MULTIBODY ANALYSIS OF LIGAMENTOUS AND BONY STRUCTURES INVOLVED IN THE ELBOW JOINT STABILITY

Supervisor: Prof. C. Bignardi
Candidate: Antonella Esposito

a.a. 2018-2019
Ringraziamenti

In primis i ringraziamenti vanno alla Professoressa Cristina Bignardi, per avermi dato l’opportunità di intraprendere questo lavoro e per avermi supportato e seguito durante il lavoro.

Ringrazio l’Ingegner Elisa Panero per la sua immensa disponibilità, gentilezza e per il tempo che mi ha dedicato.

Un ringraziamento davvero speciale va all’Ingegnere Giovanni Putame per avermi seguito in tutto il percorso, per la costante disponibilità e l’immensa gentilezza, nel guidarmi durante il lavoro.
Abstract

The elbow ligamentous together with bony structures play an essential role in the biomechanical stability of the joint. However, the specific contribution of the different structures to joint stability is not completely understood yet. Computational modelling of articular joints, based on experimental data, represents a powerful tool in order to predict the joint behaviour, also giving the possibility to simulate different pathological conditions without the drawbacks related to experimental studies performed with cadaveric specimens (i.e. specimens availability).

In this work, first, a multibody model consisting of bones and ligaments was optimized in terms of measured articular contact forces during a passive flexion. Then, a set of experimental trajectories were included in the model allowing the simulation of a clinical manoeuvre aimed at assessing the elbow stability at three different flexion angles (i.e. 30°, 60° and 90°). Applied forces, needed to accomplish the manoeuvre motion, were deduced by means of a series of PD controllers which minimized errors between experimentally acquired trajectories and computed ones. Such forces were applied to the intact elbow model as well as other two deficient elbow models created by performing a virtual resection of the posterior medial collateral ligament and of the coronoid process, also. Finally, variations of the ulna range of motion resulting from models and experiments during the manoeuvre were compared.

Even if further investigations are necessary, the implemented multibody model seems be able to predict the joint behaviour with intact and deficient stabilizing structures at each considered flexion degree. Furthermore, the PD controller strategies for deducing surgeon’s applied forces during the manoeuvre need to be validated through experimental force measurements. In conclusion, the predictive ability of the presented elbow model let foresee that future injured or surgically reconstructed conditions could be also effectively simulated.
Contents

Introduction
1.1 Anatomical body references
1.2 Articular joint
1.3 The elbow
1.3.1 The Humerus
1.3.2 The Ulna
1.3.3 The radius
1.4 The Elbow joint
1.4.1 Elbow Ligaments
1.4.2 Ligaments behaviour
1.4.3 The interosseous membrane
1.5 Movement of Elbow
1.6 Multibody approach

Materials and methods
1.7 Bones geometries
1.8 Contact force definition
1.9 Upper limit definition for the flexion movement
1.10 Flexion Simulation
1.11 Ligaments implementation
1.12 Maneuver simulation
1.13 PMCL resection
1.14 PMCL and Coronoid resection

Results and Discussion
1.15 Contact forces during flexion
1.16 Contact forces during maneuver
1.17 Forces obtained through PD controllers
1.18 Range of activation of the ligaments
1.19 Comparison between experimental and computed trajectories

Conclusion

Bibliography

Errore. L'origine riferimento non è stata trovata. Errore. L'origine riferimento non è stata trovata.
Introduction

1.1 Anatomical body references

In order to understand the study of biomechanics it is necessary to define a reference for the anatomical position [1]. All terminology refers to this particular position, regardless of what it is the body assumes when it carries out some activity. That consist in:

- Standing position
- Close heel
- Straight arm
- Palm up

![Figure 1: anatomical position](image)

From the anatomical position, three anatomical planes can be defined:

- *sagittal plane*: separates the right half of the body from the left one;
- *coronal plane*: separates the front half of the body from the back one;
- *transverse plane*: separates the upper half of the body from the bottom one;
In addition, three anatomical axes are defined perpendicularly to each plane:

- **longitudinal axis**: perpendicular to the transverse plane;
- **transverse axis**: perpendicular to the sagittal plane;
- **antero-posterior axis**: perpendicular to the coronal plane;
1.2 Articular joint

A joint or articulation is the connection made between bones in the body which link the skeletal system into a functional whole. They are constructed to allow for different degrees and types of movement. Joints are mainly classified structurally and functionally. Structural classification is determined by how the bones connect to each other, while functional classification is determined by the degree of movement between the articulating bones. In practice, there is significant overlap between the two types of classifications.

Structural classification names and divides joints according to the type of binding tissue that connects the bones to each other. There are four structural classifications of joints:

- **fibrous joint**: joined by dense regular connective tissue that is rich in collagen fibers
- **cartilaginous joint**: joined by cartilage. There are two types: primary cartilaginous joints composed of hyaline cartilage, and secondary cartilaginous joints composed of hyaline cartilage covering the articular surfaces of the involved bones with fibrocartilage connecting them.
- **synovial joint**: the bones have a synovial cavity and are united by the dense irregular connective tissue that forms the articular capsule that is normally associated with accessory ligaments.
- **facet joint**: joint between two articular processes between two vertebrae.

Joints can also be classified functionally according to the type and degree of movement they allow. Joint movements are described with reference to the anatomical planes.

- **synarthrosis**: permits little or no mobility. Most synarthrosis joints are fibrous joints
- **amphiarthrosis**: permits slight mobility. Most amphiarthrosis joints are cartilaginous joints
- **synovial joint** (also known as a diarthrosis): freely movable
Placing the attention on the synovial joints, these are formed by an outer membrane called articular capsule, formed by fibrous connective tissue, inside the joint contains ligaments (the fibrous connective tissue that connects bones to other bones), tendons, cartilage and a synovial membrane that secretes a liquid called synovial fluid that acts from lubricant and avoids the rubbing between the two bone segments.[5]

1.3 The elbow

Before going into the study of the elbow biomechanics, it is necessary to describe the anatomy and the characteristics of the bones and ligaments that allow to give established to the elbow allowing also the movement.

1.3.1 The humerus

The humerus is the longest and largest bone of the upper extremity; it is divisible in a body and two extremities.
The upper extremity, consists of a neck, which corresponds to the rounded and restricted part of the head, and by two eminences called tubercles, greater and lesser.
The head, almost hemispherical in form, joins with the glenoid cavity of the scapula. The circumference of its articular is superficially constricted and is termed the anatomical neck [3].
The anatomical neck separates the head from the tubercles and affords attachment to the articular capsule of the shoulder joint.
The greater tubercle is situated lateral to the head and lesser tubercle. Its upper surface is rounded and marked by three flat impressions.
The lesser tubercle, although smaller, is more prominent than the greater: it is situated in front, and is directed medialward and forward.
The body is almost cylindrical in the upper half of its extent, prismatic and flattened below, and has three borders and three surfaces.
The lower extremity ends in a large articulated surface divided into two parts. The projections on both sides are the lateral and medial epicondyles that extend less than the articular surface which is divided into lateral and medial, the lateral part consists of a smooth and rounded emissivity called the
capitulum of the humerus. The medial one is composed of the trochlea which is convex from the back to the front and concave from one side to the other. Above the front part of the trochlea is a small depression, the coronoid fossa, which receives the coronoid process of the ulna during flexion of the forearm. Above the back part of the trochlea is a deep triangular depression, the olecranon fossa, in which the summit of the olecranon is received in extension of the forearm.

The lateral epicondyle is a small eminence curved a little forward, which thanks to the collateral radial ligament binds to the radius.

The medial epicondyle is larger than the lateral one and allows to bind the humerus to the ulna thanks to the ulnar collateral ligament.

The bone structure of the extremities consist of cancellous tissue, covered with a thin, compact layer, the body is composed of a cylinder of compact tissue, thicker at the center than toward the extremities, and contains a large medullary canal which extends along its whole length [3, 4].

![Humerus anatomy](image)

Figure 4: humerus anatomy

1.3.2 The ulna

The ulna is a long bone, prismatic in form, placed at the medial side of the forearm, parallel with the radius. It is divisible into a body and two extremities. Its upper extremity, of great thickness and strength, forms a large part of the elbow-joint; the bone diminishes in size from above downward,
its lower extremity being very small, and excluded from the wrist-joint by the interposition of an articular disk. The upper extremity presents two curved processes, the olecranon and the coronoid process; and two concave, articular cavities, the semilunar and radial notches.

The olecranon is a large and thick eminence that is found in the superior and posterior part of the ulna, it is bent forward so as to connect to the oleocreanus humerus during the extension of the forearm. Its upper surface is quadrilateral and rough in order to bind the triceps brachial and has a slight groove for fixing the posterior ligament of the joint and the elbow. The front surface is smooth and concave and forms the upper part of the lunate notch.

The coronoid process is an emiscence found in the anterior and superior part of the ulna. its apex is pointed and is received in the coronoid fossa of the humerus during flexion. Its upper surface is smooth and concave and forms the lower part of the lunate notch. Its anterior-inferior surface is concave and marked by a rough impression for the insertion of the Brachialis while the median one allows the fixing of part of the ulnar collateral ligament [3].

The radial notch is a narrow, oblong, articular depression on the lateral side of the coronoid process; it receives the circumferential articular surface of the head of the radius.

The long, narrow medullary cavity is enclosed in a strong wall of compact tissue which is thickest along the interosseous border and dorsal surface. At the extremities the compact layer thins. The compact layer is continued onto
the back of the olecranon as a plate of close spongy bone with lamelle parallel [4].

![Figure 6: ulnar anatomy, a) posterior face, b) anterior face](image)

1.3.3 The radius

The radius is situated on the lateral side of the ulna, which exceeds it in length and size. Its upper end is small, and forms only a small part of the elbow joint; but its lower end is large, and forms the chief part of the wrist joint. It is a long bone, prismatic in form and slightly curved longitudinally. It has a body and two extremities.

The upper extremity presents a head, neck, and tuberosity. The head is of a cylindrical form, and on its upper surface is a shallow cup or fovea for
articulation with the capitulum of the humerus. The circumference of the head is smooth; it is broad medially where it articulates with the radial notch of the ulna, narrow in the rest of its extent, which is embraced by the annular ligament. The head is supported on a round, smooth, and constricted portion called the neck.

The body has the shape of a prism, narrow above and wider below and slightly curved.

The lower extremity is large, of quadrilateral form, and provided with two articular surfaces one below, for the carpus, and another at the medial side, for the ulna. The carpal articular surface is triangular and divided by anterior-posterior ridge into two parts: the lateral and the medial. The articular surface for the ulna is called the ulnar notch of the radius, it is narrow, concave, smooth, and articulates with the head of the ulna. At the base of this two surface there is a triangular articular disk, that separates the wrist-joint from the distal radioulnar articulation. The lateral surface is prolonged obliquely downward and gives attachment to the tendon of the Brachioradialis and to the radial collateral ligament of the wrist joint.

The compact tissue is often along the interosseous edge and becomes thinner at the ends excluding the head, it envelops the medullary cavity. The trabeculae of the spongy tissue are somewhat curved upwards [3].
1.4 The elbow joint

The elbow joint consists of the articulations between the distal humerus, the proximal radius, and the proximal ulna. The elbow is one of the most congruous joints in the body. There are three articulations in the elbow: the ulnohumeral joint, the radiocapitellar joint, and the proximal radioulnar joint. The ulnohumeral joint is a hinge joint between the humeral trochlea and the trochlear notch of the ulna; allowing flexion and extension. The radiocapitellar and radioulnar joints are trochoid joints, allowing axial rotation or pivoting. The trochlea is covered by articular cartilage over an arc of 300°. The capitellum is separated from the trochlea by a groove, which articulates with the rim of the radial head throughout flexion/extension and pronation/supination.

Figure 8: right elbow joint, A) anterior; B) posterior
1.4.1 Elbow Ligaments

The ligaments are made to bind a bone with another bone, transmit the loads between the bones and keep the skeleton together. They guarantee joint stability and limit freedom of movement when necessary. In general, they are composed of connective tissue with densely packed collagen fibers. The fibers therefore appear white and are relatively non-elastic. The cells that make up ligaments are called fibroblasts. They are composed of elastin and collagen which gives them a degree of hardness. This type of ligaments is called articular ligaments. Ligaments are viscoelastic so they gradually strain when are under tension, and return to their original shape when the tension is removed. However, they cannot retain their original shape when extended past a certain point or for a prolonged period of time. This is one reason why dislocated joints must be set as quickly as possible: if the ligaments lengthen too much, then the joint will be weakened, becoming prone to future dislocations. The consequence of a broken ligament can be instability of the joint. Not all broken ligaments need surgery, but, if surgery is needed to stabilise the joint, the broken ligament can be repaired. Ligaments of the elbow joint are the medial and lateral collateral, the annular and the squared one.

The medial collateral ligament (MCL) originates from the anterior inferior surface of the medial epicondyle and joins the ulna to the humerus, providing support and resistance in valgus overloads. This ligament is divided into an anterior bundle (AMCL), which is stressed during the elbow extension movement; a posterior bundle (PMCL), which is stressed during elbow flexion; and a transverse band, which joins the anterior and posterior bands. Each band of the ligament presents different functions during elbow flexion and extension movements [6][7].

The AMCL bundle is the strongest ligament with a mean load failure of 260 N and originates from the inferior anterior part of the medial epicondyle of the humerus and inserts itself on the ulnar side of the coronoid process, near the sublime tubercle. The AMCL bundle is also divided into an anterior and posterior band including a third central isometric band, called “guiding bundle”; each of these bands has a different role and angle of activation during the flexion-extension movement. Other anatomical features were
investigated in several studies and they found an origin area of 45.5 mm$^2$, an insertion area of 127.8 mm$^2$, a mean length of 28 mm; it is composed of thick parallel fibres with an increasing amplitude moving from the origin to the insertion, with a mean value of 5.4 mm[8][9].

The PMCL originates from the posterior part of the medial epicondyle and is inserted in the medial part of the olecranon. It has a fan-shaped shape of the posteromedial capsule. Compared to the front beam, the fibers here have an amplitude of less than 5-6 mm in the central part and an average of 8.8 mm with a length of 16 mm[9].

![Figure 9: MCL](image)

-the lateral collateral ligament complex (LCL): is composed by four elements, the radial collateral ligament (RCL), the annular ligament (AL), the lateral ulnar collateral ligament (LUCL) and the accessory posterior annular ligament;

The RCL originates on the surface of the lateral epicondyle and fits on the annular ligament, it also provides an origin for the supinator muscle. This ligament has an average length of about 20 mm and an amplitude of 9.7 mm[9]

The LUCL Like RCL it originates on the surface of the lateral epicondyle but is inserted on the supinating crest of the ulna, distal to the annular ligament. The length and amplitude are not defined because they vary depending on where it is genetically positioned.[10][11]
The AL is a fibrous bundle entirely covered with cartilage which fixes the posterior and anterior ends of the radial notch of the ulna and together with it forms an osteo-fibrous ring that fixes the radial head and allows the movements of rotation in particular and supination and pronation.[11]

1.4.2 Ligaments behaviour

In order to study the elbow in greater depth, anatomy is not enough, but biomechanics must also be understood in a particular way, as the activity of ligaments is essential for the stability of the elbow. From the study by Regan on the biomechanics of the elbow, it appears that the activity of the ligaments varies with the movement and the type of ligament, in particular an attempt is made to associate a biomechanical behavior to the ligaments based on their load-deformation relationship and to the angle that allows activation of the ligament, ie when the ligament is tense. In addition, tests were also performed during flexion and in a varus position, worth to understand how and how varied the activation and deactivation of ligaments. These movements are made by the immobilization of the humerus. When applying a valgus stress, the hand is supinated and it consists in an external forearm’s rotation in the transversal plan, around the vertical axis passing through the elbow joint, provided by an abducting force on the forearm; in complete
extension, this movement reduces the external angle between the forearm’s and arm’s axes. The varus stress presents the same test conditions of the valgus one, but the force is adducting, so the forearm’s rotation in the transverse plan is internal, so in complete extension the external angle between forearm’s and arm’s axes increases. From Regan’s experiments result that:

The AMCL is well defined, robust and quite distinct from the medial capsule; dividing it into three parts (anterior, medial and posterior) and carrying out movements of flexion, varus and valgus it can be observed that, in simple flexion, the anterior part is active from the beginning of the flexion up to 50 °, the medial part instead is activated at 20 ° and remains active throughout the bending range, the rear one is initially deactivated and then activated at 85 °. For the valgus movement instead the activations increase compared to the simple bending, in particular, the front part remains active up to 80 °, the medial one is always active and the rear part is anticipated, activating at 55 °. On the contrary instead in the varus the activation ranges decreasing compared to the bending in fact, the front part is active at the beginning of the movement up to 30 °, the medial is active from 50 ° up to 130 ° and the rear one is activated at 110 °.[9]

![Figure 11: tautness of AMCL](image-url)
The **PMCL**, which is also divided into three bands, is initially deactivated for all movements, and is activated only after a certain angle of flexion, in particular, for simple bending it is activated at 80° for the front band, 90° for the medial and 110° for the posterior band. For the valgus movement it is activated at 70° for the front band, 90° for the medial band and 110° for the posterior one. Finally, for the varus movement the activation is after the 90°, 100° and 110° respectively for the anterior, medial and posterior band. From these results I therefore expect not to notice any significant changes due to bending below 90° in the case where it is decided to cut this ligament.[9]

![Figure 12: Tautness of PMCL](image)

The **RCL** is a poorly demarcated fan shaped structure, that is intimately associated with the lateral joint capsule, deep to it; during a simple flexion, its anterior bundle is taut from complete extension to 40°, the posterior portion from 90° to complete flexion, and the middle portion is always taut. Applying a load in valgus the situation remains almost the same, for the front band the ligament becomes stretched after a few degrees more than the simple flexion, while for the posterior and medial band the situation remains the same. For the loading in varus instead for the front band it remains active up to 50°, for the medial it is activated after 10° and remains active for the
entire duration of the movement, for the rear band instead it is activated after a drop equal to 70°. [9]

![Figure 13: tautness of RCL](image)

The *ULC* is considered as a various capsular thickening extending from the lateral humeral epicondyle to the olecranon; with no stress application, the ligament fibres become taut at 107°, meanwhile with valgus stress they activate at 120° and with a varus stress this ligament is always taut. [9]
1.4.3 The interosseous membrane

The interosseous membrane (IOL) is composed of fibrous tissue, and divides the anterior forearm muscles from the posterior ones. It is placed at the center between the ulna and radius bone segments. The role of IOL is also to give stability and keep the ulna with the radius joining in the transverse plane providing a joint reaction force and therefore stability to the proximal joint. The lower margin of the membrane merges with the distal radioulnar joint capsule, while the upper margin stops a few centimeters before the elbow joint; under the radial tuberosity, there is the oblique chord, a fibrous bundle that is orginate and expands towards the ulnar coronoid process. The IOL comprises several components, the most important of which is the central band. Although the radial head is the primary stabilizer against longitudinal instability, the central band and the triangular fibrocartilage complex (TFCC) provide important secondary support. In fact, in the event of a radial head fracture, the TFCC contributes to the stability of the forearm. [13][14]
1.5 Movement of Elbow

Anatomically the elbow is a single articulation, in fact it has an only articular cavity. From its physiology, it is possible to distinguish two separate functions:

- The prone-supination: that makes active the proximal radio-ulnar articulation.
- The flex-extension that requires the activation of two articulations, humeral-ulnar and radial- humeral.

The first, prone-supination movement consists in a rotation of the forearm around the longitudinal axis; this movement requires the activation of two mechanically linked articulations: proximal and distal radio-ulnar.

The second, flexion and extension are movements that occur within the sagittal plane and involve anterior or posterior movements of the body or limbs. As for the elbow, flexion and extension occur on the forearm, so the bone segments that go to flex or extend are ulna and radius, up to an angle of 145° even if in reality the work done in this thesis has only bending angles equal 0°, 30°, 60° and 90° were taken into consideration. during flexion-extension movements, as has already been said before, the activation and deactivation of some ligaments that are part of the elbow joint are started.[15]
1.6 Multibody approach

After understanding the anatomy and biomechanics of the elbow thanks to the studies found in the literature, we need to combine this study with a study done through a multibody system, that is a numerical technique used in biomechanics, in order to understand how the elbow really behaves during a movement of simple bending or during a maneuver, being able then to compare the results obtained with the experimental ones. Multibody system means a mechanical system from a set of rigid bodies connected together so as to have a relative motion. The elements of a multibody system are connected to each other by kinematic pairs or joints. In other directions. The elements that make up a system are: bodies (generally rigid), kinematic constraints or pairs, (external) forces. Multibody system simulators are programs that perform the analysis of the motion of mechanical systems. More precisely, they are able to deal with both kinematics and system dynamics. This feature is characterized by a 3D animation software able to take into account only the kinematics. This thesis work, the multibody simulator is MSC Adams, the solid element is the human ulna, the humerus and radius, and the connection between the bodies. Through this approach, it's possible to study the dynamic and cinematic behavior of the human elbow when different types of forces are applied, also simulating different surgical conditions and movements.
Materials and Methods

1.7 Bones geometries

The model from which we started was a model created through various steps. The first step was to create the solids of the bones, this was done through Rhinoceros 3D, subsequently, individual Parasolid files containing ulna, humerus and radius were saved. The purpose of the preparation of the elbow pattern is to accurately reproduce every anatomical characteristic of the physiological joint particularly for the position and behavior of the ligaments, so it was important to place them in the correct position referring to the anatomical one, to realize this, a second CAD software, Solidworks by individually importing the Parasolid files containing the closed surfaces, so the solids were created, different spheres in specific points of anatomical interest to fix the origins and insertions of each elbow ligament. Parasolid files created in Solidworks® were individually imported into the MSC Adams software. The software treats these imported objects as rigid bodies and automatically sets them up as bodies of steel; to make this model accurate, it was necessary to create in Adams® a new material that could reproduce the bone features. The division of the bone into the two different types (cortical and trabecular) was not considered, but only the cortical bone was realized. To define the bone properties, the following parameters have been entered in the dialog box to define the mass:

- Young Module: 20000 MPa
- Poisson Ratio: 0.45
- Density: $1.6 \times 10^{-6}$ kg / mm$^3$

![Figure 18: Adams model](image)
1.8 Contact force definition

Another important element to set to allow the movement of the joint, are the forces of contact between the bodies to correctly describe the nature of the interaction between the bones of the upper limb; The contact force was set for each of the three long bones of the limb. To set the correct contact parameters, several data found in the literature were tested analysing which of these mainly affected the pattern and trend of the contact force. The shooting parameters to be set are stiffness, force, exponent, damping, penetration depth and friction forces it has been observed that the most influential parameter is the rigidity that as for the force of friction between joint bodies, it has been neglected to prevent penetration between bodies. In Adams, it is create a contact force between two bodies by selecting the related icon in the list of forces then selecting the affected bodies, then choosing the normal force (in this case it is the IMPACT type) and finally entering the coefficients of the contact

The values of the selected parameters are:

- Stiffness: 80000 N/mm
- Exponent Force: 2
- Dampening: 400 Ns /mm
- Depth of penetration: 0.001 mm
- No friction force

Let's create a contact for each pair of bodies:

- Ulna and radius
- Ulna and Humerus
- Radius and Humerus

1.9 Upper limit definition for the flexion movement

To achieve precise model flexion motion at a specific stop angle, it is need a motion sensor. To activate the sensors, it is need to define a set of parameters in the dialog box: the first is the "Event Definition", time-dependent expression, which is a measure to be set, then the "Event Assessment" (is an optional feature) which is useful if a certain output is desired when the sensor is active. It is important to check the "Standard action" option: in this
case it is selected the third choice and also the "Continue the simulation script or ACF file" option if the simulation should continue after the sensor is activated.

To take advantage of a sensor, it is need to set an angular measurement and when this measurement reaches the value entered in the sensor dialog box activates; an angular measure was then created for the bending by considering several markers.
1.10 Flexion Simulation

To reproduce the movements of the elbow, it is necessary to set up a script in which the instruction code must be entered, in particular, these instructions are necessary for the first time in the first phase. In the dialog box, it is need to set "Commands Adams / Solver" as the type of script and then it is possible create the script via the list of possible actions in "Add ACF command". In the case of bending, the simulation will follow the bending angle value entered in the sensor dialog box to interrupt the simulation.

![Image](image.png)

*Figure 21: simulation of flexion*

1.11 Ligaments implementation

In order to create the ligaments at the beginning it was thought to use springs, but they act both in traction and compression instead something needed to act only in traction and therefore I used single-component forces acting along the direction line between two bodies, one of action and one of reaction, defined by Adams. Unlike springs, ligaments are not a linear spring because they are produced from cartilage fibers that makes them viscous fabrics, in which the fibers are activated and progressively align along a preferential direction with force.
In the figure 24 were showed information about the ligaments angles of activation/deactivation during the three base elbow’s movements [9] values of stiffness related to the displacement between origin and insertion at the angle of activation, and the percentage of strain used to pull out the displacement activation for the transverse ligaments.
The written function to create the forces that represent the ligaments is the following:

\[
IF(\varepsilon - c \cdot VR(OR,INS) - 14 \cdot K \cdot \varepsilon^2 \varepsilon_l - c \cdot VR(OR,INS),
IF(\varepsilon - 2 \cdot \varepsilon_l - 14 \cdot K \cdot \varepsilon^2 \varepsilon_l - c \cdot VR(OR,INS), -14 \cdot K \cdot \varepsilon^2 \varepsilon_l - c \cdot VR(OR,INS) - K \cdot (\varepsilon - \varepsilon_l) - c \cdot VR(OR,INS))
\]

- The rigidity parameters (K) used were shown in the table and multiplied by the length of the zero load.
- The non-linear deformation parameter (\(\varepsilon_l\)) was assumed 0.03 for each ligament.
- The damping parameter (c) has been set to 0.5 Ns / mm.
- The term VR does not alter the linearity of the equation and the effect of high frequency.

Table 1: Mechanical characteristics of ligaments.
The length of the zero load was obtained through a point-to-point measurement between the origin and the insertion markers of each ligament, in the initial static extended position, created and then used in the following expression:

\[ l_0 = l_1 + \varepsilon \]

**First Validation**

The first validation was that of a simple movement up to 90°, as regards the activations and deactivations of the ligaments, in particular from the post-processor were plotted, the trends of each single ligament, created as forces and were compared with the activations and deactivations reported by Regan et al. [9].

- The anterior AMCL deactivates at 50° of flexion

![Figure 23: anterior AMCL](image)

- The posterior AMCL activates at 80° of flexion

![Figure 24: posterior AMCL](image)
- AMCL Regan

**Figure 25: Tautness of AMCL**

- The anterior PMCL activates at 80° of flexion

**Figure 26: anterior PMCL**

- The posterior PMCL activates at 110° of flexion

**Figure 27: posterior PMCL**
-PMCL Regan

![Figure 28: Tautness of PMCL]

- The anterior RCL deactivates at 40° of flexion

![Figure 29: Anterior RCL]

- The posterior RCL activates at 60° of flexion

![Figure 30: Posterior RCL]
Then going on to plot the contact forces between the ulna and the humerus, radio humerus, it was seen that they were too high compared to the contact forces that normally act in the elbow joint, especially for a passive flexion movement. So the first conclusion was to find a way to less contact forces, of a value below 50N. In reality, this was established based on experiments found in the literature that simulated active flexion movements, because in literature no studies and experiments were simulated that simulated passive flexions and showed values of the contact forces obtained. In order to do this it was decided to try to stretch the ligaments, doing various tests on which ligaments were stretched and by how much. In particular, what has changed is the initial length, $L_0$, of the various ligaments. Initially the modified ligaments were all those of the elbow joint, which then join the ulna with the humerus and the radius with the humerus, in particular the anterior and posterior AMCL, the anterior and posterior PMCL, the anterior and posterior RCL and finally the Ulnar ligament, because they are the only ligaments that affect the contact forces during flexion. To make this more immediate, a design variable called “perc” was created and added to each initial length of the previously mentioned ligaments.
The first test was to lengthen the initial length by 50%, then by 10% and then by 5%, with a 50% elongation there was immediately a clear reduction in contact forces with values below 10N, being that the acceptable values were those with contact forces of less than 50N, and that if we had stretched the ligaments too much we would have been detached enough from the actual physiological values, other tests have been done extending these by 10% and then 5%. And indeed the values of contact forces had decreased. Finally, the last test that was done was to understand if there was actually one of the ligaments that most influenced contact forces during flexion. After various experiments and looking also at the activations and deactivations, it was decided to lengthen only the anterior AMCL by 3%, being the ligament that most influenced.

1.12 Maneuver simulation

The next step was to simulate the experimental maneuver carried out on a cadaveric arm, the complete maneuver was carried out by first making passive flexion movements at 30 °, 60 ° and 90 °, and at each of these angles, the surgeon carried out of maneuvers, having only the trajectories of the maneuvers carried out experimentally and not knowing the forces exerted to carry out these maneuvers, these trajectories were imported on the multibody system, through the motion capture technique; and through an appropriate control operator it was possible to trace back to the forces impressed and to impose movement with these obtained forces, all this to simulate the same experimentally done movement.
To do all this the first step was to create three vectors of forces in the three directions $X, Y, Z$, placed at the distal end of the ulna, because the maneuver was carried out by making movements of the arm by imposing the force on the wrist (as you can see from the experimenental image). These three forces act according to a PD controller that measures the error of how much the motion agent attached to the ulna (which will allow the maneuver) is detached and the ulna, in order to generate this control, it is necessary to carry out displacement and velocity measurements along $x, y, z$ between the ulna reference marker and the ref agent that corresponds to the reference marker placed on the motion agent.
Precisely because the controller corresponds to a feedback circuit in which the displacement or speed is measured that corresponds to 0 in relation to the displacement and to the real speed created between the ulna and the motion agent along the three directions.

The three force vectors, along X, Y, Z were then defined through the PD.
The exact same thing was done by imposing three rotations at the same point where the forces were put, these three rotations work the same way with the PD controller except that unlike the forces, which work by measuring displacement and speed, in this case the angle of rotation and the angular velocity are measured.

The trajectories and rotations of the maneuver were then imported, obtained experimentally using the motion capture technique, in the form of splines, these trajectories being very noisy were filtered through a Butterworth low pass filter with a cutoff frequency of 10 Hz. When they were filtered it was found that they had a number of samples equal to one thousand, lower than the originals, therefore they were resampled with the actual number of samples
Figure 37: Experimental and filtered curve $X30^\circ$

Figure 38: Experimental and filtered curve $Y30^\circ$

Figure 39: Experimental and filtered curve $Z30^\circ$
Figure 40: Experimental and filtered curve $X{60}^\circ$

Figure 41: Experimental and filtered curve $Y{60}^\circ$

Figure 42: Experimental and filtered curve $Z{60}^\circ$
Once resampled they were inserted inside a motion, which allows to move the motion agent, tied to the ulna through the three force vectors and the three
rotation vectors. The motion agent, then moving causes the ulna to follow and carry out the maneuver.

Figure 46: a) motion agent b) motion agent dialogue boxe

STEP5(TIME-senval(Sensor_flex),0,0,0.1,1)*(CUBSPL(TIME-senval(Sensor_flex),0,traj_LP,0)+(first value of traj)

- CUBSPL that includes the experimental trajectory filtered in the form of splines
- FIRST VALUE OF TRAJ that must be added or subtracted to make the end of bending point coincide with the starting point of the maneuver

The commands were then written inside the ‘Simulation script’, in which initially the forces, rotations and motion of the maneuver were deactivated for bending. Having arrived at the imposed degree of flexion, motion, sensor and joint that allowed bending were deactivated and the forces, rotations and motion necessary to perform the maneuver were activated; the real time used in the experimental part was chosen as simulation time for the maneuver.
But in order to effectively compare the multibody model with the experimental one it is necessary to simulate the maneuver by imposing the right forces and rotations, which initially we did not have but that we obtained through the PD controller then, simulated the maneuver with the control, in the post processor, were plotted the curves of the forces and rotations obtained and thus created the new splines to be inserted into three other force vectors and three of rotation, used to simulate the actual maneuver. These forward vectors force and rotation have been fixed in the same points as those used with the PD controller, which, using the 'cubic fitting method' function, allows to use the previously created splines and to impose the force and rotation necessary to perform the maneuver.
By composing a new simulation script for direct forces and rotations, the maneuver was simulated with a similar trend, except for a minimum percentage error for the experimental simulation maneuver.
The cases were analyzed for a flexion of 30°, 60° and 90°, for physiological elbow, with PMCL cut and with PMCL cut also carrying out the cut of the coronoid, for each case of simulation were imported and performed simulations with the maneuver corresponding.

1.13 PMCL resection
To carry out the PMCL cut, it was enough to simply deactivate the anterior PMCL force and posterior PMCL, even if from the results it was noticed that the Posterior PMCL does not act because, according to Regan, it is activated for a flexion above 100°, while the front one activates for 80° of flexion, the simulations with the PMCL cut, the procedure is exactly analogous to that used for the physiological elbow, it was enough therefore after having deactivated the ligament, redo the simulations made for the three cases of flexion.

1.14 PMCL and Coronoid resection
To cut the coronoid, the procedure was slightly different, as a 10cm cube was created, through the 'display precision move dialog box' command, entering appropriate coordinates, it was moved to the position useful for cutting the coronoid, and using it is the coronoid has been cut from the ulna. The maneuvers for the various flexion cases were then simulated, continuing to keep the PMCL anterior and posterior deactivated ligament.

Figure 50: cube creation
Result and Discussion

By analyzing the results on the various tests done, one is able to effectively understand how the elbow joint works, in particular to understand which are the ligaments that act and influence during flexion-extension and during any maneuvers.

1.15 Contact forces during flexion

For the first validation done, having started from a model already with a base, it was noticed that by simulating the flexion, the contact forces that intervene, that is ulna-humerus and radio-humerus were too high compared to a real condition of a physiological elbow, so what we tried to do was to try to obtain contact forces between ulna-humerus and radio-humerus less than or equal to 50N, it was thought that by changing the initial length L0 (which has a set value for each of the ligaments), only of the ligaments that make up the elbow joint, the value of the contact force could change.
In order to make a comparison, it was plotted the curves of contact forces of the previous model, as shown in the figure 55, 56, the contact forces are very high.

![Figure 53: Radio-humerus contact force](image1)

![Figure 54: Ulna-humerus contact force](image2)

As can be seen from these graphs, the contact forces reach very high and improbable values, or rather if there were muscles besides the bones and, if the flexion was active, these values would be acceptable, in fact from the studies of Chadwick and Nicol on muscle and bone forces that act in the elbow joint during work activities, in particular on the basis of active flexion movements with vertical and horizontal grip of a force transducer, showed values of ulna-humer contact force of 1600 N and radio - humerus equal to 800 N. And the muscular forces highlighted in the graph. From these results, therefore, it is evident that the values of the contact forces, without the muscular bundle, due to the passive flexions, found by the previous graphs of the model, are too high, which is why we tried to find a way to reduce them. [16]
To do this, therefore, the first test was to lengthen all the ligaments that act in the elbow joint and then the anterior and posterior AMCL, anterior and posterior PMCL, anterior and posterior RCL and an ulnar 50% of their initial length.

The results show in the Figure 58 and Figure 59, show a clear decrease in strength:

Figure 55: Distribution of load across the structures at maximum joint force for the horizontal power grip

Figure 56: Radio-humerus contact force

Figure 57: Ulna-humerus contact force
Since the contact forces had clearly decreased almost to zero, but lengthening the ligament by 50% of its initial length was a very unlikely condition, then the next test was to lengthen the initial length by 10%.

![Figure 58: Radio-humerus contact force](image1)

![Figure 59: Ulna-humerus contact force](image2)

Also in this case the contact forces are very low, below 50N. It was then decided to lower further and lengthen the initial ligament length by only 5%, in this way it was the most realistic case, the contact forces remained below 50N.

![Figure 60: Radio-humerus contact force](image3)
Initially, therefore, this last test was taken into consideration, thus lengthening the initial length of the L0 ligaments by only 5%, but then other tests were done, to understand if there were ligaments that most influence the contact forces, after various simulations, it was discovered that the anterior AMCL affects more than the others and therefore it was decided to lengthen only the anterior AMCL ligament by 3% compared to its initial length. In this way, the contact forces do not exceed 50N, the ligaments, excluding the anterior AMCL, continue to have the actual length found in the literature and moreover the curves are less noisy.
1.16 Contact forces during maneuver

Ulna-humerus and radio-hume contact forces were also evaluated, for the various maneuvers carried out, for the various flexion angles of the three cases, intact elbow, PMCL and PMCL and coronoid cut. Having done more simulations for a 30 ° bending maneuver, the graphs show contact forces also in this case lower than 50 N, excluding the case of the coronoid cut. that the contact forces are slightly higher this is due precisely to the cut of the coronoid during the maneuver, the arm tends to have less stability and during the maneuver it moves and therefore the ligaments generate more traction to look for stability and for this increase the forces of contact.

Both physiological and with PMCL the contact forces are lower than 50N, with the further cut of the coronoid instead the contact forces of the ulna-humerus are slightly higher this is always due to the cut of the coronoid that causes movement during the maneuver of the arm that therefore loses stability, to find stability increases the strength of the anterior AMCL which,
being the ligament which has the greatest influence on the contact forces, consequently causes its increase.

For cases of physiological elbow and cut of the PMCL, the contact forces of the ulna-humerus and the radio-humerus are almost identical precisely because the 30° PMCL is not active.

The same condition occurs for the case of a 60° flexion maneuver in which the curves of the physiological maneuver and with PMCL cut are identical because the PMCL is not active, in the case of the coronoid cut the contact forces increase for the same reason as the 30° maneuver.
The contact forces for a 90° bend maneuver, we expect them to change with the PMCL cut because the PMCL is active.

In theory we expect that the 90° flexion contact forces vary in the case of PMCL cutting, but from the first optimization done on the ligaments, we have validated that the ligament that has the greatest influence on contact forces is the anterior AMCL. This is why we do not see obvious changes in these graphs. With the cutting of the PMCL and the coronoid, the results do
not show appreciable variations probably because the geometry of the model is different from that of a human elbow.

1.17 Forces obtained through PD controllers

- maneuver 30°

![Figure 71: force_{xPD}_{30°}](image1)

![Figure 72: force_{yPD}_{30°}](image2)

![Figure 73: force_{zPD}_{30°}](image3)
--maneuver 60°

Figure 74: force_xPD_60°

Figure 75: force_yPD_60°

Figure 76: force_zPD_60°
-maneuver 90°

Figure 77: force_xPD_90°

Figure 78: force_xPD_90°

Figure 79: force_xPD_90°
Even in the case of flexion at 30°, 60° and 90° with successive maneuvers, having extended the initial length of the front AMCL by 3%, the forces remain below 50N which was the condition we wanted to achieve. Because what it need to do is calibrate the PD parameters to minimize the error while maintaining the force on 30N. This is because the geometry of the human elbow is different from the geometry of the model, so with the PD controller if the forces increase, the error decreases, but it will come to a certain point where this error will no longer be able to decrease while the forces continue to increase and if they increase too much, due to the geometry of the model they cause the separation of the ulna from the humerus, to avoid this it is necessary to calibrate the parameters well.

-Error_30°

![Figure 80: error_30°](image)

-Error_60°

![Figure 81: error_60°](image)
1.18 Range of activation of the ligaments

A further validation was to simulate a $90^\circ$ flexion and evaluate the activations and deactivations of the ligaments, these were compared with the activations and deactivations of the Regan ligaments.

- The anterior AMCL deactivates at $50^\circ$ of flexion
- The posterior AMCL activates at 80° of flexion

![Figure 84: posterior AMCL](image1)

- AMCL Regan

![Figure 85: tautness of AMCL](image2)

- The anterior PMCL activates at 80° of flexion

![Figure 86: anterior PMCL](image3)
- The posterior PMCL activates at 110° of flexion

![Figure 87: posterior PMCL](image)

- PMCL Regan

![Figure 88: tautness of PMCL](image)

- The anterior RCL deactivates at 40° of flexion

![Figure 89: anterior RCL](image)
- The posterior RCL activates at 60° of flexion

Figure 90: posterior RCL

-RCL Regan

Figure 91: tautness of PMCL

these return with the activations and deactivations of a human elbow.
1.19 Comparison between experimental and computed trajectories

The last results that are shown are probably the most important because they represent the main purpose of this work, that is to try to build a model through a multibody system that is able to simulate and reproduce the experiments performed experimentally, in particular these results show the comparison between the trajectories obtained experimentally (blue line) and the trajectories obtained from the model (red line), during the various maneuvers made at the various degrees of flexion 30 °, 60 °, 90 °. The relative displacements of the experimental curve and the curve of the model have been plotted, in order to better compare them these curves have been repositioned so as to start from the same point.

For the maneuver 30°, 9 different graphs have been reproduced as shown in the figure [93-102] which represent the three cases analyzed, physiological elbow, with PMCL cut, PMCL cut and coronoid cut. For each of these cases the results were shown on the three anatomical planes:

- transverse plane (XY)
- coronal plane (XZ)
- Sagittal plane (YZ)

-Maneuvre 30° Transverse plane
Figure 92: Maneuvre 30° Transverse plane, Physio

Figure 93: Maneuvre 30° Transverse plane, PMCL cut
- Maneuver 30° Coronal plane
Figure 96: Maneuvre 30° Coronal plane, PMCL cut

Figure 97: Maneuvre 30° Coronal plane, PMCL and Coronoid cut
Manoeuvre 30° Sagittal plane

Figure 98: Manoeuvre 30° sagittal plane, Physio

Figure 99: Manoeuvre 30° sagittal plane, PMCL cut
As shown in the images, in the maneuver 30° for the various cases, it seems that in the physiological case the two trajectories are superimposed, this means that the displacement seems to be congruent, for the PMCL cut, the curve is not congruent with the experimental curve but is congruent with that of the model in the physiological case being the PMCL deactivated for that degree of bending. For the cut of the coronoid, the curve varies in a congruent way for the displacement with the experimental displacement, but incongruent with the amplitude.
-Manoeuvre 60° Transverse plane

Figure 101: Maneuvre 60° Transverse plane, Physio

Figure 102: Maneuvre 60° Transverse plane, PMCL cut
Figure 103: Maneuvre 60° Transverse plane, PMCL and Coronoid cut

-Maneuvre 60° Coronal plane

Figure 104: Maneuvre 60° Coronal plane, Physio
Figure 105: Maneuvre 60° Coronal plane, PMCL cut

Figure 106: Maneuvre 60° Coronal plane, PMCL and coronoid cut
-Manoeuvre 60° Sagittal plane

Figure 107: Manoeuvre 60° sagittal plane, Physio

Figure 108: Manoeuvre 60° sagittal plane, PMCL cut
For the 60° maneuver, on the other hand, there is a greater congruence of displacement for all three cases, in the physiological case the curves seem to overlap less than a very small error, in the case of the PMCL cut the displacement seems to be similar, what varies is the amplitude, in the experimental curves we also find a higher number of samples because the surgeon performed the movement twice. For the cut of the coronoid we do not find congruence with the experimental curves, but we can note a variation along X of the multibody model with respect to the physiological case.
-Manoeuvre 90° Transverse plane

Figure 110: Maneuvre 90° transverse plane, physio

Figure 111: Maneuvre 90° transverse plane, PMCL cut
Figure 112: Maneuvre 90° transverse plane, PMCL and Coronoid cut

-Maneuvre 90° Coronal plane

Figure 113: Maneuvre 90° Coronal plane, Physio
Figure 114: Maneuvre 90° Coronal plane, PMCL cut

Figure 115: Maneuvre 90° Coronal plane, PMCL and coronoid cut
Maneuvre 90° Sagittal plane

Figure 116: Maneuvre 90° sagittal plane, Physio

Figure 117: Maneuvre 90° sagittal plane, PMCL cut
Finally, for the 90° maneuver, in the physiological case the curves are superimposed, therefore the displacement of the maneuver is the same, in the other two cases in addition to an incongruity on the amplitude of the displacement, we can observe that the displacement does not seem quite congruent. Furthermore, the curve of the multibody model with the coronoid cut and PMCL has a tendency very similar to the physiological case, the optimizations made in reality are correct since it is the AMCL that influences. With the cut of the coronoid it remains the same or in any case it moves from very little to the other two and because the geometry of the model is different from the geometry of the anatomical elbow and therefore perhaps the geometry of the model does not allow to appreciate the difference with and without coronoid. Furthermore, another reason could be that the cut made, to remove the coronoid, as shown in the figure 121, is actually smaller than the one experimentally made, figure 119. But the reason why it was not possible to cut further is to go to the position of the ligament insertion points.
Figure 119: coronoid experimental cut

Figure 120: coronoid experimental cut

Figure 121: coronoid model cut
-Trajectory amplitude comparison

to highlight the different trends of the experimental trajectories with those of the model at different degrees of flexion, amplitude along the x-axis was measured, with the reference system shown in the figure. From the data shown, it is noted that the experimental trajectories have a very different pattern with the cutting of the PMCL and the coronoid compared to the physiological one. Which is not the case in the multibody model because the cut coronoid section is smaller than the experimentally cut section and so it doesn't seem to affect. On the other hand, in the 30° and 60° manoeuvre, there is a difference between the amplitude of the physiological model and the one with a cut conoid.
Conclusion

The purpose of this work was to optimize a pre-existing multibody model of a human elbow including bones and ligaments. The used multibody approach allowed to study, dynamically and kinematically, the biomechanical behaviour of an elbow under intact and deficient conditions at three different flexion angles (i.e. 30°, 60° and 90°). The presented elbow model allows simulating the application of a clinical manoeuvre, by deducing external applied forces from recorded trajectories through a PD controller algorithm. Especially, the multibody analysis focused on the PMCL and coronoid process roles in stabilizing the elbow joint. In this work, three different conditions were assessed: intact elbow condition, PMCL resected condition and PMCL resected plus coronoid resected condition.

Overall, results seem to confirm model effectiveness in predicting experimental outcomes. However, some differences emerged from the comparison between experimental and model results. These differences could be ascribed to some study limitations. Indeed, the model geometry is different from the geometry of the elbow specimen used during the considered experiments. In addition, it should be stated that manoeuvre applied experimentally among the different studied conditions is likely affected by operator dependent uncertainties, for instance, about the flexion angle extension of the elbow at the beginning of the manoeuvre and about the equality of the applied forces during the tests.

Moreover, the optimization process of the model showed that contact forces is highly influenced by the AMCL zero-load length variation. Conversely, it seems that the PMCL does not influence the flexion movement, except for flexion angles greater than 90°. As regarding the coronoid contribution in elbow stability, it seems to be more relevant at lower flexion angles by increasing the joint stability in all studied conditions. Future interesting developments could be the investigation of the deduced applied forces among simulated cases and between experimental and computed findings.
Bibliography


[12] Sarah Floris, MS, Bo S. Olsen, MD, Michel Dalstra, PhD, Jens 0. Slzrbierg, MD, and Otto Sneppen, MD, DMSc, Aarhus, Denmark, The medial collateral ligament of the elbow joint: Anatomy and kinematics.


[16] E.K.J. Chadwick*, Elbow and wrist joint contact forces during occupational pick and place activities