Orthopaedic Applications of a Validated Force-based Biomechanical Model of the Index Finger

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***Abstract*—An anatomically realistic biomechanical model of the index finger was created using a force-based approach in order to predict the isometric fingertip force or dynamic movement based on the forces of 7 index finger musculotendons. The model was validated for static forces through comparison with experimental results from 5 cadaver specimens. The model reliably simulated the isometric fingertip force produced by loading individual tendons. The average error in fingertip force direction was less than 2° and the average error in magnitude was less than 10% across finger postures for each muscle. Subsequent employment of the model to examine force transmission from the long flexors revealed a strong dependence of joint contact force on finger posture for a given tendon load. This may have ramifications for osteoarthritis as high joint contact forces are thought to contribute to the disease.**

I. Introduction

Higher joint contact force has been reported to influence the development of osteoarthritis (OA). For the patient with early stage OA, the repeated loading of the affected joint may lead to pain and progressive destruction of the joint cartilage [1]. Study of the relationship between the joint contact force and the finger functional task, finger posture and muscle activation pattern can possibly provide solutions to prevent repeated and harmfully high joint loading. However, direct *in-vivo* measurement of the joint contact force is invasive and is almost impossible to achieve in practice. Alternatively, an anatomically realistic finger model could provide feasible and reliable estimations of the contact forces.

While a number of finger models have been created [2-4], to our knowledge they have not been utilized to estimate joint contact forces. In addition, validation has often been limited. We created a three-dimensional dynamic model of the finger which describes anatomically realistic force transmission among tendons, skeleton and joints. In this study, the force-based model was tested for simulations of static fingertip force generated by loading of individual tendons across multiple finger postures over the finger working space. Its predictions were compared with the measured results in cadaveric experiments to demonstrate its accuracy. The validated model was then used to estimate the joint contact forces which would be experienced *in-vivo*.

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II. Methods and Procedures

A. Modeling

The index finger was modeled as a serial chain of four links connected by revolute joints. Both the proximal interphalangeal (PIP) and distal interphalangeal (DIP) joints were represented as hinge joints, each with one flexion-extension degree-of-freedom (DOF). The metacarpophalangeal (MCP) joint is considered a 2-DOF ellipsoid joint, with flexion-extension and abduction-adduction movement. The three flexion-extension axes are modeled as parallel axes, orthogonal to the long axes of the phalanges. To mirror the finger physiology, the abduction-adduction axis was placed at an angle of 70° with respect to the metacarpal bone within the sagittal plane, rather than 90° [5]. An open chain of four links was used for the finger, with the proximal segment (metacarpal) attached to the ground. The three phalangeal segments were represented by cylinders [2, 4] with uniform density of 1.1 g/cm^3 [4]. The lengths and cross-sectional diameters of each segment were measured on 5 fresh-frozen cadavers (mean age of 79, all female). The finger segment mass, center of mass position, and moment of inertia were calculated from the geometry.

The tendons of the seven index finger muscles were included in the model: flexor digitorum profundus (FDP), flexor digitorum superficialis (FDS), extensor digitorum communis (EDC), extensor indicis (EI), first dorsal interosseous (FDI), first palmar interosseous (FPI) and lumbrical (LUM). For the long flexors, the anatomical pulleys A1-A5 were modeled to constrain the tendon tightly against the phalanges. The flexor tendon path through each joint was described according to the geometry of Landsmeer`s Model III [6]. The extensor mechanism, which involves two long extensors EDC and EI and two intrinsic muscles FPI and LUM, were represented with a Winslow's rhombus tendon network [3, 7]. The force distribution within the extensor mechanism was estimated through the experimental validation, which is described in the next section. The FDI tendon inserts into a single point on the radial side of the proximal phalanx. A via point for FDI was also created on the radial side of the metacarpal bone. All tendon locations were obtained from measurements made by An [8]. Joint passive properties, including stiffness and damping, were included as functions of angle for each joint. These values, taken from the literature, were measured on healthy subjects by introducing pseudo-random binary sequences of position perturbations on each of the joints [9].

The model was created within Simulink using the SimMechanics toolbox (MathWorks Inc., MA). The

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phalanges were modeled as BODY elements with given dimension and moment of inertia. The proximal segment, representing a fixed metacarpal bone, was connected to the ground. The MCP joint was represented by a UNIVERSAL joint, with the abduction-adduction axis opened 70° to the longitudinal axis of the metacarpal bone. PIP and DIP joints were represented by REVOLUTE joints. Anatomical pulleys, along with all muscle insertions are represented by BODY elements attached to a finger segment without mass. Tendons were not represented directly by a structure in the model, but rather the tendon forces were well represented as external forces calculated in real-time by the MATLAB functions, which utilized real-time finger geometry and tendon load. In static simulations, for the purpose of measuring isometric fingertip force, the geometric center of the distal phalanx was affixed to a fixed ground position (Fig. 1). This kept all joints at the desired angles throughout the simulation. During the dynamic simulations, all joints and the endpoint were allowed to move freely in accordance with the given tendon load. Gravity was ignored in the model simulation. In this model, computation of forward dynamics is handled by the SimMechanics toolbox. The resulting fingertip forces, joint contact forces and kinematics during the simulation are recorded and saved for analysis.

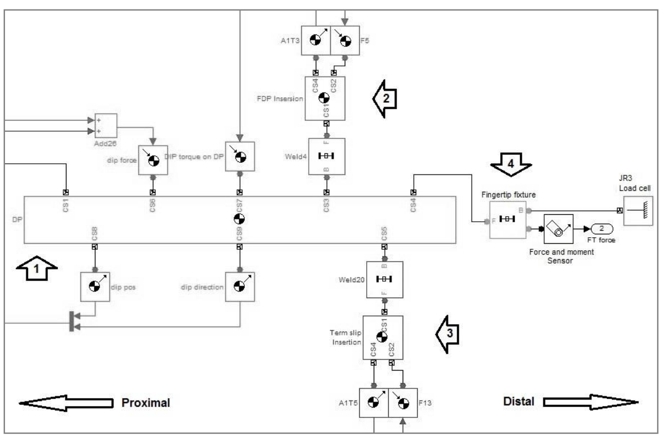

Figure 1. Part of the index finger biomechanical model in Simulink: Distal phalanx with tendon insertions and connection to JR3 load cell. As indicated by the arrows, (1) is the BODY element of distal phalanx; (2) is the BODY element of FDP insertion, which representing the connection material between end of FDP tendon and distal phalanx; similarly, (3) is the BODY element of terminal slip insertion; (4) is the JOINT element representing solid connection between distal phalanx and JR3 load cell. The force created in isometric force simulation is measured at this connection.

B. Model Validation

To test the model, the model predictions on isometric fingertip force produced by specific tendon forces were compared with the cadaveric experiment results across different finger postures. The model simulations were run to determine the fingertip forces and moments created by each of the 7 index finger musculotendons. Each was set to separately create 10% of maximum muscle force [10]. Thus, musculotendon force was used as the model input. For each musculotendon, the simulation was run at 9 different finger postures across the finger workspace. The postures were selected as the combination of MCP joint angle {0°, 30°, 60°} and PIP-DIP joint angles {(30°,0°), (45°,15°), (60°,30°)} (positive angles denote flexion). The predicted fingertip forces were compared with measurements in a cadaveric experiment (Fig. 2) with 5 specimens (mean age of 79, all female), in which each tendon was loaded separately with 10% of the muscle maximum force.

Since little is known about the force distribution between central and terminal slips within the extensor mechanism, a range of different ratios of the force distribution were tested through model simulation. The ratios were optimized to 62%:38% (Central Slip:Terminal Slip) for EDC, 59%:41% for EI, 53%:47% for FPI and 51%:49% for LUM, which resulting in minimum difference comparing with the experimentally measured forces.

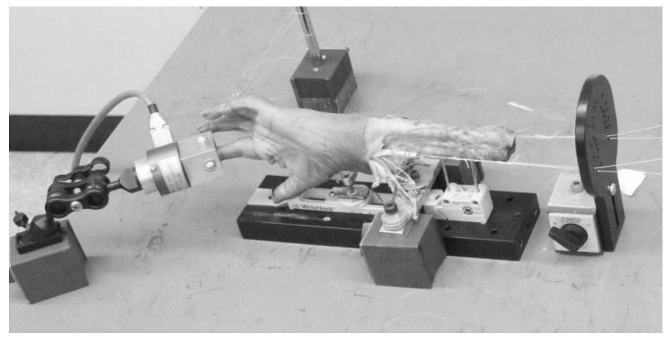

Figure 2. The cadaveric experiment setup. The hand specimen was mounted on a WristJack (Hand Biomech. Lab, CA, USA) fixation jig. The intended finger joint angles were determined with goniometer. The exposed tendons were tied to low friction steel wire and driven by stepper motors with force feedback control. Fingertip forces and moments were measured with a 6 degree-of-freedom load cell (JR3 Inc., CA, USA) secured to the distal finger segment with screws [11].

Minor adjustments were made for the tendon locations in the model. In the preliminary simulation, the fingertip force prediction for the intrinsic muscles was sensitive to their tendon position with respect to the MCP joint. Literature values [8] did not provide a good fit in terms of fingertip force in the simulation. To provide a better fit, the tendon locations of FDI, FPI and LUM that lie proximal to the MCP joint were optimized to minimize the prediction errors when comparing with the experimentally measured fingertip forces. The optimized tendon positions were used for the rest of this study.

C. Static Simulation to Predict Joint Contact Force

The model was subsequently employed to simulate the isometric force produced by activation of FDP and FDS muscles. Both of the muscles were set to 10% of their maximum force (8.2N for FDP; 6.3N for FDS). The simulation was run at thirteen MCP joint angles (0° to 60°) combined with twelve IP postures (15°-5° to 90°-30°). The joint contact force is measured as the force applied by the more distal phalanx onto the more proximal phalanx. The contact force normal to the joint surface and parallel to the long axis of the more proximal phalanx was the focus of this study, as higher contact forces can be harmful in the development of OA.

III. RESULTS

A. *Model Validation*

The model estimates of the resulting fingertip forces successfully predicted the effect of the MCP joint angle and IP posture on fingertip force direction and magnitude. The predictions were within or very close to the one standard deviation range of the cadaveric experimental results from the cadaver experiments (Figs. 3-5) across the postures. The average errors of the prediction that were outside one standard deviation of the experimental cadaver results are listed in Table I.

TABLE I. THE AVERAGE MODEL PREDICTION ERROR BEYOND ONE STANDARD DEVIATION RANGE OF CADAVERIC EXPERIMENT MEASUREMENT

	FDP	FDS	EDC	EI	FDI	FPI	LUM
Direction Err.	1.7°	0.7°	0.0°	0.0°	0.0°	0.0°	0.5°
Magnitude Err.	0.0%	2.9%	1.4%	10.0%	6.0%	1.1%	0.0%

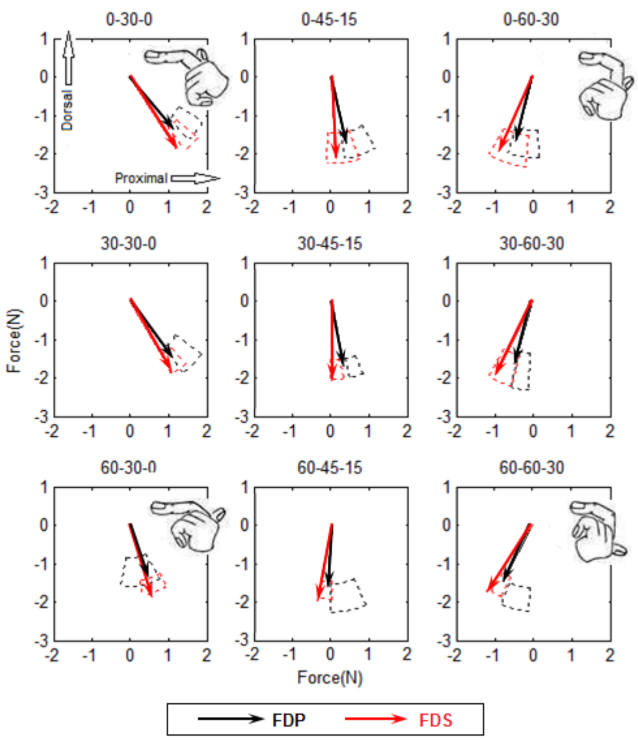

Figure 3. Comparison of model prediction (arrows) and experimentally measured fingertip force (dotted areas) in the sagittal plane as produced by FDP and FDS loading. Each tendon was loaded with 10% of the maximum muscle force, namely, 8.2N for FDP and 6.3N for FDS.

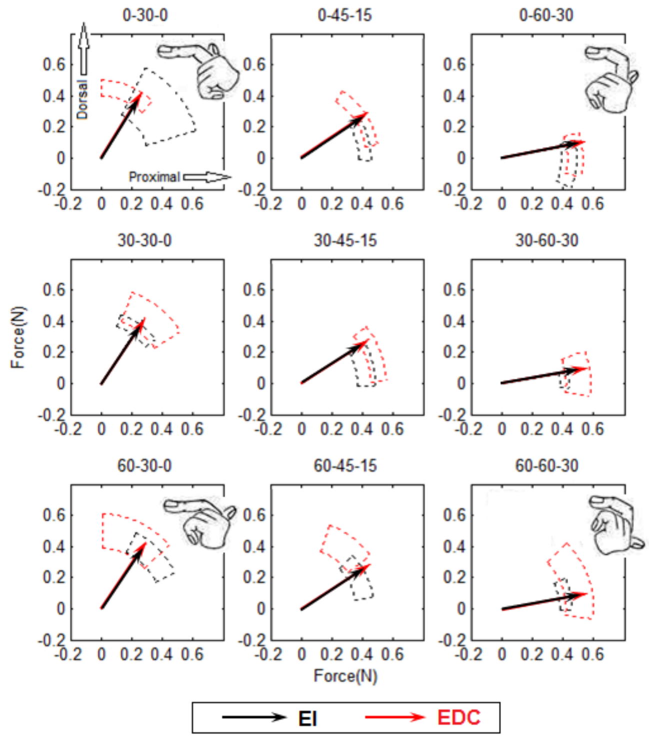

Figure 4. Comparison of model prediction and experimentally measured fingertip force in the sagittal plane as produced by EDC and EI loading. Each tendon was loaded with 10% of the maximum muscle force, namely, 3.6N for EDC and 3.4N for EI.

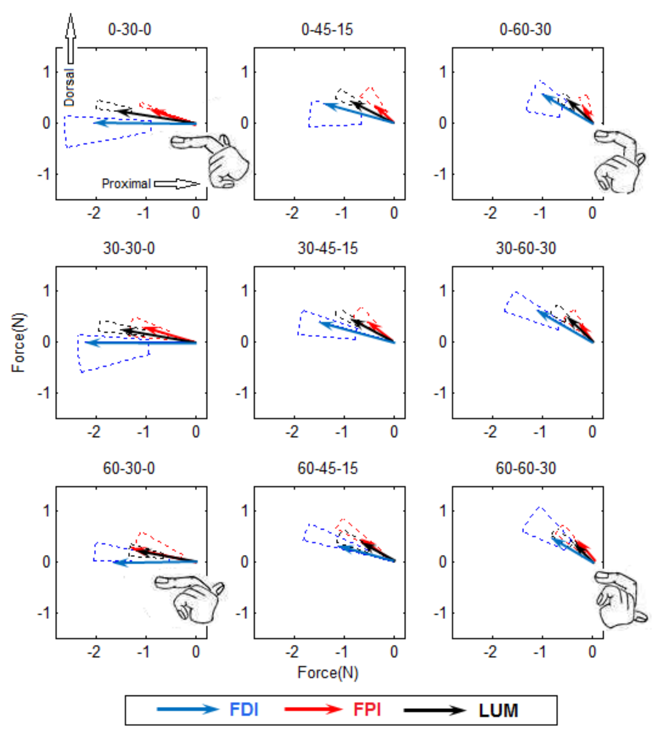

Figure 5. Comparison of model prediction and experimentally measured fingertip force in the sagittal plane as produced by FDI, FPI and LUM loading. Each tendon was loaded with 10% of the maximum muscle force, namely, 9.2N for FDI, 3.6N for FPI and 2.5N for LUM.

B. *Contact Force Prediction*

The relationship between contact force and MCP-PIP-DIP postures was revealed by the model simulations (Figs. 6-7). For both MCP and PIP contact forces, the maximum magnitudes were observed at the 60°-22.5°-7.5° posture. However, it should be noted that the contact force on PIP changed more (2.93N across postures) than that on MCP (2.3N across postures). The largest contact force on PIP (11.98N) was even larger than that observed on the MCP joint (11.09N).

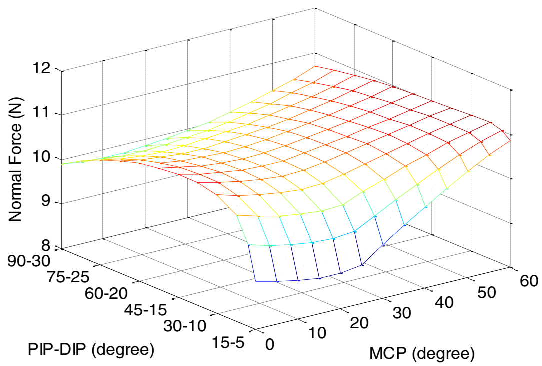

Figure 6. MCP joint contact force in normal direction created by FDP and FDS coactivation.

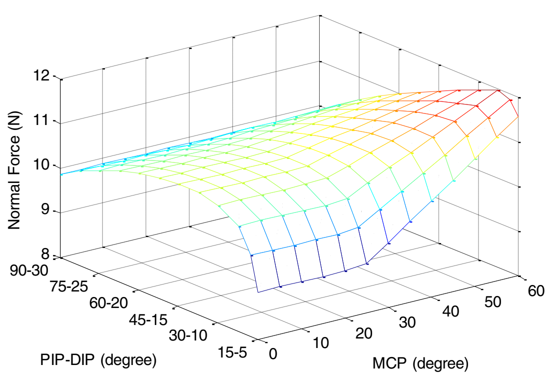

Figure 7. PIP joint contact force in normal direction created by FDP and FDS coactivation.

Generally, the contact force on DIP (with maximal value of 9.79N) is lower in magnitude compared to the MCP and PIP joints. This is largely due to the lack of FDS muscle crossing the DIP joint.

IV. DISCUSSION

The biomechanical model in this study employed an anatomically realistic force-based approach, which is able to describe all the potential force transmissions and interactions among tendons, bones and joints. The model was validated for isometric force production with each muscle across joint postures through comparison with measurements from cadaveric experiments. The model convincingly predicted the effect of the three joint angles on fingertip force direction and magnitude. Successful prediction of isometric force produced on the fingertip supports the reliability in describing the musculoskeletal mechanics within the index finger.

With the model simulation of isometric force production, the joint contact force is revealed to be related to finger posture, even though the tendon forces remain constant. As we know that high joint contact force could lead to the progressive destruction of the joint cartilage, the patient with OA may benefit from avoiding the finger postures that create higher joint contact force, such as MCP flexion. For example, palmar pinch could be performed with substantial PIP flexion and minimal MCP flexion, rather than vice versa.

It should be noted that the model was validated with the cadaveric experiment only for loading of individual tendons. The model assumed that the effects of multiple tendons combine linearly to generate the total fingertip force output. Although this has been shown to be true under some conditions [12], additional validation experiments of multi-tendon loading patterns are expected to support the reliability of the prediction. Similarly, the reliability of the model in dynamic simulation also needs to be validated by corresponding experiments.

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