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# Modeling of Multiarticular Muscles: Importance of Inclusion of Tendon—Pulley Interactions in the Finger

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## Abstract

The purpose of this study was to examine force transmission from one of the major multiarticular muscles of the finger, flexor digitorum profundus (FDP), to the index finger. Specifically, we examined whether the popular moment arm (MA)-joint torque technique of modeling muscle force transmission can accurately represent the effects of the FDP on finger movement. A dynamic finger model employing geometric MA values (model I) was compared with another model including realistic tendon force transformation mechanisms via pulley structures and joint reaction forces (model II). Finger flexion movements generated by these models were compared with those obtained from *in vivo* stimulation experiments. The model with the force transformation mechanisms (model II) resulted in more realistic joint spatial coordination (i.e., proximal interphalangeal > metacarpophalangeal distal interphalangeal) than the MA-based model (model I) in relation to the movement patterns evoked by stimulation. Also, the importance of the pulley structures and passive joint characteristics was confirmed in the model simulation; altering/ eliminating these components significantly changed the spatial coordination of the joint angles during the resulting movements. The results of this study emphasize the functional importance of the force transformation through various biomechanical components, and suggest the importance of including these components when investigating finger motor control, such as for examining injury mechanisms or designing rehabilitation protocols.

## Keywords

Dynamics; finger; hand; joint coordination; model; moment arm (MA); pulley

## I. Introduction

PRECISE understanding of the mechanism of muscle-tendon force transformation into joint torques is necessary for proper biomechanical analyses of human tasks. Generally, the

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moment arm (MA) of a muscle-tendon unit is employed to assess the moment-generating capacity of the muscle around a joint [1], [2]. Most techniques for estimating MA values rely on kinematic measurements, such as the correlation between the tendon excursion and the resultant joint rotation [3] or geometric distance between the tendon action line and the joint estimated by medical imaging techniques [4], [5]. For uniarticular muscles, the kinetic outcome of a muscle-tendon unit can be represented accurately by the MA values obtained from these methods.

Kinetic functions of the musculotendon units in the fingers and the forearm, however, are especially difficult to evaluate due to their anatomical complexity. Extensor tendons of the finger, such as extensor digitorum communis (EDC) and interosseous tendons, are interconnected within the extensor apparatus, which inserts into multiple finger phalanges. Thus, when evaluating their dynamic impacts of these muscles, the tendon force distribution within the extensor hood, as well as the force allocation to different tendon insertions, should be carefully considered [6]. Furthermore, multiarticular characteristics of the musculotendon units of the finger make evaluation of their kinetic functions particularly difficult. For instance, while the extrinsic flexor tendons of the hand, flexor digitorum profundus (FDP), and flexor digitorum superficialis (FDS) can produce concurrent flexion of the finger segments proximal to their insertion points [7], [8], these muscles do not insert into all phalanges of the fingers. The proximal phalanx (PP), in fact, does not have any apparent insertions for these tendons, and thus, no tendon pulling force is directly applied to PP to produce the observed metacarpophalangeal (MCP) joint flexion. Hence, several studies have reported conflicting views on the roles of FDP and FDS muscles in MCP joint flexion [9]–[12]. To date, it still remains unclear whether the MA values of the extrinsic finger flexor tendons (FDP/FDS), which are typically obtained from the kinematic measurements, can truly represent their contribution to the dynamics of finger movements, i.e., spatial coordination of multiple finger joints.

Several dynamic models of the finger have been developed to elucidate the kinetic functions of finger muscle–tendons. Most of these models, however, use inverse dynamics to examine the muscle force coordination patterns that generated the observed movements [13], [14] or fingertip force patterns [15]–[18]. Due to the characteristics/nature of inverse dynamics, the mechanisms of the muscle force transformation into joint torque (and the resultant movements or fingertip forces) may not be examined in detail by these models. Several forward dynamics models have also been proposed [19], [20], but most of these models simply adopted constant MA values (obtained from excursion methods) in order to estimate the magnitudes of joint torques generated by tendon forces. Thus, various tendon force transformation mechanisms that can affect the resulting spatial coordination of finger joint movements, such as direct tendon pulling forces, resultant joint reaction forces, tendon pulley constraining forces, and passive joint stiffness or damping moments, remain largely unexplored.

The purpose of this study was thus to evaluate the biomechanical components that contribute to the translation of musculotendon force from FDP into finger movement and examine the force transmission of one of the major multiarticular extrinsic musculotendon units, FDP, in finger movement generation. We examined whether the geometric MA values of the FDP,

obtained from kinematic methods, can correctly represent its impact on joint rotation by employing two different types of dynamic models of the index finger. One model (model I) used the MA values to directly calculate joint moment values, whereas the other (model II) considered realistic force transformation mechanisms in the finger biomechanical structures, such as anatomical pulleys and joint reaction forces. Model performance was evaluated by comparing the model-generated movement data with those obtained from an *in vivo* electrical stimulation experiments. Specifically, the spatial coordination of the multiple finger joint displacements, as well as their velocity characteristics, was examined. Contributions of different kinetic components to each joint moment, which resulted in the observed specific movement patterns, were identified and analyzed in detail.

## II. Methods

## A. Dynamic Models of the Index Finger

We first aimed in this study to accurately identify the contributions of different tendon force transmission mechanisms to joint moment generation, and second to evaluate the validity of using geometry-based MA values in dynamic finger models. Accordingly, two different dynamic models of the index finger were developed. The first model (model I) employed torque generators at each finger joint; in this model, each joint moment was estimated by multiplying the tendon force by the corresponding MA value. In contrast, a more detailed representation of the biomechanical structure of the finger tendons and the pulleys constraining tendon paths was considered in the second model (model II). Consequently, finger movement generated in response to a given tendon force input was compared between the two models.

1) Linkage Representation of the Finger: In both models, the index finger was modeled in the mid-sagittal plane as an open-linked chain with four segments and three single-DOF revolute joints. This model considers only the flexion–extension DOF of the MCP joint since the main function of the musculotendon unit investigated in this study (FDP) is MCP flexion rather than MCP abduction–adduction [3]. Gravitational force was assumed negligible in comparison with the tendon force used in the simulation. Length and thickness for each finger segment (Table I) were measured from one of the subjects who participated in the *in vivo* experiment (i.e., validation study) described in Section II-B. Each finger segment volume was estimated from measured segment length and thickness, and the segment mass was estimated by multiplying the calculated volume by a constant density (1.1 g/cm<sup>3</sup>) [21]. Also, passive stiffness and damping values, resulting from the contributions of passive tissue such as ligaments, skin, joint capsules, and inactive muscles and tendons, were obtained from the literature [8]. Passive stiffness was modeled as a function of joint angle. These stiffness and damping properties were included in both the models.

## 2) Calculation of the Joint Moment from the MA Values and the Tendon Force:

<u>a) Model I—Implementation of the MA values obtained from the excursion</u>
 <u>method:</u> In the first model (model I), constant MA values obtained from the excursion
 method [3] were used to calculate the joint moment generated by the tendon force. Namely,

joint moment was computed by multiplying a constant MA value [3] by the given tendon force magnitude (Table I)

$$m_{\rm J/FDP} = r_{\rm FDP/J} f_{\rm FDP}$$
 (1)

where  $m_{J/FDP}$  represents the moment generated at joint J by FDP,  $r_{FDP/J}$  the constant MA of FDP about the joint J, and  $f_{FDP}$  the FDP tendon force.

#### b) Model II—Incorporation of the biomechanics of the tendon and pulley

**<u>structure</u>**: Two types of tendon force transmission to the finger segments were considered here: tendon pulling force directly applied at the tendon insertion site and the force transmission via pulley structures constraining tendon paths.

First, the moments generated by tendon pulling forces applied to the insertion points were incorporated. Note that the FDP tendon inserts only into the distal phalanx (DP) (see Fig. 7 in Appendix). At the DP, moments generated by the FDP tendon forces were estimated by multiplying the MA vector, which is the distance from the DP center of mass to the line of action of the FDP tendon force ( $f_{\text{FDP}}$ ) to the  $f_{\text{FDP}}$ . Here, note that the MA of the FDP tendon,  $r_{\text{FDP/DP}}$ , was modeled as a function of the joint angle  $\theta_{\text{DIP}}$  and pulley geometry values (see Fig. 7).

Also, the effect of the direct pull of the FDP tendon force  $f_{\text{FDP}}$  on the proximal interphalangeal (PIP) and MCP joint moments, through transmission of the force to the middle phalanx (MP) and to PP via the reaction forces at the distal interphalangeal (DIP) and PIP joints, was included in the model. Additionally, forces transmitted from the pulley structures constraining the tendon path were considered. The pulley forces applied to the phalanges were calculated based on the normative geometric model of the finger anatomical structure [22], which was developed by examining 15 hand specimens. We considered the major annular pulley structures for the flexors: the A1, A2, A4, and A5 pulleys located on the palmar side of the finger. Details of the modeling of the flexor pulleys, similar to prior studies estimating pulley forces in sports activities [23], [24], are provided in the Appendix [for details, see Figs. 7–9 and (A1)–(A3)]. Note that the A3 pulley was not included in the model since a number of studies have shown that the absence of the A3 pulley has little effect on flexor tendon kinematics [25]–[27].

**3)** Simulation of the Finger Movements: Finger flexion movements were simulated by the two models (models I and II) using the fourth-order Runge–Kutta method (variable-step solver) on aMATLAB platform (MathWorks, Inc., Natick, MA).

FDP tendon force served as the model input. A sigmoid tangent hyperbolic function was used to model the transient temporal trajectory of the tendon force (Fig. 1). Steady-state force was achieved at approximately 1 s. Two different maximum values of FDP force magnitude were used for the generation of two flexion movements, i.e., small and large flexions; small flexion movement was defined as a movement in which the sum of all three joint flexion angles, i.e., total flexion angle, was approximately 80° at the final posture,

while for large flexion, the total flexion angle was approximately 120°. Tendon force magnitudes of small (2.5 N) and large (3.5 N) forces were selected to generate these flexion movements. The selection of the peak force values, as well as the shape of the temporal tendon force trajectory, was informed by the measurement data of an *in vivo* study that recorded FDP (and FDS) forces using buckle force transducers during active concurrent flexion of finger joints [28].

#### 4) Effects of Passive Joint Stiffness and Damping on the Joint

**Coordination:** In order to examine the passive joint stiffness and damping on the flexion movement generated by the FDP tendon, the finger movement was simulated with the passive joint stiffness and damping values of all three joints set to zero in model II. Due to the small magnitudes of segment mass and inertia values, a small magnitude of the FDP tendon force, 0.01 N, was employed in the simulation. The upper bound of each joint angle was determined from the finger geometric parameters (e.g., pulley locations; DIP ~  $100^{\circ}$ , PIP ~  $110^{\circ}$ , MCP ~  $100^{\circ}$ ).

**5) Sensitivity Analysis:** In order to examine the effects of anatomical variability in the finger pulley structures on finger movement, we performed a sensitivity analysis. Finger flexion movements were simulated in the finger pulley model (model II) in response to an FDP tendon force of 2.5 N for a range of parameter values for each of the eight pulley geometric characteristics (pulley thickness and its distal/proximal distance from the adjacent joint, i.e.,  $d_{a/PP}$ ,  $d_{b/PP}$ ,  $d_{c/PP}$ ,  $d_{a/MP}$ ,  $d_{b/MP}$ ,  $d_{c/MP}$ ,  $d_{d/MP}$ ; see Figs. 7–9). Each pulley geometric parameter *d* was changed from 0.7*d* to 1.3*d* with an increment of 0.1*d* in the model, and the finger movements were simulated accordingly. Only one parameter was changed at a time. This 30% variation from the nominal value was large in comparison with the variability described in cadaveric measurements (standard deviation ~ 10% in [22]).

6) Simulation of the Pulley Excision: In order to examine the functional importance of the finger pulley structures, two types of pulley excision were simulated, and the results were compared with literature values from studies that performed similar procedures in cadaver experiments. First, the excision of the A1 pulley was simulated by adjusting the geometrical parameters of the A1 pulley, i.e., increasing the distal tendon location around the MCP joint (see the Appendix for the details of the related geometric parameters). Then, the partial excision of distal A2 pulley (50%) was simulated by increasing proximal tendon location (i.e., increasing  $d_{b}/_{PP}$  in Fig. 9; see the Appendix) from the PIP joint. Simulation results were then compared with the literature values (for A1 excision, see [29], and for 50% A2 excision, see [30]).

#### B. Experimental Validation of the Models: Electrical Stimulation of Finger Muscles

Movement patterns generated by the two models (models I and II) were compared with the index finger movements produced by the electrical stimulation of FDP in three male subjects (all right-hand dominant). The experimental protocol was approved by the Institutional Review Board of Northwestern University (Chicago, IL), and the subjects signed informed consent to enroll in this study prior to the participation. Two 55- $\mu$ m stainless steel intramuscular fine-wire electrodes were inserted into the first compartment of the FDP [i.e.,

FDP (I)]. The wires were threaded through a 27-gauge needle, which was inserted into the muscle from the ulnar aspect of the forearm at a location approximately 4 cm distal from the olecranon, as described in [31]. The insertion was guided by audial feedback from muscle contraction. Proper electrode placement was confirmed by flexion of the DIP joint of the index finger in response to a single pulse of electrical stimulation (Digitimer stimulator; Digitimer Ltd., Welwyn Garden City, U.K.).

During the stimulation experiment, we also obtained two types of movements, i.e., small (total flexion angle ~ 80°) and large (total flexion angle ~ 120°) flexion movements, from each subject. Subjects were seated comfortably in a chair, resting the forearm on a table. The elbow was held at approximately 90° and the wrist in a slightly flexed posture (~30°). An electrical stimulator (Compex2, MediCompex SA, Switzerland) provided 500  $\mu$ s current pulses to the muscles at 50 Hz tetanus frequency. For each trial, the stimulation strength was increased in three steps (step length = 0.5 s; see Fig. 1). The peak current amplitude was initially set to 1 mA for the first stimulation trial, and was increased gradually over the trials, thereby increasing the total flexion angle of each resulting movement. The movements were recorded at each current level, and the experiment was completed when both the small and the large flexion movements were obtained (i.e., total flexion angle ~80° and 120°).

Flexion angles of the three finger joints during movements were obtained with a video capture system (OPTOTRAK 3010, Northern Digital, Inc., Waterloo, ON, Canada). Two active infrared markers were placed on each of distal, middle, and proximal segments, as well as above the index metacarpal bone. Marker locations were recorded at 150 Hz for each stimulation trial. Temporal trajectory of each marker location was then digitally filtered forward and backward using a third-order Butterworth filter (MATLAB; MathWorks, Inc., Natick, MA). Each joint angle was then computed by the law of cosines from the spatial trajectories of the corresponding pair of segments.

## III. Results

Distinct patterns of spatial coordination of the finger joint angles were obtained from the two biomechanical models (models I and II) in response to FDP force application, although both models did generate concurrent flexion of all three joints. In the MA-based model (model I), flexion of the MCP joint was the largest [see Fig. 2(a)]. PIP joint flexion was smaller than that of MCP joint, but larger than DIP joint flexion (i.e., MCP > PIP > DIP); in other words, the relative magnitudes of flexion angles of the three joints basically reflected their MA magnitudes (see Table II). In contrast, in the movements generated by model II, PIP joint flexion was the greatest, while MCP joint flexion was comparable to (or slightly larger than) the DIP flexion (PIP > MCP DIP) (see Fig. 2(b) and Table II).

In the flexion movements of model i, the rate of change of MCP joint flexion was greater than (or similar to) that of PIP flexion, while that of DIP joint flexion was the smallest among three joints. In contrast, the rate of change of PIP flexion was the greatest, and DIP and MCP joint flexions were of similar speed in model II.

Sensitivity analysis of the model pulley parameters (in model II) revealed that change in each pulley parameter mainly affects the angular profile of the most adjacent joint (e.g., the distal part of the A4 pulley affects DIP joint flexion angle, and its proximal part affects PIP joint flexion angle). Joint angles at final posture were more sensitive to the change in the pulley thickness (i.e.,  $d_a$ ,  $d_d$ ) than its distal/proximal locations ( $d_b$ ,  $d_c$ ) (see Fig. 3; see Figs. 7–9 for detailed descriptions of these geometric parameters). Across different pulley geometric parameter values, the spatial coordination pattern of the three joint angles at the final posture remains the same (PIP > MCP DIP).

During the *in vivo* validation experiments, electrical stimulation of the FDP muscle also resulted in the concurrent flexion of the three joints. For one subject (subject 3), the first compartment of the FDP muscle was difficult to locate. Thus, its second compartment (FDP II) was stimulated, and the resulting movements of the long finger were recorded. PIP joint flexion was always the largest across subjects; in most cases, MCP flexion was comparable to DIP flexion (see Fig. 2(c) and Table III). In addition to angular displacement, the rate of change of PIP flexion was greater than that of DIP or MCP joints in all observed movements, in accordance with the simulation results of model II.

Thus, the biomechanical model incorporating potential tendon force transformation mechanisms (i.e., model II) produced more realistic finger flexion movements (i.e., largest flexion at PIP joint; see Fig. 2(b) and Table II), which have similar spatial coordination of joint flexion angles to that of the flexion movements generated in the *in vivo* stimulation experiment (i.e., PIP > MCP DIP; see Fig. 2(c) and Table III).

The excision of the A1 pulley, simulated by adjusting the geometrical parameters of model II, resulted in a roughly  $10^{\circ}$  increase in MCP flexion angle [see Fig. 4(a)]. In contrast, the partial excision of distal A2 pulley increased the PIP flexion angle by  $5^{\circ}$  [see Fig. 4(b)].

The important roles of passive properties in finger movements were confirmed by removing the passive joint stiffness and damping from the model II. This resulted in rapid DIP flexion, followed by PIP flexion, while MCP joint was slightly extended (see Fig. 5).

The contributions of the different kinetic components and tendon force transmission mechanisms to the joint angular acceleration (i.e., joint moment) were examined in the movement generated by model II [Fig. 6; for the estimation process, see (A4) and (A5)]. In all three joints, contributions of the passive damping components were present in the transient phase (i.e., during joint flexion) and diminished, as anticipated, in the steady-state phase. During the steady-state phase, however, contributions from passive stiffness and from the FDP tendon force to the joint acceleration were matched. The overall effect of the passive joint stiffness (coupled by the inverse of the inertia matrix) was resistive to DIP flexion, but assistive to PIP and MCP flexion (i.e., positive joint angular acceleration). In contrast, the passive joint damping resisted DIP and PIP joint flexions, while assisting MCP flexion. Note that the contribution of the passive joint damping to MCP flexion in large flexion movement was more significant [see Fig. 6(b)] due to the higher movement speed. In all cases, the effect of the Coriolis forces was negligible.

Here, note that the impact of the kinetic components on joint accelerations is coupled by the inverse of the inertia matrix [see (A4) and (A5)]. For example, the PIP joint angular acceleration, i.e., the MP rotational moment, is affected not only by the passive moments produced at the distal end of MP, i.e., passive DIP joint stiffness/damping, but also by those at its proximal end, i.e., passive PIP joint stiffness/damping.

## IV. Discussion

The purpose of this study was to improve understanding of the transmission of musculotendon force through the complex biomechanical structure of the finger. Specifically, we sought to test if the popular MA–joint torque model of force transmission was appropriate for use with the multiarticular musculotendon units in the finger. Toward this end, two different types of dynamic models of the index finger were compared.

The results of this study emphasized the importance of the biomechanical components, e.g., pulley structures and reaction forces, on the finger dynamics involving multiarticular tendons such as FDP. The model directly incorporating kinematic MA values (model I) produced quite different spatial coordination of finger joints, i.e., larger flexion for MCP but smaller for PIP, compared to the actual movement produced by the *in vivo* FDP stimulation. In contrast, the model considering biomechanical structures (model II) generated movements with similar spatial coordination to the actual movement (PIP > MCP DIP). Sensitivity analyses on the pulley parameter values also revealed that across a range of geometric pulley parameter values, similar movement patterns, i.e., largest flexion angle at the PIP joint, were observed (see Fig. 3).

In order to examine the functional importance of the pulley structures in finger dynamics, two types of pulley excision were simulated. First, the A1 pulley excision was simulated by adjusting geometric parameters of pulley structures near the MCP joint. This resulted in an increase in the MCP flexion angle of about 10°; this increase is attributable to the increased distance between the line of action of FDP force and the MCP joint [29]. Also, partial distal A2 pulley excision (50%) was simulated, resulting in 4° increase in PIP flexion. To our knowledge, no study has directly examined the change in the individual spatial coordination of multiple finger joint flexions due to the pulley excision, but some studies indirectly examined its effects on finger dynamics. Namely, change in the work of flexion, estimated by the area under the tendon force-excursion curve, due to the pulley excision was examined in a number of studies [29], [30], [32]. Our results indicated that the total work required for the finger flexion decreased with the pulley excisions, which agrees with the results of these previous cadaver studies.

In accordance with our previous study [8], the passive stiffness and damping properties of the finger joints were key contributors to the PIP and MCP flexion during FDP force application; in other words, the resistance to the flexion of the joint distal to each segment generates the rotational (i.e., flexion) moment (e.g., see Fig. 6). Removal of passive components resulted in an abnormal movement pattern of rapid DIP flexion followed by PIP flexion, while MCP joint was slightly extended (see Fig. 5). Without the force transmission through joint passive properties, the DIP joint was first flexed until it reached its physical

flexion boundary (~ 90°). After the DIP joint was locked in its upper bound, the DP and MP acted as one rigid body; thus, the tendon force was transmitted to the PIP joint, which resulted in its flexion. Here, note that a very low level of force (~0.01 N) was enough to produce the movement, due to the very small mass and inertia properties of the finger segments, in contrast to the much higher actual force level measured during *in vivo* experiments [28]. Interestingly, overall contribution of the FDP force, i.e., the combined effects of the resultant joint reaction forces and pulley forces, to the PIP and MCP joint flexion was actually negative (see Fig. 6); this is attributable to the reaction forces transmitted from the distal and proximal segments and their interaction, coupled by the inverse of the inertia matrix [i.e., multibody dynamics; see (A5)].

Biomechanical mechanisms of the tendon force transformation into joint moments examined in this study also provide probable explanation for the conflicting views on the roles of FDP and FDS muscles in MCP flexion. For example, a cadaver study [9] found that a significant PIP flexion (>50°) always preceded MCP flexion under FDS loading, while our *in vivo* FDP stimulation resulted in virtually simultaneous PIP and MCP joint flexion. Presumably, discrepancy in the joint passive properties between *in vivo* and *in vitro* experiments greatly contributed to these conflicting results. As shown in Fig. 6, in the transient phase (i.e., flexion), passive damping contributes significantly to the MCP joint moment generation. Such joint damping might have diminished in cadaver specimens depending on their condition. Also, if the specimen were dissected at the wrist level, the contribution of inactive extrinsic muscles to the joint passive stiffness (and damping) would have been lost, thereby reducing the stiffness and damping values.

The results of this study suggest that thorough analyses of the force transformation mechanisms involving various biomechanical structures (i.e., passive joint properties, pulley structures, etc.), rather than direct application/implementation of the kinematic MAs, should be performed if accurate delineation of the kinetic functions of multiarticular musculotendon units of the finger is desired. Considering that fine coordination of finger joint flexion is essential in most manual tasks, biomechanics of different tendon force transformation processes should be carefully considered in various applications requiring tendon force-movement conversions, i.e., precise understanding of the functional outcome of the musculotendon force is expected to aid not only in the investigation of fundamental issues in human motor control, but also in practical applications, such as understanding injury mechanics or planning rehabilitation protocols.

Some limitations of the study should be acknowledged. In one subject, the second compartment of the FDP was stimulated, and the long finger movements were recorded accordingly, instead of those of the index finger. Even though the anatomical structure of the FDP and its pulleys within the index and long fingers are very similar (see [22]), this might have contributed to the variability of the experimental results. Possible discrepancy between the experimental setup and the simulation method might exist. The absolute values of the tendon forces produced by FDP stimulation are unknown. Here, potential coactivation of the antagonist muscle, i.e., EDC or extensor indicis proprius (EIP), and its effects on the movements should be acknowledged. Note that the tendon force (~3.5 N) used in our simulation to generate large flexion was smaller than the FDP force value (~5 N) that

generated similar joint spatial coordination in the *in vivo* measurement study [28], in which EDC or EIP muscles might have been coactivated to some extent. Coactivation of antagonist muscles (EDC/EIP) would require increased activity of the agonist muscle (FDP) to produce the same amount of flexion. Additionally, the temporal force profile, generated by the stimulation whose strength was increased stepwise, may have differed from the hyperbolic tangent function used in the simulation (see Fig. 1). Segment properties such as mass and inertia used in the model might differ from those of the subjects who participated in the experiment. Some biomechanical factors, such as the frictional force between pulley and flexor tendon [33], [34], were not considered in the model. The A3 pulley was excluded from the model due to its negligible impact on finger dynamics, as suggested by previous works [25]–[27]. However, some studies have suggested that the A3 pulley may have some influence on finger flexion kinetics by reducing the tendon-pulley gliding resistance during movements [35] or by restraining bowstringing when combined with adjacent sheath structures [36]. Thus, the exclusion of the A3 pulley might have affected the simulation results to some extent. Also, note that the passive joint stiffness and damping values used in the model might differ from those of the subjects who performed the *in vivo* study; such discrepancies may explain the relatively small DIP flexion angle of the simulated movement (i.e., DIP passive moment used in the model was larger than that of the subject), as well as the slight decrease in the DIP flexion angle in initiating movements (i.e., mismatch in the neutral angle between model and subject; see Fig. 2). In addition, care should be taken when comparing the results of this study with the process of voluntary finger movement generation, in which FDP and FDS muscles are usually coactivated along with intrinsic muscles. Note that this study focused on the precise identification of the effect of the individual multiarticular tendon force (i.e., FDP) on finger dynamics.

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## APPENDIX

## A. Derivation of the Equations of Motion for the Index Finger Movement Generated by the FDP Force

Proximal to the DIP joint, the A5 pulley constrains the path of the FDP tendon, and the pulley force  $f_{\text{FDP}/\text{A5}}^{\text{p}}$  can be estimated



## Fig. 7.

Finger pulley structure around/near the DIP joint. The pulley force generated by both tendons  $f_{\text{FDP}/\text{A5}}^{\text{p}}$  is applied to the MP.

## TABLE IV

TENDON PARAMETERS (ESTIMATED FROM [22], IN MILLIMETERS) FOR AN MP LENGTH = 24.5 MM

Joint	Parameters				
DID	d <sub>a/DP</sub>	d <sub>b/DP</sub> d <sub>c/MP</sub>		d <sub>d/MP</sub>	
DIP	4.5	5.4	6.1	7.3	
PIP	d <sub>a/MP</sub>	d <sub>b/MP</sub>	d <sub>c/PP</sub>	d <sub>d/PP</sub>	
	7.5	12.2 9.9		9.8	
	d <sub>a/PP</sub>	d <sub>b/PP</sub>	d <sub>c/P</sub>	d <sub>d/P</sub>	
МСР	9.3	13.4			
	d' <sub>a/PP</sub>	d' <sub>a/PP</sub>	15.9	7.3	
	11.1	10.4	•		

from the geometry of the tendon and the pulley structure considering the joint flexion angle (see Fig. 7).

Moment generated by the FDP tendon force  $f_{\text{FDP}}$  about the DP is estimated as follows:

$$\begin{split} m_{\text{DP/FDP}} &= m_{\text{DP}/f_{\text{FDP}}}(\theta_{\text{DIP}}, f_{\text{FDP}}) + m_{\text{DP}/R_{\text{DIP}}}(\theta_{\text{DIP}}, \theta_{\text{PIP}}, \theta_{\text{MCP}}, f_{\text{FDP}}) \\ &= r_{\text{FDP/DP}}(\theta_{\text{DIP}}) f_{\text{FDP}} + r_{R_{\text{DIP}}}(\theta_{\text{DIP}}, f_{\text{FDP}}) \times R_{\text{DIP}}(\theta_{\text{DIP}}, \theta_{\text{PIP}}, \theta_{\text{MCP}}, f_{\text{FDP}}) \end{split}$$
(A1)

where  $m_{\text{DP}/f_{\text{FDP}}}$  denotes the rotational moment of DP produced by  $f_{\text{FDP}}$ , and  $m_{\text{DP}/R_{\text{DIP}}}$ denotes the moment generated by the joint reaction force  $R_{\text{DIP}}$ . Here,  $r_{\text{FDP}/\text{DP}}$  denotes the MA vector indicating the distance from the DP center of mass to the line of action of the FDP tendon force, and the MA vector denoting the distance from the DP mass center to the line of action of the DIP joint reaction force  $R_{\text{DIP}}$ . As indicated in Fig. 7, the MA of the FDP tendon  $r_{\text{FDP}/\text{DP}}$  varies along with the change of the joint angle  $\theta_{\text{DIP}}$ , and can be estimated from the geometric variables, i.e.,  $d_{a/\text{DP}}$ ,  $d_{b/\text{DP}}$ ,  $d_{c/\text{MP}}$ ,  $d_{DIP}$  and  $r_{\text{DP}}$  as shown in Fig. 7 (see Table IV for the values of the geometric variables). Note that the DIP joint reaction force  $R_{\text{DIP}}$  should be calculated from the force and moment equilibrium equations that include all four segments of the finger system.

In a similar way, on the MP, pulley forces constraining FDP tendon at A4 and A5 pulleys were considered on the palmar side. The A5 pulley transmits a force  $f_{\text{FDP}/\text{A5}}^{\text{p}}$  on the MP near the DIP joint (see Fig. 7), whereas the force transmitted by the A4 pulley force  $f_{\text{FDP}/\text{A4}}^{\text{p}}$  is applied near the PIP joint (see Fig. 8). Also, the moment indirectly generated by  $f_{\text{FDP}}$  via the





Finger pulley structure around/near the PIP joint.  $f_{FDP/A4}^{p}$  is applied to MP, whereas

 $f_{\text{FDP}/\text{A2}}^{\text{p}}$  is applied to PP.

DIP joint reaction force should be considered. Then, moment generated by the FDP tendon force about the MP is

$$\begin{split} m_{\text{MP/FDP}} &= m_{\text{MP} \ / \ R_{\text{DIP}}}(\theta_{\text{DIP}}, \theta_{\text{PIP}}, \theta_{\text{MCP}}, f_{\text{FDP}}) + m_{\text{MP} \ / \ \text{A4}}(\theta_{\text{PIP}}, f_{\text{FDP}}) + m_{\text{MP} \ / \ \text{A5}}(\theta_{\text{DIP}}, f_{\text{FDP}}) \\ f_{\text{FDP}}) &+ m_{\text{MP} \ / \ R_{\text{PIP}}}(\theta_{\text{DIP}}, \theta_{\text{PIP}}, \theta_{\text{MCP}}, f_{\text{FDP}}) \,. \end{split}$$

(A2)

Here,  $m_{\text{MP}/R_{\text{DIP}}}$  denotes the moment produced by the reaction force exerted on the DIP joint capsule,  $m_{\text{MP}/A4}$  the moment generated from the A4 pulley force  $f_{\text{FDP}/A4}^{\text{p}}$ , which constrains the FDP tendon path (see Fig. 8),  $m_{\text{MP}/A5}$  the moment produced by the A5 pulley constraining force  $f_{\text{FDP}/A5}^{\text{p}}$  (see Fig. 7), and  $m_{\text{MP}/R_{\text{PIP}}}$  the moment applied to MP by the reaction force from the proximal end of the MP; as in  $R_{\text{DIP}}$  calculation,  $R_{\text{PIP}}$ , PIP joint reaction force applied to MP, is estimated from the force and moment equilibrium equations of all segments. Note that  $m_{\text{MP}/R_{\text{DIP}}}$  and  $m_{\text{MP}/A5}$  magnitudes are affected by the DIP joint angle, whereas  $m_{\text{MP}/A4}$  changes along with the PIP joint angle.

In the same way, on the PP, path constraining forces at A2 and A1 pulleys, as well as the joint reaction force transmitted from the distal end of the PP, generated by the FDP force, were considered (Fig. 9)

$$\begin{split} m_{\rm PP/FDP} &= m_{\rm PP \ / \ R_{\rm PIP}}(\theta_{\rm DIP}, \theta_{\rm PIP}, \theta_{\rm MCP}, f_{\rm FDP}) + m_{\rm PP \ / \ A2}(\theta_{\rm PIP}, f_{\rm FDP}) + m_{\rm PP \ / \ A1}(\theta_{\rm MCP}, f_{\rm FDP}) \\ ) &+ m_{\rm PP \ / \ R_{\rm MCP}}(\theta_{\rm DIP}, \theta_{\rm PIP}, \theta_{\rm MCP}, f_{\rm FDP}) \end{split}$$

(A3)

where  $m_{\text{PP}/R_{\text{PIP}}}$  denotes the moment produced by the reaction force exerted on the PIP joint capsule of PP,  $m_{\text{PP}/A2}$  the moment generated from the distal side of the A4 pulley force that constrains the FDP tendon path,  $m_{\text{PP}/A1}$  the moment produced by the A1 pulley constraining force, and  $m_{\text{PP}/R_{\text{MCP}}}$  the moment applied to PP by the reaction force from the proximal end of the PP. Force and moment equilibrium equations of all segments are incorporated to obtain reaction forces ( $R_{\text{PIP}}$  and  $R_{\text{MCP}}$ ), as in MP and DP.





## Fig. 9.

Finger pulley structure around/near MCP joint (a) without and (b) with the consideration of an A1 pulley.  $f_{FDP/A2}^{p}$  (or  $f_{FDP/A1}^{p}$ ) is applied to PP, whereas  $f_{FDP/P}^{p}$  is applied to metacarpal bone.

Here, a slight adjustment was made to the parameters of the normative model [22] in order to incorporate the effects of the A1 pulley on the MCP flexion; the A1 pulley has been found to participate significantly in the MCP flexion [29]. In that study, release of A1 pulley

resulted in the significant decrease of the FDP (or FDS) force required for the full MCP flexion, as well as an 8° bowstring of the tendon, which indicates an 8° difference in the tendon force direction ( $f_{\text{FDP}}$  in Fig. 9) between the configurations with [see Fig. 9(b)] and without A1 pulley [see Fig. 9(a)] when  $\theta_{\text{MCP}}$  is about 90°. From this, we estimated the

modified geometric parameters,  $d'_{a/PP}$  and  $d'_{b/PP}$ , which produced the observed change of 8° in the  $f_{FDP}$  direction when  $\theta_{MCP} \approx 90^{\circ}$ . Accordingly,  $m_{MP/RPIP}$  and  $m_{MP/A2}$  are modeled as functions of  $\theta_{PIP}$ ,  $f_{FDP}$ , whereas  $m_{MP/A1}$  is a function of  $\theta_{MCP}$ ,  $f_{FDP}$ .

Then, the passive stiffness and damping of each joint was incorporated, which were modeled as functions of either the joint angular displacements or velocities [8]. By combining the force and moment equilibrium equations obtained at the mass center of three segments, the matrix form of the equations of motion for the forward dynamic simulation is formed

$$\mathbf{I}(\boldsymbol{\theta}) \cdot \ddot{\boldsymbol{\theta}} = \mathbf{b}_{\mathrm{C}}(\boldsymbol{\theta}, \dot{\boldsymbol{\theta}}) + \mathbf{b}_{\mathrm{T}}(\boldsymbol{\theta}, f_{\mathrm{FDP}}) + \mathbf{b}_{\mathrm{PS}}(\boldsymbol{\theta}) + \mathbf{b}_{\mathrm{PD}}(\dot{\boldsymbol{\theta}}) \quad (A4)$$

where  $\boldsymbol{\theta} = \begin{bmatrix} \theta_{\text{DIP}} & \theta_{\text{PIP}} & \theta_{\text{MCP}} \end{bmatrix}^T$ . Here,  $\mathbf{I}(\boldsymbol{\theta})$  is the inertia matrix,  $\mathbf{b}_{\text{C}}(\boldsymbol{\theta}, \dot{\boldsymbol{\theta}})$  the Coriolis terms,  $\mathbf{b}_{\text{T}}(\boldsymbol{\theta}, f_{\text{FDP}})$  the vector denoting the moments generated by the tendon force  $(f_{\text{FDP}})$  via joint reaction forces and pulley structures,  $\mathbf{b}_{\text{PS}}(\boldsymbol{\theta})$  the passive joint stiffness vector, and  $\mathbf{b}_{\text{PD}}(\dot{\boldsymbol{\theta}})$  the passive joint damping vector. For the mathematical description of the inertia matrix and Coriolis term of the 3-DOF finger linkage system [i.e.,  $\mathbf{I}(\boldsymbol{\theta})$  and  $\mathbf{b}_{\text{C}}(\boldsymbol{\theta}, \dot{\boldsymbol{\theta}})$ ], see [37], and for the equations for the passive joint stiffness and damping [i.e.,  $\mathbf{b}_{\text{PS}}(\boldsymbol{\theta})$  and  $\mathbf{b}_{\text{PD}}(\dot{\boldsymbol{\theta}})$ ]. see [8].

From (A4), contribution of each kinetic component to the joint acceleration (i.e., joint moment) can be estimated

$$\ddot{\boldsymbol{\theta}} = \mathbf{I}^{-1} \cdot \mathbf{b}_{\mathrm{C}}(\boldsymbol{\theta}, \dot{\boldsymbol{\theta}}) + \mathbf{I}^{-1} \cdot \mathbf{b}_{\mathrm{T}}(\boldsymbol{\theta}, f_{\mathrm{FDP}}) + \mathbf{I}^{-1} \cdot \mathbf{b}_{\mathrm{PS}}(\boldsymbol{\theta}) + \mathbf{I}^{-1} \cdot \mathbf{b}_{\mathrm{PD}}(\dot{\boldsymbol{\theta}}).$$
(A5)

Here, each row of each component (i.e.,  $\mathbf{I}^{-1} \cdot \mathbf{b}^*$ ) denotes its contribution to the corresponding joint acceleration, i.e., to the joint moment generation (see Fig. 3) at a given time and configuration. For example, the first row of  $\mathbf{I}^{-1} \cdot \mathbf{b}_{T}(\boldsymbol{\theta}, f_{FDP})$  denotes the contribution of  $f_{FDP}$  to the DIP joint moment generation.

## Biographies

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#### Fig. 2.

(a) and (b) Finger flexion movements obtained from the simulation and (c) the experiment. In both models, initial joint angles were selected as (DIP, PIP, MCP) =  $(5^{\circ}, 12^{\circ}, 12^{\circ})$  to approximately match with the joint angles in neutral posture of the subject whose *in vivo* stimulation data are shown in (c). Spatial coordination of the joint flexion angles of the model-generated movements [(a) model I, (b) model II] in response to two levels of tendon force ( $f_T$  = 2.5 N versus 3.5 N) were compared with that of representative flexion movements generated in response to electrical stimulation of FDP(I) (subject 1) (c).

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### Fig. 3.

Sensitivity analysis of the model parameters. Solid line denotes the joint angle change corresponding to the change in the pulley distal/proximal distance from the center of rotation from the adjacent joint. Dotted line denotes the joint angle change corresponding to the pulley thickness change (see Figs. 7–9 for the detailed description of pulley parameters). Joint angles were more sensitive to the change in the pulley thickness than its distal/ proximal distance to the joint center. PIP flexion angle was sensitive to the distal A2 pulley thickness and the proximal A4 pulley thickness.



## Fig. 4.

Simulation of pulley excisions in model II. Compared to the intact condition [see Fig. 2(b)], (a) A1 pulley excision resulted in an increase in MCP flexion angle by  $\sim 10^{\circ}$  and (b) partial A2 distal pulley excision (50%) increased the PIP flexion angle by  $\sim 4^{\circ}$ .





## Fig. 5.

Simulation without passive stiffness and damping. FDP tendon force = 0.01 N. Removal of passive joint properties resulted in rapid DIP flexion, followed by PIP flexion; MCP joint was extended. Here, both DIP and PIP joint flexions reached their upper bound, which were estimated from the geometric parameters of tendon location given in the normative finger model [22].

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## Fig. 6.

Contribution of kinetic components to the joint angular acceleration for the flexion movements generated by model II (see Fig. 1). (a) Small flexion ( $f_T$ = 2.5 N). (b) Large flexion ( $f_T$ = 3.5 N). For each movement: 1) contribution of all kinetic components and 2) sum of these contributions, which is equivalent to the resulting joint angular acceleration [see (A5)] were presented. Rapid initial change in the angular acceleration was resulted due to the small but nonzero initial tendon force (see Fig. 1); the *y*-axis range (summed

acceleration) was adjusted to better demonstrate typical acceleration-deceleration phases of the flexion movements.

## TABLE I

## MA VALUES Employed in This Study (in Millimeters)

Joint	DIP	PIP	МСР
FDP moment arm	4.1	7.9	11.1

\* Values taken from [3].

## TABLE II

JOINT ANGLES AT FINAL POSTURE AND THEIR PROPORTION IN TOTAL FLEXION ANGLE IN SIMULATED MOVEMENTS

Monomont true	Model	Joint angles at final posture			Datia (0/)
Movement type		DIP	PIP	МСР	Kau0 (%)
Small flexion	Ι	10	29	33	14:40:46
	II	13	38	28	17:48:35
Large flexion	Ι	13	40	47	13:40:47
	II	19	56	34	18:51:31

## TABLE III

JOINT ANGLES AT FINAL POSTURE AND THEIR PROPORTION IN TOTAL FLEXION ANGLE IN SIMULATED MOVEMENTS

Movement type	Subject	Joint angles at final posture			
		DIP	PIP	МСР	Ratio (%)
Small flexion	Subject 1	16	36	25	21:47:32
	Subject 2	25	52	27	24:50:26
	Subject 3	14	45	13	20:62:18
	Mean (SD)	18 (6)	44 (8)	22 (8)	22 (2) : 53 (8) : 25 (7)
Large flexion	Subject 1	32	46	36	28:40:32
	Subject 2	40	67	30	29:49:22
	Subject 3	20	52	25	21:54: 26
	Mean (SD)	31 (10)	55 (11)	30 (6)	26 (4) : 48 (7) : 26 (5)