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The 3D Thumb-tip Forces Produced By Individual Tendons Do Not Superimpose Linearly

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Abstract— **Clinicians and biomechanists commonly assume the thumb-tip force produced by tension in individual tendons superimpose linearly. We tested this assumption experimentally in cadaver thumbs (n=13) by developing a specimen-specific linear model and quantifying the error of this model's predictive ability. We developed these models by linearly regressing 3D thumb-tip output force/torque against tendon tension. We applied linear optimization to these models to calculate the combination of tendon tensions predicted to maximize 3D thumb-tip force vectors in each of five directions (palmar, dorsal, radial, ulnar, distal) under four neurological conditions (intact, low-median paralysis, transfer A, transfer B), and applied these optimal tensions to the specimen. We quantified the error in the linear model by calculating the magnitude difference and angle between predicted and measured force vectors. This error depended on the direction of maximized thumb-tip force. The measured force vector magnitude was either significantly ($p<0.05$) lower (distal and dorsal) or not significantly different (palmar, radial, ulnar) from the predicted magnitude, and the angle between predicted and measured vectors varied greatly (distal: 20°±20°; palmar: 22°±13°; radial: 36°±13°, ulnar: 34°±20°; dorsal: 51° ± 13°). Our results suggest that the linear assumption is most accurate when the thumb-tip force is in the palmar direction.**

Keywords—**Thumb, cadaver experiment, biomechanical model**

I. Introduction

Biomechanics researchers and clinicians commonly assume that the thumb is a linear system, where thumb-tip forces produced by tension in individual tendons superimpose linearly. This assumption is implicit in biomechanical models of the thumb [1-3] that assume the orientation and location of joint axes is fixed with respect to rigid bone segments, and the moment arms (perpendicular distance from a tendon to the joint it crosses) depend only on the joint angles (thumb posture). These assumptions imply that individual tendon tensions superimpose linearly to produce a resultant thumb-tip force vector in the case of static pinch where thumb posture is constant.

Similarly, clinicians commonly use moment arm measurements, and moment-generating-capacity diagrams (which are calculated from the moment arms) to understand the functional properties of the muscles and to compare and contrast tendon transfers [4]. These moment arm measurements are obtained experimentally while nominal loads (e.g., 2N [5]) are placed on the tendons. If load-dependent changes occur, such as load-dependent motion of the trapezium, as suggested by one clinical researcher [4], these moment arms and posture may change, resulting in a change in thumb-tip force output. Biomechanically, these load-dependent changes would suggest the thumb is a non-linear system, where the thumb-tip output force due to tension in one tendon would be changed by tension in another tendon.

Our goal in this study was to test the assumption that thumb-tip force vectors produced by tension in individual tendons superimpose linearly. Testing the veracity of this assumption will advance the understanding of the thumb, and specifically help to establish if conventional biomechanical models can be used to model the thumb, or if more sophisticated models need to be employed to model the healthy, diseased, and post-surgical thumb.

To test if the thumb-tip force due to individual tendons superimpose linearly, we used a previously validated system identification/linear optimization technique [6, 7] that rigorously tests the linearity assumption on cadaver specimens. We hypothesized that the thumb is a linear system that can be modeled accurately by a set of linear equations representing the relationship between tendon tension and thumb-tip force.

II. Methodology

A.. Specimen Dissection and Mounting

As in previous studies of the index finger [6, 7] and thumb [8], we began by dissecting the intrinsic and extrinsic muscles acting in the thumb of 13 fresh-frozen forearm specimens (7 males, 6 females, Age=76±9 years). Specimens were pre-screened for blood-borne pathogens and kept moist with a mixture of bovine serum (10%, product #C6278, Sigma Chemical Company, St Louis, MO) and water during the entire dissection and experiment. We isolated and removed from their origins the extrinsic muscle bellies (*Flexor Pollicis Longus* (FPL), *Extensor Pollicis Longus* (EPL), *Extensor Pollicis Brevis* (EPB), and *Abductor Pollicis Longus* (AbPL)) and tied and glued 1-mm Nylon cords (Vetbond Tissue Adhesive, 3M Inc., St. Paul, MN) to the each tendon. Next, we isolated and removed from their origin the intrinsic muscles of the thenar eminence (*Abductor Pollicis Brevis* (AbPB), *Flexor Pollicis Brevis* (FPB), and *Opponens Pollicis* (OPP) and tied and glued Nylon cords to the AbPB and FPB at their insertions. The muscle belly of the OPP was completely excised, and an eyehook (3-mm ID) was screwed flush to the surface of

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the first metacarpal at the OPP insertion. Other eyehooks were screwed flush into the trapezium and trapezoid at the origins of the AbPB and OPP, respectively, and were used to route the Nylon cords. A floating 3-mm ring tied around the proximal end of the third metacarpal represented the origin of the FPB in the palmar fascia. We excised the belly of the *adductor pollicis* (ADD) muscle and screwed an eyehook into the proximal phalanx at the ADD origin. Because of the fan-shape of the ADD, we represented the muscle with two Nylon cords and routed the ADD oblique (ADDo) cord through an eyehook placed in the capitate, and routed the ADD transverse (ADDt) through an eyehook placed in the distal end of the palmar aspect of the third metacarpal; these eyehooks were placed at the extreme proximal and distal locations of the ADD origin to represent all of the possible actions (by vector addition) of the fan-shaped ADD. The dorsal interosseous (DIO) was isolated and excised and eyehooks were placed in the first metacarpal and in the second metacarpal at the origin and insertion of the DIO, respectively. We also routed two tendon transfers commonly used for low median nerve palsy [9]: transfer A (TRa), attributed to Burkhalter et al., is performed by transferring the *extensor indicis proprius* muscle to the insertion of the failed AbPB via the pisiform bone, and transfer B (TRb), attributed to Riordan, is performed by transferring the *flexor digitorum superficialis* (FDS) of the ring finger to the insertion of the failed AbPB via a slip in the *flexor carpi ulnaris* (FCU).

After the dissection, we fixed the hand to a custom load frame (Fig. 1) in neutral wrist flexion/extension and radial/ulnar deviation [6, 7] with a external fixation device (Agee-WristJack, Hand Biomechanics Lab, Inc., Sacramento, CA). The custom load frame (Fig. 1) was outfitted with ten linear actuators (801B-AM, American Scientific Instrument Corp., Palos Hills, IL) mounted in-line with a uniaxial load cell (SML series ± 10lb InterfaceForce, Scottsdale, AZ) (to apply tendon tension). A 6 degree-of-freedom force/torque sensor (F/T nano17, ATI/Industrial Automation, Garner, NC) was used to measure thumb-tip force. A personal computer (Celeron, Dell Computer Corporation, Round Rock, Texas) with a data acquisition card (DAQ PCI 6021E, National Instruments Corporation, Austin, TX) running custom programs in LabVIEW (National Instruments Corporation, Austin, TX) controlled the linear actuators attached to each tendon's Nylon cord and recorded the force/torque sensor.

We configured the thumb in two functional postures using a manual goniometer: key pinch (10° interphalangeal (IP) flexion, 45° metacarpophalangeal (MP) flexion, and 0° carpometacarpal (CMC) abduction and flexion) and opposition pinch (45° IP flexion, 10° MP flexion, 45° CMC abduction, and 0° CMC flexion). We formed thermoplastic molds (Omega Max, North Coast Medical Inc., Morgan Hill, CA) for each of these postures to hold the thumb in position while the thumb-tip was being rigidly coupled to the force/torque sensor via a thermo-plastic thimble.

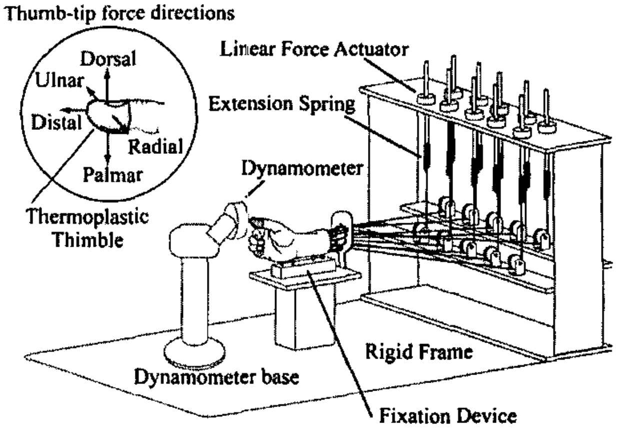


Fig. 1. Custom Load Frame. Insert: Desired force directions.

We formed this thimble around the thumb-tip and glued to the thumb pulp *only*, to mimic an *in vivo* high friction physiological boundary condition where the thumb-tip force vector is produced is through the pulp of the thumb and axial motion of the thumb is allowed to some natural extent.

B. Developing The Linear Model of the Thumb

We developed a specimen-specific linear model in both key and opposition pinch by measuring the thumb-tip output force/torque while applying tension to each tendon cord individually. We commanded each linear actuator to incrementally ramp tension up from zero to a desired value and back down to zero in ten ramp-and-hold steps in each direction under force control (precision: ± 0.05 N). We recorded thumb-tip output force/torque vector during hold periods to exclude actuator vibrations. The maximal force for each muscle was based on the mass fraction reported in the literature [4] and was scaled so that each muscle reached approximately 1/3 of its maximal predicted force based on its physiological cross-sectional area (PCSA).

After recording the data, we linearly regressed the tendon forces against the thumb-tip forces/torques in the 3 Euclidean directions, and compiled a 6x12 transfer function matrix (A) for each specimen in each posture, where each column of this matrix is composed of the six linear regression coefficients (3 force and 3 torque) for one of the 10 muscles and 2 tendon transfers. A matrix equation $\mathbf{Au}=\mathbf{b}$ can be formed where multiplying a (12x1) column-vector of tendon tensions ($\mathbf{u}=[T_1,T_2,T_3,T_4,T_5,T_6,T_7,T_8,T_9,T_{10},T_{11},T_{12}]^T$) with the transfer function results in a (6x1) column vector of predicted thumb-tip output forces/torques ($\mathbf{b}=[f_x,f_y,f_z,t_x,t_y,t_z]^T$) based on the linear model. The force/torque sensor was oriented such that its coordinate axes were aligned with the target orthogonal directions of fingertip force production (Fig. 1).

C. Testing if thumb-tip output forces superimpose linearly

We used the matrix equation to find the optimal combination of tendon tensions (u) which would maximize thumb-tip output force in five orthogonal directions (palmar, dorsal, radial, ulnar, distal), for four different neurological conditions (intact, paralyzed, Transfer A, Transfer B) using the linear programming algorithms in *Mathematica* (Wolfram Research, Inc. Champaign, IL). In the intact case all muscles were potentially active (excluding transfers) while in the paralyzed case the AbPB, FPB, and OPP were constrained to zero tension (simulating low-median nerve palsy). The Transfer cases were identical to the paralyzed case, but included either TRa or TRb tension in the optimization. We used the appropriate rows of matrix A as the cost function and constraint equations for each force direction [6, 7]. By iteratively adjusting the constraints on the force components perpendicular to the desired direction we predicted the maximal output force vector directed within 15° of the desired direction. We constrained the torques to be functionally small ($-0.2 \le t_x, t_y, t_z \le 0.2$ Nm) and the tension in *non-paralyzed* tendons (u) to be ≥ 0 and $\le 1/3$ their predicted maximum value from PCSA. We applied the predicted optimal combinations of tendon tensions to the specimen by simultaneously ramping all tendons to their optimal tension levels and measuring the thumb-tip output force/torque in ten steps. To test repeatability of the experiment, we repeated one of the trials (one direction, one case) at random.

We quantified the error of the linear model by calculating the vector magnitude difference and the included angle between the measured and predicted thumb-tip force, and performed a repeated measures ANOVA [10] in SuperAnova (SAS Institute Inc., Cary, NC) controlling for all factors (case, direction, and posture). If significant differences were found for some factors, we used a Tukey-Kramer pair-wise comparison to determine the groupings, if any, among the elements of those significant factor(s). We quantified repeatability of the experiment by calculating the included angle and magnitude difference between the thumb-tip output force vectors of repeated trials.

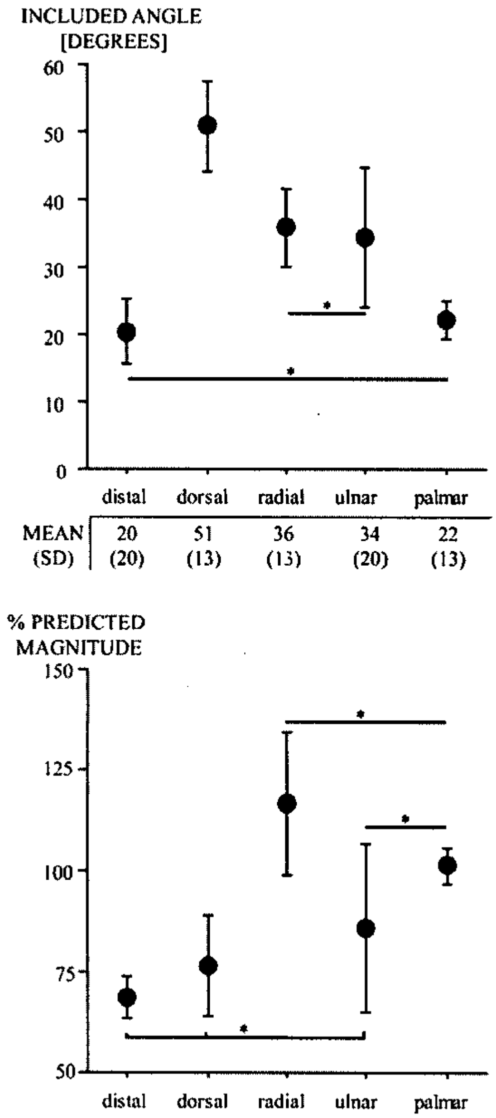
III. Results

The repeated measures ANOVA demonstrated that direction was the only significant ($p<0.05$) factor for both dependent variables: % predicted magnitude and included angle between measured and predicted force vectors.

A line plot for the included angle dependent variable (Fig. 2, top, grouped by Tukey-Kramer post-hoc), demonstrates that the distal and palmar direction were best aligned with the predicted direction (smaller angles), the ulnar and radial direction were moderately misdirected, and the dorsal direction was most misdirected.

A line plot of the % predicted magnitude dependent variable (Fig. 2, bottom), demonstrates that in the trials where the radial, ulnar, or palmar force was maximized, the measured force was not significantly different from the predicted force (i.e., 100%). Of these three directions, the palmar direction was most consistent with the predicted force magnitude, with a relatively small standard deviation and a mean of 101%±20% SD. In the trials where force was maximized in the distal or dorsal directions the % predicted magnitude was significantly below 100%.

Between repeated trials, we found that thumb-tip force vector magnitude varied minimally (% change between trials: 1% ± 4%) and had small included angles (4° ± 5°).



	distal	dorsal	radial	ulnar	palmar
MEAN	20	51	36	34	22
(SD)	(20)	(13)	(13)	(20)	(13)

	distal	dorsal	radial	ulnar	palmar
MEAN	69	76	117	86	101
(SD)	(21)	(24)	(41)	(41)	(20)

Fig. 2. Line plots of included angle (top) between measured and predicted thumb-tip force and % of predicted magnitude of thumb-tip force (bottom). Error bars are 95% confidence intervals. *Not significantly different, Tukey-Kramer post-hoc analysis. Tables below graphs report mean and standard deviation (SD) of the values in the plots above.

IV. Discussion

We found that the measured thumb-tip output force is most well aligned with the predicted thumb-tip force in the distal (20°±20) and palmar (22°±13) directions. The measured force vector magnitude in the palmar direction most accurately reproduced the predicted force magnitude (101%±20 in palmar vs. 69%±21 in distal). Thus, we argue that the thumb behaves most linearly when producing force in the palmar direction.

The moderate to large misdirection (Fig. 2, top: dorsal, ulnar, and radial directions) and/or the large magnitude difference (Fig. 2, bottom: distal and dorsal directions) between the measured versus predicted magnitude suggests the thumb behaves non-linearity when producing thumb-tip force vectors in these directions.

We noticed during the experiments slight motion of the bones of the thumb, especially the trapezium, that we argue is, at least in part, a contributor to this non-linearity. Load-dependent motion has been reported elsewhere for the trapezium [4] and carpus in general [11], but this motion has not been quantified. Even though this study was not designed to reliably measure trapezial motion, we confirmed trapezial motion in two specimens by measuring the displacement of a pin inserted in the trapezium (using digital video close-ups) as a function of FPL tension. We found this motion to be ~2.0 mm in the proximal direction at maximal FPL tension [8]. To exclude the possibility that our dissection of the intrinsic musculature led to trapezial laxity, we performed an experiment where we loaded FPL prior to the dissection of the intrinsic muscles, and noted a similar motion of a pin inserted in the trapezium. Thus, we propose the thumb does not have a rigid base at the trapezium, but in reality acts as a "floating digit" affected by motion of the carpal bones when its tendons are loaded [8]. This load-dependent displacement of the trapezium likely contributes to the nonlinear transmission of tendon tension by, at a minimum, affecting joint seating and configuration, which in turn affect the transformation of individual tendon tensions into thumb-tip force vectors.

Because we based our linear model from data collected when tendons were ramped to 1/3 of their predicted maximum muscle force, the non-linearities we report are restricted to this sub-maximal force regime. Nevertheless, because most functional manipulation occurs at sub-maximal force levels our results are representative of realistic requirements of thumb function. In addition, the routing of the Nylon cords, including the visual placement of eye-hook screws, may not have accurately reproduced the paths of some muscles. Furthermore, activity in one muscle may affect the line of action of another muscle, such as when the activity in the OPP may cause the AbPB to separate from the thumb CMC joint (increasing its moment arm [4]). Finally, the constraints on our optimization routine did not include maintaining a zero net moment at the wrist joints, which is common in functional manipulation. Thus, our experiment is analogous to the *in vivo* case where the wrist is constrained (e.g., with a brace).

V. Conclusion

Our results suggest that the assumption that the thumb acts as a linear system does not affect the work of clinicians, since they aim to reproduce pinch forces which are primarily in the palmar direction. Similarly, biomechanical models that assume of the thumb is linear will most realistically predict palmar thumb-tip force vectors (compared to other force directions, Fig. 1), but the predicted direction will be misdirected to the levels we reported (Fig. 2, top: 22±13 for palmar direction). For the other force directions (dorsal, ulnar, radial, and distal), the veracity of linear biomechanical model is reduced (Fig. 2) and their predictions should be interpreted carefully.

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