Muscular Resistance to Varus and Valgus Loads at the Elbow

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Although the contributions of passive structures to stability of the elbow have been well documented, the role of active muscular resistance of varus and valgus loads at the elbow remains unclear. We hypothesized that muscles: (1) can produce substantial varus and valgus moments about the elbow, and (2) are activated in response to sustained varus and valgus loading of the elbow. To test the first hypothesis, we developed a detailed musculoskeletal model to estimate the varus and valgus moment-generating capacity of the muscles about the elbow. To test the second hypothesis, we measured EMGs from 11 muscles in four subjects during a series of isometric tasks that included flexion, extension, varus, and valgus moments about the elbow. The EMG recordings were used as inputs to the elbow model to estimate the contributions of individual muscles to flexion-extension and varus-valgus moments. Analysis of the model revealed that nearly all of the muscles that cross the elbow are capable of producing varus or valgus moments; the capacity of the muscles to produce varus moment (34 Nm) and valgus moment (35 Nm) is roughly half of the maximum flexion moment (70 Nm). Analysis of the measured EMGs showed that the anconeus was the most significant contributor to valgus moments and the pronator teres was the largest contributor to varus moments. Although our results show that muscles were activated in response to static varus and valgus loads, their activations were modest and were not sufficient to balance the applied load.

Introduction

The ligaments of the elbow play a critical role in counteracting varus and valgus loads (i.e., loads that would produce adduction and abduction movements) and in keeping the elbow in proper alignment. Rapid, unexpected varus or valgus loading of the elbow can injure passive structures such as ligaments and bone (Hotchkiss and Weiland, 1987; Morrey, 1986; Morrey and An, 1983, 1985; Morrey et al., 1991; Regan et al., 1991; Schwab et al., 1980). Reflexive muscular responses are too slow to respond to unexpected loads (e.g., Rack, 1981) and thus the passive structures must support the full load.

Sustained or expected varus or valgus loads can also cause injuries of the elbow. However, under these loading conditions the muscles may be capable of supporting some of the load, potentially protecting the passive structures from injurious forces. For example, muscles might resist valgus loads in baseball pitchers or javelin throwers, and decrease the commonly observed injuries to the medial elbow ligaments (Andrews and Whiteside, 1993; Glousman et al., 1992; Tullos et al., 1986; Werner et al., 1993; Wilk et al., 1993).

The response of muscles to varus or valgus loads at the elbow is not well understood. Some researchers have reported electromyographic (EMG) evidence of muscular support of varus loads at the elbow (Buchanan et al., 1986; Funk et al., 1987; Glousman et al., 1992). However, no in-depth studies have been done to quantify the contributions of muscles to resistance of varus or valgus moments at the elbow. It is unknown whether any elbow muscle is capable of producing substantial varus or valgus moments. Furthermore, if a muscle is capable of resisting varus or valgus moments, it is not known whether that muscle is activated in response to these loads.

This study was conducted to address these two issues. We hypothesized that muscles about the elbow could theoretically produce substantial varus and valgus moments. Specifically, we hypothesized that the muscular contribution to varus and valgus moments could theoretically be about half of the muscular contribution to flexion-extension moment. To test this hypothesis, we developed a detailed musculoskeletal model to estimate the varus and valgus moment-generating capacity of the muscles about the elbow. We also hypothesized that the elbow muscles would be activated in response to sustained, isometric varus and valgus loads. To test this hypothesis, we performed experiments in which subjects executed a variety of isometric tasks that involved varus and valgus moments at the elbow. During each task, the EMG activity of the muscles that span the elbow were recorded. These EMG data were used as inputs to the musculoskeletal model to estimate the contributions of each muscle to the total joint moment.

Methods

A computer model of the elbow and the surrounding muscles was created using the musculoskeletal modeling software described by Delp et al. (1990, 1995). The elbow model includes three-dimensional representations of the bones, kinematic models of flexion-extension and varus-valgus deviation, and geometric representations of the muscles. Bone geometry was obtained by three-dimensional digitization of cadaver bones. The geometry of each muscle–tendon complex was represented as a series of line segments that connect the origin and insertion. Multiple line segments were used to represent each muscle–tendon path, so that the muscle wrapped over bony prominences or under soft tissue constraints. The intermediate "wrapping points" that constrain the muscle–tendon path between the origin and insertion could move as a function of the limb configuration to represent the anatomy accurately. Based on these geometric descriptions, the lengths and moment arms of the muscles were computed by taking the partial derivative of the muscle–tendon length with respect to the joint angle (Storace and Wolf, 1979; An et al., 1984, Delp and Loan, 1995).

Our model for elbow flexion-extension and pronation-supination has been described in detail by Murray et al. (1995). There we showed that moment arms calculated with the model compare well with moment arms we have measured in cadavers.

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The kinematics of varus and valgus deviation was characterized as follows. Separate axes of rotation were used for varus and valgus deviation to allow the center of rotation to translate as the joint contact position moved medially or laterally with varus and valgus motions. In the model, the center of rotation for varus deviation was located at the medial edge of the contact area between the humerus and the ulna (beneath the trochlear condyles). Similarly, the center of rotation for valgus deviation was located at the lateral edge of the contact area. The rotation axes were oriented orthogonal to both the flexion axis and the pronation axis.

The model incorporates the isometric force-generating properties of the muscles so that the moment-generating potential of each muscle can be estimated. The force-generating properties of the muscles were derived by scaling a Hill-type model of muscle (Zajac, 1989). Four parameters scale the model to represent a specific muscle–tendon complex. The four parameters are: optimal muscle fiber length, pennation angle, peak isometric force, and tendon slack length. No single source reports these parameters for all the muscles about the elbow. Therefore, optimal fiber lengths, pennation angles, and peak forces for biceps (BIC), brachialis (BRA), brachioradialis (BRD), pronator teres (PT), triceps (TRI), anconeus (ANC) were derived from An et al. (1981). Optimal fiber lengths, pennation angles, and peak forces for extensor carpi radialis longus and brevis (ECRL and ECRB), extensor carpi ulnarus (ECU), flexor carpi radialis (FCR), and flexor carpi ulnarus (FCU) were taken from An et al. (1981) and Lieber et al. (1992).

Values for the peak isometric forces of muscles were derived by multiplying physiologic cross-sectional area (PCSA) by a specific tension. Because the actual PCSA values for our subjects were not known, the specific tension values were adjusted to account for the differences. Values for the specific tensions were determined by comparing the moment-generating capacity of the model to the average maximum isometric moments measured in human subjects. To scale the physiologic cross-sectional areas reported by An et al. (1981), a specific tension of 170 N/cm^2 was used for the elbow flexors and a value of 100 N/cm^2 was used for the extensors. If the same value of specific tension had been used for the flexors and extensors (e.g., 100 N/cm^2), the moment-generating capacity of the flexors would have been underestimated by approximately 40 percent. These values for specific tension are higher than the specific tensions suggested in previous investigations, presumably because the PCSAs were measured from elderly, embalmed cadavers, whereas peak moments were measured in young men. A specific tension of 45 N/cm^2 was used to scale the PCSA values reported by Lieber et al. (1992). This value was derived by comparing the estimated moment-generating capacity of the wrist muscles to moments we have measured in ten subjects (Delp et al., 1996). This value is smaller than the values used to scale the PCSA data from An et al. (1981) perhaps in part because the cadavers used by Lieber et al. (1992) were not fully preserved (preservation is known to decrease PCSA). Buchanan (1995) discusses these issues in detail.

Tendon slack length is difficult to measure accurately because it includes both internal and external tendon (Zajac, 1989). We therefore used a three-step process to estimate tendon slack lengths for muscles about the elbow (Hoy et al., 1990). In step one, tendon slack lengths were estimated based on experimental data (An et al., 1981; Lieber et al., 1992). In step two, tendon slack lengths were adjusted so that the passive moment developed by the muscles was consistent with experimentally measured passive moments (Hayes and Hatze, 1977). In step three, the muscles were assumed to be fully activated and, if necessary, the tendon slack lengths were adjusted so that the computer model accurately represented the shapes of the average flexion and extension strength curves determined from our four subjects (see Results).

Experimental Procedure

In a previous study of seven males ranging in age from 24 to 37, maximum isometric elbow flexion and extension moments were measured through the range of motion (Buchanan, 1995). The four subjects from this previous study whose peak flexion and extension moments were closest to the mean were used in this study. These four subjects were selected to reduce the differences between the individual subjects and the computer model, which was based on the average strength of the subjects.

Each subject was seated with his shoulder at 90 deg abduction, 0 deg extension and no external rotation, the elbow at 90 deg flexion. The shoulder was secured with straps and the flexion axis of the elbow was in line with the vertical axis of a rotating table (Fig. 1). A Fiberglass cast approximately 5 cm wide was made on the subject's distal forearm just proximal to the ulnar styloid. The cast was bolted to a six degree of freedom load cell with a resolution of 0.56 N (Assurance Technologies, Inc., Model 150-600) with the subject's forearm in neutral pronation-supination.

Each subject performed a maximal flexion, extension, varus, and valgus contraction while being given visual feedback of the force produced. These maximal force values were recorded, along with maximal EMG values, which were obtained by having subjects produce combinations of elbow and wrist flexion and extension, as well as forearm supination and pronation. The smallest of these four maximum forces were used to define the parameters of the target matching procedure (described below). The maximum EMGs were used to normalize the subsequently recorded EMGs.

Each subject was then asked to match a series of targets on the computer screen by producing a combination of static flexion-extension and varus-valgus moments against the load cell, as described by Buchanan et al. (1989). The targets also indicated pronation moment and forearm axial force, which were constrained to be near zero. The loads in the flexion and varus directions (i.e., loads in the transverse plane) were characterized by a load angle and magnitude. A pure flexion force would be a load angle of 0 deg, a pure valgus force would be 90 deg, a pure extension force would be 180 deg, and a pure varus force would be 270 deg. Twenty-four load angles were used, in 15 deg increments around a circle perpendicular to the axis of the forearm. The magnitude of the desired load was a constant equal to 40 percent of the smallest of the maximum forces measured during the maximal contractions described in the previous paragraph. This desired load was represented by a target on the computer screen. The subject's current load was displayed as a movable cursor. Once the subject visually placed the cursor in

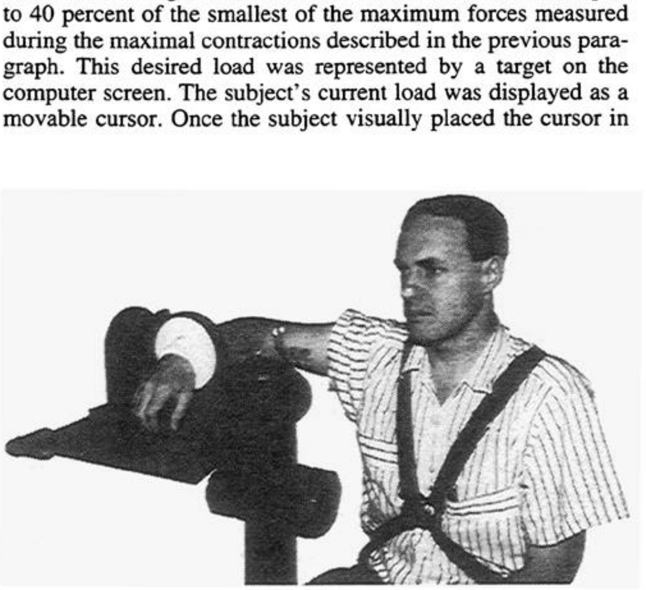

Fig. 1 Photograph of a subject in load cell apparatus. Note that the distal forearm is placed in a cast, which is rigidly attached to a load cell. EMGs are collected from eleven muscles. Shoulder straps constrain the subject to minimize shoulder movements during the experiment.

the target and remained within the target for 0.5 seconds, the data from that period were collected. The order in which the load angles were presented was randomized, and four repetitions were done at each angle. The subjects rested between trials to avoid the effects of fatigue.

EMGs were collected from eleven muscles simultaneously. Surface electrodes were placed over BIC and TRI. Intramuscular electrodes were placed into the remaining muscles. Intramuscular electrodes were used for these muscles to avoid signal contamination from nearby muscles. The intramuscular electrodes were made from 75 mm diameter Teflon-coated stainless steel wire. The Teflon coating was removed from 0.5 cm of either end of the wire, and the wire was placed using a 27 gage hypodermic needle. Two electrodes were placed in each muscle, approximately 2 cm apart, as described by Perotto (1994) and tests were performed to verify proper placement. All electrodes were connected to a custom-made pre-amplifier: a two-pole, high pass filter with a 30 Hz cut off and a gain of 1000. The signal was then sent to a filter/amplifier, an eight-pole low pass Butterworth filter with a 300 Hz cut off and a variable gain between 2.5 and 40. Myoelectric and load cell signals were sampled at 1000 Hz via an A/D converter and stored in the memory of a Macintosh Quadra 950. The EMGs were processed by taking the mean rectified EMG collected during each 0.5 second interval during which the target was matched and normalizing by the maximum EMG of that muscle.

Normalized EMG data from the target matching experiments was used as the activation (a) to the elbow model. The model was then used to estimate joint moment (M_{joint}) as defined by the following equations:

$$F_{\text{muscle}} = f(a_{\text{muscle}}, l_{\text{muscle}}(\theta), F^{\text{peak}}_{\text{muscle}}), \; M_{\text{muscle}} = F_{\text{muscle}} \times ma_{\text{muscle}}(\theta),$$

where $ma_{\text{muscle}}(\theta)$ is the moment arm of the muscle, and

$$M_{\text{joint}} = \Sigma \text{ [subscript all muscles] } \quad M_{\text{muscle}}.$$

To determine the relative contribution of each muscle, the moment produced by each muscle at both pure varus and pure valgus was normalized by the magnitude of the total joint moment, averaged across subjects, sorted by level of contribution, and plotted.

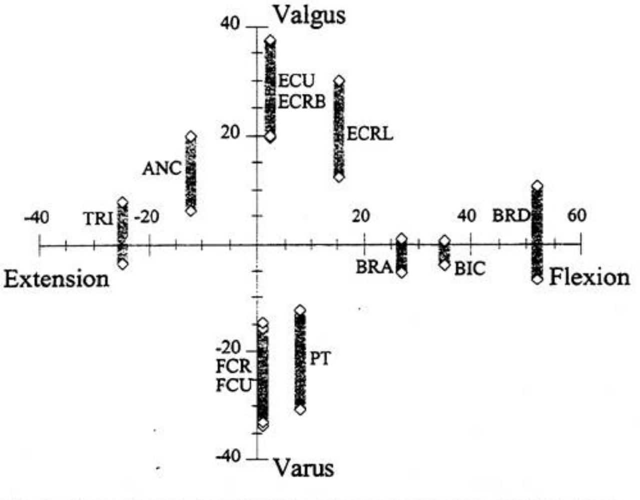

Fig. 2 Moment arms (in mm) of the muscles used in this study (abbreviations are given in Methods), computed with the elbow at 90 deg flexion and a neutral forearm position. Because there are two axes of rotation—one for varus deviation and another for valgus deviation—the varus-valgus moment arm changes as the motion shifts from one center of rotation to the other. Therefore each muscle has a range of varus-valgus moment arms.

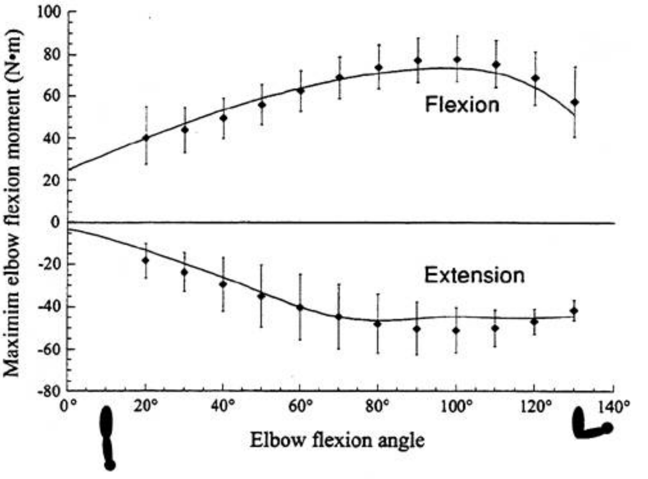

Fig. 3 Comparison of maximum isometric moments estimated with the model (curves) to moments measured during maximum voluntary contractions (points). The experimental values are the average of the four subjects. The error bars indicate one standard deviation of the experimental data.

Results

Analysis of the model revealed that several muscles about the elbow have substantial varus or valgus moment arms (Fig. 2). At 90 deg elbow flexion, extensor carpi ulnaris, extensor carpi radialis longus and brevis, and anconeus have valgus moment arms that are greater than their flexion-extension moment arms. Similarly, flexor carpi radialis, flexor carpi ulnaris, and pronator teres have varus moment arms that are larger than their flexion-extension moment arms. Because different centers of rotation were used for varus and valgus motions, each muscle has two varus-valgus moment arms. The larger varus moment arm of a muscle corresponds to the muscle resisting a valgus load; conversely, the larger valgus moment arm corresponds to the muscle resisting a varus load.

The maximum isometric flexion-extension moments estimated with the computer model correspond to the experimentally measured moments (Fig. 3). A slight notch in the extension moment curve of the model at a flexion angle of 80 deg resulted from the onset of wrapping points in the triceps muscles. Also, the model underestimated the peak flexion and extension moments at 90 deg flexion.

At 90 deg flexion, the model estimated a maximum varus moment of 34 Nm and a maximum valgus moment of 35 Nm. This maximum varus moment was computed by maximally activating all of the muscles that contributed to varus while keeping all other muscles inactive; the opposite was done to compute the maximum valgus moment. Hence, the maximum contribution of muscles to varus and valgus moments was determined to be roughly 75 percent of maximum extension and 50 percent of maximum flexion moment.

The muscle activation patterns demonstrate that all muscles were activated during loads that had varus and valgus components, although these were often driven by the need to produce flexion and extension moments (Fig. 4). During this experiment subjects were asked to produce a submaximal joint moment as previously described and no muscle was activated more than 32 percent of its maximal value. Only a few muscles showed substantial activation during varus or valgus loads that had no flexion or extension components. During valgus moments the anconeus (ANC) and biceps were activated about 50 percent of their extension and flexion levels. Likewise, the pronator teres (PT in Fig. 4) showed a tendency to produce varus moments, but it was activated over a smaller range of load angles and had a lower peak activation. The brachialis (BRA) and

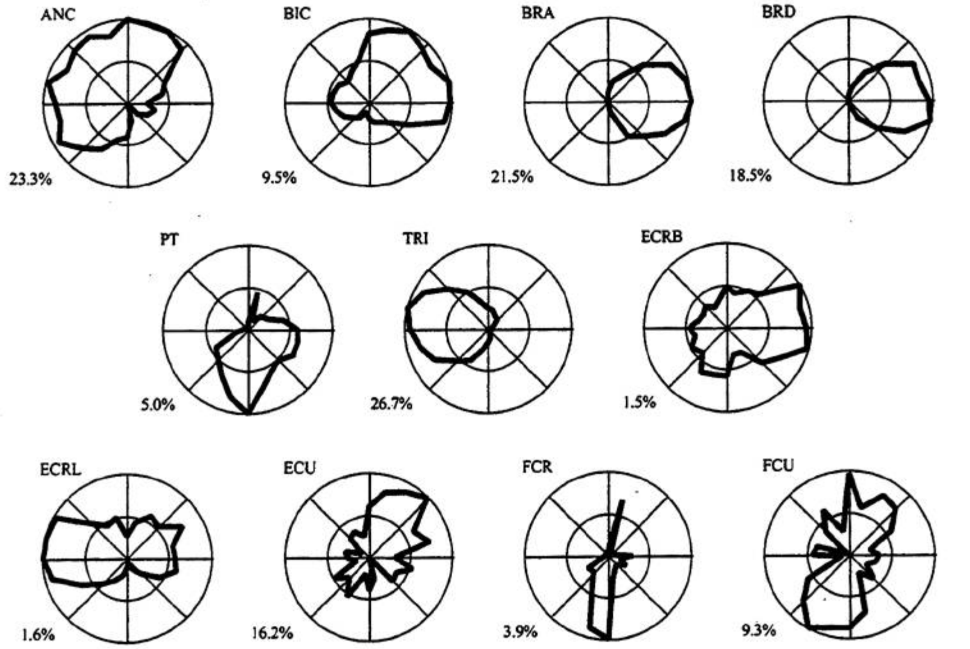

Fig. 4 EMG activation profiles for all muscles averaged across all subjects for 13 Nm joint moments applied in various directions. For each curve, the polar angle corresponds to the load direction and the magnitude corresponds to the activation of that muscle (i.e., the rectified, averaged EMG) at that load angle in the transverse plane. The magnitude at the outer ring is indicated by the percentage below and left of the circle; the magnitude of the inner ring is half of that.

brachioradialis (BRD) were activated primarily to produce flexion moments and the triceps (TRI) was activated to produce extension moments. Activations in several of the wrist muscles (ECRB and ECRL) were approximately 2 percent of their maximum voluntary activation for all load directions.

The anconeus was active over a substantial portion of the valgus load range for all four subjects (Fig. 5(A)). The pronator teres was activated to produce varus moments in three of the four subjects (Fig. 5(B)). Subject #3 produced a higher activation level (34 percent) than the other three subjects. Subject #4 activated the pronator teres primarily to produce a flexion moment.

An addition to examining the muscle activity during these tasks, the predicted joint moment contributions from each muscle were examined as well. The anconeus produced the largest valgus moment and pronator teres produced the largest varus moment in all subjects (Fig. 6). In subject #3, a large contribution of the pronator teres to varus moment caused the high standard deviation of the mean for that muscle. The order of relative contributions of the secondary muscles was similar for the four subjects.

Although the actual contribution from wrist muscles to the generation of varus and valgus moments was observed to be modest, their potential contribution was not (Table 1). Together

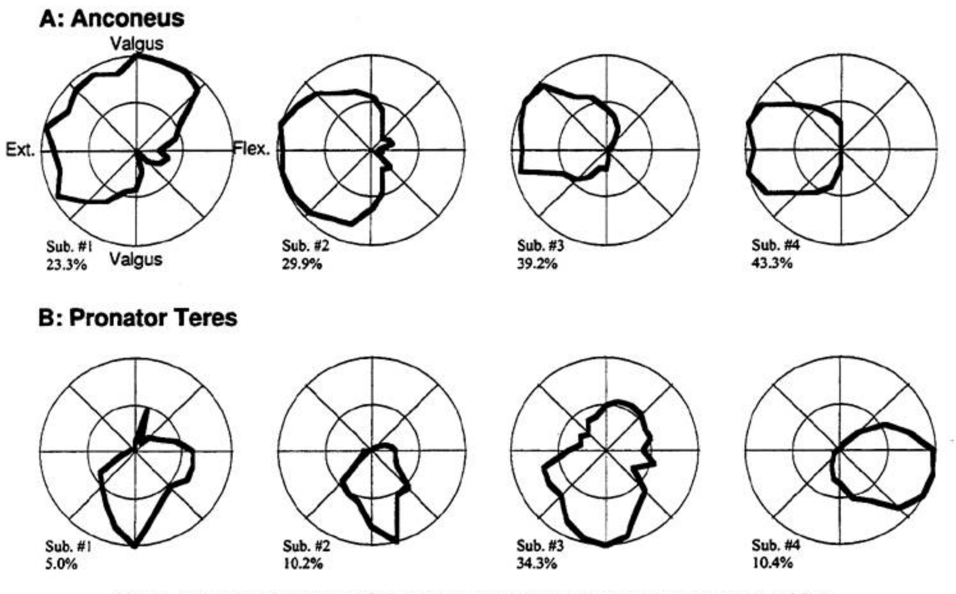

Fig. 5 Activation profiles of (A) anconeus and (B) pronator teres for each subject

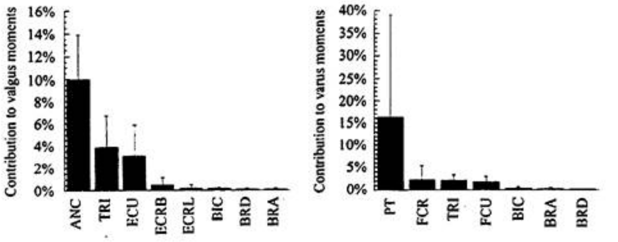

Fig. 6 Average relative percent contribution of each muscle to the production of varus and valgus moment. The large standard deviation on PT is due to high contributions of subject #3.

the three primary wrist extensors (ECRB, ECRL, ECU) could produce over twice as much valgus moment as the anconeus. The two primary wrist flexors (FCR, FCU) could contribute roughly 60 percent as much as the pronator teres to varus moment. Despite their capacity to produce varus and valgus moments, the wrist muscles were not activated to counteract these moments.

Discussion

The purpose of this study was to determine whether muscles contribute to the production of varus and valgus loads at the elbow. We found that although many of the muscles that cross the elbow are capable of producing moments to resist varus or valgus loads, only a few are used in this way. Anconeus was found to be the muscle most used for production of valgus moments, and pronator teres for varus moments.

Several assumptions and limitations should be taken into account when evaluating our results. First, we estimated varus and valgus moment arms of muscles based on separate axis for varus and valgus deviations. When resisting varus loads, the moments generated by the muscles were computed about an axis located at the medial edge of the articulating surface of the humerus and the ulna. Similarly, the lateral edge was used for valgus moments. This maximizes the moment arms of the muscles resisting the applied load and provides a reasonable upper limit for their moment-generating potential. The effect of varying the axes or rotation on varus and valgus moment arms is demonstrated by the shaded regions in Fig. 2. If a single varus-valgus axis of rotation located at the center of the articulating surface was used, the contribution of muscles to varus and valgus moments would have decreased roughly 30 percent.

A second source of error arises from using a generic musculoskeletal model to estimate joint moments generated by individual subjects. We tried to minimize this error by using subjects of similar strength; however, differences in muscle moment arms, force-generating properties, and activation patterns could not be eliminated. Differences in these first two parameters are limitations of the study. However, differences in activation patterns are due to real differences in the way subjects stabilize their joints. Hence, the standard deviations of Fig. 6 are due, to a large measure, to the different ways subjects activated their muscles (Fig. 5). We believe that the model underestimated the flexion-extension moments for two primary reasons. First, the model underestimates the maximum isometric flexion and extension moments at 90 deg flexion (Fig. 3), the elbow angle at which the target matching experiments were performed. Second, when the model was matched to the maximum flexion and extension moments shown in Fig. 3, it was assumed that no co-contraction was present. However, when EMG data from the target matching experiments were used to compute the moments, activation of the antagonist muscles reduced the flexion and extension moments estimated with the model. Assuming this activation is real (i.e., not due to current spread), we could have adjusted the model (e.g., by varying the specific tension) for each subject in order to minimize this error. Doing so would have scaled the percentage of the total varus and valgus moment provided by muscles (Fig. 6). However, muscle contribution would remain less than 50 percent of the total varus and valgus moment in all except one case (subject #3).

Table 1 Potential contribution of wrist extensors (ECRB, ECRL, ECU) and ANC to valgus moment and of wrist flexors (FCR, FCU) and PT to varus moment as calculated by model. The total joint moment required was 13.0 Nm.

Potential valgus moment (Nm)		Potential varus moment (Nm)	
ECRB+ECRL+ECU	ANC	FCR+FCU	PT
10.1	4.7	7.3	12.1

This study only examined a single posture. Had the elbow been flexed at joint angles other than 90 deg, other results would have been expected. Although the varus-valgus moment arms of most muscles change very little with flexion angle, muscle activation may be significantly different at other flexion angles due to differences in the muscles' contributions to flexion and extension moments.

Finally, we assumed a one-to-one relationship between normalized EMG and muscle activation. This ignores time delays and filtering characteristics of tissues, but these are not likely to introduce significant errors in this static analysis. Normalizing the EMG values by maximal values may introduce errors as true maximal values can be difficult to achieve.

Based on our results, the muscles about the elbow can be divided into three groups: (1) those with mainly flexion or extension moment arms—BIC, BRA, BRD, TRI; (2) those with mainly varus or valgus moment arms—ECRB, ECU, FCR, and FCU; and (3) those with relatively large varus or valgus and flexion or extension moment arms—ANC, ECRL, and PT. It is the latter two groups that are capable of producing substantial varus and valgus moments when activated as their varus or valgus moment arms are at least as large as their flexion or extension moment arms at the elbow.

Anconeus (ANC) was active throughout a great range of the valgus load angles, and pronator teres (PT) was most active in response to pure varus loads. All other primary elbow muscles had their highest activations at loads with slight valgus components.

It was interesting to find that the wrist muscles did not play a substantial role in balancing varus or valgus loads despite their large moment arms. The moment-generating capacity of the wrist muscles is on the same order of magnitude as that of the anconeus and pronator teres. Clearly, the wrist muscles had the capacity to be major contributors, but they were not consistently utilized in this way, as evidenced by the low activations in these muscles. It is difficult to determine whether this arose from our experimental protocol (in which the loads were applied at the distal forearm and did not cross the wrist) or represents normal behavior.

The purposes of this study were twofold. First, we wished to examine the potential for muscles to act as stabilizers against varus or valgus deviation. We found that many muscles that cross the elbow have some varus or valgus moment arm and therefore would be capable of acting in this role. Second, we wanted to determine whether these muscles are activated in response to varus or valgus loads. Although some muscles made only minimal contributions to these moments, most muscles showed some tendency toward increased activations with varus or valgus loads. Overall, the muscular moment contributions were generally insufficient to fully balance the varus or valgus load and a substantial amount of the moment was sustained by the ligaments.

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