Internal Mechanics of a Subject-Specific Wrist in the Sagittal versus Dart-Throwing Motion Plane in Adult and Elder Models: Finite Element Analyses

Vered Mahpari 1, Yafa Levanon 2, Yael Kaufman-Cohen 2, Meital Zilberman 1* and Sigal Portnoy 2,*

Abstract: Introduction: Most of the wrist motions occur in a diagonal plane of motion, termed the dart-throwing motion (DTM) plane; it is thought to be more stable compared with movement in the sagittal plane. However, the effect of the altered carpus motion during DTM on the stress distribution at the radiocarpal joint has yet to be explored. Aim: To calculate and compare the stresses between the radius and two carpal bones (the scaphoid and the lunate) in two wrist positions, extension and radial extension (position in DTM), and between an adult and an elder model. Methods: A healthy wrist of a 40-year-old female was scanned using Magnetic Resonance Imaging in two wrist positions (extension, radial extension). The scans were transformed into three-dimensional models and meshed. Finite element (FE) analyses in each position of the wrist were conducted for both adult and elder models, which were differentiated by the mechanical properties of the ligaments. The distal surfaces of the carpal bones articulating with the metacarpals were loaded by physically accurate tendon forces for each wrist position. Results: The von Mises, shear stresses and contact stresses were higher in the extension model compared with the radial-extension model and were higher for the radius-scaphoid interface in the adult model compared with the elder model. In the radius-scaphoid interface, the stress differences between the two wrist positions were smaller in the elder model (11.5% to 22.5%) compared with the adult model (33.6–41.5%). During radial extension, the contact area at the radius-lunate interface was increased, more so in the adult model (222.2%) compared with the elder model (127.9%), while the contact area at the radius-scaphoid was not affected by the position of the wrist in the adult model (100.9%) but decreased in the elder model (50.2%) during radial extension. Conclusion: The reduced stresses during radial extension might provide an explanation to our frequent use of this movement pattern, as the reduced stresses decrease the risk of overuse injury. Our results suggest that this conclusion is relevant to both adults and elder individuals.

Keywords: aging; radiocarpal stresses; wrist injury

1. Background

Most of the wrist motions occur in a diagonal plane of motion, termed the dart-throwing motion (DTM) plane [1], i.e., a wrist plane that ranges from radial extension to ulnar flexion. During DTM, the proximal carpal row undergoes minor sagittal translations and is therefore thought to be more stable [2]. This plane of motion is often exercised during daily activities performed at different heights [3] by both dominant and non-dominant upper limbs [4]. Since the DTM is our natural wrist motion, occupational therapists strive to incorporate it in the rehabilitation of the wrist [5–7]. However, although their interest in DTM has grown in the last decade and carpal kinematics have been documented, the stress distribution at the radiocarpal joint has yet to be explored, and
The biomechanics of the wrist was previously explored using finite element (FE) analyses to compare stresses during wrist flexion, extension, radial and ulnar deviation [10], for different purposes, e.g., to explain the mechanism of wrist injury [11] and to test the effects of surgical reconstruction techniques [12,13]. These studies showed that the load distribution changed according to the posture of the wrist; however, no data exist regarding loads during DTM.

The aim of this study was to calculate and compare the stresses between the radius and two carpal bones (the scaphoid and the lunate) in two wrist positions, extension and radial extension (position in DTM), and between an adult and an elder model.

2. Methods

Our work included four FE analyses of a healthy human wrist in two different positions of the wrist: sagittal plane (extension) and DTM plane (radial extension). The geometry of the wrist was obtained using magnetic resonance imaging (MRI) in the aforementioned wrist positions. The FE models included the distal parts of the radius and ulna, the carpal bones, cartilage, and main ligaments. Each position was analyzed once with ligament mechanical properties of a healthy adult and again with properties pertaining to age-related decline of structural properties. The stresses in the articulating surfaces of the radius bone and the two bones in the first carpal row, the scaphoid and lunate, were calculated.

2.1. Wrist Geometry

A Helsinki Committee approval was obtained pretrial (approval number 2943-16-SMC). The geometry of the right wrist of a healthy 40-year-old female was obtained using a 3T MRI (VB17A Magnetom Trio, a Total Imaging Matrix System; Siemens Healthcare, Erlangen, Germany). The distal forearm of the subject was fixed using a Three-Dimensional (3D) printed ‘arm constrainer’ (Figure 1). The arm was placed between the vertical walls, which were tied using Velcro straps in order to maintain the arm in mid-position and prevent elbow pronation and supination. The wrist was scanned twice: in extension and maintaining the same extension but with added radial deviation. The scan during wrist extension included 20 images with a slice distance of 3.3 mm (total length of 66 mm) and transverse resolution of 256 × 256. The scan during radial extension included 24 images with a slice distance of 3.3 mm (total length of 79.2 mm) and transverse resolution of 320 × 320. The scanning properties were chosen so that both wrist scans included the distal ends of the radius and the ulna, eight carpal bones and five proximal ends of the metacarpal bones.

Figure 1. The arm fixation apparatus. The distance between the walls was adjusted according to the width of the forearm and secured using two Velcro straps.
2.2. Finite Element Modeling and Analysis

2.2.1. Modeling and Meshing

The MRI scans were used to create a Solid 3D model using ScanIP software (Simpleware Ltd., Exeter, UK). The bones were identified using a Radiology Atlas in each slice [14]. Then, the pixels of each bone were identified in each slice (‘masking technique’) and each bone was formed into 3D geometry by layering up the masks. Touching surfaces were smoothed by 1 mm to allow a minimum gap for cartilage modeling. In order to minimize noise, a Recursive Gaussian filter was used, which smoothed the layers. The geometries of the bones were then meshed into 3D tetrahedral elements (Tetra 10).

The 3D model was imported into Patran 2018 software (MSC Software Corporation, CA, USA). The cartilage was modeled using wedge elements, as suggested by Gislason and Nash in the modeling of wrist cartilage [15]. It was extruded using half of the minimum distance of the articulating surfaces [16] (The cartilage thickness between the scaphoid and the radius in the radial-extension model was smaller than half of the distance in some regions due to convexities). Contact was set to surface-to-surface (touch contact). The ligaments were modeled as multi-parallel linear springs (8 CELAS elements), based on similar wrist FE studies [16,17]. The two FE models are shown in Figure 2. The total number of elements in the extension model was 770,682 and 206,645 nodes. The total number of elements in the radial-extension model was 836,842 and 221,408 nodes.

Figure 2. Finite element (FE) model of the right wrist of a healthy 40-year-old woman. Volar view of the carpal bones, cartilage elements at the articulations and ligaments in wrist posture of (a) extension and (b) radial extension. Different colors are applied to the different components in order to allow better visual differentiation. Moreover, (c) the loads and constrains on the model.

2.2.2. Mechanical Properties

The bones were modeled as a solid bulk, with linear isotropic mechanical properties [13,18]. Specifically, Young’s modulus of the bone was defined as $E = 18\text{GPa}$ with a Poisson’s ratio of $\nu = 0.2$ [15–17,19]. Although the cortical bone undergoes biomechanical
changes with age, studies showed a negligible decrease in Young’s modulus value \[20\]. Hence, we did not apply change in Young’s modulus value in the elder model.

The cartilage was defined with hyperplastic properties since it undergoes large deformations \[12,17,19,21\], using a two-parameter Mooney–Rivlin model \[22\], with the following coefficients: C10 = 4.1 MPa and C01 = 0.41 MPa. The Mooney–Rivlin strain energy density potential equation is:

\[
W = C_{10}(I_1 - 3) + C_{01}(I_2 - 3)
\]  

where \( \mu = 2(C_{10} + C_{01}) \), \( \mu \) is the material shear modulus \[23\] and the Green–Lagrange strain tensor is \( E = \frac{1}{2}(C - I) \), where \( I \) is the second order identity tensor \[23\]. The principal invariants of \( C \) are calculated by:

\[
I_1(C) = \text{trace}(C), \quad I_2(C) = \frac{1}{2} \left( (I_1(C))^2 - \text{trace}(C^2) \right)
\]

Table A1 (see Appendix A) lists the ligament’s stiffness properties as well as their origin of insertion. The ligament’s insertions were determined using an anatomic visualization software (McGrouther DA HP, Interactive Hand-Anatomy, Primal Pictures). The stiffness of the proximal scapho-lunate interosseous ligament (SLIL) was assumed to be similar to that of the scaphoid-trapezium-trapezoid ligament because there are no available data regarding its stiffness and thickness reported as variables \[24\]. Ligament decline in stiffness in the elder model was defined using a 20% stiffness reduction (compared with the adult model) \[9\]. The value presented in the Table for the transverse carpal ligament (TCL) refers to the stiffness of elder females \[25\]. We did not alter the cartilage thickness in the elder model since there is no clear documentation of cartilage thickness changes in the wrist with age, e.g., as shown by \[26\]. Furthermore, we did not alter the applied forces to the tendons because we simulated two simple and natural wrist positions, with no external forces applied to the hand. Although there is a documented decline of hand-grip force with age \[27\], the movements chosen for this study require a small percentage of the maximal muscle force, so that both an elderly individual and younger populations could produce these forces.

### 2.2.3. Loading Conditions

Werner et al. \[28\] determined the average tendon forces based on tests performed using 12 cadaver arms, set to different wrist positions. The relevant tendon forces were used as a database for our FE analyses for the following tendons: extensor carpi ulnaris (ECU), extensor carpi radialis brevis (ECRB), extensor carpi radialis longus (ECRL), abductor pollicis longus (APL), flexor carpi radialis (FCR), and flexor carpi ulnaris (FCU).

The forces (Table 1) were applied to the distal surfaces of the carpal bones articulating with the metacarpals (Figure 2c). The tendons’ loads were distributed on the surfaces of the bones. Each force was applied in the direction of a vector from the insertion point in the metacarpal relevant bone to the bone in the distal carpal row. The tendon forces reflect a wrist position of 20° in extension. Unfortunately, the radial deviation angle during the radial-extension position studied in \[28\] was not measured.

### Table 1. Forces applied to the tendons of the two models during the two conditions of extension and radial extension \[28\]. Note that the total force applied for each wrist position is 267 N.

| Wrist position | ECU [N] | ECRB [N] | ECRL [N] | APL [N] | FCR [N] | FCU [N] |
|----------------|---------|----------|----------|---------|---------|---------|
| Radial-Extension | 58.2 | 50.5 | 38.4 | 38.2 | 16.0 | 65.7 |
| 20° Extension | 54.2 | 44.1 | 31.8 | 37.6 | 12.5 | 86.8 |

ECU = Extensor Carpi Ulnaris  
ECRB = Extensor Carpi Radialis Brevis  
ECRL = Extensor Carpi Radialis Longus  
APL = Abductor Pollicis Longus  
FCR = Flexor Carpi Radialis  
FCU = Extensor Carpi Radialis Longus
2.2.4. Boundary Conditions

The proximal ends of the radius and the ulna were constrained for all translations and rotations (Figure 2c). The contact was defined as deformable-to-deformable with no friction [10,17].

2.2.5. FE Solver and Parameters

The analyses were processed by MSC Nastran software (MSC Software Corporation, Irvine, CA, USA). Implicit non-linear solver (Pure Newton Raphson) was used, and the convergence criterion was set to load equilibrium error. Separation of contacting surfaces was based on absolute stress (extrapolating integration point stresses).

The average stresses were calculated by dividing the summation of vector forces (in each grid point of the contacting surface) by the contact areas. Minor negative forces (tension) in the peripheral of the contact areas were ignored and referred to as side effects caused by mid-side nodes of the contact patch.

3. Results

The bone displacements and stresses are presented below. The forces in the ligaments relevant to the scaphoid, lunate, and radius are depicted in Appendix B.

3.1. Displacements

The displacements of the proximal carpal row are depicted in Figure 3.

Figure 3. Displacement [mm] plots in superior view of the scaphoid, lunate and the radius for the adult (left frames) and elder (right frames): extension (upper frames), and radial extension (bottom frames). The purple colored wireframes show the initial position of the wrist bones.

The pitch angles (rotation in the sagittal plane) between the lunate and scaphoid were 27.0° for the extension model and 17.2° for the for radial-extension model.

3.2. Stresses

The von Mises and maximum shear stresses in the cartilage of the radius during extension and radial extension are depicted in Figure 4. Moreover, the maximal von Mises stresses, maximal shear stresses, maximal and average contact stresses and contact area in extension and radial-extension models in the adult and elder simulations are presented in Table 2.
Table 2. Maximal von Mises stresses, maximal shear stresses, maximal and average contact stresses in the adult and elder simulations. The percent of variable in the radial-extension model in relation to the extension model is calculated and presented as “% of Extension”.

|                  | Adult Model | Elder Model |
|------------------|-------------|-------------|
| Maximal von Mises stress [MPa] | 19.9 | 3.8 |
| Maximal shear stress [MPa] | 11.2 | 2.1 |
| Maximal contact stress [MPa] | 29.7 | 5.9 |
| Average contact stress [MPa] | 8.4 | 1.0 |
| Contact area [mm²] | 6.4 | 3.7 |

The von Mises, shear stresses and contact stresses were higher in the extension model compared with the radial-extension model and were higher for the radius-scaphoid interface in the adult model compared with the elder model (Table 2, Figure 4). In the radius-lunate interface, the stress differences between the two wrist positions were smaller in the elder model compared with the adult model. During radial extension, the contact area at the radius-lunate interface was increased, more so in the adult model compared with the elder model. The contact area at the radius-scaphoid was not affected by the position of the wrist in the adult model, but decreased in the elder model during radial extension.

4. Discussions

In this study, we aimed to compare the stresses between the radius and two carpal bones (the scaphoid and the lunate) in two wrist positions: extension and radial extension (position in DTM) and between an adult and elder model. Our finding shows that the von Mises, shear stresses and contact stresses were higher in the extension model compared with the radial-extension model and were higher for the radius-scaphoid interface in the adult model compared with the elder model. In the radius-scaphoid interface, the stress differences between the two wrist positions were smaller in the elder model (11.5% to
22.5%) compared with the adult model (33.6–41.5%). During radial extension, the contact area at the radius-lunate interface was increased, more so in the adult model (222.2%) compared with the elder model (127.9%), while the contact area at the radius-scaphoid was not affected by the position of the wrist in the adult model (100.9%) but decreased in the elder model (50.2%) during radial extension.

The altered stress distribution at the radiocarpal interface during radial extension compared with extension most likely originates from the differences in loading dynamics. As depicted in Table 1, approximately 24% of the force applied by the FCU during extension is divided between the other muscles during radial extension. Consequently, kinematics of the carpal bones is altered, as previously reported by [29], which showed that during flexion-extension, the scaphoid contributes mostly in the radiocarpal, whereas during DTM, it contributes almost equally in the radiocarpal and midcarpal joints. The lunate shows slightly different kinematics, as it contributes almost equally in both joints during flexion-extension motion, but is more prominent in the midcarpal joint during DTM [29]. We surmise that due to the differences in extrinsic wrist muscle forces between the two postures, the carpus kinematics was altered, thereby affecting the stress distribution at the radiocarpal joint.

Previous 3D computation analyses suggested that ligament laxity affects carpal bone motion of the proximal row throughout radial and ulnar deviation motions [30]. By reducing the ligament stiffness in our elder model, we showed that a larger range of motion was enabled, in both extension and radial-extension positions (Figure 3). Furthermore, due to lower stresses at the radius-scaphoid interface during radial extension compared with the adult model and in the radius-lunate interface, the stress differences between the two wrist positions were smaller in the elder model compared with the adult model. The differences in contact area and stresses found in the elder model compared with the adult model may help to explain symptoms in the aging wrist. For example, in a prospective study with 96 patients with symptomatic dorsal wrist ganglions and 96 individuals without dorsal wrist ganglions, the symptoms were associated with hyperlaxity of the ligaments and a positive scaphoid shift test [31].

The magnitude of the stresses calculated in our study is comparable to similar FE analyses of wrist loading in the literature. An FE study, where 100N axial compressive force was applied to the virtual wrist, showed peak contact stresses of 5.2 MPa and 2.9 MPa at the radius-scaphoid and radius-lunate articulations, respectively, during neutral position [13]. Their model differed from our model mainly in the isotropic elastic properties attributed to the cartilage, as well as wrist position. Consequently, in the FE analyses presented here, a higher compressive force was applied (a total force of 267 N), with a peak contact stress values of 29.7 MPa and 15.5 MPa at the radius-scaphoid and radius-lunate articulations, respectively, in extension. Another study presented a peak contact pressure in the radius-scaphoid interface of 2.9 MPa during extension (maximum extension with a total 7 Kg force applied on the tendons) [10]. The FE model did not include ligaments but used almost full contact areas between the bones to distribute the loads. The FE modeling included the bones as rigid bodies and the cartilage. The contact layer of each bone was defined as initially having mutual contacting surfaces. The contact stress results here (relative to the loads) were higher than presented in the aforementioned study, since their loads were distributed over vast contacting areas due to the modeling assumptions.

There are several limitations to the study. First, the analyses presented refer to a healthy adult female, so that the results are not conclusive for males or individuals with arthritis or other pathologies that might cause bone deformities or ligament impairments. This limitation could be addressed in future studies by using statistical shape models that could allow extending results to a whole set of geometries reproducing the actual variability in population [32]. Second, several assumptions were used to simplify the model and adjust it to the computational capabilities. We assumed that cartilage thickness may have affected the cartilage contact stresses, as previously suggested for hip models [33]. Additionally, the properties of the proximal SLIL were assumed. Third, the elder was
defined by altered properties of the ligaments, whereas bone and cartilage properties were not adjusted. In matters of resembling the cartilage degradation by thickness change, a former FE study depicted rheumatoid arthritis by removing all the cartilage from the wrist model [34]. The collagen network and proteoglycans that bond to water allow the cartilage to resist compressive forces [35]. Changes in these matrix components might influence the mechanical properties of the cartilage [35]. Finally, since we used the actual wrist positions, as carried out in [28], which reported the tendon forces necessary to hold specified static positions of a cadaveric wrist, segmentation uncertainties might have affected the stress distribution in the comparison between the two positions.

In summary, this is the first study to compare the wrist FE models of adults and elders in the extension and radial-extension positions. The reduced stresses during radial extension might provide an explanation for our frequent use of this movement pattern, as the reduced stresses decrease the risk of overuse injury [36]. Further work should evaluate the effect of the DTM posture on wrist stresses in individuals with different pathologies or after fractures and ligament tears. Additionally, estimating the tendon forces through a multibody model, as simulated in [37], might advance our understanding of the biomechanics of different wrist positions under different loading conditions.

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Conflicts of Interest: The authors declare no conflict of interest.

Appendix A

Table A1. Stiffness of ligaments in the finite element model.

| Ligament            | Origin                      | Insertion                  | Stiffness [N/mm] | References |
|---------------------|-----------------------------|----------------------------|------------------|------------|
| TCL *               | Scaphoid, Trapezium         | Hamate, Pisiform           | 64.5             | [25]       |
| STT (palmar)        | Scaphoid                    | Trapezium, Trapezoid       | 150              | [38]       |
| SLIL proximal       | Scaphoid                    | Lunate                     | 150              |            |
| Palmar Capito-hamate| Capitate                    | Hamate                     | 325              | [18,38]    |
| Capito-trapezial    | Capitate                    | Trapezium                  | 300              | [18,38]    |
| Dorsal Intercarpal  | Hamate                      | Capitate                   | 325              | [18,38]    |
| Dorsal Intercarpal  | Capitate                    | Trapezoid                  | 300              | [18,38]    |
| Dorsal Intercarpal  | Hamate                      | Triquetrum                 | 300              | [18,38]    |
Table A1. Cont.

| Ligament                       | Origin | Insertion   | Stiffness [N/mm] | References |
|--------------------------------|--------|-------------|------------------|------------|
| Dorsal Intercarpal             | Hamate | Lunate      | 150              | [18,38]    |
| Dorsal Intercarpal             | Capitate | Lunate   | 150              | [18,38]    |
| Dorsal Intercarpal             | Capitate | Scaphoid | 150              | [18,38]    |
| Dorsal Intercarpal             | Scaphoid | Trapezium | 150              | [18,38]    |
| Dorsal Intercarpal             | Trapezoid | Trapezium | 110              | [16]       |
| Dorsal Intercarpal (triquetro-trapezial) | Trapezium, Trapezoid | Triquetrum | 128              | [16]       |
| Dorsal Intercarpal             | Lunate | Triquetrum | 350              | [18,38]    |
| Dorsal Radio-ulnar             | Radius | Ulna       | 50               | [18,38]    |
| Dorsal Scapho-lunate (SLIL)    | Lunate | Scaphoid   | 230              | [18,38]    |
| Long Radio-lunate              | Lunate | Radius     | 75               | [18,38]    |
| Short Radio-lunate             | Lunate | Radius     | 75               | [18,38]    |
| Piso-hamate                    | Hamate | Pisiform   | 100              | [18,38]    |
| Palmar scapho-capitate         | Capitate | Scaphoid | 40               | [18,38]    |
| Radial Collateral (radio-scaphoid) | Radius | Scaphoid | 50               | [18,38]    |
| Radio-scapho-capitate          | Radius | Capitate   | 50               | [18,38]    |
| Palmar RTq                     | Radius | Triquetrum | 15               | [39]       |
| Dorsal RTq                     | Radius | Triquetrum | 46               | [39]       |
| Scapho-triquetrum              | Scaphoid | Triquetrum | 128              | [16]       |
| Ulno-lunate                    | Lunate | Ulna       | 40               | [18,38]    |
| Triquetro-capitate (TCP)       | Capitate | Triquetrum | 36               | [39]       |
| Ulnar Collateral               | Triquetrum | Ulna | 100              | [18,38]    |
| Ulnar Collateral               | Pisiform | Ulna   | 100              | [18,38]    |
| Ulno-triquetral Volar/Dorsal   | Triquetrum | Ulna | 40               | [18,38]    |
| Volar Radio-ulnar              | Radius | Ulna       | 50               | [18,38]    |
| Volar Triquetro-hamate         | Hamate | Triquetrum | 50               | [18,38]    |
| Volar Scapho-lunate (SLIL)     | Lunate | Scaphoid   | 230              | [18,38]    |
| Volar Piso-triquetal           | Pisiform | Triquetrum | 150              | [18,38]    |
| Volar Luno-triquetral          | Lunate | Triquetrum | 350              | [18,38]    |
| Volar Trapezoid-capitate       | Capitate | Trapezoid | 300              | [18,38]    |

TCL = transverse carpal ligament; STT = scaphoid-trapezium-trapezoid, SLIL= scapho-lunate interosseous ligament. * Reference relates to elder age female and the stiffness includes an extrapolation.
Appendix B

Table A2. Forcet in the ligaments attached to the scaphoid, lunate and radius bones in extension and radial-extension models in the adult and elder simulations. The percentage of variable in the radial-extension model in relation to the extension model is calculated and presented as “% of Extension”.

|                  | Adult Model | Elder Model | % Of Extension | Adult Model | Elder Model | % Of Extension |
|------------------|-------------|-------------|----------------|-------------|-------------|----------------|
| Dorsal SLIL      | 0.16        | 0.27        | 168.8          | 0.13        | 0.37        | 284.6          |
| Volar SLIL       | 0           | 0.38        | -              | 0           | 0.46        | -              |
| Proximal SLIL    | 0.16        | 0.14        | 87.5           | 0.17        | 0.16        | 94.1           |
| Volar STT        | 0.32        | 0.58        | 181.3          | 0.35        | 0.77        | 220.0          |
| TCL              | 0.45        | 0.01        | 2.2            | 1.19        | 0           | -              |
| Dorsal Radio-Triquetrum Lunotriq | 0.09      | 0.21        | 233.3          | 0.10        | 0.24        | 240.0          |
| Dorsal Lunotriq  | 0.07        | 0.30        | 428.6          | 0           | 0.40        | -              |
| Palmar Radiotriq | 0.01        | 0.12        | 1200.0         | 0           | 0.15        | -              |
| Scapholunate     | 0.09        | 0.15        | 166.7          | 0.09        | 0.19        | 211.1          |
| Short Radioulnate| 0           | 0           | -              | 0           | 0           | -              |
| Long Radioulnate | 0           | 0.40        | -              | 0           | 0.48        | -              |
| Dorsal Radioulnar| 0           | 0           | -              | 0           | 0           | -              |
| Dorsal Cap Lun   | 0.05        | 0.26        | 520.0          | 0           | 0.34        | -              |
| Dorsal Ham Lun   | 0.04        | 0.02        | 50.0           | 0.05        | 0           | -              |
| Radioscaphoid    | 0.09        | 0           | -              | 0.11        | 0           | -              |
| Radioscaphocapitate | 0.07     | 0.21        | 300.0          | 0.09        | 0.26        | 288.9          |
| Scaphocapitate   | 0.15        | 0           | -              | 0.15        | 0           | -              |
| Dorsal Scap Tipm | 0.15        | 0.07        | 46.7           | 0.17        | 0.12        | 70.6           |

TCL = transverse carpal ligament; STT = scaphoid-trapezium-trapezoid, SLIL = scapho-lunate interosseous ligament.

References
1. Moritomo, H.; Apergis, E.P.; Herzberg, G.; Werner, F.W.; Wolfe, S.W.; Garcia-Elias, M. 2007 IFSSH Committee Report of Wrist Biomechanics Committee: Biomechanics of the So-Called Dart-Throwing Motion of the Wrist. J. Hand Surg. Am. 2007, 32, 1447–1453. [CrossRef] [PubMed]
2. Garcia-Elias, M.; Alomar Serrallach, X.; Monill Serra, J. Dart-throwing motion in patients with scapholunate instability: A dynamic four-dimensional computed tomography study. J. Hand Surg. Eur. Vol. 2014, 39, 346–352. [CrossRef] [PubMed]
3. Kaufman-Cohen, Y.; Portnoy, S.; Levanon, Y.; Friedman, J. Does Object Height Affect the Dart Throwing Motion Angle during Seated Activities of Daily Living? J. Mot. Behav. 2020, 52, 456–465. [CrossRef]
4. Kaufman-Cohen, Y.; Friedman, J.; Levanon, Y.; Jacobi, G.; Doron, N.; Portnoy, S. Wrist plane of motion and range during daily activities. Am. J. Occup. Ther. 2018, 72, 7206205080p1–7206205080p10. [CrossRef]
5. Kaufman-Cohen, Y.; Levanon, Y.; Friedman, J.; Yaniv, Y.; Portnoy, S. Home exercise in the dart-throwing motion plane after distal radius fractures: A pilot randomized controlled trial. J. Hand Ther. 2020. [CrossRef]
6. Feehan, L.; Fraser, T. Early controlled mobilization using dart-throwing motion with a twist for the conservative management of an intra-articular distal radius fracture and scapholunate ligament injury: A case report. J. Hand Ther. 2016, 29, 191–198. [CrossRef] [PubMed]
7. Feehan, L.; Fraser, T. Dart-throwing motion with a twist orthoses: Design, fabrication, and clinical tips. J. Hand Ther. 2016, 29, 205–212. [CrossRef] [PubMed]
8. Schleifenbaum, S.; Prietzel, T.; Hädrich, C.; Möbius, R.; Sichting, F.; Hammer, N. Tensile properties of the hip joint ligaments are largely variable and age-dependent–An in-vitro analysis in an age range of 14–93 years. J. Biomech. 2016, 49, 3437–3443. [CrossRef] [PubMed]
9. Woo, S.L.Y.; Hollis, J.M.; Adams, D.J.; Lyon, R.M.; Takai, S. Tensile properties of the human femur-anterior cruciate ligament-tibia complex: The effects of specimen age and orientation. Am. J. Sports Med. 1991, 19, 217–225. [CrossRef] [PubMed]
10. Varga, P.; Scheffzig, P.; Unger, E.; Mayr, W.; Zysset, P.K.; Erhart, J. Finite element based estimation of contact areas and pressures of the human scaphoid in various functional positions of the hand. J. Biomech. 2013, 46, 984–990. [CrossRef] [PubMed]
11. Oflaz, H.; Gunal, I. Maximum loading of carpal bones during movements: A finite element study. Eur. J. Orthop. Surg. Traumatol. 2019, 29, 47–50. [CrossRef] [PubMed]
12. Alonso Rasgado, T.; Zhang, Q.; Jimenez Cruz, D.; Bailey, C.; Pinder, E.; Mandaoleson, A.; Talwalkar, S. Analysis of tenodesis techniques for treatment of scapholunate instability using the finite element method. Int. J. Numer. Method. Biomed. Eng. 2017, 33. [CrossRef] [PubMed]
13. Guo, X.; Fan, Y.; Li, Z.M. Effects of dividing the transverse carpal ligament on the mechanical behavior of the carpal bones under axial compressive load: A finite element study. Med. Eng. Phys. 2009, 31, 188–194. [CrossRef] [PubMed]
14. Gosling, J.; Harris, P.; Humpherson, J.; Whitmore, I.; Willan, P. Human Anatomy, Color Atlas and Textbook, 6th ed.; Elsevier Health Sciences: London, UK, 2016.
15. Gislonson, M.; Nash, D. Finite Element Modelling of a Multi Bone Joint: The Human Wrist. In Finite Element Analysis: New Trends and Developments; Ebrahimi, F., Ed.; Intech: Rijeka, Croatia, 2012.
16. Bajuri, M.N.; Kadir, M.R.; Raman, M.M.; Kamarul, T. Mechanical and functional assessment of the wrist affected by rheumatoid arthritis: A finite element analysis. Med. Eng. Phys. 2012, 34, 1294–1302. [CrossRef] [PubMed]
17. Gislonson, M.K.; Stansfield, B.; Nash, D.H. Finite element model creation and stability considerations of complex biological articulation: The human wrist joint. Med. Eng. Phys. 2010, 32, 523–531. [CrossRef] [PubMed]
18. Carrigan, S.D.; Whiteside, R.A.; Pichora, D.R.; Small, C.F. Development of a three-dimensional finite element model for carpal load transmission in a static neutral posture. Ann. Biomed. Eng. 2003, 31, 718–725. [CrossRef]
19. Bajuri, M.N.; Abdul Kadir, M.R.; Murali, M.R.; Kamarul, T. Biomechanical analysis of the wrist arthroplasty in rheumatoid arthritis: A finite element analysis. Med. Biol. Eng. Comput. 2013, 51, 175–186. [CrossRef]
20. Courtney, A.C.; Hayes, W.C.; Gibson, L.J. Age-related differences in post-yield damage in human cortical bone. Experiment and model. J. Biomech. 1996, 29, 1463–1471. [CrossRef]
21. Edwards, W.B.; Troy, K.L. Finite element prediction of surface strain and fracture strength at the distal radius. Med. Eng. Phys. 2012, 34, 290–298. [CrossRef] [PubMed]
22. Brown, C.P.; Nguyen, T.C.; Moody, H.R.; Crawford, R.W.; Oloyede, A. Assessment of common hyperelastic constitutive equations for describing normal and osteoarthritic articular cartilage. Proc. Inst. Mech. Eng. Part H J. Eng. Med. 2009, 223, 643–652. [CrossRef] [PubMed]
23. Madireddy, S.; Sista, B.; Vemaganti, K. Bayesian calibration of hyperelastic constitutive models of soft tissue. J. Mech. Behav. Biomed. Mater. 2016, 59, 108–127. [CrossRef] [PubMed]
24. Berger, R.A. The gross and histologic anatomy of the scapholunate interosseous ligament. J. Hand Surg. Am. 1996, 21, 170–178. [CrossRef]
25. Brett, A.W.; Oliver, M.L.; Agur, A.M.R.; Edwards, A.M.; Gordon, K.D. Quantification of the transverse carpal ligament elastic properties by sex and region. Clin. Biomech. 2014, 29, 601–606. [CrossRef]
26. Miyamura, S.; Oka, K.; Lans, J.; Sakai, T.; Shioide, R.; Kazui, A.; Tanaka, H.; Shimada, S.; Murase, T. Cartilage and subchondral bone distributions of the distal radius: A 3-dimensional analysis using cadavers. Osteoarthr. Cartil. 2020, 28, 1572–1580. [CrossRef]
27. Abe, T.; Theibaud, R.S.; Loenneke, J.P. Age-related change in handgrip strength in men and women: Is muscle quality a contributing factor? Age 2016, 38, 1–7. [CrossRef] [PubMed]
28. Werner, F.W.; Palmer, A.K.; Somerset, J.H.; Tong, J.J.; Gillison, D.B.; Fortino, M.D.; Short, W.H. Wrist joint motion simulator. J. Orthop. Res. 1996, 14, 639–646. [CrossRef] [PubMed]
29. Goto, A.; Moritomo, H.; Murase, T.; Oka, K.; Sugamoto, K.; Arimura, T.; Masumoto, J.; Tamura, S.; Yoshikawa, H.; Ochi, T. In vivo three-dimensional wrist motion analysis using magnetic resonance imaging and volume-based registration. J. Orthop. Res. 2005, 23, 750–756. [CrossRef] [PubMed]
30. Best, G.M.; Zec, M.L.; Pichora, D.R.; Kamal, R.N.; Rainbow, M.J. Does Wrist Laxity Influence Three-Dimensional Carpal Bone Motion? J. Biomech. Eng. 2018, 140, 041013. [CrossRef]
31. McKeon, K.E.; London, D.A.; Osei, D.A.; Gelberman, R.H.; Goldfarb, C.A.; Boyer, M.L.; Calfee, R.P. Ligamentous hyperlaxity and dorsal wrist ganglions. J. Hand Surg. Am. 2013, 38, 2138–2143. [CrossRef] [PubMed]
32. Pascoletti, G.; Cali, M.; Bignardi, C.; Conti, P.; Zanetti, E.M. Mandible Morphing Through Principal Components Analysis. In Proceedings of the International Conference on Design Tools and Methods in Industrial Engineering, ADM 2019, Modena, Italy, 9–10 September 2019; Springer: Cham, Switzerland, 2020; pp. 15–23.
33. Anderson, A.E.; Ellis, B.J.; Maas, S.A.; Weiss, J.A. Effects of idealized joint geometry on finite element predictions of cartilage contact stresses in the hip. J. Biomech. 2010, 43, 1351–1357. [CrossRef] [PubMed]
34. Bajuri, M.N.; Abdul Kadir, M.R.; Amin, I.M.; Ochsner, A. Biomechanical analysis of rheumatoid arthritis of the wrist joint. Proc. Inst. Mech. Eng. Part H J. Eng. Med. 2012, 226, 510–520. [CrossRef]
35. Cooke, M.E.; Lawless, B.M.; Jones, S.W.; Grover, L.M. Matrix degradation in osteoarthritis primes the superficial region of cartilage for mechanical damage. Acta Biomater. 2018, 78, 320–328. [CrossRef] [PubMed]
36. Terzini, M.; Zanetti, E.M.; Audenino, A.L.; Putame, G.; Pastorelli, S.; Panero, E.; Sard, A.; Bignardi, C. Multibody modelling of ligamentous and bony stabilizers in the human elbow. Muscles. Ligaments Tendons J. 2017, 7, 493–502. [CrossRef] [PubMed]
37. Schuind, F.; Cooney, W.P.; Linscheid, R.L.; An, K.N.; Chao, E.Y.S. Force and pressure transmission through the normal wrist. A theoretical two-dimensional study in the posteroanterior plane. J. Biomech. 1995, 28, 587–601. [CrossRef]
38. Savellberg, T.H.C.M.; Kooloo, J.G.M.; Huiskes, R.; Kauer, J.M.G. Stiffness of the ligaments of the human wrist joint. J. Biomech. 1992, 25, 369–376. [CrossRef]