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Prediction of elbow joint contact mechanics in the multibody framework

Munsur Rahman[a], Akin Cil[a,b,c], Antonis P. Stylianou[a,*]

[a] Department of Civil and Mechanical Engineering, University of Missouri-Kansas City, 5110 Rockhill Road, Kansas City, MO 64110, USA
[b] Department of Orthopaedic Surgery, University of Missouri-Kansas City, 2411 Holmes Street, Kansas City, MO 64108, USA
[c] Department of Orthopaedics, Truman Medical Centers, 2301 Holmes Street, Kansas City, MO 64108, USA

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ABSTRACT

Computational multibody musculoskeletal models of the elbow joint that are capable of simultaneous and accurate predictions of muscle and ligament forces, along with cartilage contact mechanics can be immensely useful in clinical practice. As a step towards producing a musculoskeletal model that includes the interaction between cartilage and muscle loading, the goal of this study was to develop subject-specific multibody models of the elbow joint with discretized humerus cartilage representation interacting with the radius and ulna cartilages through deformable contacts. The contact parameters for the compliant contact law were derived using simplified elastic foundation contact theory. The models were then validated by placing the model in a virtual mechanical tester for flexion-extension motion similar to a cadaver experiment, and the resulting kinematics were compared. Two cadaveric upper limbs were used in this study. The humeral heads were subjected to axial motion in a mechanical tester and the resulting kinematics from three bones were recorded for model validation. The maximum RMS error between the predicted and measured kinematics during the complete testing cycle was 2.7 mm medial-lateral translation and 9.7° varus–valgus rotation of radius relative to humerus (for elbow 2). After model validation, a lateral ulnar collateral ligament (LUCL) deficient condition was simulated and, contact pressures and kinematics were compared to the intact elbow model. A noticeable difference in kinematics, contact area, and contact pressure were observed for LUCL deficient condition. LUCL deficiency induced higher internal rotations for both the radius and ulna during flexion and an associated medial shift of the articular contact area.

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1. Introduction

The elbow joint, recognized as the most important joint of the upper extremity serves as a fulcrum of the forearm lever that greatly enhances the spatial positioning of the hand. This compound joint is formed by dual articulations of the humerus with the radius and ulna. Stabilization of the joint is achieved through the interactions of bone geometries, ligament constraints and muscular contractions [1–3]. Articular cartilage within the elbow joint withstands repetitive mechanical forces which are about 50% body weight during activities of daily living and may reach up to 3 times body weight at about 90° of elbow flexion [4,5]. The elbow is the most commonly dislocated joint in children and second most commonly dislocated joint in adults often resulting in significant damage to bones and ligaments [6]. Forty-nine percent of these dislocations are complex (dislocation associated with a fracture) which often result in long-term loss of function, chronic stiffness, instability, and posttraumatic osteoarthritis [1,7]. Loss of elbow function can cause significant deficits in upper extremity mobility and jeopardize independence.

Comprehensive knowledge of the in vivo loading environment of the elbow structures is essential in understanding the biomechanical causes associated with elbow diseases and injuries, and in finding appropriate treatments. Currently, measuring the in vivo ligament, tendon and articular contact forces during elbow activities is not possible therefore, computational models have to be employed for predictions. Models can also enhance our understanding of the interrelationships between joint structures and the musculature, facilitating the development of patient specific surgical and conservative treatment strategies, and refining elbow prosthetic design.

Computational models of the elbow have been developed to study joint behavior [8–12], but most of these models have limited applicability because the joint structure was modeled as an

* Corresponding author. Tel.: +18162351252.
E-mail address: stylianoua@umkc.edu (A.P. Stylianou).

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idealized joint (e.g. hinge joint) rather than a true anatomical joint. Although in some circumstances such simplification is helpful for understanding joint kinematics and muscle function, it is not always appropriate to assume a human joint as a generalized mechanical joint [13]. For example, three-dimensional measurements of simulated active elbow motion revealed the amount of potential varus–valgus laxity that occurs during elbow flexion to average about 3–4°, which is ignored in an idealized joint definition [14]. Omission of this normal laxity into the implant design was one of the reasons behind the failure of fully constraint elbow replacement implants due to increased transfer of stresses to the implant-cement-bone interfaces resulting in aseptic loosening [15]. Recently, there has been an effort to develop and validate a computational model of the elbow whereby joint behavior is dictated by the three-dimensional articular contact, ligament constraints, muscle loading [16,17]. In this model, articular cartilage was not included, and bone-to-bone contact and ligament tension was used to constrain joint motion. A recent study has shown that the coronoid cartilage height at the tip of the bony coronoid was 2.96 mm, and the thickness at the tip was 2.63 mm which are significant for varus stability and coronoid fracture fixation, so cartilage is too significant to exclude [18]. Moreover, joint contact forces can be erroneously predicted since the extra conforming cartilage surface is omitted. In the previous models, the ligament tension was assumed to vary linearly with elongation and may decrease the accuracy of the model since the ligament force-length relationship is not linear.

A review of the literature also reveals few models where elbow joint cartilage contact area and contact forces have been examined. Traditional techniques such as using pressure sensitive films in a cadaver joint can give some indication of cartilage contact mechanics [19,20], but that technique has difficulty in measuring the contact of curved surfaces and also measurements may be compromised by joint fluid exposure. Several other techniques such as dynamic MRI and CT imaging [21–23], stereophotogrammetric (SPG) analysis [24,25], fluoroscopy and biplane radiographic imaging [26], and tracking systems [27,28] have been employed to measure the cartilage contact area. Recently, Willing et al. (2013, 2014) created a finite element (FE) model of the elbow to investigate articular contact mechanics [29,30]. The model was validated in a static condition. Finite element models are computationally expensive and are typically used to study isolated tissues or joints in static or quasi-static conditions.

A computational musculoskeletal model with an anatomical elbow joint capable of concurrent predictions of muscle, ligament, and cartilage contact forces in dynamic conditions can be immensely useful [31]. Such models can be effectively used to predict joint loads during activities of daily living, to study the mechanisms of joint elbow injuries such as terrible triad injuries, and to assist in designing better prosthetic implants. The multibody framework is the ideal computational platform to be used for such concurrent, dynamic simulations because of its computational efficiency. In general, contact mechanics in multibody models are greatly simplified and do not allow predictions of contact pressure and contact areas. Detailed knowledge of the contact mechanics during dynamic activities can provide insight into the mechanics of both acute and chronic injuries. The purpose of this study was to develop an anatomically correct elbow joint model with non-linear ligaments that include wrapping around bony structures, and discrete cartilage in the multi-body framework, and evaluate the performance of this model in predicting bone segment kinematics against experimental measurements. After kinematic validation, a lateral ulnar collateral ligament (LUCL) deficient condition was simulated and; contact pressures and kinematics were compared to the intact elbow. This model is the first step in the development of a full musculoskeletal model of the elbow joint capable of contact pressure estimation under dynamic conditions.

2. Method

2.1. Cadaver elbow measurements

Two fresh frozen cadaver elbow specimens were used in this study (Elbow#1, 61 year old, male, right arm; Elbow#2, 42 years old, male, right arm). The elbows were thawed at room temperature for 24 h before testing. The elbow donors had never been diagnosed with major elbow diseases and the elbows appeared normal and intact during visual inspection. The elbows were imaged with both computed tomography (CT) scans and magnetic resonance imaging (MRI). The entire arm was CT scanned to obtain the complete bone lengths. Three mutually perpendicular CT sequences were taken using Siemens SOMATOM definition flash CT scanner (Siemens, Siemens Medical Solutions, PA) with the following parameters: slice thickness of 2 mm, imaging frequency 63.68 Hz, image resolution 512×512, and group lengths 192. MRIs were obtained using a Siemens 3T machine with a narrow field fine resolution setting. The parameters used for MRI were: TR:1200, TE:38, image resolution 320×320, slice thickness 0.5 mm, imaging frequency 123.17 Hz, and group lengths 178. Before imaging, a custom made ABS plastic localizer containing two perpendicular tubes and packed with mustard (visible during medical imaging) was rigidly attached with titanium screws to each bone segment (humerus, ulna and radius) to assist in global coordinate registration later in the experiment [32]. Following medical imaging, the joint capsule, ligaments, interosseous membrane, brachialis tendon, biceps tendon, triceps tendon, wrist joint and hand were kept intact, and the remaining tissues were removed by a shoulder and elbow fellowship trained orthopaedic surgeon. After dissection, the elbows were mounted in a bi-axial Instron 8821 (Instron, Norwood, MA, USA) mechanical testing machine (Fig. 1a). The humerus head was cemented inside a cup that was attached by a hinge joint to the top ram of the mechanical tester. The intact hand was placed and secured on a slider that could slide horizontally in a single axis. Three rigid-body motion markers (each containing three infrared emitting diodes) were firmly attached to the humerus, radius, and ulna localizers. The slider plate also had a rigid motion marker added to it to measure its movement and to aid in computational model alignment. An Optotrak Certus motion capture system (Northern Digital Inc, Waterloo, Ontario, Canada) was used to track the motion of each bone segment during experimental testing.

A laxity test was then performed to calculate ligament bundle zero-load lengths (the lengths at which ligament bundles first become taut). The humerus was held in a fixed position while the ulna and radius were manually moved through their full range of motion with minimal force applied (as judged by the experimenter) [33]. The kinematic envelope of motion (KEM) was measured from the corresponding bone segments using the attached Optotrak markers and camera system. The zero-load length for each ligament is determined by calculating the maximum straight-line distance between insertion and origin sites of the individual ligaments throughout the range of motion and then multiplying by a correction factor of 0.8 [34]. The purpose of the correction factor is to reduce the error inadvertently introduced by the experimenter during the laxity test when a small amount of force was applied to the ligaments.

After completion of the laxity test, two 100 lb load cells were rigidly attached to the humerus cylinder to measure force in the tendons. The brachialis and triceps tendons were sutured for elbow 1 and biceps and triceps tendon were sutured for elbow 2, and attached to the load cells with a threaded nut and bolt. The sutures were pulled taut and secured to the load cell to provide passive

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