Loading at the distal radius and ulna during active simulated dart throw motion

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ABSTRACT

Loading at the distal forearm during dart throw motion (DTM) has been examined under static loads but there is no consensus on how loading is affected by active motion. In this work two implants were designed to measure forearm loading in a cadaveric model of wrist motion. Loads through the radius and ulna were significantly greater in reverse DTM than forward DTM. Radius loads were greatest in extended and radial deviated positions, and ulnar loads were greatest in flexed and ulnar deviated position. This work gives insight into the biomechanics of loading of the forearm to guide further studies.

1. Introduction

The wrist is capable of a diverse range of motion while serving as a stable platform for gripping. In order to maximize function, force transmission from the hand to the forearm must be painless and coordinated. In the native wrist under static conditions the force transferred to the forearm is divided 82/18 between the radius and ulna.^{1–6} Although classically this ratio was reported as dogma, load sharing at the distal forearm has been demonstrated to be a dynamic process that changes with wrist position and motion.^{1,2} These native biomechanics may be altered by traumatic injuries and degenerative diseases that often lead to weakness, stiffness and arthritis, due at least in part to altered bone loading. This emphasizes the need for a comprehensive understanding of native load sharing at the wrist and distal forearm, as well as changes in biomechanical factors when a disorder arises in an effort to target clinical interventions.

The wrist has many planes of motion, but dart throw motion (DTM) is considered to best represent typical motions that occur during activities of daily living (ADLs).^{4,7} The wrist naturally radial deviates during extensions and ulnar deviates during flexion. Dart throw motion is a combination flexion-extension motion (FEM) and radioulnar deviation

(RUD) and occurs at the radiocarpal and midcarpal joints around the sagittal plane of the hand. There are varying opinions on which range of motion best represents a DTM, but it is agreed that this composite movement is critical to normal carpal function.

The purpose of this cadaver-based study was to quantify the magnitude of bone loads and bone load sharing during dart throw motions to allow for a more complete understanding of loading at the distal forearm. Axial loads through the distal radius and ulna were measured during simulated multiplanar DTM. Differences in forearm bone load were examined based on the direction of wrist joint motion and wrist joint angle. It was hypothesized that the direction and angle of motion will have effects on axial forearm bone loads and load sharing.

2. Materials and methods

2.1. Specimen preparation

Nine (9) fresh frozen cadaveric upper extremities resected mid-humerus were used. Computed tomography (CT) scans were examined and medical histories were reviewed for all specimens to rule out previous forearm injuries and disease. The specimens were thawed for 18 h

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Fig. 1. Implanted Distal Radius and Ulna Load Sensors: (A) Computational depiction of custom made experimental load cells were implanted in the distal ulna and radius to allow for continuous monitoring of load experienced by each bone during active dart throw motion. (B) Radiograph demonstrating implanted device within cadaveric specimen.

Fig. 2. Active Motion Wrist Simulator: The in vitro active motion simulator is capable of loading seven forearm muscles to generate wrist flexion-extension and radioulnar deviation: (A) structural base of simulator (B) Biceps servomotor (C) servomotor manifold, (D) cable guide, (E) suture cables, (F) ulnar support tower acts to maintain forearm in 90° of flexion (G) humeral clamp, (H) optical trackers.

prior to testing. Optical trackers were placed under fluoroscopic imaging and braided nylon line (45 kg test strength) was sutured into the tendons of interest proximal to the wrist including the biceps, pronator teres (PT), flexor carpi radialis (FCR), flexor carpi ulnaris (FCU), extensor carpi ulnaris (ECU), extensor carpi radialis brevis (ECRB), and extensor carpi radialis longus (ECRL). Finger flexors and extensors were not included in order to maintain visualization of the optical trackers. Guides were secured to the medial and lateral humeral epicondyles and tendon lines were directed through them to maintain physiological lines of action.

2.2. Load measurement apparatus and implantation

Specialized implants incorporating in-line load cells were developed for the measurement of loads in the ulna and radius independently. The implants were custom machined from stainless steel and contained miniature uniaxial load cells. They were surgically implanted in the distal forearm bones, being careful to preserve the soft tissues (Fig. 1). A volar radial osteotomy was performed, and a dorsal bone bridge was left intact to maintain the alignment of the distal and proximal bone segments during insertion of the device. Polymethylmethacrylate bone cement was injected into the distal trabecular bone and proximal intramedullary canal and the distal plate and proximal stem were secured in place. A similar procedure was followed on the medial side of the ulna for device implantation. Ulnar device fixation was achieved with intramedullary stems both proximally and distally. The osteotomies were performed as distal as possible while still proximal to the

DRUJ to avoid joint disruption.

2.3. Testing simulator and experimental protocols

Specimens were mounted on the wrist simulator using a humeral clamp and two perpendicular ulnar pins with the elbow fixed at 90° of flexion (Fig. 2). The tendon sutures were connected to smart motors located at the base of the wrist simulator.⁸ Optical trackers were secured to the dorsal distal aspect of the third metacarpal, the radius proximal to the PT insertion, the proximal ulna, the distal radius and ulna distal to the osteotomies to track wrist and forearm motion and collect kinematic data throughout testing. Neutral wrist position was defined by the International Society of Biomechanics guidelines and used to calculate the angle of wrist flexion-extension and radial-ulnar position.⁹ Anatomical landmarks visible through soft tissues were used to establish clinically relevant coordinate systems for the radius, ulna and metacarpal to track joint angles. Wrist angle was calculated as the angle between the long axis of the forearm and the long axis of the 3rd metacarpal with respect to the radial coordinate system. Neutral position was defined as 0° of wrist flexion, 0° of radioulnar deviation and 0° of forearm rotation. Joint motions were controlled by a machine vision algorithm that applied varying tendon loads required to reach the programmed joint angle. Joint angle was recorded using the relative motion of the radial, ulnar and metacarpal coordinate systems. Two cyclic motions were completed at the rate of 5° of joint angulation per second for DTM (combined –30° of extension and –10° of radial deviation to 30° of flexion and 10° of ulnar deviation). The first (pre-conditioning) cyclic movement was

Fig. 3. Axial Bone Loads through the Distal Radius and Ulnar during Active Forward and Reverse DTM: Mean axial load through the distal radius and distal ulna throughout fDTM and rDTM -30° of extension and 10° of radial deviation to 30° of flexion and 10° of ulnar deviation. Standard deviations have been omitted for clarity (radius fDTM range: $+10.5$ N to $+55.3$ N; radius rDTM range: $+20.2$ N to $+43.1$ N; ulna fDTM range: $+6.7$ N to $+6.7$ N; ulna rDTM range: $+8.2$ N to $+15.3$ N).

disregarded and the second cycle was analyzed. Individual motions were defined as a full range of motion in one direction, e.g. forward DTM (fDTM) was defined as motion from 30° of wrist extension to 30° flexion, with reverse DTM (rDTM) being defined as the opposite path.

Axial bone loads were continuously collected from the load cells implanted in the distal radius and ulna throughout simulated active motion. Absolute bone loads were measured, and the percentage of total bone load was calculated for both forearm bones to examine their individual contributions. Individual motion pathways and the correlating radial and ulnar load cell measurements were discretized in 5° increments of wrist motion for data analysis and statistical purposes. Loads applied to the FCR, FCU, ECU, ECRB and ECRL tendons were also recorded for the purposes of interpretation of the bone loads, but not analyzed statistically as tendon loads were not a specific outcome variable.

2.4. Statistical analysis

A two-way (wrist angle, forearm bone) repeated measures analysis of variance (RM-ANOVA) was used, with a level of significance set at $\alpha = 0.05$. The independent variables included: motion direction and wrist joint angle. The dependent variable was axial bone load. Individual analyses were completed for the radius and ulna throughout DTM. Additional pairwise comparisons were completed using the Bonferroni adjustment to examine the differences between individual joint angles.

3. Results

The average age of specimens was 74 ± 7 years (avg $\pm$ std). All specimens were male and right sided and able to achieve the desired range of DTM (30° extension and 10° radial deviation to 30° flexion and

Table 1

Forces Through the Distal Radius and Ulna During Reverse and Forward DTM at given Wrist Positions. Loads through the radius were always greater than the ulna. Forces peaked in the radius when the wrist was initiating fDTM from an extended and radially deviated position. Forces peaked in the ulna when the wrist was initiating rDTM from a flexed and ulnarly deviated position.

Magnitude of Load Through the Distal Forearm Bones (Mean $\pm$ SD)					
Wrist Angle		Radius Load (N)		Ulna Load (N)	
FEM	RUD	Flexion DTM	Extension DTM	Flexion DTM	Extension DTM
-30°	-10°	109.6 ± 55.2	105.7 ± 30.7	5.4 ± 9.7	8.7 ± 8.7
0	0	57.4 ± 18.9	35.3 ± 20.7	11.5 ± 8.3	19.2 ± 7.7
30°	10°	64.4 ± 16.4	74.1 ± 20.1	25.2 ± 9.8	34.1 ± 15.9

10° ulnar deviation). The repeatability of the radial and ulnar axial load cell measurements had an ICC of 0.984 and 0.996 respectively for the five cycles of fDTM.

3.1. Forearm loading during active DTM

The direction of DTM had a significant effect on load transmission through the distal radius and ulna ($p = 0.003$, $p = 0.02$). Loads through the distal radius were significantly greater in rDTM than those in fDTM at -10° – 30° of wrist flexion ($p < 0.032$). Loads through the distal ulna were significantly greater in rDTM than in fDTM at 15° – 30° of wrist flexion ($p < 0.047$). The magnitude of load transmitted through the distal radius and ulna changed significantly throughout the forward and reverse DTM ($p = 0.016$, $p = 0.001$) (Fig. 3). The magnitude of distal radial loading was greater than ulnar loading throughout all DTM ($p <$

Fig. 4. Proportion of Total Forearm Bone Load through the Distal Radius and Ulna During Active Forward DTM: Proportion of total forearm bone load through the distal radius and distal ulna throughout forward and reverse dart throw motion from -30° of extension and 10° of radial deviation to 30° of flexion and 10° of ulnar deviation. Standard deviations have been omitted for clarity (radius range: $+7.9\%$ to $+14.2\%$; ulna range: $+7.9\%$ to $+14.2\%$).

Table 2

Proportion of load transferred through the ulna for a given position of the wrist in unconstrained active DTM. Maximal load sharing through the ulna occurred during flexion and ulnar deviation of the wrist. Previous position of the wrist had an impact on maximal force with rDTM requiring a larger force to initiate motion than fDTM.

Wrist Angle		Proportion of Total Bone Load Through the Ulna (%)	
FEM	RUD	Forward DTM	Reverse DTM
-30°	-10°	5.2 ± 9.3	6.8 ± 6.6
0	0	18.2 ± 14.2	16.3 ± 6.8
30°	10°	24.8 ± 8.7	29.2 ± 9.5

0.001) (Table 1).

3.2. Forearm load-sharing during active DTM

The direction of motion had no effect on distal forearm bone load sharing ($p = 0.602$). Load sharing between the distal radius and ulna changed significantly throughout DTM ($p < 0.001$) (Fig. 4). The proportion of load through the distal radius was higher than that through the ulna throughout fDTM ($p < 0.001$) (Table 2).

3.3. Tendon loading during active DTM

Tendon loads for forward and reverse DTM are demonstrated in Fig. 5 and Fig. 6. Ulnar sided tendons exerted maximal force during the maximal ulnar extent of DTM and similarly radial sided tendons exerted maximal force at the radial extreme of motion. In each case, extensor tendon forces were consistently greater than analogous flexor tendon forces. 4.

4. Discussion

The carpus is a dynamic system that transfers force from the hand to the distal forearm. This process is dynamic with load sharing between the radius and ulna varying with wrist position and motion.^{1,2} When injuries and degenerative pathologies alter normal anatomy joint kinematics may be altered leading to a painful, dysfunctional grip. Many authors have described load sharing at the distal forearm during static states, as well as simplified planar wrist motion, but there has been little investigation on the effects of complex multiplanar wrist motion, such as DTM, during active wrist motion while measuring in vitro bone loading.¹⁻⁶

Classic literature reports load sharing between the radius and ulna to be 82/18 when the wrist is statically loaded in a neutral position, with changes to this ratio occurring that are dependent on wrist position.¹⁻⁶ Harley et al. (2015) examined the impact of RUD on forearm load sharing and demonstrated peak ulnar loading of 23.5% at maximal ulnar deviation. This force diminished with radial deviation, reaching 8% at the radial endpoint of motion. Similar results were obtained by Ekenstam et al. (1984) using a static tendon loading protocol that demonstrated a radius/ulna load sharing ratio of 91/9 at maximal radial deviation and 76/24 at maximal ulnar deviation. This group of authors also examined the effect of FEM on distal forearm loading and demonstrated that maximal loading of the ulna occurred with wrist extension, peaking at 20%, and was minimal during flexion, reaching 10%.^{2,5,10} Greenberg et al. (2013) investigated ulnar loading and produced similar results regarding RUD and FEM but also reported that during dynamic fDTM that a combination of wrist flexion and ulnar deviation produced the maximum proportion of load sharing through the distal ulna at $26 \pm 10\%$.

The purpose of this study was to examine the relationship of DTM

Fig. 5. Mean applied tendon loads of wrist flexors and extensors throughout forward DTM. Range of motion was from -30° of extension and 10° of radial deviation to 30° of flexion and 10° of ulnar deviation. Standard deviation bars have been omitted for clarity (FCR range: +1.9 N to +18.9 N; FCU range: +2.3 N to +14.6 N; ECU range: +3.4 N to +18.3 N; ECRB range: +1.5 N to +6.4 N; ECRL range: +2.8 N to +25.6 N).

with loading of the forearm, as DTM is thought to best reflect the range of motion of ADLs.^{1,2} Load through the ulna was minimized at 5% during combined wrist extension and radial deviation and maximum during combined wrist flexion and ulnar deviation, reaching 29%. Comparing these results to current literature, loading of the ulna was more similar to values reported during RUD than those of wrist FEM, although in comparison to prior studies of simple planar motion, there does appear to be an additive effect combining these motions.

Loads through the distal forearm were significantly higher in rDTM compared to fDTM. The muscle moment arms of the wrist flexors are greater than the extensors and both vary with wrist position.¹¹ During rDTM extensors must exert more force than flexors to achieve the same amount of joint motion. The increased extensor muscle force may account for the increased loading through the distal radius and ulna during rDTM. Tendon loads required to initiate and achieve these motions may also contribute to differences in distal forearm loading DTM. In particular, there was an increase in ECU loads to initiate rDTM, but as the wrist moved toward a neutral position ECU tensions tapered and ECRL loads increase to complete the combined extension-radial deviation motion. In fDTM, FCR loads peaked to initiate the motion and then quickly decreased to allow for ulnar deviation. The ECU and FCU then increased in parallel to generate the flexion-ulnar deviation position that terminates fDTM. The tendon loads required to produce forward and reverse DTM more similarly mimic those of RUD than those of FEM.

The increase in ulnar loading during fDTM is thought to be caused by the compression of the triquetrum on the triangular fibrocartilage complex (TFCC) induced active tendon loading. Clinically this increased compression of the triquetrum against the TFCC would also explain the pain induced clinically when performing the ulnocarpal stress test for ulnar sided wrist pain.¹² Similarly, the increased load bearing of the radius during rDTM may be due in part to the compression of the

scaphoid against the scaphoid facet on the distal radius, which is induced by the wrist flexors and extensors that span the radiocarpal joint. Our results are in agreement with Greenberg et al.(2013) demonstrating that the radius accounted for the largest proportion of forearm bone load DTM regardless of path, peaking in extension and radial deviation (94%) and reaching a minimum in flexion and ulnar deviation (72%). The increased loads and proportion of total forearm load through the ulna during DTM may explain ulnar-sided wrist pain that often occurs during repetitive wrist motions and power gripping.¹³ High ECU activation in dart throw motion may also explain the increased incidence of wrist tendonitis in workplaces that require repetitive wrist motions.¹³ This aspect of our study is particularly novel and provides a better understanding of the effect of the combined extension-radial deviation and flexion-ulnar deviation generated by DTM.

The study has limitations. Forearm bone loads were collected strictly through the long axis of the radius and ulna, and as such, off axis loading was not measured. While physiologic loads occur in all directions of the wrist joint, the axial bone loads are more indicative of radiocarpal joint forces than they are of distal radioulnar joint loads. Wrist motions were performed actively through applied physiologic tendon loads; but these loads are an estimation of in vivo loads generated during simple loading scenarios and not resisted joint motion or simulated object lifting. Additionally, no forces were transmitted through digital flexors or extensors which would contribute to overall bone loading. This study also has several strengths. Highly accurate optical motion tracking was employed, allowing for real time joint angle feedback during a multi-planar dart throw motion. The experimental devices were discrete, located anatomically, and permitted the retention of all soft tissues, excluding the distal interosseous membrane.^{1-3,5,6,10} The direction of motion was accounted for and load magnitudes were reported in both directions of DTM.

Fig. 6. Mean applied tendon loads of wrist flexors and extensors throughout reverse DTM. Range of motion was 30° of flexion and 10° of ulnar deviation to 30° of extension and 10° of radial deviation. Standard deviation bars have been omitted for clarity (FCR range: +1.3 N to +14.1 N; FCU range: +1.1 N to +17.5 N; ECU range: +2.6 N to +28.9 N; ECRB range: +2.7 N to +11.5 N; ECRL range: +8.4 N to +18.4 N).

In summary, the current study supports the hypotheses that axial loads through the distal radius and ulna fluctuate quite markedly during active wrist motions. This study provides a more detailed insight into the loads occurring in the healthy wrist during active wrist motion and may have implications for further investigations of wrist biomechanics.

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