

Finger Flexor Tendon Forces Are a Complex Function of Finger Joint Motions and Fingertip Forces

Jack Tigh Dennerlein, PhD

*Department of Environmental Health
Harvard School of Public Health
Boston, Massachusetts*

ABSTRACT: In vivo tendon forces provide a view inside the musculoskeletal system revealing muscle function and potential injury etiologies. The studies presented here measured the in vivo tendon force of the flexor digitorum superficialis of the long finger during open carpal tunnel release surgery in ten adult patients. Forces were measured during passive movement of the finger, isometric pinch, and dynamic tapping of the finger. The tendon forces during passive movement of the finger were the largest with the finger fully extended. During isometric pinch, tendon force was linearly related to fingertip force, and was on average 3.3 times larger than the fingertip force. During dynamic activities, however, the relationship between tip and tendon force was nonlinear and often remained elevated when the finger was moving but with no applied force. Tendon forces were the highest with the isometric finger pinch. In conclusion, tendon force is a completed function of both fingertip load and motion of the joints that the tendons cross. A comparison of these results with others published in the literature indicated that rehabilitation processes need to incorporate a systems approach rather than rely on one specific physiologic relationship to minimize finger flexor tendon forces. *J HAND THER.* 2005;18:120-127.

Force is a fundamental mechanism associated with injury. If the forces in a tissue exceed the tolerance of the tissue, then the tissue will fail and be injured such as an excessive strain, tear, or fracture. The concept is simple enough. Most activities of daily life are well within these tolerances for most healthy adults. However, injuries are still quite prevalent. For example, in one prospective study of computer workers, 50% of all new employees at computer-intensive jobs experienced musculoskeletal disorders or discomfort.¹ Submaximal forces can create damage in muscles² and tendons,^{3,4} indicating smaller tolerance levels than those associated with acute trauma. One hypothesized model for these injuries is that the rate of repair of the tissue lags behind the rate of tissue damage.^{5,6} This assumes that the tissues of the body undergo a damage-repair process during exposure to submaximal repetitive and sustained loads.

During rehabilitation following surgery to the tendons of the upper extremity, forceful actions can

also rupture repair sutures at levels below maximal forces. Rupture failure of surgical tendon repairs have been studied in vitro⁷ and in animal models.⁸ These tensile forces at which failure occurs vary greatly depending upon the suturing technique used and other factors, such as the curvature of a tendon as it travels around and through a bent wrist.⁷

The question is, therefore, what are the direct loads on these internal tissues for simple activities associated with injury and also during activities associated with rehabilitation from surgery? Fundamental biomechanical and physiological models can be used to predict the internal loads associated with activities. However, external forces and motions are not uniquely related to exposure of the internal tissues (e.g., muscles, tendons, bones) to stresses.⁸ External forces and motion cause internal exposure through interactions with musculoskeletal elements, which can be complex and vary across individuals. They include both passive and dynamic mechanisms. Many models do not account for passive force elements of the musculoskeletal system.⁹ Furthermore, the dynamics of segmented systems are inherently nonlinear, and the potential for differences in internal state (i.e., tendon forces) could result in differences in internal exposure even if the external exposure is the same.¹⁰

Correspondence and reprint requests to Jack Tigh Dennerlein, PhD, Department of Environmental Health, Harvard School of Public Health, 665 Huntington Avenue, Boston, MA 02115.

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Therefore, alternative methods to determine the forces, role, and function of muscles have incorporated direct in vivo measure of tendon forces in both human and animal studies.¹¹ Direct measures bypass many of the limitations associated with modeling efforts; however, they are quite invasive and difficult to perform on humans. Combining measurements of force with kinematic data, such as muscle length, can define specific function and role of muscles through various terrestrial activities through biomechanics that define the energy and mechanical work completed by a muscle throughout a repetitive cycle.¹²

The upper extremity is quite prone to injury associated with submaximal activities. Repetitive submaximal work with the hands has been associated with carpal tunnel syndrome, which may be the result of forces transferred from the finger flexor tendons or inflammation and swelling of the tendons, compressing the nerves that travel through the carpal tunnel. The finger flexors are hence of interest to study. Furthermore, they are quite accessible during open carpal tunnel release surgery.

Therefore, we undertook studies to examine the in vivo tendon force of a human finger flexor muscle, the flexor digitorum superficialis (FDS), with the goal of defining the forces during simple activities. These activities included passive movement of the finger, isometric tip pinch of the finger, and dynamic tapping simulating typing. The main objectives of these studies were primarily descriptive in nature, but they did test hypotheses relating the external tip forces and the internal tendon force. The relationships were, as expected, complex and depended upon other factors such as posture and kinematics. While many of these results have been presented elsewhere,^{13,14} they are brought together here in one article in the context of injury prevention and rehabilitation from tendon injury. Specific issues and implications for rehabilitation procedures will be discussed.

MATERIALS AND METHODS

Tendon force transducers come in many forms and technologies, from image-based to implantable force transducers (see Ravey et al.¹¹ for a review of different technologies and their advantages and disadvantages). Buckle transducers (BT) are commonly used implantable force transducers, which are usually mounted onto tendons distal to and in series with the muscle. BTs, similar to a belt buckle, encompass a complete tendon, and are based on systems developed to measure forces in cable stays for antennae tower supports and sailboats. Buckle transducers work by displacing the path of the tendon (Figure 1). As the tension of the tendon increases the tendon wants to follow a straight path, but the buckle transducer prevents this. As the tension increases, the tension on the tendon increases its three-point loading on the transducer. The long sides of the transducer are hence loaded as a beam of a bridge. Strain gauges mounted on this bending-beam provide an electronic measurement of the force.

To mount the BT onto the tendon, the tendon is looped up through the frame and the removable fulcrum is placed underneath the tendon. The idea is very similar to a belt buckle, where the strap goes under then over and then back under the buckle of a standard belt. Unlike a belt where the one end of the belt is loose and is thus threaded through the buckle, the ends of the tendon are fixed. Instead of being fixed, the fulcrum is first removed to loop the tendon up through the rectangular frame and then replaced inside the loop between the tendon and the frame. Buckle transducers will lengthen the muscle-tendon unit; however, their design is to minimize this lengthening and is usually only a fraction of a percent change.

The calibration of buckle transducers depends upon the geometry of the tendon within the transducer, and because tendons vary in shape and size,

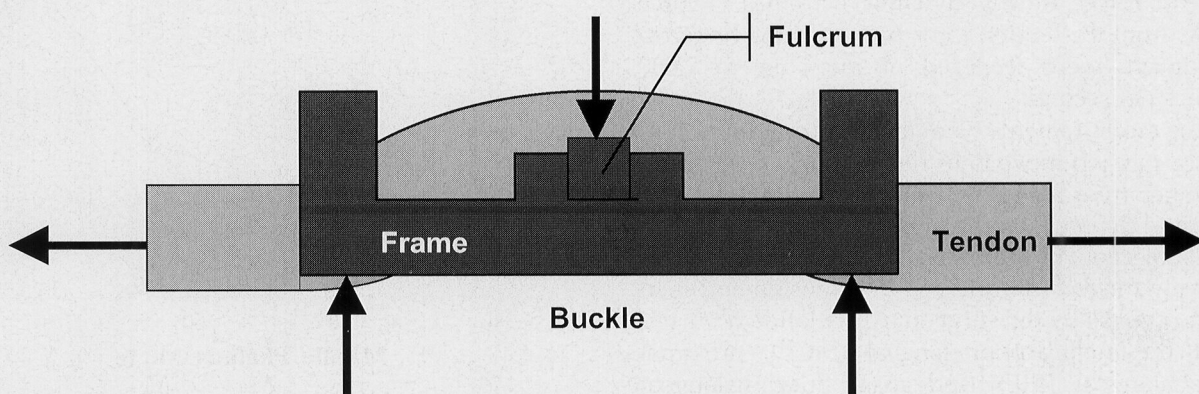


FIGURE 1. A side view of a buckle transducer with tendon illustrating the geometry and mechanics. The buckle transducer works by displacing the path of the tendon. As the tendon tension increases it wants to straighten, but the buckle prevents this. As a result, the transducer is loaded at three points bending the beam, which has strain gauges mounted to measure the strain.

the calibration varies with each tendon measured.¹⁵ Ideally, calibrating each tendon/buckle combination would be preferred. For animal studies, this is standard practice. After an experiment, the tendon is harvested with the transducer still mounted, and the transducer is calibrated through direct loading of the tendon. This is unacceptable for human studies. Calibration of a tendon is inferred based upon the thickness of the tendon as it resides in the transducers.^{15,16}

For our experiments, the *in vivo* tendon force of the flexor digitorum superficialis of the long finger was measured within ten human participants (nine female, one male) during their prescribed open carpal tunnel release surgery performed at the University of California, San Francisco. The University of California, San Francisco Committee on Human Research approved the procedures and consent forms. The measurements were made during open carpal tunnel surgery and utilized the buckle transducer described in Dennerlein et al.¹⁵ Surgery was performed, as usual, with the participants under local anesthesia at the incision site. The contents of the carpal tunnel were exposed through a 5-cm longitudinal carpal tunnel incision. The FDS tendon of the longer finger was identified and a gas-sterilized tendon force transducer was then mounted onto the FDS. The tendon thickness was then measured *in situ* to determine the transducer calibration factor. With the upper arm abducted at the shoulder and the elbow fully extended, the subject's forearm was rotated 90 degrees from full supination toward a neutral forearm posture, with the thumb upward and the palm open facing toward the subjects' feet as they lay prone on the operating table. A video camera mounted above the surgical field recorded the sagittal view of the finger posture at 30 frames per second. A load cell mounted on a rigid surface measured fingertip force for isometric conditions (Greenleaf Medical™ Pinch Meter, Palo Alto, CA). An instrumented keycap containing a custom-designed load cell measured fingertip forces during dynamic activities.¹⁷ Force signals from the tendon transducer and the fingertip transducers were recorded at rates up to 2,000 samples per second.

Force measurements were made during three conditions: passive movement flexion and extension of the finger (two subjects), isometric pinch (ten subjects), and dynamic tapping on a computer keyswitch (five subjects). For the passive forces, the finger was passively moved from flexion to extension and then back to neutral by the surgeon. Articulation occurred about the metacarpophalangeal (MCP), proximal interphalangeal (PIP), and distal interphalangeal (DIP) joints with the wrist fixed in a neutral position. The sum of the MCP and PIP joint angles derived from the video images indirectly represented the excursion of the FDS tendon.

For the isometric contraction condition, subjects gradually (~10 seconds) increased the fingertip force from 0 to 10 N (~1,000 g) and then decreased it to 0 N using a visual monitor of the fingertip force to guide them. For the dynamic tapping tasks, subjects tapped on a keyswitch mounted on a rigid surface at a rate of one tap per second. The key was connected to a speaker that emitted an audio signal when it was pressed. Subjects were instructed to minimize contact time with the keycap, and hence the duration of the audio feedback.

RESULTS

Passive forces of the finger flexor (FDS) muscle increased as the finger extended and decreased as the finger flexed (Figure 2). The maximum tendon forces observed were below 10 N and occurred when the finger joints were fully extended. The tendon showed some level of force for most of the range of motion of the finger. Forces were below 2 N (~0.5 lb) when the middle phalanx is flexed more than 60 degrees compared to the metacarpal bone.

For the isometric pinch, tendon forces increased, and then decreased monotonically with fingertip forces (Figure 3). With the fingertip force increasing to 10 N, the tendon force reached maximum levels up to 70 N (~7 kg); however, the maximum tendon forces varied across subjects. The ratio of tendon force to fingertip force averaged within a subject ranged from 1.7 to 5.8 (average = 3.3, SD = 1.4).

For the dynamic tapping task, the maximum tendon forces ranged from 8 to 17 N (average = 13, SD = 3 N [~3 lb]). The ratio of the maximum tendon force to fingertip force ranged from 3.5 to 6.8 (aver-

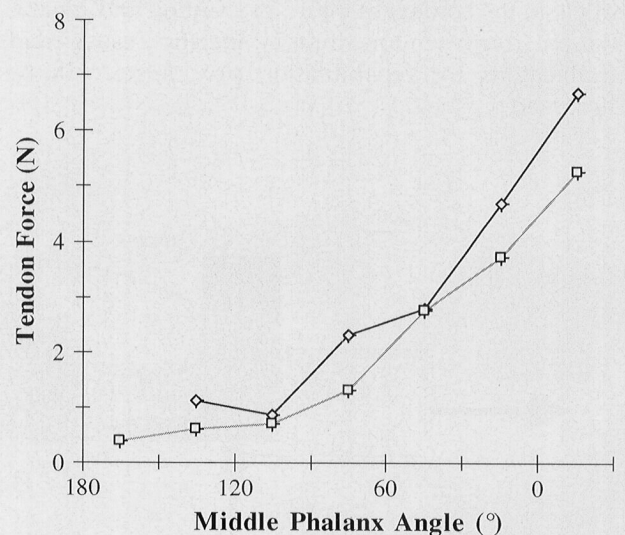


FIGURE 2. Passive muscle-tendon force of two subjects. As the finger extends, the passive force of the FDS increases (Modified from Dennerlein et al.¹⁴). The angle of the middle phalanx relative to the metacarpal bone is used as a surrogate measure for FDS muscle length.

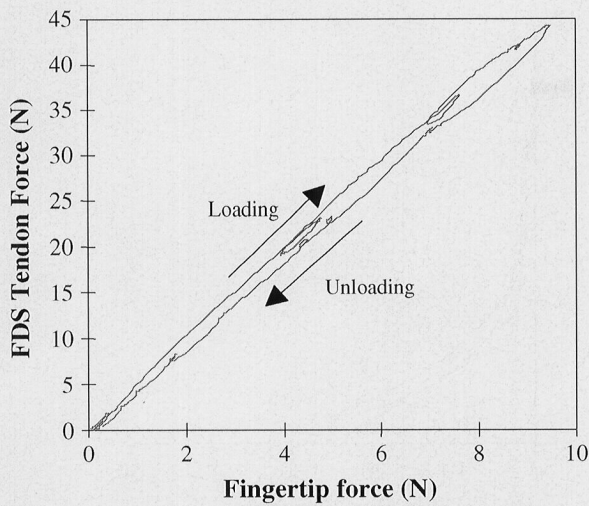


FIGURE 3. Isometric tendon force versus fingertip force. The relationship is linear and one-to-one, with little to no hysteresis. (Reprinted with permission from Dennerlein et al.¹³)

age = 5.4, SD = 1.4). Unlike the isometric task, the relationship between tendon and fingertip force differed between the loading and unloading portions of the keystroke (Figure 4). During loading, the tendon force increased with fingertip force. During unloading, however, the force of the tendon remained elevated as the fingertip force was removed. Tendon forces were often elevated during unrestrained motions; however, they seemed to vary with fingertip motion. For example, the fingertip of one subject followed a simple up/down motion (Figure 5). When the tendon force was plotted versus fingertip position, we observed a simple concentric contraction with the fingertip moving in the flexion direction as

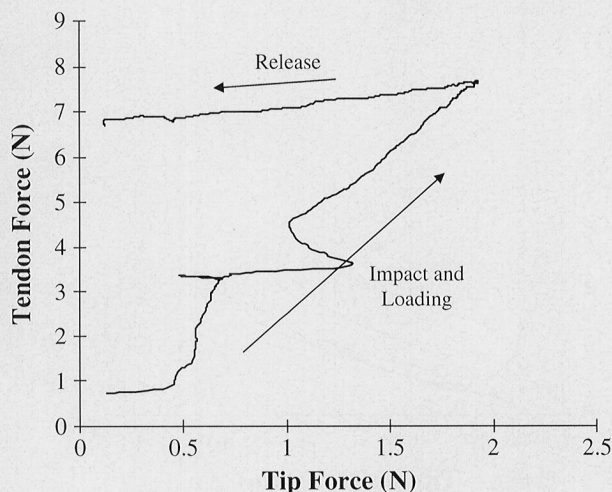


FIGURE 4. Dynamic finger tapping. The relationship here is complex and not simply one-to-one; rather, the tendon load often remains elevated while the fingertip unloads.

the force in the tendon increased. When contact with the keyswitch was achieved, the fingertip stopped moving, and the contraction became isometric. A very different type of force-fingertip position was observed for another subject (Figure 6). The tendon force was often elevated before and after the fingertip force associated with the key strike. The fingertip also moved in a different pattern extending before and after the key strike. The forces here are often associated with the fingertip being extended and are similar in nature to the forces observed during passive extension of the finger (Figure 2).

DISCUSSION

The goal of these in vivo tendon force measurements was to provide descriptive nature of the forces that exist in the finger during three simple activities. In summary, we observed that tendon forces are a complicated function of fingertip force and motion. Hence, the development of rehabilitation process needs to approach minimizing tendon forces with a systematic approach. This article presents only three components of tendon forces illustrating the complex relationship.

First, the passive muscle forces measured in the tendon reached levels of 8 N (~1.8 lb) with the finger fully extended, but required significant passive flexion to reach zero forces. There was the expected passive force, which increased with lengthening of the muscle through extension of the MCP and PIP joints. The relationship is not purely linear, illustrating the expected exponential increase with the lengthening of the flexor muscles through extension of the MCP and PIP joints. With the middle phalanx flexed 60 degrees or more, the muscle forces are less than 50% of the flexor force observed at full extension.

Second, during pinch activities, the forces in the finger flexor tendon can be as large as five times the force applied at the fingertip. For example, in some cases when someone created a tip pinch requiring 10 N of force, the resulting tendon forces was observed to reach levels of 50 N (5.1 kg). This ratio varied greatly across the ten subjects of this study. These variations appeared to be related to different motor control strategies utilized across subjects.¹⁸ Dennerlein et al.¹³ accounted for these differences by the different postures of their subjects, where the DIP joint was either flexed (creating a tip pinch posture and lower tendon-to-tip force ratio) or extended (creating a pulp pinch and a higher tendon-to-tip force ratio). In that study, the tip pinch posture was thought to lead to a more balanced distribution of forces between the FDS and the flexor digitorum profundus (FDP); however, the difference in forces of the FDP between postures was not investigated in these studies.

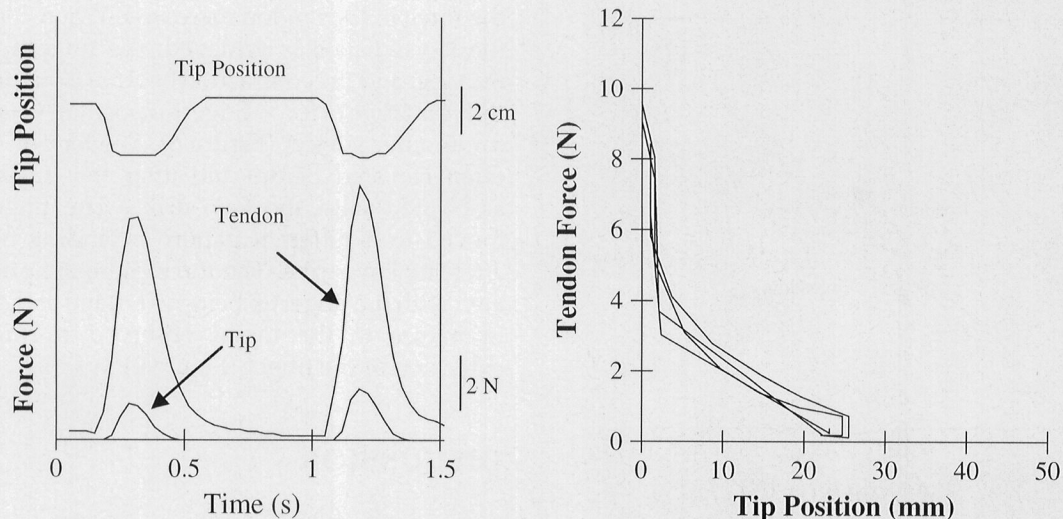


FIGURE 5. This subject displayed a motion pattern where the fingertip pushed the keyswitch. Tendon force simply increased as the motion of the finger went down and then decreased as the fingertip released the key.

Third, during dynamic finger activities requiring both unrestrained motion and force generation to press a keyswitch, such as during typing, the tendon forces are complicated and involve both passive and active muscle force components. During the fingertip force-generation portions of a key strike, the tendon forces can reach levels of five or six times the fingertip force. The finger flexor tendon reached levels as high as 20 N (~4 lb). During the unrestrained motions of the tap, the tendon force was observed to reach levels as high as 4 N (~1 lb). These force patterns varied with fingertip motion patterns (Figures 5 and 6).

Other studies have also examined the finger flexor tendon forces, and the forces observed here are in good agreement with their findings (Table 1). A large

limitation of our previous studies is that the force of only one flexor tendon was measured (i.e., the FDP was not examined). However, the other two studies did examine the forces of the FDP.^{19,20} These studies also illustrate the complex relationships between finger tendon forces and joint motions. Schuind et al.,¹⁹ for example, examined the passive forces of the finger tendons as the wrist joint was manipulated, illustrating that some wrist movements create passive forces in the finger flexor tendons. Kursa²⁰ examined unrestrained finger movements with the wrist in two postures to observe the interaction of the two joints on finger flexor forces.

The goal of this article is to use these in vivo tendon force measurements to guide rehabilitation methods

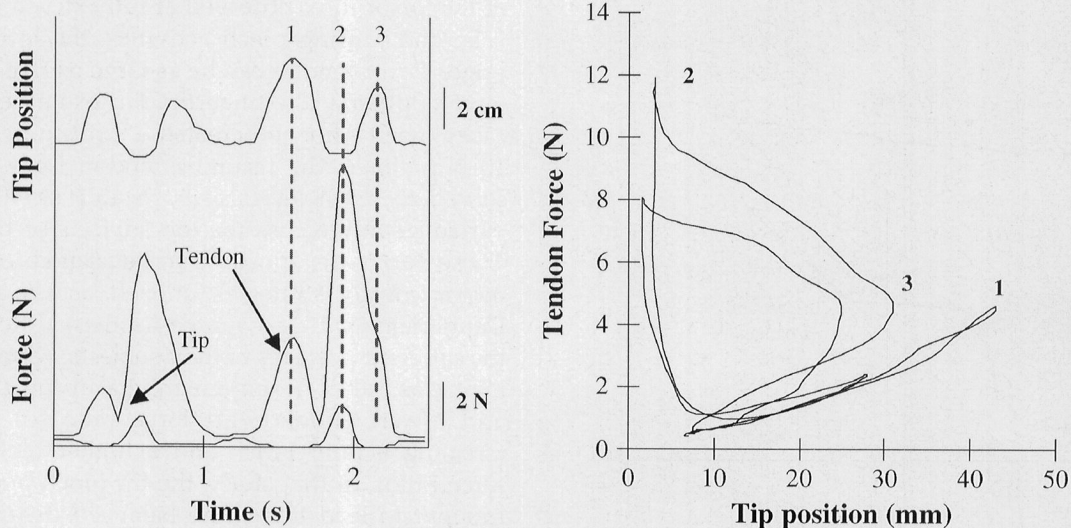


FIGURE 6. This motion pattern of one subject is different than that of the subject shown in Figure 5. The subject pulls the fingertip back away from the keyswitch before depressing the key, which is accompanied by an increase of tendon force that looks a lot like passive muscle forces. The fingertip then decelerates downward and then lifts up again after the key strike. The numbers in the figure on the right denote the vertical lines in the figure on the left.

TABLE 1. Comparison of Different Finger Flexor Tendon Force Measurement Studies

	<i>Schuind et al., 1992</i>	<i>Dennerlein et al., 1998, 1999</i>	<i>Kursa, 2004</i>
Passive tendon force			
FDS	0-3 N	0.3-8 N	
FDP	0-6 N		
Isometric tendon-to-tip force ratio			
FDS	1.7 ± 1.5	3.3 ± 1.4	1.6 ± 0.9
FDP	7.9 ± 6.3		2.2 ± 0.75
FDS tendon force during tapping		8-16 N (4.6 ± 1.7) ratio	
Unrestrained (wrist neutral)			
FDS	0-3 N	0-4 N*	5 ± 6 N
FDP	0-28 N		7.6 ± 5 N
Unrestrained (wrist flexed)			
FDS			9 ± 13 N
FDP			7 ± 5 N

*Recorded during tapping.

FDS = flexor digitorum superficialis; FDP = flexor digitorum profundus.

after surgical repair of the flexor tendons. Published tensile strengths (i.e., the force at which the tendon and suture rupture) range from 30 to 80 N depending on the suture technique used.²¹⁻²⁵ Muscle mechanics and animal models suggest that the wrist be flexed to minimize passive forces of the finger flexor muscle.²⁶ We observed here that the passive forces are associated with a neutral wrist are well below these levels. If no force is desired, then flexing the wrist will achieve the goal; yet, it should be noted that we did not examine the effects of wrist posture on passive finger forces.

If the rehabilitation process requires unrestrained movement of the finger, which is common to prevent adhesion to surrounding tissues during the healing process, then flexing the wrist increases the forces observed in the tendon.²⁰ During unrestrained movements of the fingers with the wrist flexed, the forces of the finger flexors can reach levels as high as 16 N (~3-4 pounds). This is twice as high as 8 N with the wrist straight. This is due in part to overcoming the increased passive forces of stretching the finger extensors, which increase in length with wrist flexion. With their increased length, their passive forces increase. Other connective tissues may play a role in these observed phenomena. To move the finger, the flexors have to increase their forces to overcome the passive force of these antagonist muscles, which increase with wrist flexion. Furthermore, tensile strength of repair decreases when they travel and bear load across a pulley with a small radius of curvature.⁷

Based on these observations, we recommend that the wrist be in a neutral posture when finger movement exercises are employed to minimize forces in the tendons. As an alternative, only extension movement exercises can be used if the wrist is in a flex posture. This will move the flexor tendons, but the driving force will be from the extensors rather than from the flexors themselves. The movement may not be enough to control adhesion formation, but the forces will remain low. Nonetheless, appropriate

rehabilitation methods need to consider the complete biomechanical system of the musculoskeletal systems along with the goals of the rehabilitation.

These data illustrate that measuring force alone does provide insight into the relationships involving external force and internal forces; however, internal forces often exist without direct external forces and depend on the motion of the finger. To fully understand the dynamic components and the role of a muscle during specific tasks, one has to relate muscle forces with the muscle length changes and muscle activity to access the full mechanical context of muscle function. The work of Biewner et al.¹² illustrates these points. For example, they described the relationship between force and muscle length changes for two different types of locomotive muscles, the avian pectoralis muscle of a pigeon used for flight and the plantaris muscle of the wallaby used for hopping. These two muscles operate at very different points on the force-velocity curve. For the wallaby plantaris muscle, the force generated is fairly isometric, with very little change in the length of the muscle. Therefore, the muscle's output in this case is force and it outputs very little power. For the avian pectoralis muscle, the muscle operates with force being generated as the muscle shortens, operating out on the force-velocity curve where power output is often maximized. Therefore, the pectoralis muscle is associated more with outputting power. As a result, the energetics of the movement and locomotion are very different. For the flexors of the finger during a keystroke, the muscles of the fingers work in both areas of the force-velocity curve. During the downswing of a keystroke, the muscle shortens and operates in the middle of the force-velocity curve (Figure 7). Once the fingertip impacts the bottom of the keystroke, the muscle stops shortening and the muscle is now in an isometric contraction, creating more force.

In vivo force measurements of the human finger flexor tendons are difficult and quite invasive, as one can imagine, which creates several limitations. The

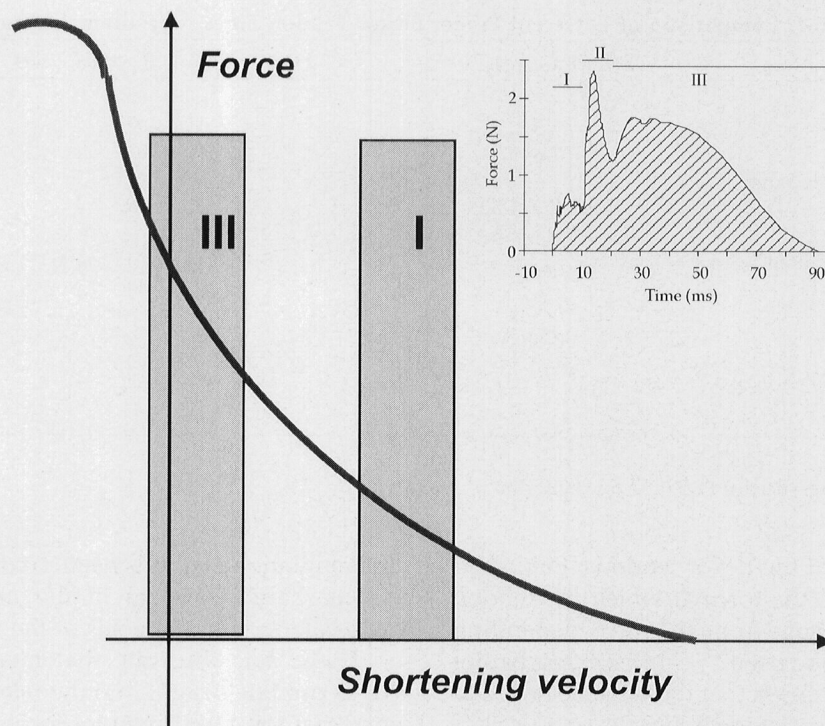


FIGURE 7. The force–velocity relationship for a shortening muscle and the hypothesized operating points of a finger flexor muscle during a key strike. During the unrestrained movement, the muscle shortens and operates with reduced force (Phase I from Rempel et al.¹⁷). As the fingertip impacts the key, the muscle–tendon stops shortening and the operating points move up the force–velocity curve force levels associated with isometric forces. As a result, we see an increase in the fingertip force.

difficulties range from simply lengthening the muscle–tendon unit to taking measurements during a surgical procedure where the contents of the carpal tunnel are exposed through the incision. Voluntary control of the finger is also impaired by the awkward postures and the use of local anesthesia. These make testing specific measurements about rehabilitation difficult. However, based on these measurements, we can infer specific trends observed and gain insight about general procedures. It is difficult to measure friction forces *in vivo* because we usually require measuring forces in two places on a tendon, usually distally and proximal of a joint during movement. That means more incisions are needed. We rely on animal models to assess friction forces associated with different repair processes.²⁷ There are also limits in terms of time of the processes and data collection associated with the device. Nonetheless, there is still a great amount of data that can be collected for different rehabilitation hypotheses.

CONCLUSION

In vivo tendon force measures provide a view inside the musculoskeletal system illustrating the complex relationships between external exposures of forces and motions to the internal tissue loads. Tendon forces vary greatly for both unrestrained

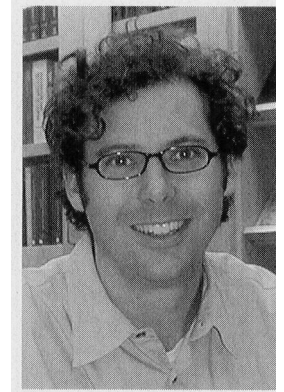
motions and isometric activities, increasing in force from passive movements at the lowest levels and pinch activities at larger activities forces. As a result of the complex relationships, rehabilitation processes after tendon repairs require creative approaches that minimize forces in the tendons to a greater degree than what the literature suggests. The relationships of applied force and motion with the internal loads are complex and require a systematic approach to fully understand them. The force in a tendon is not a singular unit in the musculoskeletal system; rather, it works in concert with other internal tendons, and recommendations for rehabilitation need to integrate these different concepts.

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Jack Tigh Dennerlein, PhD, is currently Associate Professor of Ergonomics and Safety at the Harvard School of Public Health in Boston, Massachusetts, where he has been a member of the Department of Environmental Health's faculty since 1999. He obtained his PhD in mechanical engineering from the University of California, Berkeley in 1996 for his work in the biomechanics of office ergonomics. Previously he received a BS degree from the State University of New York at Buffalo (1986) and an MS degree from MIT (1989). Dr. Dennerlein's research goals are to understand the neuromuscular biomechanics and physiology of work-related musculoskeletal disorders and traumatic injuries, and to improve approaches to exposure assessment and epidemiology for work-related injuries.



Susan V. Duff, EdD, OTR/L, PT, CHT. Susan's clinical expertise is in pediatric upper limb rehabilitation. She is particularly interested in how prehension develops and how prehensile skill can be fostered after injury. Susan received her doctorate in movement science from the Department of Biobehavioral Sciences at Teachers College, Columbia University. She currently is a postdoctoral fellow with Robert L. Sainburg, PhD, OTR, in the Department of Kinesiology at Pennsylvania State University.

David Elliot, MA, FRCS, is currently the Senior Hand Surgeon in the St. Andrew's Centre for Plastic Surgery, Chelmsford, UK. He is the Editor of the *British Journal of Hand Surgery* and President of the British Society for Surgery of the Hand (2005). He is a past winner of the BSSH Pulvertaft prize. He is a past Examiner of the FESSH Diploma, President of the British Association of Hand Therapists, Chairman of the Flexor Tendon Committee of the IFSSH, Foundation Visitor of the Royal Australasian College of Surgeons, and Venkataswami Orator of the Indian Society for Surgery of the Hand.



Roslyn B. Evans, OTR/L, CHT, is the owner and director of Indian River Hand and Upper Extremity Rehabilitation in Vero Beach, Florida. She has developed innovative ideas and published clinical studies on early motion for extensor and flexor tendon injuries, conservative management of the trigger finger, and postoperative Dupuytren's fasciectomy, and has published chapters on wound healing and carpal tunnel, flexor, and extensor tendon injuries. Her clinical studies have been awarded the R.L. Petzolt Prize by the American Society of Hand Therapists (ASHT) three times, and the Best Scientific Paper, the Nathalie Barr Lectureship, and the Lifetime Fellowship Award by ASHT. She was the first recipient for the American Foundation Research Award of the American Hand Therapy Foundation. She has served on the editorial board of the *Journal of Hand Therapy* since 1990, the editorial board of the *Yearbook of Hand Surgery* from 1990 to 2001, and has edited two special issues for the *Journal of Hand Therapy*.