Moment Arm and Force-Generating Capacity of the Extensor Carpi Ulnaris After Transfer to the Extensor Carpi Radialis Brevis

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Tendon transfers to the extensor carpi radialis brevis (ECRB) are often performed to augment wrist extension. This study was conducted to analyze how transfer of the extensor carpi ulnaris (ECU) to the ECRB affects the moment arms, force-generating capacity, and moment-generating capacity of the ECU over a range of wrist flexion-extension. A graphics-based computer model was developed from anatomic measurements of the muscle–tendon paths before and after transfer. This model calculates the lengths and moment arms of the muscles over a range of wrist flexion-extension and represents the muscles' force-generating characteristics from previous measurements of their physiologic cross-sectional areas, fiber lengths, and pennation angles. Analysis of the computer model revealed that the maximum isometric extension moment of the ECU at the neutral wrist position increased from 0.50 N-m to 1.72 N-m after transfer to the ECRB. The deviation moment shifted from 2.72 N-m ulnar deviation to 1.42 N-m radial deviation. The extension moment generated by the ECU varied more with wrist flexion angle after transfer due to its broadened operating range on the muscle force–length relationship. The simulations highlight the need for proper intraoperative tensioning of the ECU to maximize the force-generating potential of the transferred muscle over the functional range of motion. (J Hand Surg 1999;24A:1083–1090. Copyright © 1999 by the American Society for Surgery of the Hand.)

Key words: Biomechanics, muscle, tendon transfer, surgery simulation strength.

Tendon transfers are often performed to improve functional capacity of the hand and wrist after spinal cord injury. The restoration of voluntary active grasp, comprising both palmar and lateral prehension, is one of the major goals of these procedures.[1] In higher-level quadriplegics, voluntary control is often limited but can be augmented through functional electrical stimulation (FES) of muscles with intact lower motor neurons.[2,3] Tendon transfers may be used in conjunction with FES to simplify the control of movement and to compensate for the lost function of muscles with lower motor neuron damage.[1,3–5]

Finger flexion is a critical component of lateral pinch.[6] To provide a stable platform for digital flexion, strong active wrist extension is necessary.[7] Two studies have demonstrated that pinch strength depends on available wrist extension moment.*[6,8] Wrist extension is typically aided by transfer to the extensor carpi radialis brevis (ECRB) because of its

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* The moment generated by a muscle is equal to the product of its force and its moment arm.

large extension moment arm, and the extensor carpi ulnaris (ECU) has been suggested as a good candidate for transfer if its function is not severely impaired.[9,10] Keith et al[3] have reported the effects of this transfer, combined with electrical stimulation, on the moment generated by the ECU at the neutral wrist position in one spinal cord injury patient. However, the effects of this procedure on the moment-generating capacity of the ECU throughout its operating range of motion remain unclear.

The objective of this study was to analyze how the transfer of the ECU to the ECRB affects the moment arms, force-generating capacity, and resulting moments generated by the ECU over a range of wrist flexion-extension. A computer model of the wrist was developed that accurately represented the musculoskeletal geometry and enabled a detailed theoretical examination of changes that arise from the tendon transfer. This computer model also provides a basis for an inquiry into the elements of a successful tendon transfer design.

Materials and Methods

We developed a computer model of the wrist[11] to examine how muscle architecture and moment arms affect the moments generated by the muscles. This model specifies bone geometry, joint kinematics, and origin-to-insertion paths of all the major muscles crossing the wrist. In this model, muscle–tendon paths are represented as a series of line segments delineated by the origins, insertions, and intermediate points that represent anatomic constraints. The moment arms, lengths, and force-generating capacities of individual muscles may be calculated over the full range of flexion-extension. The location of points that define the muscle–tendon paths may be altered interactively using a computer graphics workstation to simulate the changes that arise from a tendon transfer.[12,13]

To accurately represent the musculoskeletal geometry of the transfer surgery, the procedure was performed and muscle–tendon paths were digitized on a fresh-frozen cadaver. An incision was made from the lateral epicondyle of the humerus to the middle of the third metacarpal. The skin was reflected to expose the full length of the wrist extensors for digitizing, but the fascia was left intact to maintain normal anatomic constraints on the muscles. Before transferring the ECU, steel tacks were inserted every 2 cm along the paths of both the ECU and the ECRB from origin to insertion to mark the points to be digitized. An indentation was made in the head of each tack to facilitate repeatable positioning of the digitizer tip. The elbow was maintained at 90° flexion and the wrist remained in 0° flexion and 0° deviation (the "neutral" position), with fingers naturally in slight flexion. After measuring preoperative geometry, the ECU was freed from its insertion and surrounding fascia for the distal 10 cm. Without length alteration, the ECU was transferred to 5 cm proximal from the insertion of the ECRB and attached using the Pulvertaft side-weave technique.[14]

A SpaceArm digitizer (Faro Inc, Lake Mary, FL) coupled to a Macintosh II computer (Apple Computer Inc, Cupertino, CA) was used to collect the 3-dimensional coordinates that described the muscle paths before and after transfer. Calibration tests determined that the position of the probe's tip could be resolved to within 1 mm. The base of the digitizer was attached to the countertop on which the specimen rested. Before and after digitizing the muscle paths, anatomic landmarks on the lateral epicondyle of the humerus and on the radial and ulnar styloid processes were digitized to establish a reference frame for the upper extremity and to confirm that the specimen remained immobile during the experiment.

Digitized coordinates describing the paths of the ECU and the ECRB before transfer and of the ECU after transfer were transformed into the reference frames of the upper extremity model (Fig. 1). To ensure proper orientation and location of the muscle points, the digitized anatomic landmarks were matched to corresponding locations on the computer model. Coordinates were not scaled due to the similarity of the dimensions of the specimen and the model. Several points distal to the wrist did not correspond exactly with the model's bone geometry, necessitating the definition of new points in this region. Each point was attached to the appropriate bony reference frame. Origins were attached to the humerus; insertions were attached to the third and fifth metacarpals, as appropriate; and intermediate points were attached to the radius, ulna, and carpal bones.

Flexion-extension of the wrist was represented as 2 revolute joints to account for the motion of the radiocarpal and midcarpal joints.[15] The approximate axes of rotation were at the center of the lunate for the radiocarpal joint and at the center of the capitate for the midcarpal joint. The motion of radial–ulnar deviation was also represented as a dual revolute

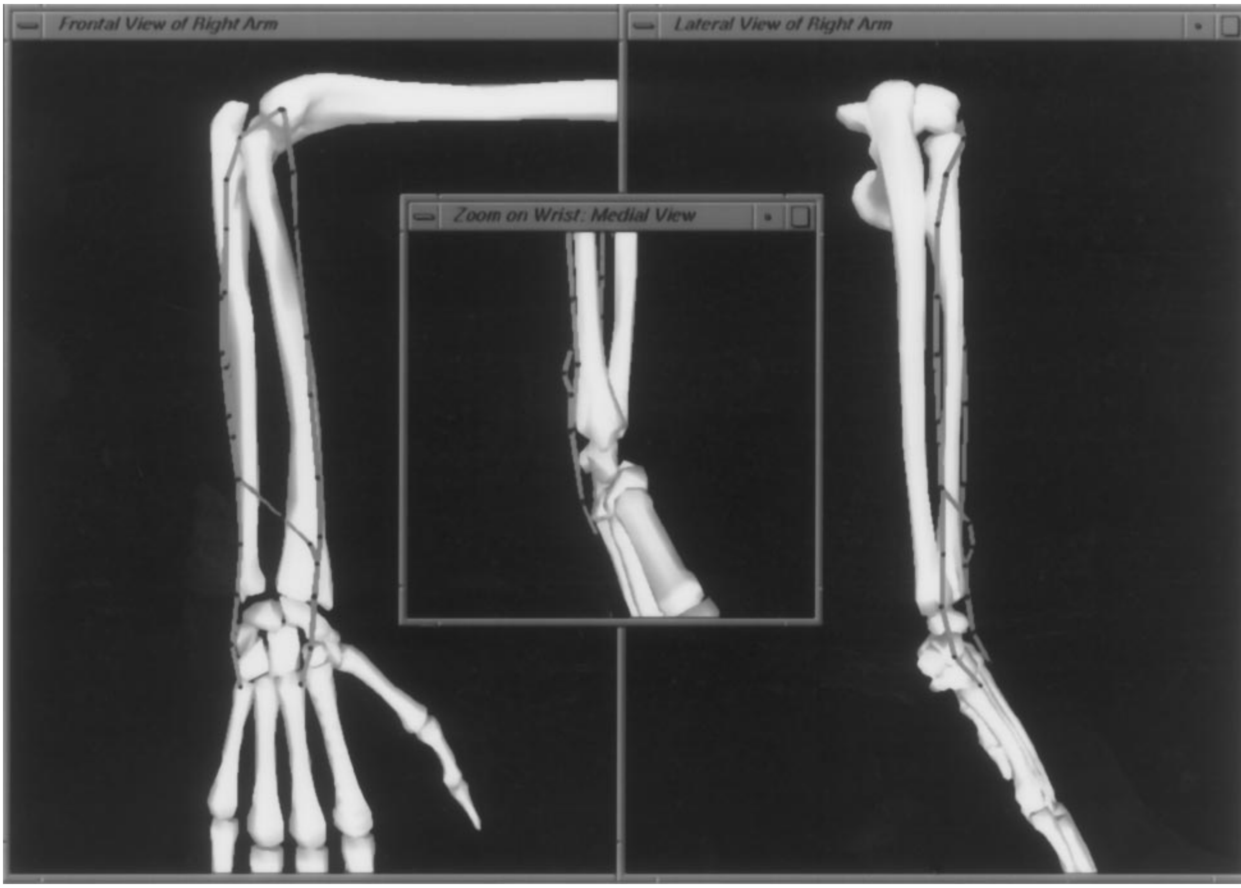

Figure 1. Biomechanical model of the ECRB and the ECU before and after transfer to the ECRB. The left window shows a frontal view in the neutral wrist position, the inset shows a medial view (wrist in 25° flexion), and the right window shows a lateral view (wrist in 25° extension).

joint, with axes of rotation between the lunate and the scaphoid for the proximal joint and at the center of the capitate for the distal joint. The ranges of motion of the model were from 70° flexion to 70° extension and from 25° radial deviation to 35° ulnar deviation.

The maximum force generated by each muscle over a range of motion was determined from measurements of the muscles' physiologic cross-sectional areas, fiber lengths, and tendon lengths.[13,16] The physiologic cross-sectional areas, fiber lengths, and tendon lengths for the ECU and the ECRB were taken from detailed anatomic measurements,[17,18] as reported previously.[11] The physiologic cross-sectional area and fiber length of the transferred ECU were not changed from the normal values. The tendon length of the transferred ECU was adjusted in the model so that the force generated by the ECU peaked with the wrist at 0° flexion. Simulations were also performed with the tendon length increased by 1 cm (a "slack" transfer) and decreased by 1 cm (a "tight" transfer).

The model was used to calculate muscle moment arms, maximum isometric forces, and maximum isometric moments generated by the muscles. Moment arms of the ECU and the ECRB were computed for both flexion-extension and radial–ulnar deviation as the change of muscle–tendon length with joint angle.[13,19,20] The maximum isometric force generated by a muscle over a range of flexion was estimated by assuming the muscle was maximally activated and calculating the force corresponding to each flexion angle. The maximum moment generated by each muscle was calculated by multiplying its moment arm–joint angle relationship and its force–joint angle relationship.

Results

The maximum extension moment generated by the ECU at the neutral wrist position increased from 0.50

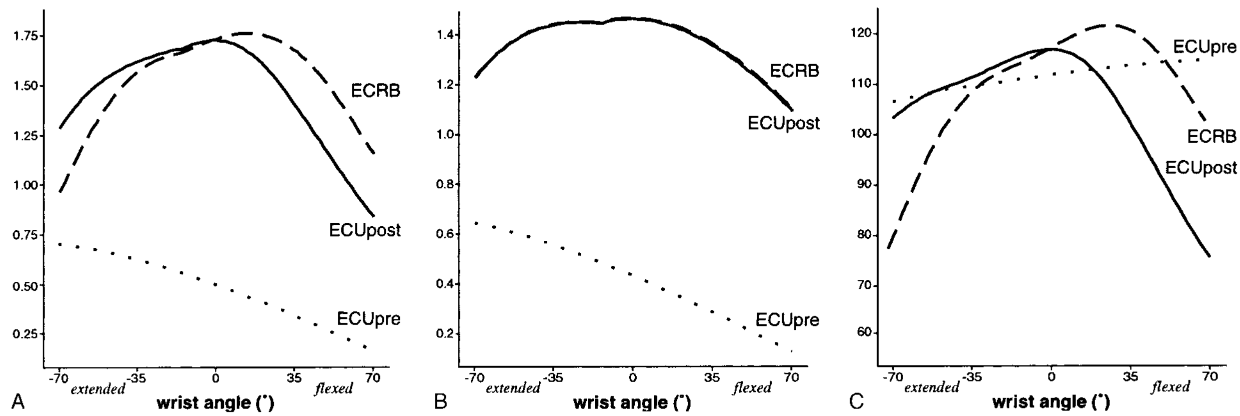

Figure 2. Moment, moment arm, and force-generating potential of the ECRB before transfer and of the ECU before (ECUpre) and after (ECUpost) transfer, as calculated with the computer model. Positive values of the flexion angle represent wrist flexion; negative values represent wrist extension. The graphs represent (A) maximum isometric wrist extension moment versus wrist flexion angle, (B) wrist extension moment arm versus wrist flexion angle, and (C) maximum isometric muscle force versus wrist flexion angle. The wrist remained in neutral deviation. The moment-generating capacity of the ECU greatly increases through transfer to the ECRB.

N-m to 1.72 N-m when transferred to the ECRB (Fig. 2A). The extension moment increased primarily because of an increase in the extension moment arm after tendon transfer. The extension moment arm of the ECU increased from 0.45 cm to 1.47 cm at the neutral wrist position after the transfer (Fig. 2B). The transferred ECU has essentially the same anatomic configuration across the wrist as the normal ECRB; thus, the transfer of the ECU resulted in a moment arm–joint angle relationship mimicking that of the ECRB.

The force-generating capacity of the ECU was nearly constant before the simulated tendon transfer (see ECUpre in Fig. 2C). However, the force-generating capacity varied substantially with wrist angle after the simulated surgery. This occurred because the greater moment arm of the ECU after transfer to the ECRB causes the musculotendon unit to undergo greater excursion with flexion of the wrist. Increasing the muscle excursion caused the ECU to operate on a broader portion of its force-length curve; this resulted in a greater variation of force over the range of motion.

Alteration of tendon length had a sizable effect on the force-generating capacity of the ECU after tendon transfer (Fig. 3). Before surgery, the force generated by the ECU was nearly constant (see PRE in Fig. 3). However, the force-generating capacity varied substantially with wrist angle after tendon transfer. Increasing or decreasing tendon length by 1 cm (SLACK and TIGHT, respectively, in Fig. 3) caused a 35° shift in the angle at which peak force was developed. In the slack transfer, the active force-generating capacity of the ECU decreased with wrist extension by up to 27% compared with the neutral transfer. In the tight transfer, the active force-generating capacity decreased by up to 89% with flexion because the muscle fibers extended beyond their optimal length. In the tight transfer, the passive muscle force contributed substantially in the total force generated by the ECU with the wrist in flexion, accounting for 84% of the total force-generating capacity in full flexion.

Transfer of the ECU changed the maximum deviation moment from 2.72 N-m ulnar deviation to 1.42 N-m radial deviation at the neutral wrist position. This was anticipated due to the alteration of the insertion of the ECU from the fifth to the third metacarpal and the resulting change in deviation moment arm.

Discussion

The objective of this study was to quantify the effects of a transfer of the ECU to the ECRB. Our simulations suggest that the transfer greatly increases the capacity of the ECU to generate an extension moment and alters its deviation moment from a relatively large magnitude in the ulnar direction to a comparatively smaller one in the radial direction. Although the radial deviation moment is relatively small, the change in deviation moment arm indicates

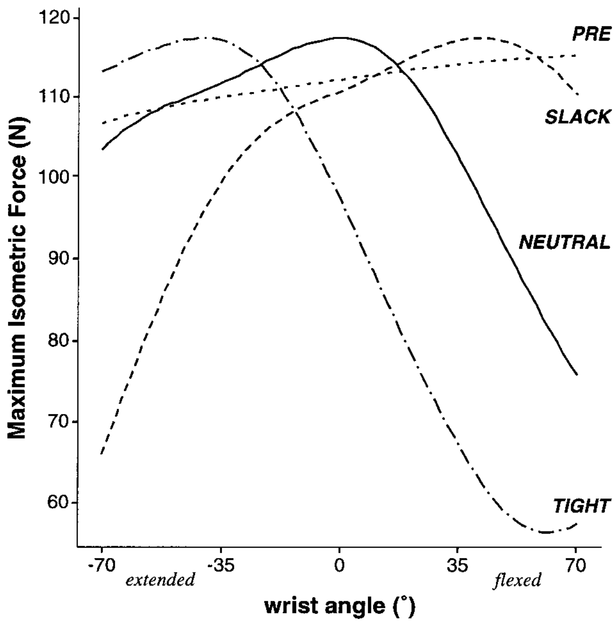

Figure 3. Postoperative force versus wrist flexion angle of the ECU after adjustment of tendon slack length to yield maximum force at the neutral wrist position (solid line) and after a 1-cm increase (dashed line) and decrease (dot-dashed line) from this value. The preoperative force profile of the ECU (dotted line) is provided for comparison. Intraoperative alteration of the muscle–tendon length has a sizable effect on the isometric force-generating characteristics.

the removal of the ECU from its role as an ulnar deviator, shifting the overall muscle balance further toward radial deviation. The increased extension moment of the ECU results from an increase in extension moment arm. This increase in extension moment arm causes a greater variation of the force developed by the ECU with wrist flexion-extension.

Limitations of the Computer Model

Before discussing the implications of these results, several limitations of the computer model should be considered. First, the results presented here represent values obtained using a computer model that represents a single adult subject with average muscle–tendon properties. There are, however, wide variations in musculoskeletal geometry and muscle force-generating capacity among individuals. In addition, the model does not account for the influence of disease, which may induce spasticity or increase tissue stiffness. The presence of scar tissue, for instance, could greatly alter the transmission of muscle force and have a large effect on the moments generated by the muscles. The absolute values presented here should not be applied to a particular individual, who may greatly differ from the model. Rather, the results should be used to understand the general relationships between the alteration of moment arms after the tendon transfer and the variation of force with joint motion.

The simulations kept constant the peak isometric force and muscle-fiber length of the ECU. However, these parameters may change through adaptation of the muscle–tendon complex, either before or after the tendon transfer. For example, the number of sarcomeres in a muscle may decrease, changing the fiber length, as a muscle–tendon complex adapts to altered conditions.[21] Muscle atrophy may produce changes in physiologic cross-sectional area and decrease the force-generating capacity of a muscle. By keeping the properties of the ECU constant, this study isolated the effects of changing the muscle–tendon path on the moments generated by the muscle.

Constant maximal activation was assumed to estimate the maximum force output of the muscles in the simulations. When activation of the ECU was reduced to 30% of maximum, the shape of the force–wrist flexion angle relationship was altered notably (Fig. 4). With this change, passive force accounted for approximately 15% of the total force generated by the ECU with the wrist in 35° flexion. The passive and active forces were nearly equal in full flexion. Muscle force output is strongly influenced by the electrodes used to stimulate muscles in paralyzed subjects,[2] and it is unknown how the activation levels in this study compare with those obtained using electrical stimulation. In spite of the limitations outlined above, this study presents a comprehensive examination of the biomechanical effects of the transfer procedure.

Comparison With Other Studies

The moment arms calculated with the model of the ECRB and the ECU before surgery correspond well with reported experimental values at the neutral wrist position[22,23] (Fig. 5). Our finding that the muscle force–joint angle relationship of the ECRB is steeper than that of the ECU is consistent with the observations of Loren et al,[23] who showed that the ECRB operates on a broader range of its force–length relationship. After transfer, the ECU also operates on a broad range of its force–length relationship, a direct result of its greater extension moment arm.

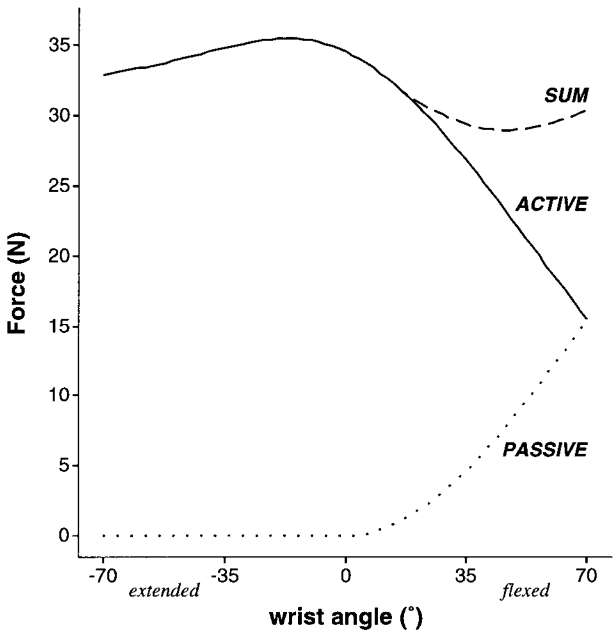

Figure 4. Contributions of active (solid line) and passive (dotted line) force to the total force-generating capacity of the ECU after transfer (dashed line), assuming 30% of maximum muscle activation. Note the increased dependence of the force profile on passive force at high flexion angles.

The moments calculated with the computer model of the ECU were compared with the preoperative and postoperative moments reported by Keith et al,[3] which were measured during electrical stimulation of the ECU in a subject with tetraplegia. The absolute magnitudes of the moments computed with the model were larger than the moments measured in the subject, most likely due to the difficulty of maximally stimulating the ECU and to the possibility of atrophy in the subject with spinal cord injury. When the moments calculated with the model were scaled by the experimental data, the relative magnitudes of the preoperative and postoperative extension moments were similar, but the model indicated greater radial and ulnar deviation components (Fig. 6). This discrepancy may be explained by the fact that stimulated muscles may not act independently. During electrical stimulation, current spillover (the simultaneous recruitment of fibers from multiple muscles) is common, and it is difficult to cause a single muscle to generate force independently.[24] Because the ECU is the most ulnar muscle, current spillover to other muscles, such as the extensor digitorum communis, before transfer is likely to reduce the ulnar deviation component. Conversely, since the wrist extensors in closest proximity to the belly of the ECU (and thus most likely to be unintentionally stimulated) generate ulnar deviation moments, current spillover after surgery could reduce the radial component. Although simple anatomic differences between the transfer model and the human subject could account for the discrepancies observed, it is important to note that

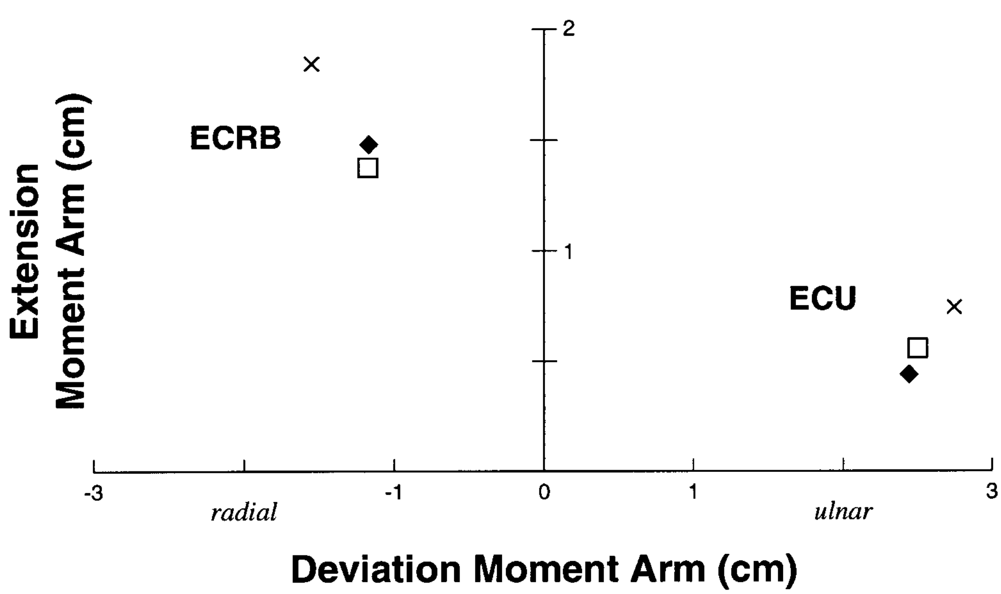

Figure 5. Comparison of the model moment arms (◆) with the experimentally determined moment arms at the neutral wrist position (□, Brand and Hollister[22]; ×, Loren et al[23]). Extension and ulnar deviation are positive; radial deviation is negative.

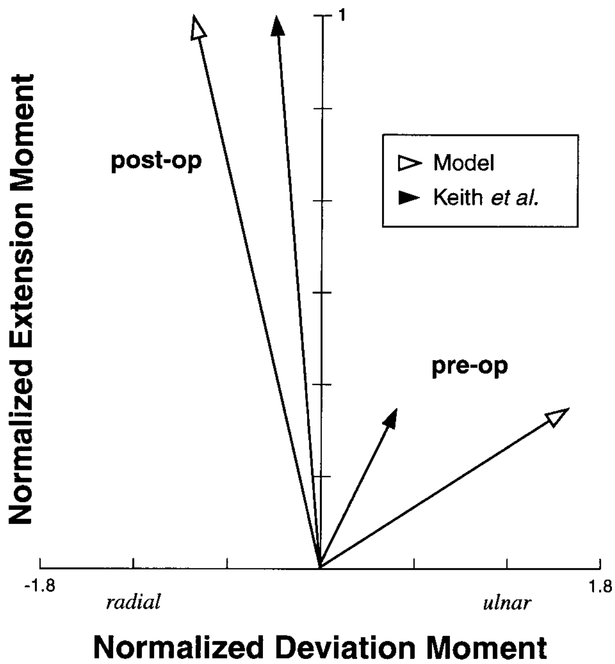

Figure 6. Moment generated by the ECU at the neutral wrist position before and after transfer, as calculated from the computer model (▷) and as reported by Keith et al[3] (▶). The data have been scaled such that the postoperative extension moment = 1.0.

the computer model calculates each muscle's operating characteristics independently without accounting for recruitment of other muscles.

Clinical Implications

The results of this study indicate that the transfer of the ECU to the ECRB yields a muscle–tendon unit that closely replicates the moment-generating characteristics of the normal ECRB. This occurs because the ECU acquires the moment arm of the ECRB and has similar architectural characteristics (Table 1). Zajac[25] noted that 2 quantities should be similar between muscles to reproduce the same moment-generating capacities: (1) the product of the moment arm and physiologic cross-sectional area and (2) the ratios of fiber length to moment arm. Table 1 shows that these values are similar for the ECRB and the transferred ECU. Lieber et al[17] reported that the "difference index" (a quantitative representation of the difference between muscles based on 5 architectural parameters) between ECU and ECRB is small, suggesting that the ECU is an excellent substitute for the ECRB.

The significant effect of tendon length alteration on muscle force output (Fig. 3) suggests that establishing proper muscle–tendon length is essential to provide maximum force over the desired range of motion. Although Brand[21] argued that muscles adapt sarcomere numbers to biomechanical requirements, suggesting that the resting tension of muscle fibers is ultimately immutable, the mechanics of muscle remodeling and the factors that determine the degree to which it occurs are not fully understood. Although clinical observations of patients with stroke, head injury, or cerebral palsy have associated muscle shortening with chronic motor unit discharge, no study has demonstrated alteration of fiber length or sarcomere number with chronic functional electrical stimulation. If muscle fiber length is preserved, then surgical alteration of muscle–tendon length to yield the desired force-generating profile during surgery seems to be a more reasonable option than reliance on muscle–tendon remodeling.

Our results indicate that the transfer of the ECU to the ECRB is an effective means of increasing wrist extension moment, a prerequisite for the strong finger flexion needed in both lateral and palmar grasp. When the tendon length of the ECU is adjusted to cause maximum force-generating potential to occur

Table 1. Architectural Parameters of Extensor Carpi Radialis Brevis and the Extensor Carpi Ulnaris

	PCSA	F_o^m	l_o^m	l^m	MA^n	$PCSA*MA^n$	l_o^m/MA^n	l_o^m/l^m
ECRB	2.7	123	5.9	18.6	1.6	4.3	3.7	0.32
ECU								
post-op	2.6	117	6.2	21.0	1.6	4.2	3.9	0.30

The physiologic cross-sectional area (PCSA) and optimal muscle fiber length (l_o^m) values were taken from Lieber et al.[17] The muscle lengths (l^m) are from Loren and Lieber.[18] Moment arms at the neutral wrist position (MA^n) of the ECRB and the postoperative ECU were calculated from the computer model. Ratios of l_o^m to MA^n and of l_o^m to l^m were calculated with the muscle length and optimal fiber length data above. Peak isometric muscle force (F_o^m) was scaled from PCSA using 45 N/cm^2 for maximum muscle stress.[26] See Gonzalez et al[11] for a detailed description of these calculations. Muscle force is given in newtons (N); MA^n, l_o^m, l_o^m, and l^m are given in centimeters (cm); and PCSA is given in square centimeters (cm^2).

in slight extension, the resultant moment–wrist joint angle relationship duplicates that of the normal ECRB. This transfer can serve to augment wrist extension when voluntary control of the ECU is present and may be among the most effective means of restoring wrist extension in conjunction with functional electrical stimulation when direct activation of the ECRB is untenable because of lower motor neuron damage.

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