INTRODUCTION: Laboratory testing plays an important role in advancing patient care by helping to develop new treatment techniques and improved rehabilitation therapies. By simulating active elbow motion in-vitro, injury mechanisms can be explored and new clinical approaches can be studied prior to implementation in vivo. Several devices capable of simulating such elbow motion currently exist, and incorporate either load-controlled tendon loading (1,2) or motion-controlled bone loading (3-5). However, a motion-controlled tendon loading device capable of simulating active elbow motion has not been previously described. The purpose of this study was to develop such a motion-controlled testing system to study kinematics in the stable and unstable elbow in various orientations, and to evaluate its performance.

METHODS: The motion-controlled elbow simulator is shown in Figure 1. Simulated active joint motion is achieved using pneumatic actuators to apply loads via cables sutured to the tendons of biceps, brachialis, brachioradialis, triceps and pronator teres. The loads are applied in accordance with muscle loading ratios developed from EMG (6) and pCSA (7) data, as previously described (2). The tendon that supplies the largest percentage of the required load (i.e. brachialis) is designated as the ‘prime mover’, and is connected to a pneumatic actuator containing a linear resistive transducer (LRT). The LRT provides position-feedback, incorporated into a real-time PID controller, such that the prime mover tendon can be displaced at a constant velocity (i.e. motion control). An inline load cell continually measures the load applied by this motion-controlled actuator. This load feedback is used to apportion load to the remaining actuators (i.e. operated under load control) according to the loading ratio.

Eight fresh-frozen upper extremities (mean age 67 ± 19 years) were tested. Elbow flexion was performed with the arm in the vertical (Figure 2A). In the vertical orientation, there was no effect of forearm position (p=0.5) or LCL sectioning (p=0.8). The results were less repeatable in valgus (p=0.008) and varus (p=0.023), compared to vertical, but cutting the LCL had no effect (p_{diff}=0.6; p_{repe}=0.6).

Repeatable V-V and I-E motion pathways were also achieved with this simulator. Average data (n=12) indicated the maximum standard deviations in these values over the five successive trials ranged from 0.11 to 0.19 degrees. There was no effect of LCL sectioning (p=0.5) or forearm position (p=0.6) on repeatability of the motion pathways.

RESULTS: The motion-controlled pneumatic actuator was able to reach its targeted velocity with only small RMS errors (Figure 2). A typical position versus time curve for one specimen is shown in Figure 2A. Figure 2B plots the average RMS errors (n=12) in each of the testing configurations. In the vertical orientation, RMS errors was smaller in supination compared to pronation (p=0.002), but there was no effect of LCL sectioning (p=0.8). RMS error increased in valgus (p=0.005) but decreased in varus (p=0.006), compared to vertical orientation. Error increased with LCL sectioning in valgus (p=0.004) but not varus (p=0.6).

The simulator produced repeatable flexion angle versus time curves, as shown for one representative specimen in the vertical orientation in Figure 3A. The average velocity of flexion (n=12) in this orientation, calculated from the slope of this curve between 20-120° of flexion, was 35.6±8.5°/sec (i.e. inter-specimen variability = 16%). Repeatability for all 12 specimens was quantified by determining the maximum standard deviation in the flexion-time data from the five successive trials performed in each loading position (Figure 3B). In the vertical orientation, there was no effect of forearm position (p=0.5) or LCL sectioning (p=0.8). The results were less repeatable in valgus (p=0.008) and varus (p=0.023), compared to vertical, but cutting the LCL had no effect (p_{diff}=0.6; p_{repe}=0.6).

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DISCUSSION: This study details the development of the first elbow simulator to achieve motion-controlled tendon loading, and to use pneumatic actuators as the motion source. The system performed with small values of RMS error and a high degree of repeatability.

Compared to a previous load-controlled design (1,2), the motion-controlled simulator produces kinematic pathways that are of equal or improved repeatability. The resulting motions are also more consistent, with the current inter-specimen velocity variability of 16% compared to 82% reported in our previous work (1). Unlike the load-controlled device, this simulator is capable of producing simulated active motion in the varus and valgus orientations. Although statistical differences were observed in RMS error between the vertical and varus/valgus orientations, these differences were very small and quite comparable.

This simulator is not without limitations. Although motion in the varus and valgus orientations is possible, the repeatability of these motions is less than in the vertical orientation. In addition, it is currently not possible to actively maintain the forearm in pronation in the valgus position or in supination in the varus orientation. The magnitudes of the loads that must be applied to the pronator teres and biceps, respectively, to maintain these forearm positions are too large to withstand the current suture-cable interface. Future studies will address these limitations.

In summary, the motion-controlled simulator is an important tool for in-vitro studies aimed at increasing our knowledge of elbow mechanics.