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Subject-Specific Finite Element Analysis of the Carpal Tunnel Cross-Sectional to Examine Tunnel Area Changes in Response to Carpal Arch Loading

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Abstract

Background—Manipulating the carpal arch width (i.e. distance between hamate and trapezium bones) has been suggested as a means to increase carpal tunnel cross-sectional area and alleviate median nerve compression. The purpose of this study was to develop a finite element model of the carpal tunnel and to determine an optimal force direction to maximize area.

Methods—A planar geometric model of carpal bones at hamate level was reconstructed from MRI with inter-carpal joint spaces filled with a linear elastic surrogate tissue. Experimental data with discrete carpal tunnel pressures (50, 100, 150, and 200 mmHg) and corresponding carpal bone movements were used to obtain material property of surrogate tissue by inverse finite element analysis. The resulting model was used to simulate changes of carpal arch widths and areas with directional variations of a unit force applied at the hook of hamate.

Findings—Inverse finite element model predicted the experimental area data within 1.5% error. Simulation of force applications showed that carpal arch width and area were dependent on the direction of force application, and minimal arch width and maximal area occurred at 138° (i.e. volar-radial direction) with respect to the hamate-to-trapezium axis. At this force direction, the width changed to 24.4 mm from its initial 25.1 mm (3% decrease), and the area changed to 301.6 mm² from 290.3 mm² (4% increase).

Interpretation—The findings of the current study guide biomechanical manipulation to gain tunnel area increase, potentially helping reduce carpal tunnel pressure and relieve symptoms of compression median neuropathy.

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Keywords

Carpal Tunnel; Cross-sectional Area; Mechanics

1. Introduction

The carpal tunnel is a fibro-osseous structure comprised of the transverse carpal ligament (TCL) at the volar side and the carpal bones at the dorsal, radial, and ulnar sides. The tunnel has an elliptical cross section extending approximately 20 mm in the proximal-distal direction (Cobb et al., 1993; Gabra et al., 2015; Pacek et al., 2010). The nine flexor tendons and median nerve pass through the confined carpal tunnel, making the nerve prone to compression neuropathy, i.e. carpal tunnel syndrome. While surgical treatment of carpal tunnel syndrome may effectively relieve symptoms, the invasive approach results in a number of complications (e.g. scar tissue, pillar pain, and weakness), making non-surgical alternatives desirable.

The biomechanical properties of the carpal tunnel have been investigated with respect to the carpal arch width (CAW). It has been shown that the carpal tunnel has a decreased CAW during wrist flexion and extension in comparison to the neutral wrist position (Garcia-Elias et al., 1992) and an increased CAW after carpal tunnel release surgery (Garcia-Elias et al., 1992; Gartsman et al., 1986). This CAW change is attributable to the mobility of carpal bones which is guided by bone/cartilage configuration and inter-carpal ligaments (Gabra et al., 2012; Xiu et al., 2010). Attempts to widen (Sucher et al., 2005) or narrow (Li et al., 2013; Marquardt et al., 2016; Marquardt et al., 2015) the CAW have been made to alter the carpal tunnel morphological structure as a means to increase the carpal tunnel cross-sectional area (CSA) for median nerve decompression. Widening the CAW can be limited due to the mechanical constraint imposed by the thick band of the transverse carpal ligament. Narrowing of the CAW has been examined by applying forces to the carpal bones in cadaveric hands (Gabra and Li, 2016) and to the radio-ulnar aspects of the wrist in human subjects (Marquardt et al., 2016; Marquardt et al., 2015). Decreasing the CAW at the distal row was demonstrated to increase the volar arch area of the carpal tunnel (Fig. 1), i.e., the area formed by the TCL divided by a reference line from the ridge of the trapezium and hook of the hamate (Li et al., 2013; Li et al., 2011; Li et al., 2009; Marquardt et al., 2016; Marquardt et al., 2015). However, previous studies were limited to the examination of effects of CAW on volar area without the inclusion of the dorsal arch area formed by carpal bones. In addition, the forces applied to the carpal tunnel in previous studies were prescribed along the CAW, and the optimal direction of force application to achieve maximum volar area increase remains unknown, as does the associated total changes in the carpal tunnel area.

Computational modeling is an effective tool to perform parametric analysis of multiple factors that might be challenging in an experimental study. Finite element (FE) representations, representative of patient-specific anatomy, have been utilized to investigate mechanics of different tissues (Martin et al., 2015; Sun et al., 2015). Specific to the wrist, FE analysis has been used to investigate the load transmission through the carpal bones (Carrigan et al., 2003; Gislason et al., 2009), the effects of pathological conditions on wrist

joint (Bajuri et al., 2012), wrist mechanics in response to tendon loading (Majors and Wayne, 2011), and changes in carpal load transmission associated with carpal tunnel release (Guo et al., 2009). FE modeling was also used to analyze stress of the carpal tunnel contents and showed that median nerve compression is attributable to structural contact force rather than fluid pressure (Ko and Brown, 2007; Mouzakis et al., 2014). The FE model of the carpal tunnel presented in the current study permits the exploration of different force conditions applied between the trapezium and hamate bone to evaluate the change in CSA using individualized model. This computational approach allows for virtual testing of models with subject-specific bone geometries, where the influence of any force direction can be evaluated to determine desired outcome parameter, e.g. carpal tunnel area. Nonetheless, a comprehensive understanding of the specimen-specific biomechanics of carpal tunnel morphology depends not only on individualized anatomy but also specimen-specific material properties of tissue representations, which in turn dictate the emergent response of the CAW and carpal tunnel CSA changes when subjected to an intervention.

The overall goal of this study was to develop a simplified, specimen-specific model of the wrist to evaluate the role of varying force direction application between the hamate and trapezium bone on CAW and CSA of the distal carpal tunnel. The distal region of the carpal tunnel was selected for this study because it is the narrowest region in the carpal tunnel (Cobb et al., 1993). The first aim was to establish the FE model by determining the individualized material properties of tissue representations using previously collected pressure and bone kinematics data (Li et al., 2011). It was hypothesized that the individualized FE model would agree with the experimental data from the same specimen within 5% error in CSA correspondence. The second aim was to conduct clinically relevant simulations of mechanical manipulation of CAW and carpal tunnel area. Specifically, the optimal force direction, i.e., the force direction which maximized carpal tunnel CSA, was determined by quantifying CSA change associated with varying force directions. It was hypothesized that the amount of CSA change would be dependent on the orientation of the applied force, increasing area with inward force and decreasing area with outward force.

2. Methods

The overall strategy to achieve the goals of the study relied on finite element analysis of the distal carpal tunnel to predict the CAW and carpal tunnel CSA under various loading and boundary conditions. Technical steps included the model development, inverse finite element analysis to obtain specimen-specific tissue properties, and clinically relevant simulations to understand the influence of the external force orientation on the CAW and carpal tunnel CSA. Data from a previously conducted experimentation on pressure-area relationship were utilized for development of the model and for inverse finite element analysis. All simulations were conducted using Abaqus CAE (v6.10, Simulia, Rhode Island, USA).

The specimen used for finite element analysis was a left wrist from a 42 years old female donor. Experimental data were collected in a previous study (Li et al., 2011). The specimen was a representative closer to the mean CAW and CSA of all. In the experiment, a balloon was inserted in the tunnel and pressurized at 50, 100, 150, and 200 mmHg. For each pressure level, including the unloaded state of 0 mmHg, magnetic resonance images (MRI) of the

carpal tunnel cross section were collected. An extremity MRI system with a strength of 1 Tesla (OrthOne, ONI Medical Systems Inc., MA, USA) was used to obtain carpal tunnel cross sections with a slice thickness of 1 mm and resolution of 3.4 pixels per mm. Axial images were obtained by 3D gradient echo (TR = 30 ms, TE = 8.9 ms, flip angle = 358) with a $150 \times 150 \text{ mm}^2$ field of view and 260×192 matrix. The protocol was adjusted such that (1) the axial imaging plane was obliquely oriented at 13° so that the slice plane was perpendicular to the dorsal side of the carpal tunnel (the sagittal scout images were used to determine the angle of oblique orientation for specimen); and (2) the number of image slices was set 32 for the specimen based on the measured length of the carpal tunnel during dissection. The data were processed to characterize the movement of the carpal bones relative to the trapezium as a function of balloon pressure. The coordinates of two points on each bone were extracted for the modeling purpose.

A two-dimensional plane strain model of the distal carpal tunnel cross section with a unit thickness was developed and utilized for this study. The cross-sectional MRI for the unloaded state (0 mmHg) at the level of the hook of the hamate was used to define a reference geometrical configuration of the transverse carpal ligament and carpal bones (the trapezium, trapezoid, capitate, and hamate; Fig. 1). Manual segmentation of the bones and ligament was performed using the polygon selection tool in ImageJ (v1.43, National Institutes of Health, USA). The mechanics of bones are dictated by the combined mechanical response of cartilage and the surrounding intercarpal ligaments. In the model, the cartilage and the intercarpal ligaments were not modeled explicitly. Instead, the joint spaces between bones were filled with a surrogate tissue to represent the combined mechanical constraints of these structures, both in tension and compression. Then the geometries of the bones and surrogate tissue were reconstructed using Rhinoceros software (v.5, Robert McNeel & Associates, USA) (Fig. 2) and were imported into Abaqus CAE to create the FE model. The geometrical components were meshed with element type C3D8R; the mesh density was updated by conducting preliminary simulations to ensure that CAW predictions did not change by more than 0.5% with increase in the mesh size. The carpal bones were modeled as a co-rotational linear material with Young's modulus and Poisson's ratio of 10,000 MPa and 0.33, respectively (Guo et al., 2009). The surrogate tissue was modeled as a co-rotational linear material with unknown material properties (Young's modulus, E and Poisson's ratio, ν) which were obtained by the inverse finite element analysis described below. A non-linear finite element analysis was performed, which incorporated co-rotational formulation of linearly elastic materials to accommodate for large rotations. Bone and surrogate tissue were connected to each other using a tie constraint (gluing to two surfaces). The trapezium was fixed with its node displacements constrained (Fig. 2). The TCL was not implemented as an explicit structure in the model, but its mechanical function was included through loading boundary conditions corresponding to various carpal tunnel pressures.

The model was first used in an inverse finite element analysis procedure to obtain specimen-specific material properties (E and ν) of the surrogate tissue. The model's loading and boundary conditions were updated with a quasi-static incremental pressure change in the carpal tunnel space. The loading and boundary conditions were used to describe forces and movements acting on the model. Based on the previously obtained experimental data (Li et

al., 2011), four balloon pressures (50, 100, 150, 200 mmHg) were simulated by applying uniform pressure to all internal nodes of the bones and surrogate tissue (Fig. 2A). An additional point force represented the influence of TCL tensioning. The point force was directed along the approximate orientation of the TCL, as obtained from MRI and was applied at the TCL insertion site on the hamate bone based on the assumption that the TCL acts as a tension only spring. The TCL tension was calculated as a function of pressure relying on its approximation as the section of a thin walled pressurized cylinder (Zhang et al., 2011). This assumption allowed the calculation of TCL tension as the multiplication of carpal tunnel pressure and the TCL radius of curvature derived from fitting a circle of the TCL arc. TCL force applied to the model was calculated by scaling of TCL tension relying on the assumption of total TCL out-of-plane width and its corresponding footprint on the hamate. The TCL's out-of-plane width was assumed to be 20 mm, covering both proximal and distal bones (Cobb et al., 1993). It was also assumed that the TCL load would be distributed equally at both distal and proximal bones, i.e. 10 mm of the tunnel's length. The insertion length at the hook of hamate level was 7.74 mm (measured from MRI). This provided the ratio of the half of TCL's out-of-plane width and hamate insertion length as 1.3, which was then used to scale TCL tension to obtain the TCL force applied to the model with unit out of plane thickness.

The inverse finite element analysis was carried out by iteratively changing material properties (E and ν) of the surrogate tissue to match the model prediction of bone motion (calculated at pressure levels of 50, 100, 150, and 200 mmHg) to those measured experimentally on the same specimen. Experimental bone motion at different pressure levels was obtained by manually matching the geometry of template bones (acquired at unloaded reference state of 0 mmHg) to bones at different pressure levels using Rhinoceros. Bone motion was represented by the movement of two bone markers on the template bones while coordinates of the trapezium were fixed. To determine the material property values an optimization procedure was implemented that updated surrogate tissue properties by minimizing the objective function (f), which was defined as summed squares of the differences in markers coordinates obtained from inverse FEA model and from experiment at each pressure level (Eq. 1).

$$f = \sum_{P=1}^4 \sum_{n=1}^6 [(X_{P,n}^e - X_{P,n}^m)^2 + (Y_{P,n}^e - Y_{P,n}^m)^2] \quad \text{Eq. 1}$$

Where X^e and Y^e are the x and y coordinates of the bony landmarks obtained from the experimental study (Li et al., 2011), and X^m and Y^m are the model-predicted x and y coordinates of the same bony landmarks. The subscripts P and n denote a pressure level (i.e. 50, 100, 150, or 200 mmHg) and a bony landmark, respectively.

The optimization was implemented using *Python*'s SciPy library with differential evolution algorithm for a global optimization routine (*Python* v2.7, SciPy v0.15.1, Anaconda 2.4). Bounds for E and ν of the surrogate tissue were set as 0.5–100 MPa and 0–0.49, respectively. After the optimal values of the material properties for the surrogate tissue were determined, the total CSA values for the model were calculated and were compared with the

experimental data. The total carpal tunnel CSA was the sum of the dorsal area and the volar area. Coordinates from the node set representing the bone boundary were used to compute dorsal area under the CAW width line. Calculations were performed using a custom *MATLAB* program (MathWorks, Natick, MA). As the TCL was not explicitly modeled in this study, the volar area was calculated using a geometric model driven by CAW. The geometric model from our previous study (Li et al., 2009) was adapted and assumed a constant TCL length (26.1 mm) where its deformation was approximated by formation of the arc of a circular segment with the CAW as the chord length. The volar area was defined as the area between the TCL arc and the chord between the ridge of trapezium and hook of hamate (i.e. the CAW, Fig 1.).

The established FE model was also used to conduct clinically relevant simulations. The goal of these simulations was to evaluate the CSA changes associated with applying a manipulation condition consisting of an external force (F) of 1 N at the hook of hamate (Fig. 2B). The 1 N force for this planar model (with a unit thickness) was chosen to represent a 20 N force applied across the whole tunnel out-of-plane thickness of approximately 20 mm (Cobb et al., 1993). The 1 N point force (F) was applied at the hamate in varying directions, counterclockwise from 0° to 359° in increments of 1° , with respect to the CAW axis (Fig. 2B). This force was a follower force as it moved along with the bone. Simulations were analyzed for 360 force directions to determine the optimal direction of force application that resulted in the maximum increase of the carpal tunnel CSA.

3. Results

The inverse finite element analysis revealed material properties of the surrogate tissue for the specimen as $E = 21.6$ MPa and $\nu = 0.43$, respectively. The minimum value of the objective function was 5.341. The global optimization procedures converged to a consistent solution. When the predictions of the model (with optimized material properties) were compared against the experimental CSA, a similar trend as a function of pressure level was observed (Fig. 3). The model-predicted CSA values were within 1.5% error of the experimental data.

Clinically relevant simulations to understand the influence of the directionality of an external force revealed that the CAW decreased between angles of 51° and 231° as compared to the reference value (Fig. 4). The reference value of CAW was 25.1 mm as calculated for the unloaded state of 0 mmHg. The manipulation force directed at radial-volar direction (138°) led to a maximum shortening of CAW, which was found to be 24.4 mm (a 3% reduction from the reference value). At all other angles, the CAW increased with a maximum value of 25.9 mm at 321° .

The model simulation predicted the dorsal area as 238.1 mm^2 for the unloaded tunnel. The volar area increased for force orientations between 51° and 231° (Fig. 5a). The reference volar area at no force application was 52.2 mm^2 . The maximum volar area (66.9 mm^2) was achieved at 138° , which was 28% increase from the reference area value. During widening of CAW, the volar area decreased (Fig. 5A). In contrast to the change of volar area, the dorsal area decreased for force direction between 51° and 231° . The decrease in bony area at 138° (radial-volar direction) was found to be 3.3 mm^2 . An increase in dorsal area occurred at

remaining angular directions with a maximum increase of 3.3 mm² at 321° (Fig. 5B). The total CSA increased for force orientations between 51° and 231° (Fig. 5C). The reference CSA was found to be 290.3 mm² without force application. A maximum total CSA increase of 11.3 mm² occurred at an angle of 138° which had a value of 301.6 mm² (a 4% increase from baseline). Sample simulation results of the carpal tunnel areas at 138° and 321° are shown in Fig. 6.

4. Discussion

This study presented a planar FE model of the distal carpal tunnel with individualized material properties of the surrogate tissue at the joint spaces between carpal bones. It was observed that the model predicted a maximum increase in total CSA of the carpal tunnel with the force applied in the radial-volar quadrant, specifically at 138° direction (Fig 2.). The individualized analysis also identified that an external force application oriented between 51° and 231° resulted in narrowing of CAW and therefore increase in tunnel CSA, albeit with variable effectiveness. The increase in total CSA was predominantly attributed to an increase in volar area as the change in the dorsal area was relatively small (~ 1%).

Inverse finite element analysis allowed the determination of specimen-specific properties to obtain a virtual specimen model for identifying manipulation strategies that is most beneficial to the specimen. In this study, the individualized material properties of the surrogate tissue for the FE model were determined by utilizing the pressure and area relationship observed during a previous experimental study that described the bone movement (Li et al., 2011). The area predicted by the model agreed well with those obtained from experimental data at each pressure condition (error less than 1.5%). In future, it may be desirable to explore alternative, non-invasive experimentation to acquire bone kinematics and/or CAW under known loading to develop patient-specific models. Such experimental data and derived models will further realize the potential for clinical application of the modeling and simulation workflow.

During the simulation of force directions applied at the hook of the hamate, half of the orientations were associated with the narrowing of the CAW and other half orientations led to the widening of the CAW. CAW values during widening always remained lower than the length of the TCL itself, indicating the TCL did not assume a tensioned state. Narrowing of CAW has been shown to increase the volar area of the carpal tunnel in previous biomechanical studies (Li et al., 2013; Marquardt et al., 2016; Marquardt et al., 2015). The reduction in CAW was maximized in radial volar direction (at 138° from 25.1 to 24.4 mm). The increases in volar area and total CSA with force directed at radial-volar direction was 14.7 and 11.4 mm², respectively. These results are consistent with a previous in-vivo study which showed that 10 N of transverse compressive force results in a volar area increase of 5.65 mm² (Marquardt et al., 2015). Similarly, another *in vivo* study evaluating the pinching of the index finger against thumb reported a decrease of 0.8 mm in CAW and an increase of 5.1 mm² in volar area (Shen and Li, 2013). Several other studies also have shown that an increase in volar area results from the narrowing of the CAW (Li et al., 2013; Li et al., 2011; Li et al., 2009; Marquardt et al., 2016; Marquardt et al., 2015). The current FE model demonstrated that changes in the dorsal area were smaller than those of volar area,

indicating that during narrowing the formation of the ligament arch contributes to an increase in the overall area of the tunnel. Nonetheless, the model of the bony arch (dorsal area) provides the prediction of the CAW that drives the calculation of volar area. Moreover, it has also been shown in a previous study that during transverse compression of the wrist the median nerve shape became rounder and is displaced in a radial-volar direction (Marquardt et al., 2015). The observed increase in volar and total area supports the utilization of biomechanical manipulation to help alleviate the compression of median nerve.

Increases in CAW and the dorsal area were observed with an outward force and the amount of increase varied with the direction of force application. The maximal CAW increase (from 25.1 to 25.9 mm) occurred with a force application at 310°. This change was similar to those reported in the previous biomechanical studies (Sucher et al., 2005; Xiu et al., 2010). The largest decrease of 22.1 mm² (8%) in the total CSA was found at the force direction of 310° (dorsally directed). This decrease in total CSA corresponded to the decrease of 25.3 mm² in the volar area and a small increase of 3.2 mm² in the dorsal area. These findings demonstrate that during the CAW widening, a small increase in the bony area is overshadowed by the decrease in volar area (Fig. 6). This can be explained by the collapse of the ligament arch associated with CAW widening, resulting in a negative change in volar area. In addition, a previous study (Li et al., 2009) showed that the ligament did not become elongated even with a 200 N load applied volarly from within the carpal tunnel, signifying the ligament arch tends to flatten and does not gain area increase from an outward force. In general, the area of the bony portion of the carpal tunnel was relatively insensitive to force application and CAW variation, either widening or narrowing.

One of the limitations of this study is that the FE model includes a surrogate tissue includes a surrogate tissue which serves a substitution for the cartilage and intercarpal ligaments. This surrogate tissue represented the combined mechanical effects of the cartilage and ligaments and it permitted the parametric analysis of force direction to evaluate changes in the area of the carpal tunnel in a mechanically equivalent structure, which is computationally efficient to obtain correct bone kinematics. A second limitation of this study is that a two-dimensional FE model focusing at the distal level of the carpal tunnel was utilized. Future work can expand the modeling framework developed in this study to be 3-D for determining individualized and location dependent material properties and conduct clinically relevant simulations with improved realism. For example, forces distributed in a larger area (rather than a point force) on the radial and ulnar sides of the wrist can be simulated to improve clinical utilization of the model. Third, the volar area was calculated using a geometric model due to the absence of the TCL in the model. The geometric model assumes the shape of the ligament arch to be a sector of an arc, which well approximates the carpal arch. Any error associated with this assumption is expected to be systematic, which still allow for examining the relative changes in area in relation to CAW variations. Fourth, the results obtained in this study are specific to the tested specimen, and generalization of the results requires population-based modeling and simulation with more specimens. Furthermore, the current modeling methods and results provide a basis for future studies to understand the sensitivity of CSA to modeling parameters such as material properties of the involved tissues and geometrical configuration of the carpal tunnel.

5. Conclusions

In conclusion, this study utilized the compliance characteristics of the carpal tunnel to develop an individualized model, confirming its predictive capacity, and explored performance of carpal tunnel manipulation strategies. An application of narrowing force directed towards the volar-radial direction resulted in the maximum increase in CSA. Further studies are needed to fully understand how force application affects the area of the carpal tunnel at different levels along the longitudinal axis of the wrist.

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Highlights

1. Carpal tunnel compliance allows for geometrical change with biomechanical manipulation.
2. Narrowing of the carpal arch width results in an increase in carpal tunnel area.
3. Patient-specific finite element modeling predicted optimal force direction to achieve maximum area increase.
4. Increase in overall area of the carpal tunnel potentially relieves median nerve compression.

Fig. 1.

Segmentation of the transverse carpal ligament (TCL) and four distal carpal bones from a magnetic resonance image. The dashed line along the CAW divides the total carpal tunnel into volar and dorsal areas. Tm = Trapezium, Td = Trapezoid, C= Capitate, H = Hamate, and TCL = transverse carpal ligament.

Fig. 2.
The planar finite element model of the distal carpal tunnel for inverse finite element analysis (A) and simulation (B). The carpal bones are dark shaded, and surrogate tissue filling in the joint space is light shaded. The optimal force (F) was determined to be at 138°, which resulted in maximum increase in CSA.

Fig. 3.
Total carpal tunnel cross-sectional areas (CSAs) predicted by the finite element model and experimental data.

Fig. 4.
Simulated changes of CAW as a function of force direction. The horizontal dashed line shows the baseline CAW value at the unloaded reference state.

Fig. 5.

The volar area as a function of force direction. The maximum area occurred at 138° (A), the dorsal area as a function of force direction (B), and the total carpal tunnel cross-sectional area (CSA) as a function of force direction. The maximum area occurred at 138° (C). The horizontal dashed line shows the baseline value at the unloaded reference state.

Fig 6.
Carpal tunnel CSA with unloaded state (solid line) and force applications at 138° (dotted line) and 321° (double dashed line).