Wrist and Tendon Dynamics as Contributory Risk Factors in Work-Related Musculoskeletal Disorders

Yongku Kong, Hyunkook Jang, and Andris Freivalds
Harold and Inge Marcus Department of Industrial and Manufacturing Engineering, 310 Leonhard Building, The Pennsylvania State University, University Park, PA 16802

ABSTRACT

Many studies have shown that repetitive wrist motion is a major risk factor for work-related musculoskeletal disorders (WMSDs). Specific contributory factors include wrist and tendon dynamics. The authors present recent methodological advances, epidemiological studies, and biomechanical models estimating the effects of wrist dynamics on internal tendon force as a theoretical basis for the risk of incurring a WMSD. These biomechanical models utilize either the reduction method or the optimization method to solve the indeterminate problem resulting from too many internal variables. Generally, the optimization methods show the best agreement with direct in vivo tendon force studies. For the models of pinch grips, the average ratio of tendon forces to external forces ranges from 1.8 to 3.5, while for direct tendon measurements, the ratio ranges from 1.73 to 7.92. Similarly, high contributions of flexor tendons for pinches and grasps are found in both the models and direct tendon measurements. These high tendon forces combined with wrist dynamics may be a significant factor in the development of WMSDs. © 2006 Wiley Periodicals, Inc.

1. INTRODUCTION

Repetitive, sustained, forceful movements of the wrist have been found to be an important risk factor associated with work-related musculoskeletal disorders (WMSDs) such as carpal tunnel syndrome (CTS), tenosynovitis, tendinitis, and DeQuervain’s disease (Armstrong, Foulke, Joseph, & Goldstein, 1982; Armstrong, Radwin, & Hansen, 1986; Silverstein, Fine, & Armstrong, 1986; Moore, Wells, & Ranney, 1991). Other epidemiological studies have further identified that awkward hand postures and highly dynamic wrist motions have a strong positive association with the prevalence of hand and wrist tendon disorders (Armstrong & Chaffin, 1979; Marras & Schoenmarklin, 1993; Schoenmarklin, Marras, & Leurgans, 1994). In fact, tendon disorders at the hand–wrist account for 55% of all WMSDs (Bureau of Labor Statistics [BLS], 1996). The dynamic aspects of wrist motion are especially important in the etiology of WMSDs because tendon force is directly affected by wrist acceleration through Newton’s second law, force = mass × acceleration, and friction. Also, empirical and in-the-field studies have shown that wrist acceleration...
dramatically increases risk of CTS (Schoenmarklin & Marras, 1990; Marras & Schoenmarklin, 1993).

To understand the mechanisms of these disorders, the study of the relationship between the finger tendon (internal tissue) forces and the externally applied forces to the fingers is essential. These tendon force studies can help us to understand how external forces are transmitted to the internal tendons and identify which tendons are highly exposed to these external forces; this can lead to tendon repair techniques, rehabilitation procedures, and joint replacement designs (Komi, 1990; Schuind, Garcia-Elias, Cooney, & An, 1992; Soejima, Diao, Lotz, & Hariharan, 1995; Weightman & Amis, 1982).

Therefore, in this review we examine recent methodological advances and epidemiological studies assessing static and dynamic wrist movements and compare various biomechanical tendon force models that assess the increased risk for WMSDs.

2. ASSESSMENT OF WRIST MOTION

2.1. Measurement Devices

Various technologies for measuring wrist motions have been developed, but none has received universal acceptance. One such approach is based on videotape analysis of wrist motion. Developed by Armstrong et al. (1982), it utilizes a frame-by-frame analysis of videotape that records wrist flexion/extension in one of five categories and radial/ulnar deviation in one of three categories. This method requires considerable time and effort because each individual frame has to be analyzed manually, and yields absolute resolutions of only $30^\circ$ for flexion/extension and $20^\circ$ for radial/ulnar deviation. In addition, dynamic variables of the wrist, such as angular velocity and acceleration, are not easily obtainable with this type of analysis.

Logan and Groszewski (1989) developed another method of wrist motion measurement; it utilizes an electromagnetic three space-digitizer sensor system to obtain real-time 6 degrees of freedom position information. Sensors determine specific x, y, z coordinates and $\theta$, $\phi$, $\psi$ orientation angles with respects to a coordinate system based on a low-frequency magnetic field. Further analysis of this data provides flexion/extension, radial/ulnar deviation, and pronation/supination angles. Although this system provided useful information for a number of work tasks in the food-processing industry, the system appeared to require an excessive amount of time for data acquisition and reduction and had limitations due to its instrumentation and magnetic noise.

Another very common approach is the simple goniometer with its output assessed electrically or mechanically. Typically, goniometers are relatively small and lightweight offering quick and objective measurements of wrist joint motions (Buchholz & Wellman, 1997; Hansson, Balogh, Ohlsson, Rylander, & Skerfving, 1996; Nicole, 1987; Ojima, Miyake, Kumashiro, Togami, & Suzeki, 1991). Numerous studies have already been performed using electrogoniometers (goniometers with electrical output) in the laboratory, the factory, and in clinical fields. For example, Smutz, Serina, and Rempel (1994) employed an electrogoniometer for measuring wrist posture in their ergonomic assessment of keyboard design. Moore et al. (1991) used the electrogoniometer to quantify wrist motion of ergonomic risk factors, and Ojima et al. (1991) performed a dynamic analysis of wrist circumduction using the electrogoniometer in the clinical field. However, in all these studies extensive calibration techniques were required.
Schoenmarklin and Marras (1993) developed an electromechanical goniometer to collect online data of wrist movements of flexion/extension, radial/ulnar deviation, and pronation/supination planes simultaneously. Further analyses of the data yielded angular velocity and acceleration. This wrist monitor was composed of two thin metal strips, placed on two adjacent segments with a rotary potentiometer placed at the center of the joint. This system produced relatively accurate and repeatable results; however, it was uncomfortable and obtrusive for the subject to move freely in the normal environment. Furthermore, this wrist monitor is not generally available because it must be custom built.

There are several commercial goniometers (Biometrics Ltd., Gwent, UK, and BIOPAC Systems Inc., Santa Barbara, CA) currently available for measuring both wrist flexion/extension and radial/ulnar deviation, and forearm rotation of pronation/supination. These devices consist of two plastic end blocks that are separated by a flexible spring protecting a strain wire. The goniometers incorporate gauge elements that measure bending strain along or around a particular axis. Biaxial goniometers measure orthogonal rotational axes simultaneously (e.g., wrist flexion/extension and radial/ulnar deviations), while torsimeters are used to measure angular twisting (e.g., forearm pronation/supination) as opposed to bending. Hansson et al. (1996), Rawes, Richardson, and Dias (1996), Buchholz and Wellman (1997), Spielholz (1998) and Marshall, Mozrall, and Shealy (1999) used biaxial goniometers and torsimeters to continuously measure wrist motions and forearm rotations. However, in many cases, problems of cross talk and zero drift errors occurred. When the forearm rotates, the distal and proximal end blocks do not rotate together, causing twist in the goniometer wire. The twist is primarily the result of the kinesiology of forearm rotation, in which the proximal end block of goniometer is attached toward the middle of the forearm so that it rotates less than does the distal end block. The resulting twist leads to cross talk and zero drift errors. Such common measurement errors happen because of the complexity of human joints; they should be continually corrected. A summary of these measurement devices, with respective advantages and disadvantages is given in Table 1.

<table>
<thead>
<tr>
<th>Measurement device</th>
<th>Reference</th>
<th>Advantages</th>
<th>Disadvantages</th>
</tr>
</thead>
<tbody>
<tr>
<td>3D Digitizer</td>
<td>Logan &amp; Groszewski (1989)</td>
<td>Computer control</td>
<td>Problem with noise</td>
</tr>
<tr>
<td>Goniometer Electromechanical</td>
<td>Schoenmarklin &amp; Marras (1993)</td>
<td>(F/E, R/U, S/P) planes simultaneously</td>
<td>Uncomfortable</td>
</tr>
<tr>
<td>Biaxial, Torsiometers (Biometrics Ltd, BIOPAC Systems Inc.)</td>
<td>Nicole (1987)</td>
<td>Small, light, and Unobtrusive</td>
<td>Problem with noise</td>
</tr>
</tbody>
</table>

\(F/E = \) flexion/extension; \(R/U = \) radial/ulnar; \(S/P = \) supination/pronation.

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2.2. Calibration Methods

Calibration accuracy typically occurs at several levels: the accuracy of the instrument in the measurement of a known fixed angle, the application errors, repeatability of measurements, and correction procedures for any errors. Buchholz and Wellman (1997) constructed a calibration fixture to allow accurate measurement of true angles in one wrist plane for various forearm rotations. The fixture consisted of a protractor element for measuring true wrist angles and a rotary element for controlling forearm rotation. They also developed correction procedures that included a slope transformation and zero drift transformation to adjust for errors due to cross talk and zero drift. The equations of the slope-transformed values of flexion/extension \( F/E_S \) and radial/ulnar deviation \( R/U_S \) from goniometer measurements of flexion/extension \( F/E_M \) and radial/ulnar deviation \( R/U_M \) were:

\[
F/E_S = (F/E_M^2 + R/U_M^2)^{1/2} \left[ \sin^{-1} \left( \frac{F/E_M}{R/U_M} \right) + \phi \right],
\]
\[
R/U_S = (F/E_M^2 + R/U_M^2)^{1/2} \left[ \cos^{-1} \left( \frac{F/E_M}{R/U_M} \right) + \phi \right].
\]

The angle \( \phi \) is the twist in goniometer wire due to forearm rotation.

The second transformation of zero drift was made for corrected flexion/extension \( F/E_C \) and radial/ulnar deviation \( R/U_C \) angles from slope-transformed zero point averages of flexion/extension \( F/E_0 \) and radial/ulnar deviation \( R/U_0 \) angles as follows:

\[
F/E_C = F/E_S + F/E_0,
\]
\[
R/U_C = R/U_S + R/U_0.
\]

Based on the slope transformation and zero drift equations, they performed a nonlinear optimization minimizing the error (as seen below) and calculated true flexion/extension \( F/E_T \) and radial/ulnar deviation \( R/U_T \) angles.

\[
\min \sum (|F/E_T - F/E_C| + |R/U_T - R/U_C|)
\]

Hansson et al. (1996) introduced a test jig allowing independent setting of three planes of wrist angles, thus simulating the biomechanics of the wrist and forearm, and developed correction equations to induce actual wrist angles from recorded wrist angles as follows:

\[
F/E' = S \times \cos (\phi - C \times R),
\]
\[
R/U' = S \times \sin (\phi - C \times R),
\]
\[
S = (F/E^2 + R/U^2)^{0.5} \quad \text{and}
\]
\[
\phi = \sin(R/U) \times \arctan \left( \frac{F/E}{R/U} \right)
\]

where \( F/E' \) is the recorded flexion/extension angle; \( R/U' \) is the recorded radial/ulnar deviation angle; \( F/E \) is the actual flexion/extension angle; and \( R/U \) is the actual radial/
ulnar deviation angle. \( R \) is the forearm rotation angle and \( C \) is a transducer-dependent constant.

Ojima et al. (1991) measured the output of the electrogoniometer during wrist circumduction using a specially developed calibration apparatus. It consisted of a horizontal flat board and table, a universal joint at the end of the board, a metal straight bar joined to a universal joint, and a vertical board with a round hole and a sliding side bar. The results showed that the measurement errors for flexion/extension and radial/ulnar deviation were within 4°, and 1°, respectively, and that hysteresis was barely measurable, especially when subjects’ forearms were fixed to eliminate forearm rotation.

2.3. Static Assessment: Range of Motion

The static components of the wrist joint typically include postural information such as position and range of motion (ROM) measured in each plane of the wrist joint.

The range of motion (ROM) of body joints is obviously an important factor in assessment of body mobility (Webb Associates, 1978). Table 2 shows static ROM data for the wrist joint except for the data of Schoenmarklin and Marras (1993), which were measured dynamically.

These reported maximal wrist angles are very similar to each other. Generally, maximum wrist angles of flexion/extension are much larger than those of radial/ulnar deviation are, and maximum ulnar deviation and supination angles are much larger than maximum radial deviation and pronation angles, respectively. Wrist angles under static conditions are much larger than those found under dynamic conditions are, when measured from the beginning to the end of maximal dynamic movement. Schoenmarklin and Marras (1993) rationalized that the subjects might have focused more on the exertion and less on the maximal range of motion in dynamic movements.

Another consideration of static wrist joint measurements is forearm posture. Maximum wrist angles increase as the arm and shoulder muscles are used to rotate the forearm and hand. Therefore, when measuring the maximum range of motion in the wrist joint, the location and fixation of the arm and shoulder must be carefully constrained and documented.

2.4. Dynamic Assessment: Angular Velocity and Acceleration

The dynamic components of the wrist typically include angular velocity and angular acceleration. Unfortunately, these have not been studied in detail until fairly recently.

<table>
<thead>
<tr>
<th>Previous studies</th>
<th>Flexion</th>
<th>Extension</th>
<th>Radial deviation</th>
<th>Ulnar deviation</th>
<th>Pronation</th>
<th>Supination</th>
</tr>
</thead>
<tbody>
<tr>
<td>Boone &amp; Azen (1979)</td>
<td>76</td>
<td>75</td>
<td>19</td>
<td>33</td>
<td>—</td>
<td>—</td>
</tr>
<tr>
<td>Bonebrake et al. (1990)</td>
<td>86</td>
<td>62</td>
<td>34</td>
<td>68</td>
<td>105</td>
<td>120</td>
</tr>
<tr>
<td>Schoenmarklin &amp; Marras (1993)</td>
<td>62</td>
<td>57</td>
<td>21</td>
<td>28</td>
<td>81</td>
<td>101</td>
</tr>
</tbody>
</table>

TABLE 2. Maximum Range of Motion Data of the Wrist Joint

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Schoenmarklin and Marras (1993) first established an important database on maximum dynamic capability in the three planes of the wrist joint. Table 3 shows the maximum angular velocities and accelerations as a function of direction of movement.

Table 3 reveals that maximum angular velocity and acceleration of pronation/supination plane are much higher than those for flexion/extension are, and maximum angular velocity and acceleration of flexion/extension movements are much greater than those for radial/ulnar deviation movements are. The magnitude of dynamic wrist capability depends on the movement direction in each plane as shown in Table 3. Flexion movement (from extreme extension angle to flexion, E→F), ulnar movement (R→U), and supination movement (P→S) generate greater maximum angular velocities and accelerations than opposing movements. Schoenmarklin and Marras (1993) indicated that high dynamic capabilities of the flexion and supination movements were probably due to the greater biomechanical potentials of flexor and supinator muscles rather than of the extensor and pronator muscles, respectively. Also, the greater peak velocity and acceleration of ulnar movement were probably attributable to the effect of gravity.

Several studies have attempted to use the angular velocity and acceleration variable as potential risk factors in their epidemiological research (Hansson et al., 1996; Marklin & Monroe, 1998; Marras & Schoenmarklin, 1993; Serina, Tal, & Rempel, 1999). These values are summarized in Table 4.

Marras and Schoenmarklin (1993) performed a quantitative surveillance study on the factory floor. Based on 40 subjects from eight industrial plants, wrist deviation, angular velocity, and acceleration variables were measured in three planes of wrist movements using dichotomous WMSD risk levels (low and high risk). Table 4 indicates that the mean and maximum position, angular velocity, and acceleration values of high-risk tasks are generally much higher than those of low-risk tasks in all three planes of wrist movements. Besides, angular velocity and acceleration appear to separate WMSD risk levels more reliably than wrist deviation. The angular velocity and acceleration measures in high-risk tasks showed increases of 46.2% and 67.1%, respectively, over those in low-risk tasks. These results show the importance of dynamic components on assessing WMSD risk.

Hansson et al. (1996) investigated the position and angular velocity variables for tasks in the fish-processing industry as means for characterizing static and dynamic properties
<table>
<thead>
<tr>
<th>Previous studies</th>
<th>Variable</th>
<th>Task</th>
<th>Mean</th>
<th>Peak</th>
<th>Mean</th>
<th>Peak</th>
<th>Mean</th>
<th>Peak</th>
</tr>
</thead>
<tbody>
<tr>
<td>Marras &amp; Schoenmarklin (1993)</td>
<td>Position</td>
<td>High risk</td>
<td>−12.0</td>
<td>6.6</td>
<td>−6.7</td>
<td>4.7</td>
<td>8.3</td>
<td>47.4</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Low risk</td>
<td>−10.1</td>
<td>4.4</td>
<td>−7.6</td>
<td>10.1</td>
<td>2.5</td>
<td>37.4</td>
</tr>
<tr>
<td></td>
<td>Angular velocity</td>
<td>High risk</td>
<td>42</td>
<td>174</td>
<td>26</td>
<td>116</td>
<td>91</td>
<td>449</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Low risk</td>
<td>29</td>
<td>120</td>
<td>17</td>
<td>77</td>
<td>68</td>
<td>300</td>
</tr>
<tr>
<td></td>
<td>Angular acceleration</td>
<td>High risk</td>
<td>824</td>
<td>4,471</td>
<td>494</td>
<td>3,077</td>
<td>1,824</td>
<td>11,291</td>
</tr>
<tr>
<td></td>
<td></td>
<td>Low risk</td>
<td>494</td>
<td>2,588</td>
<td>301</td>
<td>1,759</td>
<td>68</td>
<td>300</td>
</tr>
<tr>
<td>Hansson et al. (1996)</td>
<td>Position</td>
<td>Fish processing industry</td>
<td>−1</td>
<td>n.a.</td>
<td>12</td>
<td>n.a.</td>
<td>n.a.</td>
<td>n.a.</td>
</tr>
<tr>
<td></td>
<td>Angular velocity</td>
<td>Fish processing industry</td>
<td>61</td>
<td>142</td>
<td>36</td>
<td>84</td>
<td>n.a.</td>
<td>n.a.</td>
</tr>
<tr>
<td>Marklin &amp; Monroe (1998)&lt;sup&gt;a&lt;/sup&gt;</td>
<td>Angular velocity</td>
<td>Bone trimming</td>
<td>45</td>
<td>239</td>
<td>30</td>
<td>156</td>
<td>100</td>
<td>540</td>
</tr>
<tr>
<td></td>
<td>Angular acceleration</td>
<td>Bone trimming</td>
<td>844</td>
<td>4,895</td>
<td>578</td>
<td>3,593</td>
<td>1872</td>
<td>12,522</td>
</tr>
<tr>
<td>Serina et al. (1999)</td>
<td>Position</td>
<td>Typing</td>
<td>−21.7</td>
<td>n.a.</td>
<td>−16.7</td>
<td>n.a.</td>
<td>86.8</td>
<td>n.a.</td>
</tr>
<tr>
<td></td>
<td>Angular velocity</td>
<td>Typing</td>
<td>24</td>
<td>n.a.</td>
<td>12</td>
<td>n.a.</td>
<td>13</td>
<td>n.a.</td>
</tr>
<tr>
<td></td>
<td>Angular acceleration</td>
<td>Typing</td>
<td>306</td>
<td>n.a.</td>
<td>134</td>
<td>n.a.</td>
<td>168</td>
<td>n.a.</td>
</tr>
</tbody>
</table>

*Note.*  *F/E* = flexion/extension; *R/U* = radial/ulnar; *S/P* = supination/pronation; n.a. = not available. “−” denotes extension and ulnar deviation.

<sup>a</sup>Data are maximum values among three intervals.
of wrist movements. The results indicated that wrist deviations in fish-processing tasks are much smaller than in the low-risk tasks reported by Marras and Schoenmarklin (1993). However, the angular velocities for F/E and R/U planes were 45% and 39%, respectively, higher than those of high-risk tasks.

Marklin and Monroe (1998) measured wrist motions in bone-trimming tasks using angular velocity and acceleration, and compared the results to those reported by Marras and Schoenmarklin (1993). Most bone-trimming tasks for both left and right hands fell in the high-risk category.

Serina et al. (1999) conducted a laboratory study to continuously measure wrist and forearm postures and motions while typing. The results indicated that mean angular velocities and accelerations of typing task were similar to those of industrial tasks reported by Marras and Schoenmarklin (1993).

In a clinical study, Ojima et al. (1991) obtained preliminary data on the angular velocity–wrist angle loci of healthy men. The loci represented the continuous change of angular velocity during wrist circumduction, and the distance from the origin to a locus means the angular velocity at the direction. The results showed that the angular velocity is faster in flexion/extension but slows at maximum flexion and extension angles during wrist circumduction. In addition, the loci of the healthy men were oval and the long axis of each locus was inclined to the radiodorsal-ulnovolar direction from ordinate, while the loci of patients were smaller than those of the healthy men.

In conclusion, these studies demonstrated that dynamic components of wrist motions such as angular velocity and acceleration are major contributing risk factors for work-related musculoskeletal disorders. Also, it is necessary to measure the wrist motion continuously and simultaneously in a dynamic state in the three planes of wrist motions (flexion/extension, radial/ulnar deviation, and pronation/supination) to extensively investigate the risk of WMSDs.

3. BIOMECHANICAL MODELS OF THE WRIST AND TENDONS

3.1. Wrist Mechanics

The seven main muscles involved in wrist and hand motion are flexor carpi radialis (FCR), flexor carpi ulnaris (FCU), flexor digitorum profundus (FDP), flexor digitorum superficialis (FDS), extensor carpi radialis brevis (ECRB), extensor carpi radialis longus (ECRL), and extensor carpi radialis ulnaris (ECU) (Garcia-Elias, Horii, & Berger 1991). The FCR, FCU, ECRB, ECRL, and ECU’s primary function is to move the wrist, while FDP and FDS are secondary wrist movers. The FDP and FDS’s primary function is to flex and extend the fingers and secondarily to rotate the wrist. The FDP and FDS pass through carpal tunnel. The primary muscles and tendons involved with wrist movements of flexion, extension, radial/ulnar deviation, and pronation/supination are listed below (An, Hui, Morrey, Linscheid, & Chao, 1981):

- **Flexion:** FCR and FCU
- **Extension:** ECRB, ECRL, and ECU
- **Radial deviation:** FCR, ECRB, and ECRL
- **Ulnar deviation:** FCU, and ECU
The parameters that have been commonly used to describe the muscles are muscle fiber length (FL) and physiological cross sectional (PCSA). Muscle length is related to mechanical potential for tendon excursion, and the maximum tension of the muscle to its PCSA (An, Horii, & Ryu, 1991). In moving the wrist, each tendon across the wrist joint slides a certain distance to execute the movement, and the tendon excursion and moment arm at various wrist joint angles can be measured and derived through experiments (Armstrong & Chaffin, 1978; An, Horii, & Ryu, 1991). Muscle parameters, tendon excursions, and moment arms at wrist joint (An et al., 1981, 1991; Lieber, Fazeli, & Botte, 1990) are summarized in Table 5. The magnitudes of tendon excursion were measured over a 100° range of motion in the flexion-extension plane and a 50° range of motion in the radial-ulnar deviation plane in the forearm neutral position.

In Table 4 it is shown that the FCR, FCU, and ECRB provide larger tendon excursion during flexion and extension movement than ECRL and ECU, while ECRL and ECU have greater tendon excursion during radial and ulnar deviation movement. The results also demonstrate that the FCR and FCU are prime muscles for flexion, ECRB for extension, ECRL for radial deviation, and ECU for ulnar deviation. Despite the three-dimensional orientation of the wrist tendons to the rotation axes and the complexity of carpal bone motion, the data in Table 5 indicates that the moment arms of wrist motion are maintained consistently; they correspond well with the anatomical location of the tendons. According to An et al. (1991), these findings are related to the anatomical considerations; the extensor retinaculum ensures a consistent relationship of the wrist extensors (ECRB, ECBL, and ECU) to the rotation axes, while the FCR is firmly fixed in the fibro-osseous groove, and the FCU infixed on the pisiform.

### 3.2. Static Tendon Pulley Models

Landsmeer (1960, 1962) developed the most comprehensive set of biomechanical models for finger flexor tendon displacements, in which the tendon-joint displacement relationships are determined by the spatial relationships between the tendons and joints. In the

### TABLE 5. Physiological and Mechanical Properties of Wrist Joint Muscles and Tendons

<table>
<thead>
<tr>
<th>Muscle and tendon</th>
<th>Physiological size</th>
<th>Tendon excursion (mm)</th>
<th>Moment arm (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Length (cm)</td>
<td>PCSA (cm²)</td>
<td>F/E plane</td>
</tr>
<tr>
<td>FCR</td>
<td>10.9~12.4</td>
<td>2.0</td>
<td>25 ± 4</td>
</tr>
<tr>
<td>FCU</td>
<td>15.2~15.4</td>
<td>3.2~3.4</td>
<td>28 ± 4</td>
</tr>
<tr>
<td>ECRB</td>
<td>13.8~15.8</td>
<td>2.7~2.9</td>
<td>20 ± 3</td>
</tr>
<tr>
<td>ECRL</td>
<td>11.8~18.3</td>
<td>1.5~2.4</td>
<td>12 ± 3</td>
</tr>
<tr>
<td>ECU</td>
<td>13.6~14.9</td>
<td>2.6~3.4</td>
<td>10 ± 2</td>
</tr>
</tbody>
</table>

Note. F/E = flexion/extension; R/U = radial/ulnar. FCR = flexor carpi radialis; FCU = flexor carpi ulnaris; ECRB = extensor carpi radialis brevis; ECRL = extensor carpi radialis longus; ECU = extensor carpi ulnaris. “+” denotes flexion and radial deviation; “−” denotes extension and ulnar deviation.
first model, he assumed that the tendon is held securely against the curved articular surface of the proximal bone of the joint, and the proximal articular surface can be described as a trochlea. Landsmeer showed that the tendon and joint displacement relationship for such a joint is described by Equation 1. However, if the tendon is not held securely, it may be displaced from the joint when the joint is flexed. Landsmeer (1960, 1962) considered this tendon configuration in his second model described by Equation 2:

\[ x = R_1 \theta \]  
\[ x = 2R_2 \sin \left( \frac{\theta}{2} \right) \]  

where \( x \) is the tendon displacement past the joint; \( R_1 \) is the distance from the joint center to the tendon (tendon moment arm); \( R_2 \) is the distance from the joint center to the geometric tendon constraint; and \( \theta \) is the angle of joint rotation from neutral position.

The biomechanical model for wrist joint proposed by Armstrong and Chaffin (1979) is a static model that is based on Landsmeer’s (1960, 1962) tendon model. Armstrong and Chaffin (1978) found that, when the wrist is flexed, the flexor tendons are supported by flexor retinaculum on the volar side of the carpal tunnel. When the wrist is extended, the flexor tendons are supported by the carpal bones. Thus, deviation of the wrist from neutral position causes the tendons to be displaced against and past the adjacent walls of the carpal tunnel. Based on the mechanical principles of LeVeau (1977), they assumed that a tendon sliding over a curved surface is analogous to a belt wrapped around a pulley as shown in Figure 1.

\[ F_R = 2 \times F_t \times \sin \left( \frac{\theta}{2} \right) \]

The forces acting normal to the tendon, tendon force, and radius are shown in Figure 1 and defined in equations as follow:

\[ F_L = \frac{F_t e^{\mu \theta}}{R} \]

where \( F_L \) is the supporting forces exerted on the tendon; \( F_t \) is the tendon force or belt tension; \( \mu \) is the coefficient of friction between tendon and supporting tissues; \( \theta \) is the wrist deviation angle (in radians); and \( R \) is the radius of curvature around supporting tissues.

According to LeVeau (1977), with the coefficient of friction \( \mu \) being small (0.0150), it can be approximated by zero. Therefore, the equation is changed to following equation:

\[ F_L = \frac{F_t}{R} \]

The above equation indicates that \( F_L \) is a function of the tendon force and radius of curvature. As the tendon force increases or the radius of curvature decreases, the normal supporting force exerted on tendon increases.

The resultant reaction force of the ligament and carpal bones, \( F_R \), depends on the tendon force \( F_t \) and the wrist deviation angle \( \theta \) as follows:

\[ F_R = 2F_t \sin\left(\frac{\theta}{2}\right) \]

The above equation indicates that \( F_R \) is independent of the radius of the curvature. As the tendon force and wrist deviation angle increase, the resultant force increases linearly. This model provides a relatively simple mechanism for calculating the normal supporting force exerted on tendons that are a major factor in WMSDs. But this model does not include the dynamic components of wrist movements such as angular velocity and acceleration, which might be risk factors in work-related musculoskeletal disorders.

### 3.3. Dynamic Tendon-Pulley Models

Schoenmarklin and Marras’ (1990) dynamic biomechanical model extended Armstrong and Chaffin’s (1979) static model to include the dynamic component of angular acceleration. The dynamic model is two-dimensional (2D) in that only the forces in the flexion and extension plane are analyzed. This model investigates the effects of maximum angular acceleration on the resultant reaction force that the wrist ligaments and carpal bones exert on tendons and their sheaths. The free body diagram and mass × acceleration diagram of the model are shown in Figure 2.

Figure 2 illustrates the reaction force on the center of the wrist \( (W_x \text{ and } W_y) \), the couple or moment \( (M_w) \) required to flex and extend the wrist, and the inertial force \( (M \times A_n \text{ and } M \times A_t) \) and inertial moment \( (I \times \ddot{\theta}) \) acting around the hand’s center of mass. In these relationships, the magnitude of moment around the wrist in FBD must equal the magnitude of moment acting around the hand’s center of mass in MAD. Therefore,
Thus, the hand is assumed to accelerate from a stationary posture, so, the angular velocity is theoretically zero, resulted in zero centripetal force $\frac{F_t}{R}$. Then,

$$F_t \times R = (M \times A_t + M \times A_n) \times D + I \times \ddot{\theta}$$

Thus, the hand is assumed to accelerate from a stationary posture, so, the angular velocity is theoretically zero, resulted in zero centripetal force ($A_n = \frac{V^2}{R} = 0$). Then,

$$F_t \times R = (M \times A_t) \times D + I \times \ddot{\theta}$$

$$F_t \times R = (M \times D \times \ddot{\theta}) \times D + I \times \ddot{\theta}$$

$$F_t = \frac{(M \times D^2 + I) \times \ddot{\theta}}{R}$$

$$F_R = 2 \times \left( \frac{(M \times D^2 + I) \times \ddot{\theta}}{R} \right) \times \sin \left( \frac{\theta}{2} \right)$$

(from $F_R = 2F_t \sin \left( \frac{\theta}{2} \right)$)

where $R$ is the radius of curvature of the tendon; $D$ is the distance between the center of mass of hand and wrist; $M$ is the weight of the hand; $I$ is the moment of inertia of the hand in flexion and extension; $\ddot{\theta}$ is the angular acceleration; and $\theta$ is the wrist deviation angle.

The above equations indicate that the resultant reaction force, $F_R$, is a function of angular acceleration, the radius of curvature, and wrist deviation. Thus, exertion of the wrist and hand with greatly angular acceleration and deviated wrist angle would result in greater total resultant reaction forces on the tendons and supporting tissues than exertions with small angular acceleration and neutral wrist position. According to Armstrong and Chaffin (1979), increases in resultant reaction force would increase the supporting force that the carpal bones and ligaments exert on the flexor tendons, therefore increasing the chance...
of inflammation and risk of CTS. Therefore, these results might provide theoretical support to why the angular acceleration variable can be considered a risk factor of WMSDs. The advantage of Schoenmarklin and Marras’ (1990) model is that it does include the dynamic variable of angular acceleration into assessment of resultant reaction force on the tendons. But the model is two-dimensional, and it does not consider the coactivation of antagonistic muscles in wrist joint motions. This points to the need for further model developments to account for additional physiological factors.

3.4. Complex Tendon Models

Any model that incorporates more than the one muscle-tendon unit of the above models, all of sudden, becomes much more complicated because the number of unknown muscle forces exceeds the number of equilibrium or constraint equations. This is known as the statically indeterminate problem. The two main approaches utilized in solving this problem are reduction methods and optimization methods.

3.4.1. Reduction methods. The main objective of the reduction method is to reduce the number of excessive variables until the number of unknown forces are equal to the number of required equilibrium equations eliminating static indeterminacy.

Smith, Juvinall, Bender, & Pearson (1964) initiated a mathematical analyses of the finger tendon forces to find the effects of the flexor tendons acting on a metacarpal phalangeal (MP) joint deformed by rheumatoid arthritis. They used a 2D model to analyze the MP joint and muscle forces of the index finger during tip pinch. To reduce the number of unknown muscle forces, the following assumptions were applied (a) the sum of the interosseous (I) forces is treated as a single force 2I; (b) half of the interosseous forces of I act at the proximal interphalangeal (PIP) joint and the other half act at the distal interphalangeal (DIP) joint; and (c) the lumbrical (L) is much smaller than the interosseous (I), as much as 1/3I. They solved the three moment equations using these assumptions and anthropometrical data of the index finger obtained from a cadaver hand in a tip pinch position. They reported the tendon forces normalized to the external force P, as 3.8P, 2.5P, 0.9P, and 0.3P for the FDP, FDS, I, and L, respectively. They also found a value of 7.5P for the MP joint force. The results indicate that the flexor tendons are dominant and the forces are many times larger than the intrinsic muscle forces during tip pinch.

Chao, Opgrande, and Axmear (1976) presented a comprehensive analysis of the three-dimensional (3D) tendon and joint forces of the fingers in pinch and power grip functions. Kirschner wires (K-wires) were drilled through the phalanges to fix the finger configuration in the desired position and different surgical wires were inserted into the tendon and muscles of hand specimens of the cadaver to identify different tendons on x-ray film. The exact orientations of finger digits and the locations of the tendons were defined by bi-planar x-ray analysis. Through a free-body analysis, 19 independent equations were obtained for 23 unknown joint and tendon forces. Using the permutation-combination principle of setting any four of the nine tendon forces equal to zero solved the indeterminate problem. The selection of these tendons was based primarily on electromyographic responses (EMG) and physiological assessment. They found that high constraint forces and moments at the DIP and PIP joints were found during pinches, whereas large magnitudes of constraint forces at the MP joint were found during power grips. The total of the intrinsic muscles (RI, UI, and LI) produced a greater force than the total of the flexor tendons (FDP and FDS) in both pinch and power grip actions.
3.4.2. **Optimization methods.** An alternate method using a typical optimization technique was suggested by Seireg and Arvikar (1973) and Penrod, Davy, and Singh (1974). In this approach, force equilibrium equations and anatomical constraint relationships were used for the equality constraints and the physiological limits on the tendon, muscle and joint forces were applied as the inequality constraints. In addition, the most important factor in this method is optimal criteria that correspond to the objective function of the formulation. The possible solutions can vary based on the optimal criteria selected.

Chao and An (1978a) studied the middle finger during tip pinch and power grip actions, with an aid of three-dimensional analysis. They analyzed the same problem using the optimization and linear programming (LP) technique of Chao et al. (1976) instead of the previously described EMG and permutation-combination method. The predicted middle finger muscle and joint forces were very similar to those of the previous study (Chao et al., 1976), except for the intrinsic muscle forces whose predicted values were considerably lower. They found that the highest joint contact forces for all three joints occurred for pinch grip rather than power grip. They also found that the main flexors (FDP and FDS) were most active in both pinch and power grip functions, whereas the intrinsic muscles were less active in power grip than in pinch.

An, Chao, Cooney, and Linscheid (1985) also applied LP optimization techniques to solve the indeterminate problem of a 3D analytic hand model. The ranges of muscle forces of the index finger under isometric hand functions, such as tip pinch, lateral key pinch, power grip, and other functional activities were analyzed. The FDP and FDS carried high tendon forces compared with other muscles in most hand functions, although the predicted FDS force was zero in a pinch grip. The long extensors (LE) and two intrinsic muscles contributed large forces in the key pinch. The large force of these intrinsic muscles in pinch action can be explained by the role of these muscles maintaining balance and stabilization of the MP joint. The joint constraint forces for each finger were also studied. The Chao et al. (1976) study showed a trend for joint constraint forces in which the DIP joint had the lowest force and the force progressively increased for the PIP joint and was largest at the MP joint. An et al. (1985) showed the same trend in lateral pinch functions.

3.4.3. **Combined approaches.** Chao and An (1978b) used a graphical presentation with a combined permutation/combination and optimization technique to solve the statically indeterminate tendon force problem. They analyzed the maximum tip pinch force of the index finger as a function of external force directions (0°, 30°, and 45°) and the DIP joint flexion angles (10° to 50°). The results showed that the pinch strength relied on the direction of applied external force as well as the finger joint configuration. The tendon forces of the index finger were also studied with the same finger posture as that in the Chao et al. (1976) study but only one angle (45°) of the external force was assumed. Also, the predicted extrinsic extensor tendon force was considerably larger than in their previous studies.

Weightman and Amis (1982) presented a good critical review for the previously published studies and applied their 2D finger model to the analysis of resultant joint forces and muscle tensions in various pinch actions. To create a statically determinate problem, all joints were assumed pin joints with a fixed center of rotation during flexion. The relationships of the intrinsic muscle forces were assumed identical to those of Chao and An (1978a), except that the long extensor muscles forces dropped to zero. They also used the physiological cross-sectional area (PCSA) of the muscles to define...
the force distributions in the intrinsic muscles. Their results compared to other previously published studies with a good correlation of both muscle and joint force predictions. Based on these comparisons, they verified that a 2D finger model could be valid for analyzing 2D finger actions, even though realistically any finger motion is still three-dimensional.

### 3.5. Direct Measurement Validation Studies

Directly measured tendon forces under isometric finger function were first reported by Bright and Urbaniak (1976). They developed a strain gauge to measure the tendon forces in both tip pinch and power grip actions during operative procedures. Flexor tendon forces were found to be in the range of 4.0 to 20.0 kg and 1.25 to 15.0 kg for the flexor digitorum profundus (FDP) and flexor digitorum superficialis (FDS), respectively, in power grip action, while 2.5 to 12.5 kg for the FDP and 1.0 to 7.5 kg for the FDS in pinch action. Because they directly measured the tendon forces only, they did not report the actual applied pinch and power grip force and the ratio of tendon force to the externally applied force.

Schuind et al. (1992) directly measured the flexor tendons (flexor pollicis longus [FPL], FDP, and FDS) during various finger functions. They developed an s-shaped tendon force transducer and measured the flexor tendon forces in pinch and power grip functions. Also, a pinch dynamometer was used to record the applied loads in pinch action. The tendon forces showed proportionality to the externally applied forces. To compare their results with the previously published mathematical finger models, they normalized their tendon forces, as a ratio of the tendon force to the applied forces. In tip pinch, the ratios were 3.6P, 7.92P, and 1.73P for the externally applied force P for the FPL, FDP, and FDS, respectively. In lateral pinches, the ratios were 3.05P, 2.9P, and 0.71P for the FPL, FDP, and FDS, respectively. Although the FDP and FPL showed high forces during tip and lateral pinch, the maximal values recorded are probably on the lower side of the potential forces and it could be explained by the significantly weaker pinch and power grip forces during carpal tunnel surgery due to the denervation or partial anesthesia of the sensory area of the median nerve. However, the magnitude of tendon forces was similar to values reported by Bright and Urbaniak (1976), although direct comparison is not possible because the applied force was not recorded in their study.

In another in vivo tendon force measurement study, Dennerlein, Diao, Mote, and Rempel (1998) measured only FDS tendon forces of the middle finger at three finger postures, which ranged from extended to flexed pinch postures using a gas-sterilized tendon force transducer (Dennerlein, Diao, Mote, & Rempel, 1997) and a single axis load cell (Green-Leaf Medical Pinch Meter, Palo Alto, CA). The investigation was centered upon the average ratio of the FDS tendon tension to the externally applied force. The average ratio ranged from 1.7P to 5.8P, with a mean of 3.3P, in the study. Tip pinches with the DIP joint flexed were also studied with the tendon-to-tip force ratio being 2.4P. These ratios were compared with the results of their own three finger models as well as other contemporary published isometric tendon force models. These ratios were larger than those of other studies. The average values were also slightly higher than that (1.73P) of Schuind et al.’s (1992) in vivo tendon force measurement study. It was found that the tendon force ratios and muscle strength varied substantially from individual to individual, although the ratio of force from tendon to tendon was relatively constant within the same limb for all studies (Brand, Beach, & Thompson, 1981; Dennerlein et al., 1998; Ketchum et al., 1978). A summary of these tendon force models is given in Table 6.
4. CRITICAL EVALUATION AND DISCUSSION

Although the intrinsic muscles are more active in pinch action than in power grip action, the relative magnitudes of the main flexor tendon forces (such as FDP and FDS) are usually high in both actions. These in vivo tendon forces of the flexors are presented in Table 7 based on the previous studies.

In general, these averages and ranges of tendon forces are very similar with a few exceptions. Schuind et al. (1992) showed lower FDS tendon forces in power grip action than those of other types of grips. The tendon force ranges of Brand et al. (1981) show

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TABLE 6. Summary of Tendon Force Models

<table>
<thead>
<tr>
<th>Model</th>
<th>Reference</th>
<th>Key features</th>
</tr>
</thead>
<tbody>
<tr>
<td>Static</td>
<td>Landsmeer (1960, 1962)</td>
<td>Simplest tendon pulley model</td>
</tr>
<tr>
<td></td>
<td>Armstrong &amp; Chaffin (1978, 1979b)</td>
<td>Pulley model with tendon force</td>
</tr>
<tr>
<td>Dynamic</td>
<td>Schoenmarklin &amp; Marras (1990)</td>
<td>Pulley model with acceleration</td>
</tr>
<tr>
<td>Complex tendon forces</td>
<td>Reduction</td>
<td>2D, scaled tendon forces</td>
</tr>
<tr>
<td></td>
<td>Smith et al. (1964)</td>
<td>3D, some forces set equal to zero</td>
</tr>
<tr>
<td></td>
<td>Smith et al. (1964)</td>
<td></td>
</tr>
<tr>
<td>Optimization</td>
<td>Seireg &amp; Arvikar (1973)</td>
<td>Objective function with constraints</td>
</tr>
<tr>
<td></td>
<td>Penrod (1974)</td>
<td>Objective function with constraints</td>
</tr>
<tr>
<td></td>
<td>Chao &amp; An (1978)</td>
<td>3D, linear programming</td>
</tr>
<tr>
<td></td>
<td>An et al. (1985)</td>
<td>3D, linear programming</td>
</tr>
<tr>
<td>Combined</td>
<td>Chao &amp; An (1978)</td>
<td>Graphical, optimization</td>
</tr>
<tr>
<td></td>
<td>Weightman &amp; Amis (1982)</td>
<td>Some F = 0, some F ∝ muscle area</td>
</tr>
<tr>
<td>In vivo studies</td>
<td>Bright &amp; Urbaniak (1976)</td>
<td>Strain gage</td>
</tr>
<tr>
<td></td>
<td>Schuind et al. (1992)</td>
<td>S-shape tendon force transducer</td>
</tr>
<tr>
<td></td>
<td>Dennerlein et al. (1998)</td>
<td>Gas-sterilized tendon force transducer</td>
</tr>
</tbody>
</table>

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TABLE 7. In Vivo Tendon Forces (kg) of Flexor Digitorum Profundus (FDP) and Flexor Digitorum Superficialis (FDS)

<table>
<thead>
<tr>
<th>Finger configuration</th>
<th>FDP</th>
<th>FDS</th>
</tr>
</thead>
<tbody>
<tr>
<td>Brand et al., 1981</td>
<td>14.9 (13.5–17.0)</td>
<td>10.4 (4.5–17)</td>
</tr>
<tr>
<td>Ketchum et al., 1978</td>
<td>5.7 (5.27–6.18)</td>
<td>6.12 (3.73–7.63)</td>
</tr>
<tr>
<td>Bright et al., 1976</td>
<td>2.5–12.5</td>
<td>1.0–7.5</td>
</tr>
<tr>
<td></td>
<td>4.0–20.0</td>
<td>1.25–15.0</td>
</tr>
<tr>
<td>Schuind et al., 1992</td>
<td>8.3 (2.0–12.0)</td>
<td>1.9 (0.3–3.5)</td>
</tr>
<tr>
<td></td>
<td>4.0 (1.9–6.4)</td>
<td>0.6 (0.0–0.9)</td>
</tr>
</tbody>
</table>

*Average tendon forces for all fingers.

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similar magnitudes with Bright and Urbaniak (1976) power grips. Ketchum et al. (1978) and Bright and Urbaniak (1976) tip pinch action show similar ranges of tendon forces. Schuind et al. (1992) represented the significant differences between FDP and FDS tendon forces in both pinch and power grip, whereas the others showed that the force of FDP tendon was only slightly larger than that of the FDS tendon. These discrepancies can be explained by the different finger postures utilized in each study because each finger could have various functional muscle capacities depending upon its joint configuration (Chao & An, 1978b).

In all these in vivo tendon force studies, the muscle and tendon forces were proportional to the externally applied forces. However, these predicted maximum tendon forces are probably lower than the true potential forces because these experiments were performed during carpal tunnel surgery under local anesthesia in the median nerve innervation area. In such cases, the muscles are partially inactive and produce lower pinch and power grip forces. To normalize these tendon forces, the ratios of the tendon force to the applied force, FDP to FDS, and joint forces are studied for both pinch and power grip functions (Tables 8 and 9).

Average ratios of tendon forces to the applied forces in the tendon force prediction models were, for an external force of P, $3.5P (SD: 0.74)$, $1.8P (SD: 1.03)$, and $3.67P (SD: 2.29)$ for the FDP, FDS, and I (intrinsic) tendons in pinch, while $3.14P (SD: 0.29)$, $3.48P (SD: 0.72)$, and $11.4P (SD: 6.6)$ were for FDP, FDS, and I tendons in power grip, respectively. Generally, all data agreed with high contributions of flexor tendons (FDP and FDS) for both pinch and power grip actions, although intrinsic tendons showed high variations among those data. The average ratios of FDP to FDS were also obtained, $2.92:1$ and $0.93:1$ for pinch and power grip, respectively. These data showed the significant strength contribution of the FDP tendon to the pinch, as opposed to the equal contributions of these two flexors in the overall power grips.

Although they did not measure the externally applied force with the tendon forces in a power grip, Schuind et al. (1992) attempted to validate these mathematical solutions experimentally. They only used the pinch dynamometer for measuring the amount of the applied force for pinch functions. Thus, the mean tendon forces were applied for validating power grip functions in this study.

In pinch actions, Schuind et al. (1992) reported the higher ratio of FDP to the applied force ($7.92P$) than the result ($3.5P$) of mathematical tendon force prediction models. The FDS ratio to the applied force ($1.73P$), however, was similar to the average ratio ($1.8P$) in the finger model studies. Because of the large force measurement for the FDP tendon, a higher ratio of FDP/FDS ($4.6P$) was presented in their study than that of finger model studies. Dennerlein et al. (1998) also found higher ratios of the FDS tendon to the applied force ($3.3P$) than those of mathematical finger model studies.

In power grip actions, there were no in vivo tendon force data for the ratio of tendon force to the externally applied force. Thus, only the FDP:FDS ratio of direct measurement study can be used for the comparison with the finger model studies. They presented a $6.67:1$ ratio based on their mean tendon forces of FDP and FDS. The force of FDP showed significantly larger than that of FDS in a direct measurement study, whereas both FDP and FDS showed similar contributions to power grip ($3.14P$ for FDP and $3.48P$ for FDS) and a FDP:FDS ratio of $0.93:1$ was also calculated in finger force prediction models. The variability of these results may be expected because not all researchers used the same finger characteristics: moment arms, finger configurations, and angles of the applied forces to the fingertip or pulp area.

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<table>
<thead>
<tr>
<th>Grip type</th>
<th>Muscle force</th>
<th>Joint force</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>FDP</td>
<td>FDS</td>
</tr>
<tr>
<td>Bright et al., 1976</td>
<td>Tip</td>
<td>2.5–12.5 kg(^a)</td>
</tr>
<tr>
<td>Schuind et al., 1992</td>
<td>Tip</td>
<td>7.92</td>
</tr>
<tr>
<td></td>
<td>Lateral</td>
<td>2.90</td>
</tr>
<tr>
<td>Dennerlein et al., 1998</td>
<td>Tip</td>
<td>—</td>
</tr>
<tr>
<td>Smith et al., 1964</td>
<td>Tip</td>
<td>3.8</td>
</tr>
<tr>
<td>Chao et al., 1976</td>
<td>Tip, radial, ulnar</td>
<td>4.32</td>
</tr>
<tr>
<td>Chao &amp; An, 1978</td>
<td>Tip(^c)</td>
<td>3.3</td>
</tr>
<tr>
<td>Chao &amp; An, 1978</td>
<td>Tip</td>
<td>3.97</td>
</tr>
<tr>
<td>Weightman &amp; Amis, 1982</td>
<td>Tip</td>
<td>2.4(^b)</td>
</tr>
<tr>
<td>An et al., 1985</td>
<td>Tip</td>
<td>1.93–2.08</td>
</tr>
</tbody>
</table>

*Note.* FDP = Flexor digitorum profundus; FDS = flexor digitorum superficialis; DIP = distal interphalangeal joint; PIP = proximal interphalangeal joint; MP = metacarpal phalangeal joint. \(^a\)Tendon force. \(^b\)Average value for various finger configurations. \(^c\)Middle finger tip pinch.
<table>
<thead>
<tr>
<th></th>
<th>Muscle force</th>
<th>Joint force</th>
</tr>
</thead>
<tbody>
<tr>
<td>Finger</td>
<td>FDP</td>
<td>FDS</td>
</tr>
<tr>
<td>Bright et al., 1976</td>
<td>4.0–20.0(^a)</td>
<td>1.25–15.0(^a)</td>
</tr>
<tr>
<td>Schuind et al., 1992</td>
<td>4.0(^b)</td>
<td>0.6(^b)</td>
</tr>
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<td>Index</td>
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<tr>
<td></td>
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<tr>
<td></td>
<td>Little</td>
<td>3.37</td>
</tr>
<tr>
<td>Chao and An, 1978</td>
<td>Index</td>
<td>3.37–3.47</td>
</tr>
<tr>
<td>An et al., 1985</td>
<td>Index</td>
<td>3.17–3.47</td>
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</tbody>
</table>

*Note.* FDP = Flexor digitorum profundus; FDS = Flexor digitorum superficialis; DIP = distal interphalangeal joint; PIP = proximal interphalangeal joint; MP = metacarpal phalangeal joint. \(^a\)Tendon force, unit: kg. \(^b\)Mean tendon forces, unit: kg.
There are controversial issues for the functions of tendons (intrinsic muscles vs. flexors) during pinch and power grip actions in biomechanical finger models. Based on the solutions from the three moment equations, Smith et al. (1964) found that flexor tendons usually carry larger forces than other intrinsic muscle during tip pinch. Chao and An (1978b) also supported this result in their study. They showed the flexors were most active and produced high tendon forces in both pinch and power grip actions. However, Chao et al. (1976) and An et al. (1985) suggested contradictory results for the contributions of intrinsic muscles in finger actions. They presented higher contributions of intrinsic muscles than those of flexors did in pinch and power grip functions. In general, although the magnitude of the intrinsic muscle force was less than that of the flexors, the intrinsic muscles were more active in pinches than in power grip. An et al. (1985) also agreed with high intrinsic muscle forces in pinches and explained it by the need for these intrinsic muscles to balance and stabilize the large MP joint forces.

Most of these studies showed similar trends for joint forces. Small constraint forces and moments were seen at both the DIP and PIP joints, while the constraint forces and moments were considerably higher at the MP joint in both actions. DIP- and PIP-joint forces of the power grip actions were relatively lower than were those of the pinch actions. It may be why hands are more adaptable in performing powerful grip actions rather than with pinches: It is more difficult to maintain the proper stability requirements at the distal joints (Chao et al., 1976).

5. CONCLUSIONS

Through these 2D or 3D models, many researchers have tried to understand how the externally applied forces are transmitted across the finger joint to the internal tendons of the human hand and how these tendon forces relate to the applied forces. Generally, most studies agreed with high contributions of flexor tendons (FDP and FDS) in power grips and good proportionality tendon forces to the external forces. However, generally, the in vivo direct flexor tendon forces and the FDP to FDS ratios were found to be larger than those predicted from the finger models.

Despite the relatively low tendon forces predicted in biomechanical finger models, FDP and FDS tendon forces were 3.14–3.5 times and 1.8–3.48 times larger than the applied forces in pinches and power grips, respectively. Thus, large and repeated tendon forces can be a contributory factor in tendon disorder, especially in hand-intensive tasks. In addition, these tendon forces vary because of the finger joint configurations or finger postures utilized in each task. Therefore, using the properly designed hand tools, which optimize the hand posture to minimize tendon forces can reduce and prevent work-related musculoskeletal disorders.

The essential part of the finger model is the basic assumption to simplify the finger mechanism. From a mathematical point of view, the simpler the finger, the simpler the finger model. However, these simplifications of the finger mechanism can lead to the misleading information on the joint and tendon forces calculated from the finger model. In addition, most of these finger models have been developed for clinical and rehabilitation purposes (Chao et al., 1976; Cooney & Chao, 1977; Toft & Berme, 1980) rather than for industrial applications.

In summary, four critical needs must be met to promote a better understanding of WMSD risk factors:
1. Better measurement devices for wrist dynamics.
2. More accurate validation studies of internal tendon forces.
3. More valid and accurate biomechanical tendon models.
4. More industrial applications.

REFERENCES


