circ, 60^\circ)$, were tested for each specimen (Table 1). Tendon force was applied to each tendon separately by hanging weights from the end of the cable in 3.92 N (i.e. 400 gf) increments. During each tendon loading, no nominal load was applied to the other tendons. The potential effects of frictional force (between the string and the table) on the tendon force magnitudes were minimized by utilizing the edge of the table as a pulley. The table edge was rounded, resembling the shape of a regular pulley, and covered with a soft plastic sheet to minimize the friction between the surface and the string. Additionally, the string was checked for tautness to ensure that stiction was not limiting displacement. The maximum magnitude of the applied force to each tendon was set to 11.8 N, which corresponds to approximately 30% of the estimated maximal force capability of the muscle of interest (EDC or EIP) [16,17]. Three levels of tendon forces were applied in each experimental condition (i.e. each tendon loading and finger posture), and the experimental data (i.e. fingertip force/moment vectors and marker movements) measured under the maximum tendon force loading (11.8 N) were analyzed and used in the further analysis (i.e. tendon slip force calculation and strain estimation).

2.2 Calculation of the forces in tendon slips from the force/moment equilibrium

Forces transmitted to the two tendon slips were estimated based on the measured fingertip force and moment data, utilizing the knowledge of the anatomical structure of the extensor hood. The magnitudes of the EDC/EIP tendon forces transmitted into two slips (f_{TS} and f_{CS}) can be defined by two parameters (r, c):

$$\begin{aligned} f_{TS} &= \|\mathbf{f}_{TS}\| = r \cdot c \cdot \|\mathbf{f}_T\| \\ f_{CS} &= \|\mathbf{f}_{CS}\| = (1 - r) \cdot c \cdot \|\mathbf{f}_T\| \end{aligned} \quad (0 \leq r \leq 1, 0 \leq c \leq 1) \quad (1)$$

where $\mathbf{f}_{TS}$ denotes the force vector transmitted to the terminal slip, $\mathbf{f}_{CS}$ the force to the central slip, and $\mathbf{f}_T$ the force applied to the tendon (EDC/EIP) in the experiment. $\|\cdot\|$ denotes the norm (i.e. magnitude) of the vector. c denotes the portion of the tendon force actually conveyed into the extensor apparatus ($0 \leq c \leq 1$). Of that, r represents the fraction of the extensor hood force transmitted/distributed to the terminal slip ($0 \leq r \leq 1$). Thus, $(1-r)$ represents the fraction delivered to the central slip. Several studies have reported that extensor tendons (EDC/EIP) have connections with the MCP joint capsule or with palmar aspects of the proximal phalanx through sagittal bands, which have significant functional outcomes when forces are applied to the tendons [18]; c explains the tendon force reduction due to such force dissipation/

transmission to the surrounding structures. The parameter, c , can also account for the probable energy dissipation/loss within the carpal tunnel through the frictional force.

In each experimental condition, tendon forces transmitted to the two slips ($\mathbf{f}_{TS}$ and $\mathbf{f}_{CS}$) were calculated from the force and moment equilibriums by employing measured fingertip force and moment data. Force equilibrium at the three finger segments and moment equilibrium at the three finger joints were employed to define a cost function (or an objective function), f_c , to be minimized (for the detailed derivation of f_c , see Appendix A). In order to estimate the unknown variables r and c , a non-linear optimization was employed (MATLAB optimization toolbox, MathWorks, Inc., Natick, MA) (Eq. 2), namely:

$$(r, c) = \arg \min_{(r, c) \in \mathbb{R}} f_c \quad (2)$$

In order to find a global rather than a local minimum solution, the optimization process was performed multiple times using different initial values. These initial values covered the entire vector space for r and c ($r \in \{0, 1\}$, $c \in \{0, 1\}$) with 10% increments for each of these variables. The solution that generated the overall minimum cost function value was selected. From the estimated parameters r and c in each condition, the magnitudes of the tendon slip forces ($\|\mathbf{f}_{TS}\|$ and $\|\mathbf{f}_{CS}\|$) were calculated (Eq. 1).

2.3 Strain estimation

Strains on the specific regions of the extensor hood were estimated based on the distances between markers attached to the hood surface. Gauge lengths, from which the elongation (i.e. strain) is measured to define the tensile property, were defined from the distances between neighboring pairs of the surface markers. Markers yielded 37 to 53 gauge lengths in distal-proximal direction, and 29 to 41 in radial-ulnar direction; the number of gauge lengths varied across specimens depending on the size of each specimen. In order to focus on the degree of deformation at the terminal and proximal slips, two subsets of distal-proximal (or longitudinal) gauge lengths were selected for each specimen (see Fig. 1).

2.4 Slip force estimation from the strain measurements

Measured strain values were employed to estimate the correlation between the tendon slip force (f_{TS} or f_{CS}) and the finger posture. If the applied force lies within the linear region of the stress-strain curve of the material, the change in strain values produced by the tendon loading is linearly correlated with the change in stress. Under the assumption that the cross-sectional area of the segment of interest remains relatively constant, the change in the strain magnitudes at the terminal and central slips should reflect the tendon slip force change patterns.

Note that in-situ strains, i.e. zero-load reference gauge lengths, should be measured for the accurate strain calculation. In the strain measurements of human soft tissues, such as the medial collateral ligaments [19], in-situ strains are usually estimated by dissecting the soft tissue of interest from its attachment to surrounding structures. However, the ratio of the force magnitudes (or relative magnitudes of the forces) (force change) in different experimental conditions, i.e. postures, can be estimated without the in-situ strain measurements as follows.

Let l_r denote the zero-load reference gauge length (which only can be measured with the detachment of the extensor hood from its bony insertion sites) of the segment of interest, and l_o^i and l_f^i the gauge lengths before and after tendon loading in posture i ($i = 1, 2, \dots, 9$), respectively. Then, the corresponding strain values, ε_o^i and ε_f^i , are:

$$\varepsilon_o^i = \frac{l_o^i - l_r}{l_r}, \quad \varepsilon_f^i = \frac{l_f^i - l_r}{l_r} \quad (3)$$

The increase in the strain value under tendon loading: $\delta\varepsilon^i$

$$\delta\varepsilon^i = \varepsilon_f^i - \varepsilon_o^i = \frac{l_f^i - l_r}{l_r} - \frac{l_o^i - l_r}{l_r} = \frac{l_f^i - l_o^i}{l_r} \quad (4)$$

We defined a relative strain at posture i , $\hat{\varepsilon}^i$, as the gauge length under tendon loading at posture i divided by the zero-load gauge length in posture 1 (in which presumably the least deformation is produced compared to other postures):

$$\hat{\varepsilon}^i = \frac{l_f^i - l_o^i}{l_o^1} \quad (5)$$

Let f_r^i be the force applied to the component under zero-loading condition in posture i , producing in situ strain of the component (i.e. strain induced by the posture change) in the given posture i , and f_o^i and f_f^i the forces transmitted to the component from the applied force before and after tendon loading, respectively (i.e. $f_o^i = 0$). Then, the stress values before and after tendon loading are:

$$\begin{aligned} \sigma_o^i &= \frac{f_o^i + f_r^i}{A} = \frac{f_r^i}{A} \\ \sigma_f^i &= \frac{f_f^i + f_r^i}{A} \end{aligned} \quad (6)$$

where σ_o^i and σ_f^i denote the stress values before and after tendon loading, respectively, and A the cross-sectional area of the component of interest. The increase in the stress value under tendon loading is:

$$\delta\sigma^i = \sigma_f^i - \sigma_o^i = \frac{f_f^i + f_r^i}{A} - \frac{f_r^i}{A} = \frac{f_f^i}{A} \quad (7)$$

We desire to identify how the f_f^i magnitudes, i.e. force transmitted to the segment of interest, increase or decrease along with the postural change. From the stress-strain relationship,

$$\delta\sigma^i = \sigma_f^i - \sigma_o^i = E\varepsilon_f^i - E\varepsilon_o^i = E\delta\varepsilon^i \quad (8)$$

where E denotes the Young's modulus of the element/segment of interest.

From Eq. (6) – Eq (8),

$$\begin{aligned} \delta\sigma^i &= \frac{f_f^i}{A} = E\delta\varepsilon^i = E \frac{l_f^i - l_o^i}{l_r} \\ \therefore f_f^i &= \frac{EA}{l_r} (l_f^i - l_o^i) = EA \frac{l_o^1}{l_r} \cdot \frac{(l_f^i - l_o^i)}{l_o^1} = C \frac{(l_f^i - l_o^i)}{l_o^1} = C \hat{\varepsilon}^i \end{aligned} \quad (9)$$

Note that the C value is constant if the cross-sectional area A remains constant. Since the magnitude of f_f^i , i.e. the force transferred to the segment in posture i , is proportional to that of $\hat{\epsilon}^i$, the force transmitted to the segment in different postures can be inferred from the $\hat{\epsilon}^i$ values, which can be computed from the segment lengths measured before and after tendon loading (Eq. 5).

Thus, we examined the $\hat{\epsilon}^i$ values, i.e. the relative strain values at posture i , in different postures in order to indirectly estimate the change in f_f^i magnitudes along with the posture change. In other words, the change in strains at two tendon slips in different postures was compared to that of the tendon slip force magnitudes estimated from the fingertip force and moment data (section 2.2).

2.4 Statistical analysis

After estimating the force and the relative strain values at terminal and central slips, a two-way repeated measures analysis of variance (ANOVA) was performed to elucidate the effects of the posture on the estimated force and strain values. Two variables describing finger configurations, i.e. MCP and IP postures, were employed as independent variables, and the estimated force and strain values at two slips, i.e. f_{TS} , f_{CS} , $\hat{\epsilon}_{TS}$, and $\hat{\epsilon}_{CS}$, served as dependent variables. MATLAB statistics toolbox (MathWorks, Inc., Natick, MA) was used for the analysis.

3. Results

3.1 Tendon slip force estimation from the fingertip force/moment data measured in different postures

Both the direction and magnitudes of the fingertip force and moment vectors, in response to the same tendon force, varied significantly with postural changes (Fig. 3). Accordingly, the magnitudes of the estimated tendon slip forces, i.e. f_{TS} and f_{CS} , also varied considerably across posture. Both tendon slip forces (f_{TS} and f_{CS}) were significantly affected by the IP posture ($p < .01$), but MCP posture was not a significant factor for the terminal slip forces in both tendons ($p > .1$) (Table 2). The magnitudes of the forces transmitted to tendon slips generally decreased along with the IP flexion, while the f_{TS}/f_{CS} ratio remained approximately constant (Table 3); in other words, f_{TS} and f_{CS} tended to covary (Fig. 4). MCP flexion exhibited a trend toward negative correlation with the tendon slip forces in many cases (as shown in Fig. 4), but this negative correlation was not consistently observed throughout different IP postures and specimens. For example, in some specimens, the magnitude of the estimated tendon slip force remained the same or even increased when the MCP joint was flexed.

The postural effects on the model parameters (r and c), which were employed to estimate the slip forces, were also found to be significant (Table 4). Specifically, a similar relationship between IP posture and the parameter c was observed (i.e. negative correlation) in all the specimens tested, whereas the effects of MCP posture on the parameter values were found to be rather inconsistent across specimens; r was found to be dependent upon MCP posture to some degree. The mean (SD) values of the model parameter values are summarized in Table 5.

3.2 Estimation of the tendon slip force magnitudes based on the strain values measured in different postures

In all experimental conditions, significant marker movements upon tendon loading were recorded, and the resultant longitudinal and lateral strain values were calculated (Fig. 5). Estimated strain values at the central slip were generally larger than those at terminal slip, in contrast with the force results. In concurrence with the slip force estimation, the magnitudes

of the estimated (longitudinal) strains at terminal and central slips varied significantly with the postural change. Similar to the results of the slip force estimation (Table 2), IP posture was found to be a significant factor for strains at both slips (ε_{TS} , ε_{CS}) (Table 6). Strain values were also found to be affected by the change in MCP posture in one case (EDC terminal slip), and the interaction between MCP and IP postures were significant in three out of four conditions (Table 6). Note that sphericity violation was not found by Mauchly's test ($p > 0.01$; SPSS, SPSS Inc.) in all statistical analyses performed in this study (i.e. Table 2, Table 4, Table 6).

As with the slip force, the magnitudes of both strains decreased along with IP flexion (Fig. 6). While IP flexion had a consistent effect on strain values (i.e. negative correlation between IP flexion and strain magnitudes) across different MCP postures, the correlation pattern between MCP posture and strain magnitude varied across different IP postures (Fig. 6). Also, MCP posture-strain correlation patterns were inconsistent across specimens. For both force and strain the interaction between MCP-IP is significant; for slip force, this is due to the effect of MCP angle on total force for IP extension of (0° , 0°).

4. Discussion

Recently, the notion that peripheral anatomical structure can facilitate and reinforce, or possibly even replace, some functions of the central nervous system in motor control was presented [5]. Some peripheral structures of the hand (such as soft tissue adhesions between tendons) have been shown to contribute to finger movement generation [20,21]. Yet, the role of these passive mechanical structures on different aspects of motor control, such as the mapping from muscle forces into multi-body dynamics of the finger, has remained largely undefined.

The results of this study strongly suggest that passive tissues in the finger, such as the extensor hood, have a profound impact on the translation of muscle forces to the finger. In particular we noted that force transmission to the tendons terminating on the finger phalanges exhibited a strong postural dependence. Postural effects on the transmission of tendon force to the tendon slips were found to be significant in both methods of assessment employed in this study, i.e. the slip force estimation from the fingertip forces and moments (Table 2) and the measured strain (Table 4). Similar relationships between slip force (or strain) magnitudes and the finger posture were observed in both methods; specifically, IP joint flexion was found to be a significant factor that influenced the force transmission to the tendon slips. Generally, the magnitudes of both tendon slip forces decreased along with the IP flexion (Table 3). MCP flexion was also found to affect the tendon slip force magnitudes in some conditions, but the relationships between the MCP flexion and the tendon slip force (or strain) were not as consistent across specimens as were the IP flexion-slip force relationships. The MCP-IP interaction term was significant, however, due to the impact of MCP flexion angle on tendon slip forces (and strains) when the PIP and DIP joints were both in extension (0° , 0°) (Table 3).

The modeling methodology employed in this study can differentiate two different types of tendon force transmission: the tendon force distribution into two tendon slips within the extensor apparatus, and the dissipation of tendon force into surrounding structure through the connective tissues. The force distribution ratio between two tendon slips, r in Eq. (1), was found to remain relatively constant across different postures within each specimen; the force magnitudes at the two tendon slips decreased concurrently. As shown in the Results section, the f_{TS} : f_{CS} ratios for both tendons (EDC and EIP) were found to remain constant across different postures in all specimens tested (Table 3; Fig. 4 and Fig 6). Thus, our results support prior assumptions that the ratio of the force distribution of the extensor tendon into the two tendon slips, represented by the ratio between forces in the central band and lateral offshoots in the Winslow's rhombus, are constant [17,22]. Our mean magnitude for this ratio, however, differs from the commonly employed prior estimations of 25%: 50%: 25% (radial lateral

offshoot: central band: ulnar lateral offshoot), which results in a ratio of terminal slip force: total force of 0.5. We observed a ratio of terminal slip force: total force of 0.63 (± 0.09), implying relatively larger force is transmitted into the terminal slip than into the central slip. Note that, in contrast to the estimated terminal: central slip force ratio, the magnitudes of the central slip strains (ϵ_{CS}) were generally larger than those of the terminal slip strains (ϵ_{TS}). Here, it should be acknowledged that the strain values cannot be used to compare the magnitudes of the terminal and central slip forces (i.e. $f_{TS} : f_{CS}$), since these slips may have different cross-sectional area (A in Eq. 9) and possibly different material properties (i.e. Young's modulus; E in Eq. 9). Our own observation of the specimens confirmed that the terminal slips generally have a larger cross-sectional area than the central slips, which can explain the smaller strain values at terminal slips.

The amount of force actually transmitted into the extensor hood, represented by the magnitude of the parameter c , however, was found to vary significantly across finger postures. Particularly, the magnitude of c decreased considerably as IP flexion increased. Thus, the variation in the total force transmitted into the extensor apparatus seems largely responsible for the observed negative correlation between IP flexion and the individual slip force magnitudes f_{CS} and f_{TS} . Beekman et al. [23] reported several potential mechanisms for tendon force dissipation, specifically for the EDC tendon, into surrounding structures such as sagittal bands, tendon insertion on the MCP joint capsule, and the juncturae tendinae. We surmise that similar mechanisms contribute to the tendon force dissipation patterns observed in this study. Indeed, our own dissection of selected specimens confirmed that soft tissue connections between the extensor hood and the dorsum of the middle and proximal phalanges are present along the length of the hood. These connections are especially strong near the PIP joint capsule. When the IP joints were flexed, the connective tissue appeared to become taut and consequently, little deformation (i.e. strain) was observed near the terminal slip in response to the EDC tendon force, in agreement with the strain measurements of this study (Fig. 6). However, after this soft tissue was excised, noticeably larger deformation around the terminal slip was observed in response to the same magnitude of the EDC tendon force. The results of our study, i.e. the negative correlation between the IP flexion and the strain magnitudes, are consistent with the observed relationship between the finger posture and the tendon force dissipation mechanisms. Flexion of the MCP joint seemed to produce relatively less pull on the connective tissue, in accordance with our findings that IP flexion had a more consistent impact on force translation. It should be noted, though, that when the IP joints were not flexed, MCP flexion had a large effect on the forces realized in the central and terminal slips. Reductions in slip forces of over 30–50% were measured as the MCP angle increased from 0° to 60° of flexion.

The effects of MCP posture alone were not as consistent as the IP and MCP-IP interaction effects. Some of this variability may be due to experimental and methodological limitations and assumptions. For example, in estimating the slip force from the fingertip forces/moments, measurement errors of the load cell could have decreased the accuracy of the inverse kinetics calculation. Certain measurement errors involved in the geometric properties, e.g., segment length or the joint angle measurement, could also have affected the slip force calculation. Strain estimation was limited by camera resolution (7.4 μm per pixel). We performed the experiment once for each condition, thus the issue of repeatability was not directly addressed. Furthermore, a simplified representation of the extensor apparatus as a network of interconnected tendon strings was employed in performing the inverse kinetics that estimated the slip force magnitudes. Although such a tendon network model has been commonly employed in a number of previous studies [5,22,24], the extensor apparatus is actually a heterogeneous structure consisting of tendons and fibers with variable mechanical characteristics. Also, the tendon force estimation was performed in the sagittal plane rather than three-dimensional space, which might have affected the precision of the slip force estimation to a certain degree. For the strain

measurements, we selected a group of gauge lengths from the surface markers (Fig. 1) based on the observation of the extensor hood geometry. Although careful selection was made, some of these gauge lengths may not properly represent the deformation of the two tendon slips. It should be acknowledged that the postural effects on the kinetic functions of the intrinsic extensor muscles of the finger, i.e. dorsal/palmar interosseous and lumbrical muscles, were not examined in this study. This remains as a future study that should be accomplished in order to provide a comprehensive understanding of the finger extensor mechanism and its behavior.

In summary, the finger extensor apparatus is a complex anatomical structure that integrates and transmits multiple tendon forces into the finger segments. It is thus of great functional importance in various finger movements or tasks involving fingertip force exertion. The results of this study corroborate the significant impact of finger posture on tendon force transmission characteristics, reported in both *in vivo* [14] and *in vitro* [25] studies. Consequently, the finger posture in relation to the tendon force transmission should be carefully considered when examining finger motor control or analyzing injuries to the extensor apparatus.

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Appendix

Appendix

A. Cost function derivation from the force and moment equilibrium

Fig. A1.

Free-body diagram of the four-segment finger model. Only extensor tendon force is applied in this model. All force and directional vectors are projected to the sagittal plane.

Fig. A1 shows the free-body diagram of the four-segment finger model. All forces and geometry vectors are projected to the sagittal plane. Each joint is modeled as a revolute joint,

and force and moment equilibrium at each segment was defined. The cost function to be minimized in the optimization process was derived from these equilibrium equations.

In Fig. A1, $\mathbf{f}_T$ denotes the applied tendon force vector, which was assumed to be parallel to the metacarpal bone. $\mathbf{f}_{TIP}$ and $\mathbf{m}_{TIP}$ correspond to the force and moment vectors measured at the fingertip in response to the applied tendon force. $\mathbf{r}_{DP}$ denotes the distal phalanx (DP) segment vector connecting the fingertip and DIP joint, $\mathbf{r}_{MP}$ the middle phalanx (MP) vector connecting the DIP and PIP joints, and $\mathbf{r}_{PP}$ the proximal phalanx (PP) vector linking the PIP and MCP joints.

The tendon (EDC or EIP) is modeled to insert into distal and middle phalanges via two slips (terminal and central), but not into the proximal phalanx [26]. The distance between the tendon line of action at distal slip and the DIP joint (i.e. moment arm of $\mathbf{f}_{TS}$) were denoted as $\mathbf{r}_{TS}$, whose magnitude was obtained from the measured DIP joint thickness (i.e. the half of the measured DIP joint thickness). In the same way, $\mathbf{r}_{CS}$, the moment arm of $\mathbf{f}_{CS}$ with respect to PIP joint, was obtained from the measured PIP joint thickness. Note that $\mathbf{f}_T$ at MCP joint was not considered since the tendon is not inserted into the proximal phalanx. $\mathbf{f}_{DIP}$ denotes the joint reaction force acting on the distal phalanx from the middle phalanx, resulting from the tendon force application. Note that $-\mathbf{f}_{DIP}$ is applied on the middle phalanx. In the same way, $\mathbf{f}_{PIP}$ represent the PIP joint reaction force applied on the middle phalanx, and $\mathbf{f}_{MCP}$ the MCP joint reaction force on the proximal phalanx.

The direction of each tendon force changes over each joint, when the joint is flexed, and the resultant force ($\mathbf{f}_{T/MP}$, $\mathbf{f}_{T/PP}$, $\mathbf{f}_{T/CP}$) due to the force direction change over the joint was assumed to be applied to the distal joint capsule of each bone segment. In other words, each joint was modeled as a pulley system attached to the segment proximal to the joint. The resultant pulley force due to the direction change over the DIP joint ($\mathbf{f}_{T/MP}$) was modeled to be applied to the distal end of the MP segment, $\mathbf{f}_{T/PP}$ to the distal end of the PP segment, and $\mathbf{f}_{T/CP}$ to the distal end of the metacarpal bone. Also, the direction of the resultant pulley force, pointing toward the center of each joint, can be defined from the segment direction vectors; for example, $\mathbf{f}_{T/MP}$ is parallel to the $\mathbf{r}_{DP}/\|\mathbf{r}_{DP}\| - \mathbf{r}_{MP}/\|\mathbf{r}_{MP}\|$.

The magnitudes of the tendon force at terminal and central slips are determined by the parameters r and c , and the tendon force magnitude $\|\mathbf{f}_T\|$:

$$\|\mathbf{f}_{TS}\| = r \cdot c \cdot \|\mathbf{f}_T\| \text{ and } \|\mathbf{f}_{CS}\| = (1 - r) \cdot c \cdot \|\mathbf{f}_T\| \quad (\text{A1})$$

Also the $\mathbf{f}_{TS}$ vector was assumed to be parallel to the $\mathbf{r}_{DP}$ vector (opposite direction), and $\mathbf{f}_{CS}$ to $\mathbf{r}_{MP}$. Then,

$$\begin{aligned} \mathbf{f}_{TS} &= \|\mathbf{f}_{TS}\| \cdot \mathbf{r}_{DP}/\|\mathbf{r}_{DP}\| = r \cdot c \cdot \|\mathbf{f}_T\| \cdot \mathbf{r}_{DP}/\|\mathbf{r}_{DP}\| \\ \mathbf{f}_{CS} &= \|\mathbf{f}_{CS}\| \cdot \mathbf{r}_{MP}/\|\mathbf{r}_{MP}\| = (1 - r) \cdot c \cdot \|\mathbf{f}_T\| \cdot \mathbf{r}_{MP}/\|\mathbf{r}_{MP}\| \end{aligned} \quad (\text{A2})$$

The magnitude of the $\mathbf{f}_{T/MP}$ vector, which corresponds to the force exerted on MP from the tendon caused by the tendon force direction change (due to the DIP flexion angle), can be estimated from the geometry of the finger.

$$\begin{aligned} \|\mathbf{f}_{T/MP}\| &= \sin(\theta_{DIP}/2) \cdot \|\mathbf{f}_{TS}\| \\ \|\mathbf{f}_{T/PP}\| &= \sin(\theta_{PIP}/2) \cdot \|\mathbf{f}_{TS} + \mathbf{f}_{CS}\| \end{aligned} \quad (\text{A3})$$

Thus, these force pulley vectors can be defined as follows:

$$\begin{aligned}\mathbf{f}_{T/MP} &= \|\mathbf{f}_{TS}\| \cdot \sin(\theta_{DIP}/2) \cdot (\mathbf{r}_{DP} - \mathbf{r}_{MP}) / \|\mathbf{r}_{DP} - \mathbf{r}_{MP}\| \\ \mathbf{f}_{T/PP} &= \|\mathbf{f}_{TS} + \mathbf{f}_{CS}\| \cdot \sin(\theta_{PIP}/2) \cdot (\mathbf{r}_{MP} - \mathbf{r}_{PP}) / \|\mathbf{r}_{MP} - \mathbf{r}_{PP}\|\end{aligned}\quad (A4)$$

Then, we used the force and moment equilibrium at each joint to derive the cost function. Here, we represent the moment m generated by force vector $\mathbf{f}$ and position vector $\mathbf{r}$, the distance between the center of rotation and the position where the force is applied, using the operation $\otimes$.

$$m = \mathbf{r} \otimes \mathbf{f} = r_x f_y - r_y f_x \quad \text{where } \mathbf{r} = (r_x, r_y), \quad \mathbf{f} = (f_x, f_y) \quad (A5)$$

First, from the force equilibrium in the distal phalanx (DP):

$$\sum \mathbf{f}_{DP} = \mathbf{0}: \quad \mathbf{f}_{TIP} + \mathbf{f}_{DIP} + \mathbf{f}_{TS} = \mathbf{0} \quad (A6)$$

From the moment equilibrium in the DP at DIP joint:

$$\sum m_{DIP} = 0: \quad m_{TIP} + \mathbf{r}_{DP} \otimes \mathbf{f}_{TIP} + \mathbf{r}_{TS} \otimes \mathbf{f}_{TS} = 0 \quad (A7)$$

From the force equilibrium in the middle phalanx (MP):

$$\sum \mathbf{f}_{MP} = \mathbf{0}: \quad -\mathbf{f}_{DIP} + \mathbf{f}_{PIP} + \mathbf{f}_{CS} + \mathbf{f}_{T/MP} = \mathbf{0} \quad (A8)$$

From the moment equilibrium in the MP at PIP joint:

$$\sum m_{PIP} = 0: \quad \mathbf{r}_{MP} \otimes (-\mathbf{f}_{DIP}) + \mathbf{r}_{CS} \otimes \mathbf{f}_{CS} + \mathbf{r}_{MP} \otimes \mathbf{f}_{T/MP} = 0 \quad (A9)$$

From the force equilibrium in the proximal phalanx (PP):

$$\sum \mathbf{f}_{PP} = \mathbf{0}: \quad -\mathbf{f}_{PIP} + \mathbf{f}_{MCP} + \mathbf{f}_{T/PP} = \mathbf{0} \quad (A10)$$

From the moment equilibrium in the PP at MCP joint:

$$\sum m_{MCP} = 0: \quad \mathbf{r}_{PP} \otimes (-\mathbf{f}_{PIP}) + \mathbf{r}_{PP} \otimes \mathbf{f}_{T/PP} = 0 \quad (A11)$$

Since each joint was modeled as a revolute joint, the sum of the moments about each joint should be zero. Thus, the cost function f_c was defined as the squared sum of the total moments at DIP, PIP and MCP joints (i.e. moment equilibrium at three joints). The force equilibrium equations were used to represent the unknown joint reaction forces (i.e. $\mathbf{f}_{DIP}$ and $\mathbf{f}_{PIP}$) with known (or measured) geometric or force vectors. By combining (A1) – (A11):

$$\begin{aligned}
f_c = & [m_{TIP} + \mathbf{r}_{DP} \otimes \mathbf{f}_{TIP} + \mathbf{r}_{TS} \times \mathbf{f}_{TS}]^2 + [\mathbf{r}_{MP} \otimes (-\mathbf{f}_{DIP}) + \mathbf{r}_{CS} \otimes \mathbf{f}_{CS} + (\mathbf{r}_{MP} + \mathbf{r}_{TS}) \otimes \mathbf{f}_{T/MP}]^2 \\
& + [\mathbf{r}_{PP} \otimes (-\mathbf{f}_{PIP}) + (\mathbf{r}_{PP} + \mathbf{r}_{CS}) \otimes \mathbf{f}_{T/PP}]^2 \\
= & [m_{TIP} + \mathbf{r}_{DP} \otimes \mathbf{f}_{TIP} + \mathbf{r}_{TS} \otimes (r \cdot c \cdot \|\mathbf{f}_T\| \cdot \mathbf{r}_{DP} / \|\mathbf{r}_{DP}\|)]^2 \\
& + [\mathbf{r}_{MP} \otimes \{(r \cdot c \cdot \|\mathbf{f}_T\| \cdot \mathbf{r}_{DP} / \|\mathbf{r}_{DP}\| + \mathbf{f}_{TIP})\} + \mathbf{r}_{CS} \otimes \{(1-r) \cdot c \cdot \|\mathbf{f}_T\| \cdot \mathbf{r}_{DP} / \|\mathbf{r}_{DP}\|\} \\
& + (\mathbf{r}_{MP} + \mathbf{r}_{TS}) \otimes \{\sin(\theta_{DIP}/2) \cdot r \cdot c \cdot \|\mathbf{f}_T\| \cdot (\mathbf{r}_{DP} - \mathbf{r}_{MP}) / \|\mathbf{r}_{DP} - \mathbf{r}_{MP}\|\}]^2 \\
& + [\mathbf{r}_{PP} \otimes \{(r \cdot c \cdot \|\mathbf{f}_T\| \cdot \mathbf{r}_{DP} / \|\mathbf{r}_{DP}\|) + \mathbf{f}_{TIP} + (1-r) \cdot c \cdot \|\mathbf{f}_T\| \cdot \mathbf{r}_{MP} / \|\mathbf{r}_{MP}\| \\
& + \sin(\theta_{DIP}/2) \cdot r \cdot c \cdot \|\mathbf{f}_T\| \cdot (\mathbf{r}_{DP} - \mathbf{r}_{MP}) / \|\mathbf{r}_{DP} - \mathbf{r}_{MP}\|\} \\
& + (\mathbf{r}_{PP} + \mathbf{r}_{CS}) \otimes \{\sin(\theta_{PIP}/2) \cdot \|\mathbf{f}_{TS}\| \cdot (\mathbf{r}_{MP} - \mathbf{r}_{PP}) / \|\mathbf{r}_{MP} - \mathbf{r}_{PP}\|\}]^2
\end{aligned} \tag{A12}$$

Here, the geometry vectors ($\mathbf{r}_{DP}$, $\mathbf{r}_{MP}$, $\mathbf{r}_{TS}$, $\mathbf{r}_{CS}$) are given from the finger configuration information (i.e. joint angles) and the specimen geometry measurements (i.e. segment length and thickness), m_{TIP} and $\mathbf{f}_{TIP}$ are obtained from the experimental data, and $\|\mathbf{f}_T\|$ is given from the experimental condition (i.e. applied tendon force magnitude). Thus, the only unknown variables are r and c in the defined cost function f_c .

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Fig. 1. Marker placement on the finger extensor apparatus for the strain estimation (Specimen #5). Small squares around markers denote their locations as recognized by the motion capture system. Two subsets of markers, S_{TS} and S_{CS} , were employed to estimate the longitudinal strain values at the terminal and central slips, respectively.

Fig. 2.
Experimental setup for the measurement of fingertip force/moment and strains in the cadaveric specimen.

Fig. 3.

Representative plot of fingertip force and moment vectors in response to the EDC tendon force measured in nine postures (Specimen #1: right hand): (a) fingertip force (b) fingertip moment. The force and moment vectors are projected onto the sagittal plane. The effects of the IP posture on the fingertip force and moment vectors appear to be more significant than that of MCP posture.

Fig. 4. Mean and standard deviation values of the estimated terminal and central slip force magnitudes across all specimens: (a) EDC (b) EIP. Magnitudes of both slip forces decreased with IP flexion.

Fig. 5. Strain measurement. (a) Measured marker movements under EDC tendon loading (Specimen #2, Posture 1: all joint angles = 0° , tendon: EDC, tendon force (f_T) = 11.8N) (b) longitudinal and (c) lateral strain distributions at maximum loading condition (f_T) estimated from the recorded marker movements.

Fig. 6.

Mean and standard deviation values of the estimated terminal and central slip strain values across all specimens: (a) EDC (b) EIP. As observed in the calculated slip forces, the magnitudes of the both slip forces decreased with the IP flexion.

Table 1

Joint angles for all nine postures (unit: °)

Joint	Posture								
	1	2	3	4	5	6	7	8	9
DIP	0	20	30	0	20	30	0	20	30
PIP	0	30	45	0	30	45	0	30	45
MCP	0	0	0	30	30	30	60	60	60

Table 2
Two-way measures ANOVA results: *p*-values for the effect of each term on tendon slip force magnitudes

	EDC		EIP	
	f_{TS}	f_{CS}	f_{TS}	f_{CS}
MCP posture	.431	.004	.176	.010
IP posture	.001	.005	< .001	.009
MCP $\times$ IP	.003	.025	.008	.027

Table 3

Mean (SD) values of the estimated tendon slip forces across all five specimens in nine postures (unit: N)

(PIP, DIP)	EDC terminal slip force (f_{TS})		
	MCP = 0°	MCP = 30°	MCP = 60°
(0°, 0°)	5.05 (1.98)	4.66 (1.50)	3.40 (1.26)
(30°, 20°)	2.30 (0.63)	2.31 (0.41)	2.38 (0.44)
(45°, 30°)	1.78 (0.45)	1.85 (0.56)	1.93 (0.50)
(PIP, DIP)	EDC central slip force (f_{CS})		
	MCP = 0°	MCP = 30°	MCP = 60°
(0°, 0°)	3.44 (0.33)	2.67 (1.06)	1.43 (0.84)
(30°, 20°)	1.22 (0.28)	1.30 (0.20)	1.31 (0.23)
(45°, 30°)	1.55 (1.14)	1.50 (0.83)	1.04 (0.23)
(PIP, DIP)	EIP terminal slip force (f_{TS})		
	MCP = 0°	MCP = 30°	MCP = 60°
(0°, 0°)	5.43 (2.34)	4.94 (1.50)	3.66 (1.38)
(30°, 20°)	2.51 (1.01)	2.43 (0.67)	2.42 (0.47)
(45°, 30°)	1.83 (0.65)	1.89 (0.53)	1.88 (0.54)
(PIP, DIP)	EIP central slip force (f_{CS})		
	MCP = 0°	MCP = 30°	MCP = 60°
(0°, 0°)	3.31 (0.83)	2.70 (1.22)	1.49 (0.79)
(30°, 20°)	1.31 (0.45)	1.33 (0.32)	1.32 (0.21)
(45°, 30°)	1.60 (1.30)	1.40 (0.83)	1.02 (0.25)

Table 4
Two-way measures ANOVA results: *p*-values for the effect of each term on estimated parameter values

	EDC		EIP	
	<i>r</i>	<i>c</i>	<i>r</i>	<i>c</i>
MCP posture	.020	.072	.023	.039
IP posture	.601	<.001	.496	<.001
MCP × IP	.334	.002	.424	.002

Table 5 Mean (SD) values of the model parameter r and c

	r	c
EDC	0.63 (0.10)	0.39 (0.19)
EIP	0.64 (0.09)	0.40 (0.21)

Table 6
Two-way measures ANOVA results: *p*-values for the effect of each term on strain values at tendon slips

	EDC			EIP		
	ϵ_{JS}	ϵ_{CS}	ϵ_{JS}	ϵ_{CS}	ϵ_{JS}	ϵ_{CS}
MCP posture	<.001	.056	.061	.065		
IP posture	.001	.002	.003	.007		
MCP $\times$ IP	.001	.015	.003	.007		