



TESIS DOCTORAL

SIMULACIÓN DEL COMPORTAMIENTO MECÁNICO DE UNA ARTICULACIÓN HUMANA MEDIANTE EL MÉTODO DE ELEMENTOS FINITOS.

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Que el trabajo titulado " SIMULACIÓN DEL COMPORTAMIENTO MECÁNICO DE UNA ARTICULACIÓN HUMANA MEDIANTE EL MÉTODO DE ELEMENTOS FINITOS" ha sido realizado por D. Emilio Lechosa Urquijo, bajo nuestra dirección; reuniendo, a nuestro juicio, los requisitos y méritos suficientes para que el autor de este pueda optar al Grado de Doctor por la Universidad de Salamanca.

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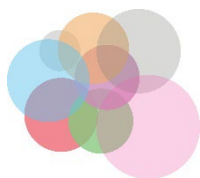
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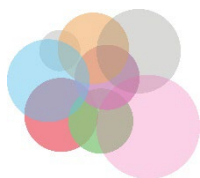
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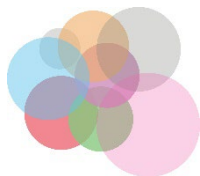


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DECLARACIÓN DE AUTORÍA

Declaro que he redactado el trabajo titulado: “SIMULACIÓN DEL COMPORTAMIENTO MECÁNICO DE UNA ARTICULACIÓN HUMANA MEDIANTE EL MÉTODO DE ELEMENTOS FINITOS” para la obtención del Grado de Doctor por la Universidad de Salamanca. Ha sido realizado, de forma autónoma, con la ayuda de mis directores, siguiendo las directrices del artículo 14.1 en su capítulo II del Reglamento de Doctorado de la Universidad de Salamanca, sobre la elaboración y defensa de la Tesis Doctoral.

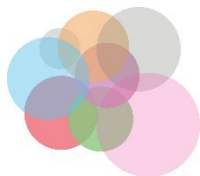
Los resultados de este trabajo de Tesis Doctoral han sido publicados en diferentes revistas de carácter internacional con elevado índice de impacto; además se han presentado algunos de los resultados en Congresos Internacionales.

En Salamanca, 23 de julio de dos mil veinticuatro.

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AGRADECIMIENTOS

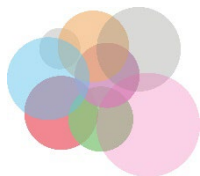
Quiero expresar mi más sincero agradecimiento a mis directores D. Juan Antonio Juanes Méndez, D. Fernando Blaya Haro y a D. Roberto D'Amato, por su aceptación en la dirección de esta Tesis Doctoral, seguimiento, y las facilidades prestadas en todo el transcurso de este trabajo. Sin su gran esfuerzo y acierto en la ejecución de este proyecto de Tesis Doctoral este trabajo no hubiera sido posible. También al profesor Juan David Cano quien me ha aconsejado, enseñado y ayudado durante todo el proceso.

Especial mención y agradecimiento al Departamento de Ingeniería Mecánica y Química y Diseño Industrial, de la Universidad Politécnica de Madrid, y a D. Fernando Blaya por brindarme todos los recursos y herramientas que han sido necesarias para llevar a cabo el proceso de investigación. No hubiese podido alcanzar estos resultados de no haber sido por su incondicional ayuda.

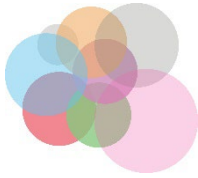
Quiero agradecer igualmente a todos aquellos que me han acompañado en el camino de elaboración de este trabajo, soportando ausencias y largas jornadas de encierro en el ordenador. En especial, quiero mencionar a mis padres y mi hermano, que siempre me apoyaron, de modo incondicional, en las decisiones que he tomado y que, con su trabajo, paciencia y esfuerzo, han hecho posible que supere los retos a los que me he enfrentado y me formase correctamente hasta este día de presentación de mi propia Tesis Doctoral.

A todos mis amigos, colegas y compañeros que, cada uno a su manera y dentro de sus posibilidades, me alentaron y ayudaron para que esta Tesis fuera terminada.

A todos, muchas gracias.



1. INTRODUCCIÓN



Es un hecho que los métodos de simulación han sido, y siguen siendo, herramientas indispensables en el campo de la investigación. Gracias a la resolución de las ecuaciones que rigen el comportamiento de un fenómeno se pueden llevar a cabo simulaciones que permiten predecir resultados con un menor coste, económico y en tiempo, que, si se tuvieran que realizar los estudios con elementos reales, prototipos o modelos a escala [1], [2], [3], [4], [5], [6], [7], [8], [9]

El método de los elementos finitos consiste en discretizar el dominio (lo que comúnmente se conoce como mallar la geometría) analizado mediante geometrías conocidas en las que el comportamiento está regido por las ecuaciones de estado. Estas geometrías son conocidas como *elementos*, y existen diferentes tipos de elementos en función de que geometría se está simulando:

Elementos planos, 2D, para simular geometrías de pequeño espesor, en las que la variación del parámetro estudiado a lo largo del espesor de este se considera irrelevante.

Elementos volumétricos, 3D, para representar geometrías con dimensión en los 3 ejes coordenados, con volumen. La variación del parámetro estudiado a lo largo del elemento considerado es tenida en cuenta.

Elementos lineales, 1D, para simular aquellas geometrías en las que una dimensión es mucho mayor que las otras dos.

Elementos especiales, 0D, para simular elementos característicos; masas concentradas, amortiguadores, etc....

La siguiente imagen, Figura 1, ilustra, de modo gráfico, diferentes geometrías malladas empleando estos elementos

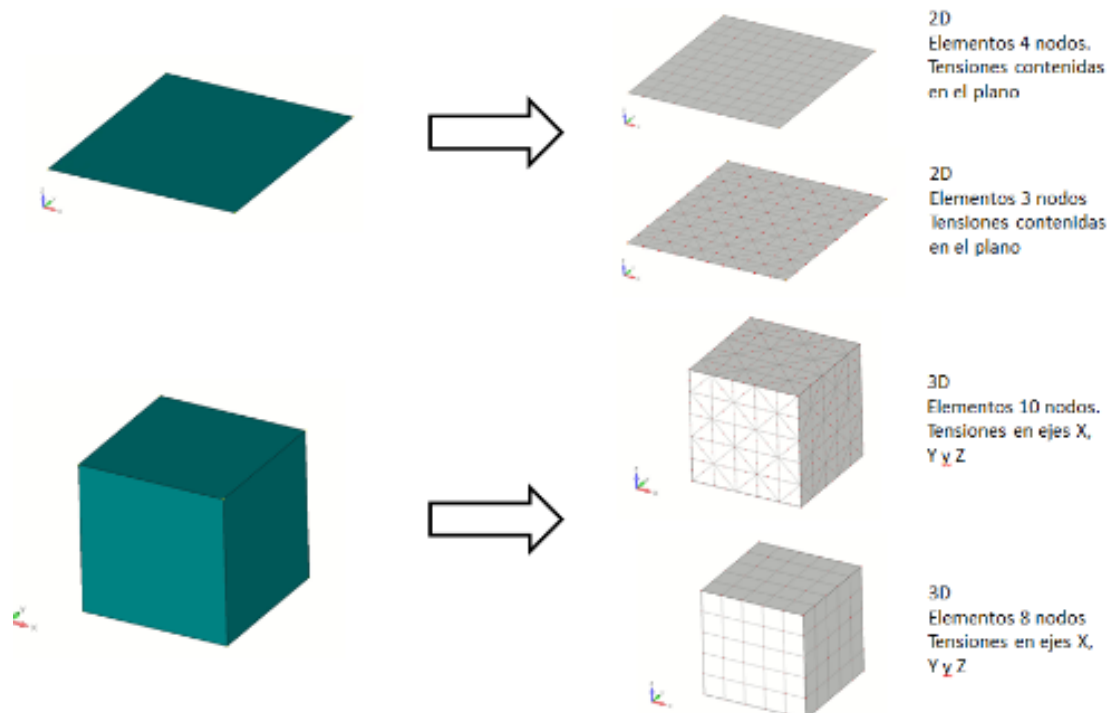
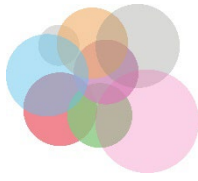


Figura 1. Tipología y descripción de los elementos finitos más comúnmente empleados en simulaciones mecánicas.



Dentro el campo de las simulaciones el método de los elementos finitos (FEA) en aplicaciones de tipo mecánico es, tal vez, de los más empleados y conocidos, con aplicación en numerosos, y variados, estudios; análisis estáticos estructurales, predicción de rotura por fatiga en materiales, análisis dinámicos y de deformaciones en condiciones de linealidad o no linealidad,... la lista sería interminable, pero cabe mencionar especialmente la aplicación de estas técnicas de simulación al campo de la biomecánica, y concretamente al campo de estudio de esta tesis, el estudio y análisis del tejido muscular y de los músculos.

La simulación de elementos anatómicos como; huesos, ligamentos, cartílagos, músculos.... mediante el método de elementos finitos (análisis FEM) se ha llevado a cabo en diversas ocasiones [1], [2], [10], [11], [12], [13], [14], [15].

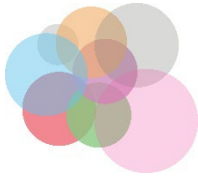
La finalidad de estas simulaciones es diversa. Una de las posibilidades que brinda un modelo FEM es la de poder prever el comportamiento del elemento considerado (músculo, hueso, etc...) en diversos estados de carga para así estimar las tensiones, [1], deformaciones y esfuerzos a los que se puede ver sometido, [10], estudiar la fatiga que se puede producir en el mismo, [9], [16] comprobar las deformaciones admisibles en un elemento antes de que se produzca la lesión en el mismo, en definitiva, predecir el comportamiento del elemento estudiado. De los elementos que intervienen en una articulación se pueden diferenciar tres tipos fundamentales, con la simplificación que se aplica en los modelos de elementos finitos a cada uno de ellos:

Huesos y tendones. Pueden aproximarse de un modo válido a elementos de rigidez elevada, en comparación al resto de elementos que intervienen en la articulación, por lo que el empleo de elementos 3D para su mallado es aceptado [1], [7], [9], [17], [18], [19], [20]. La caracterización del tejido esta aceptada como un material con un módulo de elasticidad constante.

Ligamentos. Su comportamiento puede asimilarse al de un muelle, con una constante elástica definida, asumiendo que únicamente trabajarían a tracción,[7].

Músculos. El elemento más complejo de simular dentro de la articulación, su comportamiento depende de numerosos factores y su módulo de elasticidad no es constante ya que son los elementos activos de la articulación. Su deformación no responde únicamente a la carga aplicada al elemento. El comportamiento de un músculo debe el desarrollo de la fuerza a dos elementos, lo que se considera elemento activo (fibras musculares) y elementos pasivos (tendones, fascia) [21]. Las fuerzas contráctiles en los músculos se desarrollan en las fibras musculares, sin embargo, la fuerza total que un músculo puede desarrollar cuenta con la colaboración de los elementos pasivos.

Los estudios realizados hasta ahora han empleado elementos 3D para el mallado de los músculos de una articulación, [17], [19], [22], siendo esta una buena aproximación, presenta un inconveniente importante para aprovechar los resultados. Al analizar una articulación en una posición diferente de la estudiada es necesario volver a mallar los músculos de nuevo, puesto que su geometría habrá cambiado. Esto impide elaborar una



metodología para el análisis de una articulación en diferentes posiciones ya que habría que repetir todo el trabajo realizado para el resto de las posiciones.

El comportamiento biomecánico de la musculatura humana se ha estudiado en varias ocasiones, en términos de la morfología del músculo, la excitación recibida, la velocidad del movimiento, etc... [1], [2], [4], [8], [9], [10], [12], [15], [16]. Sin embargo, ninguno ha abordado la tarea de integrar los resultados obtenidos en una herramienta, de uso común, que permita el estudio completo de una articulación como los análisis FEM.

El problema que presenta un estudio FEM de estas características es la obtención de una metodología que pueda reutilizarse de un modo sencillo. Al ser los músculos elementos con geometría variable, durante el movimiento de la articulación, se hace necesario rehacer el trabajo completo de análisis, y de generación de malla, para cada posición que desee estudiarse.

Debido a todo lo anterior, este trabajo de investigación se centrará en la obtención de un modelo matemático, basado en el modelo de Hill [23], [24] de simulación de la fuerza producida en un músculo mediante elementos 1D (de dos nodos) que permita simular la articulación en diferentes posiciones.

La articulación elegida para este estudio es la del codo, debido a que:

Presenta un único grado de libertad [25].

Los músculos del codo, fusiformes y con inserciones en dos puntos, pueden ser simplificados por un elemento 1D definido a lo largo de su línea de acción [11], Figura 1.

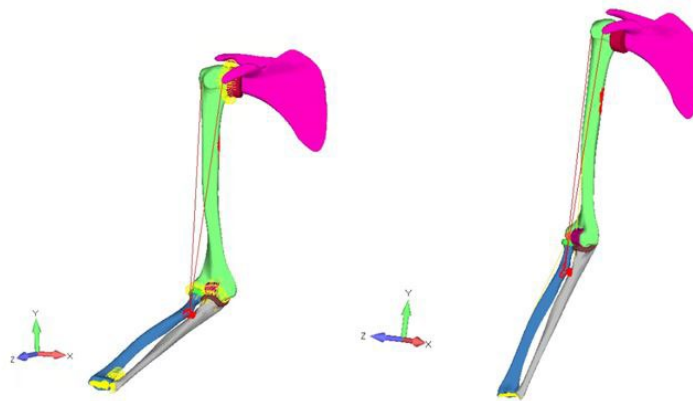
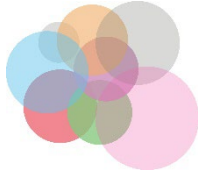


Figura 2. Elemento 1D, simplificación de músculos en la articulación en dos posiciones de ésta.

Al simplificar el músculo a un elemento de 2 nodos el problema de reutilizar el modelo FEM en diferentes estudios desaparece, ya que el músculo quedaría siempre definido por esos dos nodos y, a pesar de que estos se desplazasen, el músculo seguiría unido en los mismos puntos y se alinearía con su nueva posición de modo automático [11].

El objetivo de este estudio es de poder caracterizar un músculo de modo que la malla generada en la articulación pueda ser reutilizada en diferentes posiciones sin repetir el proceso de mallado. Esto permite abordar la investigación propuesta con un menor esfuerzo, en términos de tiempo computacional o recursos personales y económicos.



1.1. Compendio de publicaciones realizadas

Como resultado de este estudio, se han realizado las siguientes publicaciones en **Revistas Internacionales Indexada en el JCR** que pueden ser encontradas en los Anexos:

Revista: Journal of medical systems, Journal Rank: ICR - Q1 (Instruments & Instrumentation) / CiteScore - Q1 (Instrumentation):

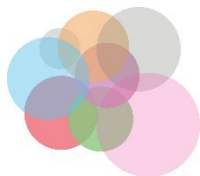
- Islan, M., Blaya, F., Pedro, P. S., D'Amato, R., Urquijo, E. L., & Juanes, J. A. (2018). Analysis and FEM Simulation Methodology of Dynamic Behavior of Human Rotator Cuff in Repetitive Routines: Musician Case Study. *Journal of Medical Systems*, 42(3), 55. <https://doi.org/10.1007/s10916-018-0908-7>
- Islán Marcos, M., Lechosa Urquijo, E., Blaya Haro, F., D'Amato, R., Soriano Heras, E., & Juanes, J. A. (2019). Behavior under Load of A Human Shoulder: Finite Element Simulation and Analysis. *Journal of Medical Systems*, 43(5), 132. <https://doi.org/10.1007/s10916-019-1248-y>

Revista: Mathematics, Journal Rank: ICR - Q1 (Mathematics) / CiteScore - Q1 (Mathematics):

- Lechosa Urquijo, E., Blaya Haro, F., David Cano-Moreno, J., D'Amato, R., & Antonio Juanes Méndez, J. (2022b). Mechanical Model and FEM Simulations for Efforts on Biceps and Triceps Muscles under Vertical Load: Mathematical Formulation of Results. *Mathematics* 2022, Vol. 10, Page 2441, 10(14), 2441. <https://doi.org/10.3390/MATH10142441>

Por último, se han participado en los siguientes **Congresos Internacionales**:

- Marcos, M. I., D'Amato, R., Urquijo, E. L., Juanes, J. A., Haro, F. B., & Heras, E. S. (2018). Finite Element Simulation and Analysis of the behavior under load of a human shoulder. *ACM International Conference Proceeding Series*, 454–461. <https://doi.org/10.1145/3284179.3284256>
- Lechosa, E., D'Amato, R., Haro, F. B., & Juanes, J. A. (2019). 1D Finite Element model using rod element applied to muscle behavior simulation. *ACM International Conference Proceeding Series*, 411–415. <https://doi.org/10.1145/3362789.3362868>.
- Lechosa Urquijo, E., Blaya Haro, F., D'Amato, R., & Juanes Méndez, J. A. (2020). Finite Element model of an elbow under load, muscle effort analysis when modeled using 1D rod element. *ACM International Conference Proceeding Series*, 8(20), 475–482. <https://doi.org/10.1145/3434780.3436612>.



2. PLANTEAMIENTO DEL TRABAJO



Se han realizado, con anterioridad, simulaciones para predecir el comportamiento del tejido muscular en articulaciones humanas. La metodología seguida suele ser considerar el músculo como un elemento tridimensional y mallarlo de acuerdo a su volumen para realizar la simulación mediante herramientas de análisis por elementos finitos [2], [12], [14], [16], [19], [20], [21]

Estas mallas están vinculadas a una geometría del músculo y habitualmente emplean elementos 3D tetraédricos de segundo orden, [9] o bien hexaedros [26].

La finalidad de estas simulaciones es diversa. En algunos estudios se busca entender el comportamiento del músculo [27], comprobando la distribución de tensiones en el interior de éste y verificar los efectos mecánicos generados en las estructuras tridimensionales de las unidades músculo-tendón implicadas.

Otros trabajos han modelado tejido con elementos 3D para comprobar las deformaciones que se producen en un seno femenino durante los procesos de imagen de mama [22], o bien las que se producen en el tejido muscular, así como las tensiones en los elementos involucrados en el proceso de la marcha humana, especialmente en el tendón de Aquiles, [8].

Esas simulaciones persiguen, fundamentalmente, explicar el estado de carga del músculo en la situación concreta analizada. Fuera de esa situación de carga analizada no se obtienen resultados.

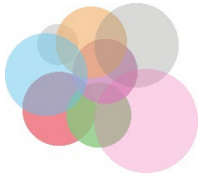
Estos estudios presentan la limitación de que el músculo cambia de geometría durante el recorrido de la articulación, por lo que, en caso de querer analizar la articulación, y por lo tanto el músculo, en diferentes configuraciones de ésta se hace necesario repetir el proceso de análisis tantas veces como posiciones quieran analizarse.

Esta limitación se hace más evidente si se pretende ejecutar un análisis de tipo dinámico, es decir, a lo largo de todo el recorrido de la articulación de modo continuo y no en posiciones discretas, ante la imposibilidad de emplear la misma malla para el FEM de la articulación completa. Para solventar esta limitación sería necesario caracterizar el tejido muscular, o un músculo concreto, de modo que pueda incluirse en el modelo de análisis FEA.

El desarrollo del trabajo en esta investigación parte de la creación de un modelo CAD de la articulación del codo a partir de imágenes obtenidas mediante técnicas de diagnóstico por imagen por resonancia magnética. Mas detalles del proceso se encuentran en el apartado 4, Material y métodos de esta memoria.

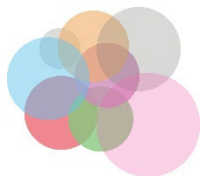
El análisis realizado en el presente estudio, parte de la simplificación de los músculos en una articulación por un elemento 1D que permita ser reutilizado en diferentes posiciones sin necesidad de modificar la malla empleada en el modelo de elementos finitos completo de la articulación.

Con los diferentes modelos FEM de la articulación (para diferentes ángulos de apertura de la articulación) se llevarán a cabo análisis en situación de carga, que permitan obtener las fuerzas necesarias en los músculos de la articulación del codo (tanto en flexión como en extensión).

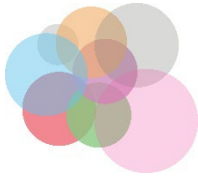


Una vez obtenidas las fuerzas necesarias en cada músculo de la articulación, se pueden extrapolar los resultados para diferentes estados de carga.

Por otro lado, en base a la literatura existente, se puede conocer la fuerza máxima que un músculo de la articulación puede desarrollar por lo que, comparando la fuerza obtenida mediante los modelos FEM y las fuerzas máximas teóricas de cada músculo se puede valorar si el músculo se encuentra en situación de sobre esfuerzo o está por debajo del límite de carga.



3. OBJETIVO Y ALCANCES



3.1. Objetivos principal y alcances

El objetivo principal de la tesis es obtener la siguiente relación, que pueda emplearse en un análisis mediante elementos finitos, del comportamiento de un músculo.

$$F_{musc} = F_{musc}(L, A, V, f, T)$$

Dónde:

- F_{musc} representa la fuerza desarrollada por el músculo
- L representa la Longitud del músculo respecto a su longitud en reposo
- A representa la Sección transversal del músculo
- T representa el tipo de músculo

El alcance del estudio se restringe a las siguientes condiciones.

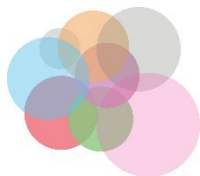
- Estará basado en el modelo de Hill.
- Se centrará en el estudio de los músculos involucrados en la articulación del codo.
- El tipo de contracción muscular analizado será isométrica, sin desplazamiento en la articulación. Se analizarán únicamente casos estáticos.
- La validación del modelo matemático se realizará comparando los resultados obtenidos al analizar la articulación, con los resultados en un ensayo con participantes.

Una vez definido el modelo matemático teórico que explique el comportamiento de un músculo en determinadas situaciones de carga, este podrá ser extrapolado a diferentes tipos de músculos mediante la modificación del modelo matemático de acuerdo con las variaciones fisiológicas entre diferentes tipos de músculos o estados de carga.

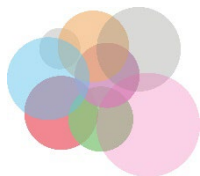
3.2. Objetivos secundarios

Como parte del proceso, los objetivos secundarios que se irán contemplando durante el proceso y desarrollo del estudio son:

1. Obtención de la evolución de la fuerza en los músculos del codo, en flexión y extensión, para diferentes ángulos de esta.
2. Caracterización de la fuerza en los músculos de la articulación del codo en diferentes estados de carga
3. Obtención de un modelo de elementos que pueda emplearse en diferentes posiciones sin necesidad de tener que modificar la malla.
4. Caracterización de la fuerza desarrollada por los músculos de la articulación del codo en un grupo poblacional diverso, en base a parámetros biométricos de los participantes (altura, peso, dimensiones del brazo...)



4. MATERIALES Y MÉTODO



Los materiales y la metodología utilizada para alcanzar los objetivos planteados se muestran detalladamente en los anexos adjuntos a esta Memoria. A continuación, se proporcionará un resumen de la metodología y herramienta utilizada para lograr los resultados esperados.

En el artículo “Analysis and FEM Simulation Methodology of Dynamic Behavior of Human Rotator Cuff in Repetitive Routines Musician Case Study” (Journal of Medical Systems, 2018) se obtiene un modelo CAD completo de la articulación glenohumeral a partir de imágenes generadas mediante tomografía 3D. A partir de éste se genera un modelo de elementos finitos de la articulación y se lleva a cabo el análisis de esta en un caso de carga asociado a la práctica del violín.

En esta investigación se calcula una predicción del número de ciclos que puede soportar la articulación durante la práctica musical sin que se produzca daño en ninguno de sus elementos. La determinación de este número de ciclos se lleva a cabo comparando los valores de tensión en el análisis de elementos finitos con los valores publicados de tensión admisible a fatiga de los tejidos.

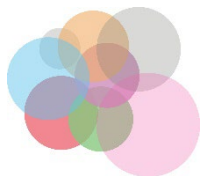
Tras esto, en los artículos “Finite Element Simulation and Analysis of the behavior under load of a human shoulder” (Technological Ecosystems for Enhancing Multiculturality, TEEM, Salamanca 2018) y “Behavior under Load of A Human Shoulder: Finite Element Simulation and Analysis (JOURNAL OF MEDICAL SYSTEMS 2019) se realizan modificaciones en el estudio de la misma articulación, glenohumeral, incluyendo en el estudio el comportamiento de cartílagos y ligamentos de la articulación.

En esta ocasión los casos de carga considerados tienen como finalidad comprobar los resultados globales del modelo, distribución de tensiones en los tejidos, comportamiento de los cartílagos y transferencia de esfuerzos en los contactos entre ellos. Los resultados se consideran adecuados en lo que a distribución de tensiones se refiere, dándose por válido el modelo FEM considerado en lo que a proceso de mallado y aplicación de condiciones de contorno se refiere.

En el artículo “1D Finite Element model using rod element applied to muscle behavior simulation” (Technological Ecosystems for Enhancing Multiculturality, TEEM, León 2019) se plantea por primera vez el modelo de un elemento 1D para simular los músculos, de acuerdo con el modelo de Hill.

La caracterización de músculo se lleva a cabo mediante un elemento tipo CROD, caracterizado por presentar esfuerzos únicamente en la dirección axial del elemento. Al que se asigna un material con un módulo de elasticidad no lineal.

Una vez comprobado que el modelado de los músculos mediante elementos 1D responde correctamente a las necesidades del estudio se desarrolla el primer modelo de elementos finitos completo de la articulación, en esta ocasión del codo. Dado que los músculos de la articulación glenohumeral son planos y no fusiformes se considera que la simulación de los músculos del codo, grupo bíceps y tríceps es más sencilla.



Aislar los músculos responsables de la flexión y la extensión del codo es más sencillo y por lo tanto permitirán una validación del modelo matemático más fiable que si se mantuviese la articulación glenohumeral.

En el artículo “Finite Element model of an elbow under load, muscle effort analysis when modeled using 1D rod element” (Technological Ecosystems for Enhancing Multiculturality, TEEM, 2020) se desarrolla y calcula el modelo de elementos finitos de la articulación del codo en dos situaciones de carga:

- 150 N aplicados en la muñeca en sentido vertical ascendente para obtener el comportamiento de los músculos del tríceps.
- 150 N en sentido vertical descendente para obtener la respuesta de los músculos del bíceps.

Los casos de carga considerados responden a contracciones isométricas en ambos casos. A la hora de modelar el tejido muscular se comprueba como el módulo de elasticidad del músculo no es solamente No lineal, también se detecta que la fuerza desarrollada por un músculo no solamente depende de las características del tejido, o de la posición de la articulación, sino que la carga aplicada también influye a la hora de evaluar la fuerza desarrollada en un músculo.

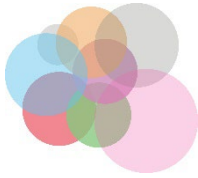
Es fácil entender esto si se imagina la articulación del codo en una posición dada, 60 grados, por ejemplo, y se aplican diferentes cargas en la muñeca. Se comprobará como, en la medida que los músculos involucrados en el ejercicio sean capaces, no se producirá desplazamiento alguno en la articulación, por lo tanto, sin haber cambiado las características del tejido, ni la geometría de la articulación, la fuerza desarrollada por los músculos es diferente en ambos casos.

La obtención de las ecuaciones de comportamiento de los músculos, las que permiten calcular la fuerza en cada uno de ellos, se han de obtener de un modo indirecto. Partiendo de diferentes análisis FEM para la articulación en diferentes posiciones se puede obtener la fuerza que cada músculo ha de desarrollar en esa condición particular (ángulo y carga).

Posteriormente con todos los cálculos en diferentes posiciones se obtiene la ecuación que indica la fuerza que un músculo ha de desarrollar en una posición determinada con una carga aplicada dada.

En el artículo “Mechanical Model and FEM Simulations for Efforts on Biceps and Triceps Muscles under Vertical Load: Mathematical Formulation of Results” (Mathematics, 2022, <https://doi.org/10.3390/math10142441>) se desarrolla el cálculo de las ecuaciones que asocian la fuerza en cada músculo de la articulación del codo respecto al ángulo que ésta forma y la carga aplicada en la muñeca.

Estas ecuaciones se han obtenido mediante un modelo FEM en el que los músculos han sido modelados empleando elementos 1D CROD, dada la validez de esta modelización estudiada en las publicaciones anteriores.



En este mismo modelo se ha simulado la inserción de los tendones mediante un elemento de reparto, dado que el músculo se ha modelado como un elemento de dos nodos, insertar el músculo simplificado en un solo punto concentraría los esfuerzos en éste y alteraría los resultados notablemente.

Por ello se ha empleado un elemento rígido para el reparto del esfuerzo en ese nodo en un grupo de nodos que simulan el área ocupada por la inserción de los tendones. Igualmente se ha llevado a cabo un estudio comparativo entre dos metodologías diferentes para simular el área de contacto en los cartílagos.

En un modelo de elementos finitos se ha simulado éste mediante un contacto Hertziano (con un tiempo de cálculo de 67 horas aproximadamente) en el otro se ha reemplazado este contacto por una unión rígida en la zona de contacto de los cartílagos mediante elementos rígidos RBE2 (con un tiempo de cálculo de 20 minutos).

Crear un nuevo tipo de elementos finitos que responda a estas ecuaciones permitiría generar modelos FEM de la articulación del codo válidos para predecir las fuerzas desarrolladas para cada músculo.

En las secciones siguientes se expondrán los métodos y técnicas empleados en la generación de los modelos utilizados en la investigación, así como los datos que han sido recogidos en las publicaciones y artículos que generan esta Tesis Doctoral.

4.1. HERRAMIENTAS EMPLEADAS EN EL ESTUDIO

El estudio se plantea en diferentes etapas, cada una de las cuales requiere de diferentes métodos de trabajo y herramientas específicas.

4.1.1. HERRAMIENTAS PARA LA GENERACIÓN Y SOLUCIÓN DEL MODELO DE ELEMENTOS FINITOS

4.1.1.1. *Imágenes 3D anatómicas*

El trabajo parte de la obtención de imágenes 3D de los elementos de la articulación, huesos, músculos, cartílagos, etc.... Estas imágenes obtenidas mediante técnicas de imagen, se utilizó como técnica de imagen la Resonancia Magnética, utilizando un equipo de 1.5 Teslas (marca Signa General & Electric).

Mediante estas técnicas se pueden obtener una nube de puntos que representa la geometría del cuerpo. Estas nubes de puntos están compuestas por puntos ubicados en la superficie exterior del cuerpo, identificados por sus coordenadas referidas a un origen de coordenadas. Puesto que los puntos están referidos en el espacio a un origen de coordenadas, pueden representarse en un software adecuado y tener una representación tridimensional del cuerpo considerado.

En la Figura 4 se puede observar la nube de puntos obtenida de la escapula, que forma parte de una imagen de un esqueleto completo, Figura 3.

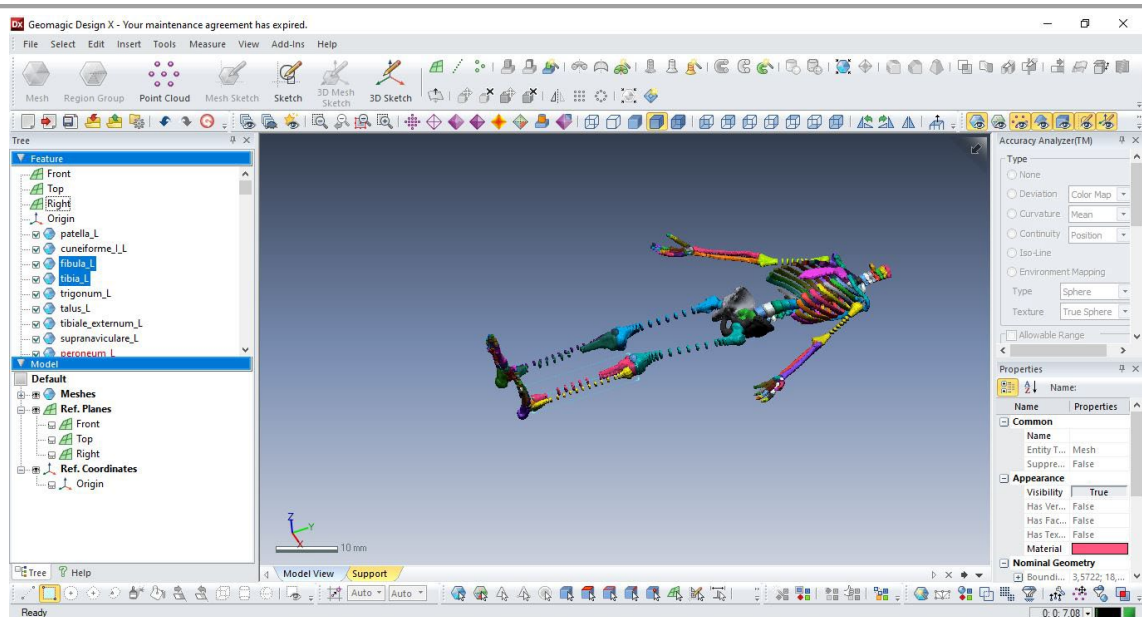
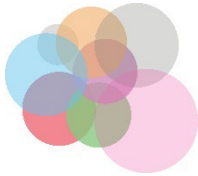


Figura 3. Esqueleto completo. Geomagic Design X.

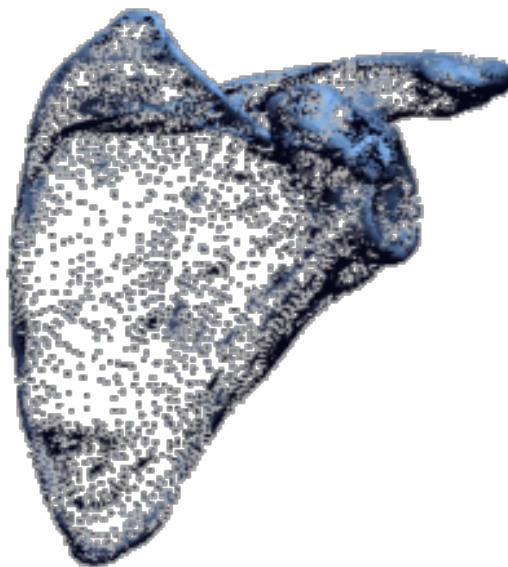


Figura 4. Nube de puntos asociada a la escapula.

4.1.1.2. Herramienta Software para la reconstrucción de los cuerpos

Dado que las imágenes se componen únicamente de puntos en el espacio, no se tiene información sobre el volumen contenido por el cuerpo, su masa, etc.... Es necesario para ello emplear herramientas software de reconstrucción de nubes de puntos, se emplea el software **Geomagic Design X (V 5.1.0.0)** (Figura 5), que, a partir de la nube de puntos, es capaz de generar las superficies exteriores del cuerpo.

En la Figura 6 se observa el resultado de la regeneración de una nube de puntos para obtener un sólido asociado a la escapula.

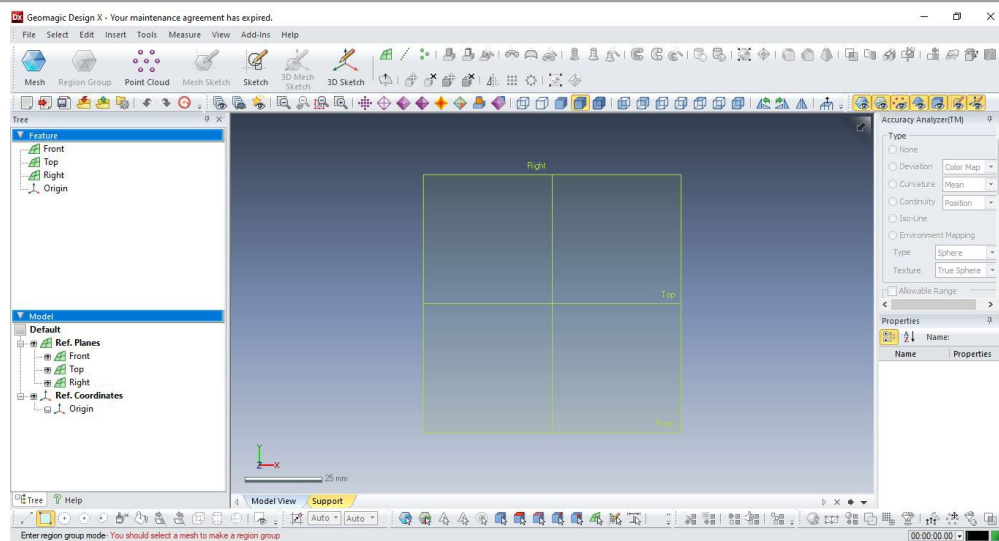
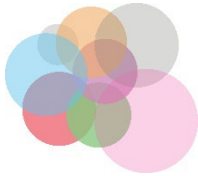


Figura 5. Entorno del software Geomagic Design X.



Figura 6. Reconstrucción de la geometría, mediante el software Geomagic Design X, a partir de la nube de puntos asociada a la escapula.

No todas las herramientas lo incluyen, pero Geomagic Design X incluye una potente herramienta que convierte un conjunto de superficies que formen un volumen cerrado en un sólido. La diferencia, a nivel software, entre un volumen y un sólido es que el volumen no tiene masa, la superficie representada en este tipo de piezas está “vacía” en su interior. Al no tener material interior no es posible llevar a cabo sobre ella las modificaciones necesarias para poder generar con ella un modelo válido de elementos finitos. Por el contrario, las piezas sólidas sí tienen material interior y permiten realizar sobre ellas las operaciones necesarias para adaptarlas a un modelo de elementos finitos válido.



4.1.1.3. Herramienta Software para la generación de sólidos

A pesar de que la herramienta Geomagic Design X, esta permite obtener sólidos, a priori, válidos para generar un modelo de elementos finitos. Puede ser necesario realizar sobre estos cuerpos determinadas operaciones encaminadas a asegurar que el FEM dé una solución válida, estas se han de realizar un software específico, denominados CAD (Computer Aided Design). En el caso de este estudio se empleará el software **NX de Siemens, en su versión 10** (Figura 7).

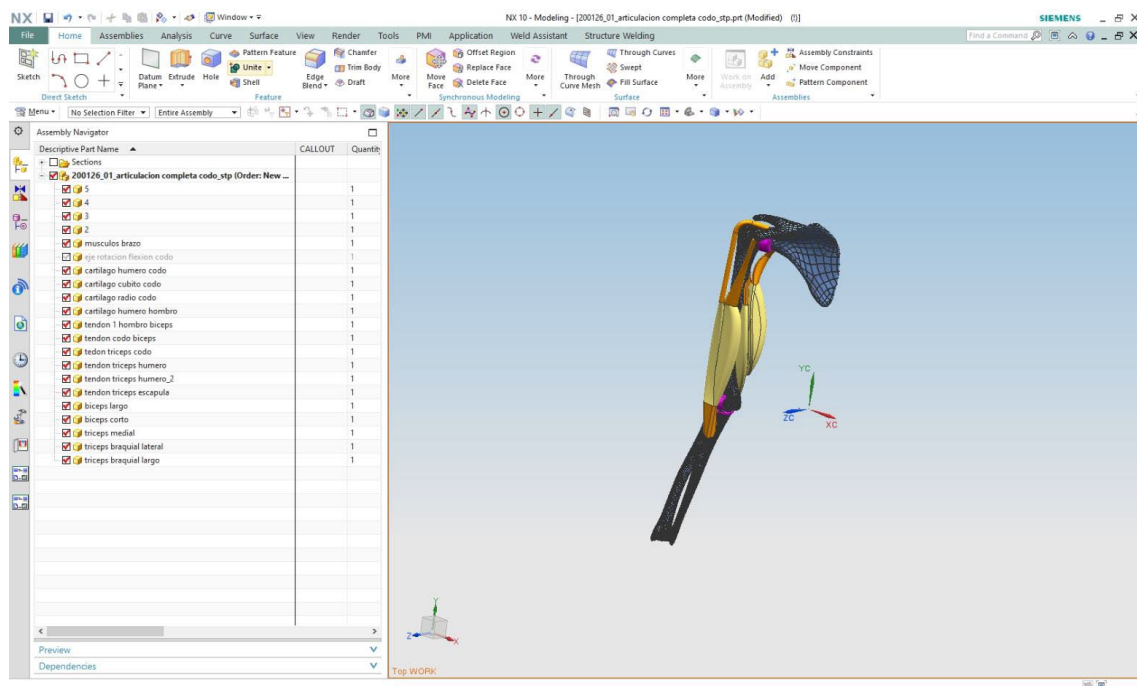


Figura 7. Modelo CAD, articulación del codo, Siemens NX.

El motivo de tener que trabajar estos cuerpos antes de procesar el modelo de elementos finitos es que aspectos como los indicados a continuación son fuentes de problemas de cara a la obtención de resultados validos en el análisis.

- Superficies de tamaño infinitesimal. Las herramientas CAD tienen una precisión matemática de varios órdenes de magnitud, pudiendo medir distancias del orden de $1e-6$ mm (micrómetros). A nivel matemático estas superficies de tamaño infinitesimal no suponen un problema para la herramienta CAD, sin embargo, son focos de numerosos problemas a la hora de generar una malla válida para elementos finitos. La simplificación y limpieza del modelo CAD generado por Geomagic Design X es una de las tareas que han de realizarse.
- Piezas con auto-interferencias. Definimos una interferencia entre dos piezas a aquella situación en la que dos cuerpos comparten la misma posición (total o parcialmente) en el espacio. Es como si parte de uno de ellos estuviera dentro del otro. Si esta situación se produce en un mismo cuerpo, se produce una auto-interferencia, hará



imposible generar una malla con dicho sólido. Se han de localizar y corregir estas posibles situaciones antes de comenzar el proceso de mallado, Figura 8.

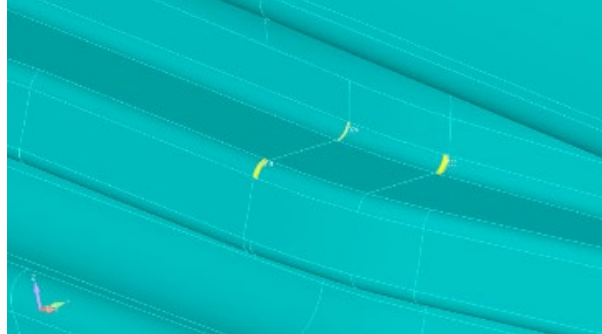


Figura 8. Superficie de tamaño reducido (en amarillo).

4.1.1.4. *Herramienta Software para la generación del modelo de elementos finitos*

Una vez obtenidos, y preparados convenientemente, los sólidos que forman el modelo de análisis, se puede generar el modelo de elementos finitos, para su posterior análisis. Las herramientas de análisis por elementos finitos requieren de dos herramientas diferentes;

Preprocesador. Herramienta SW en la que se realiza el proceso de mallado. Para el mismo es necesario definir diferentes parámetros;

- Tamaño del elemento. Cuanto menor sea el elemento definido (lado del cuadrado empleado en el caso de elementos 2D, o lado del cubo empleado en el caso de elementos 3D, por ejemplo) más precisos serán los resultados, pero más tiempo de cálculo será necesario. Es importante alcanzar un equilibrio entre el tamaño de elemento empleado y precisión requerida. Es posible emplear diferentes tamaños de elemento en diferentes zonas del cuerpo, Figura 9.

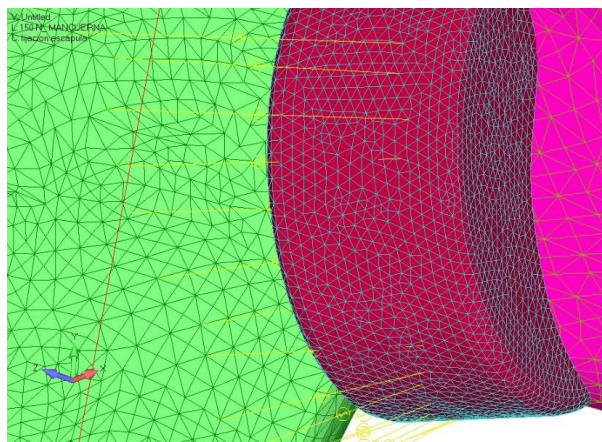
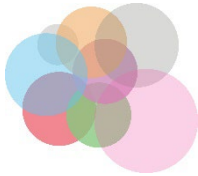


Figura 9. Piezas anexas, diferente tamaño de elemento en la malla.

- Características del material empleado. Una vez creados los elementos, la malla, del cuerpo a analizar, es necesario indicar al SW de que material esta hecho ese cuerpo.



Propiedades mecánicas como el módulo de elasticidad, el módulo de Poisson o la densidad del material, son necesarias para poder solucionar el problema, Figura 10.

- Cargas aplicadas. Ya sean estas fuerzas, temperaturas, presiones, etc... se han de definir las mismas en el modelo del modo más aproximado a como se van a producir en la realidad. Para definir las es posible, en la mayoría de las herramientas comerciales para análisis, apoyarse en la geometría del cuerpo (superficies, aristas, etc...) así como en los elementos de la propia malla (elementos y/o nodos), Figura 11.

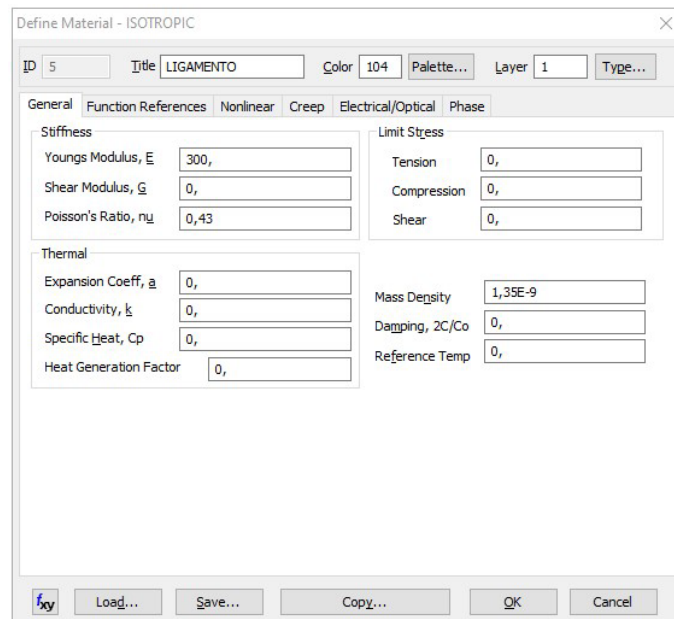


Figura 10. Definición de las características del material.

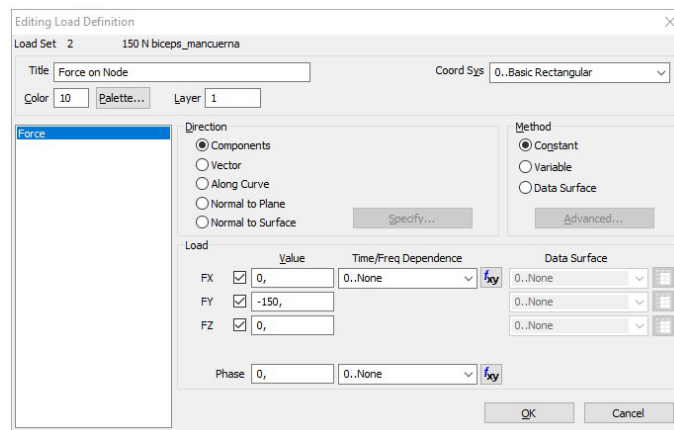
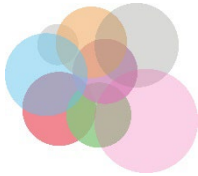


Figura 11. Definición de las cargas aplicadas.

- Restricciones. Se definen así las limitaciones que se aplican al modelo de cara a realizar el cálculo; zonas fijas, zonas deslizantes, piezas en contacto, etc... Igual que en el caso de la aplicación de cargas en el modelo estas restricciones pueden aplicarse en la geometría del cuerpo mallado o bien en los elementos de la propia malla, Figura 12.



- Tipo de análisis a realizar. Las posibilidades a considerar son muchas, cálculos estructurales lineales, no lineales, con contactos, etc..., Figura 13.

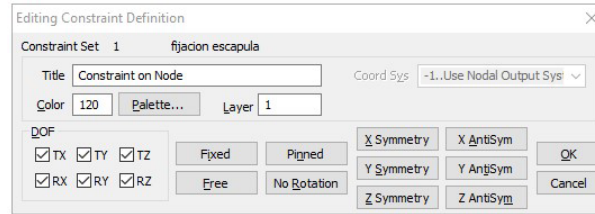


Figura 12. Definición de las restricciones en el modelo.

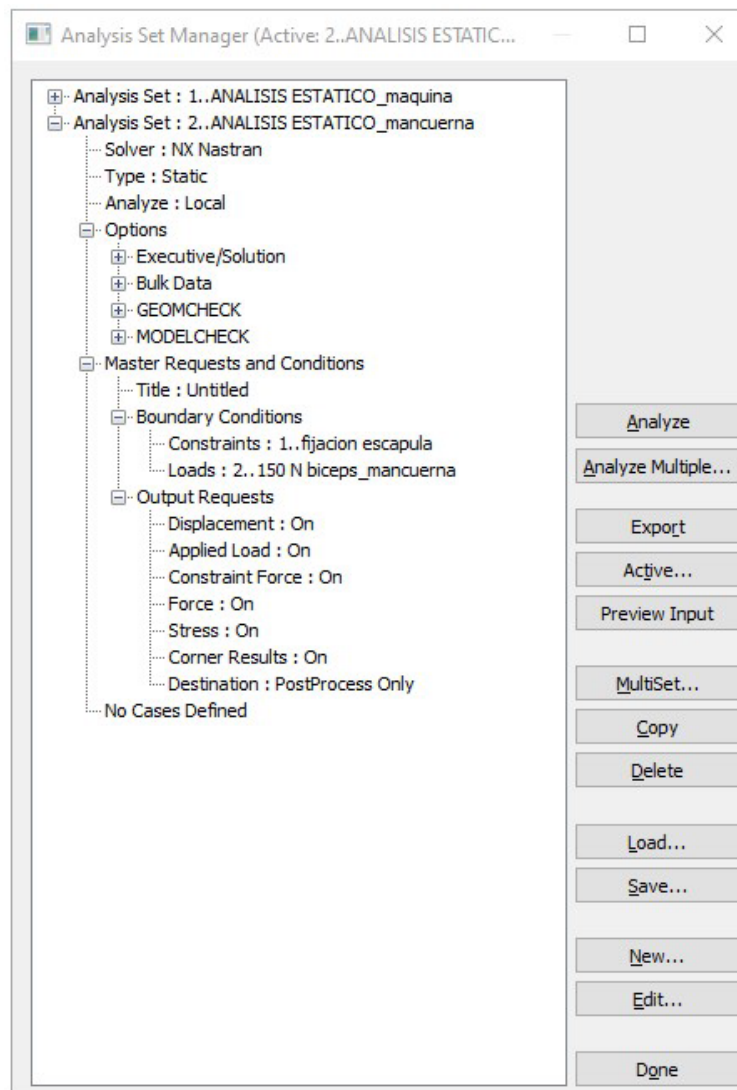


Figura 13. Definición de un caso de carga.

El resultado de las tareas de generación del modelo de elementos finitos es, como se observa en la Figura 14, una simulación de las piezas involucradas en el modelo a analizar, con las cargas aplicadas.

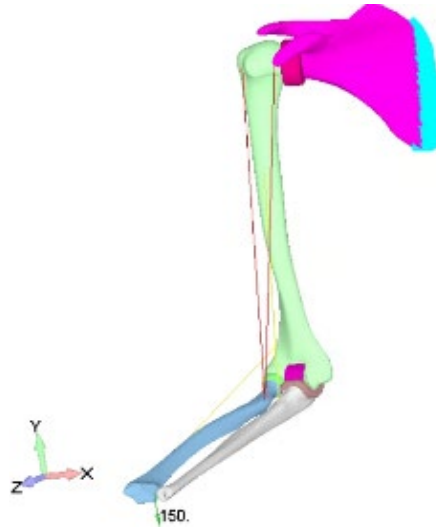
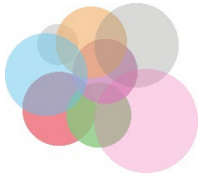


Figura 14. Modelo de elementos finitos completo.

Post procesador. Herramienta empleada para representar (en pantalla, en listados de hoja de cálculo, ...) los resultados obtenidos en el cálculo. Fundamentalmente se obtienen los resultados indicados a continuación.

- Tensiones. Expresadas como fuerza / superficie.
- Desplazamientos. En los nodos, lo que indica la deformación que presenta el cuerpo bajo el estado de cargas considerado.
- Esfuerzos. Correspondiente al valor de fuerza que aparece en cualquier zona del cuerpo.
- Strain o deformación unitaria, lo que da información valiosa acerca de la calidad de malla empleada.

La visualización de los resultados con estas herramientas se presenta como se observa en la Figura 15

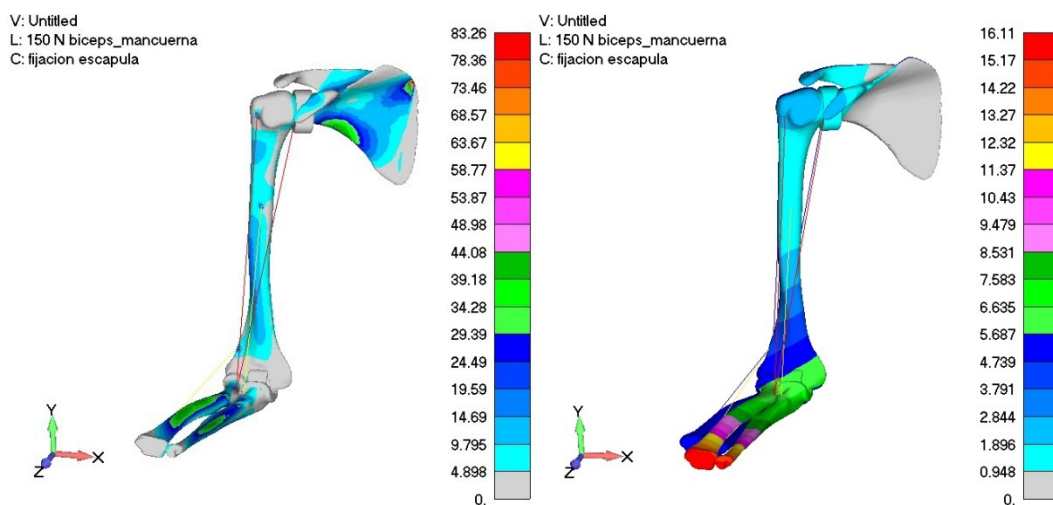
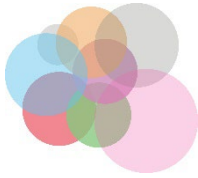


Figura 15. presentación de resultados, tensiones ó desplazamientos.



4.1.1.5. Estación de trabajo

Se denominan estación de trabajo al PC en el que se instalan y ejecutan los softwares indicados anteriormente. En nuestro caso la estación tiene las siguientes características.

- Procesador: Intel Core i9 – 10900 X @ 3.70 GHz
- Memoria: 128 GB
- Sistema operativo. Windows 10 Pro de 64 bits.
- Disco SSD de 1 TB de capacidad.

4.1.2. HERRAMIENTA PARA EL TRATAMIENTO MATEMATICO DE LOS RESULTADOS

Puesto que los resultados obtenidos en los modelos de elementos finitos han de procesarse para obtener ecuaciones de comportamiento, graficas, etc.... Se han de emplear herramientas adecuadas que permitan llevar a cabo estas tareas de modo eficiente.

En nuestro caso se emplean:

4.1.1.6. Software Microsoft Excel

Herramienta de cálculo de amplia difusión. A partir de los resultados obtenidos de los diferentes modelos de elementos finitos empleados, y exportándolos en forma de archivo de texto, es posible, en Excel, conseguir generar curvas de tendencia, estudios estadísticos y solución de diferentes ecuaciones de modo sencillo.

4.1.1.7. Software Matlab

Herramienta de programación con la que es posible solucionar aquellos problemas, más complejos, en los que la hoja de cálculo Microsoft Excel no resulta eficaz, Figura 16.

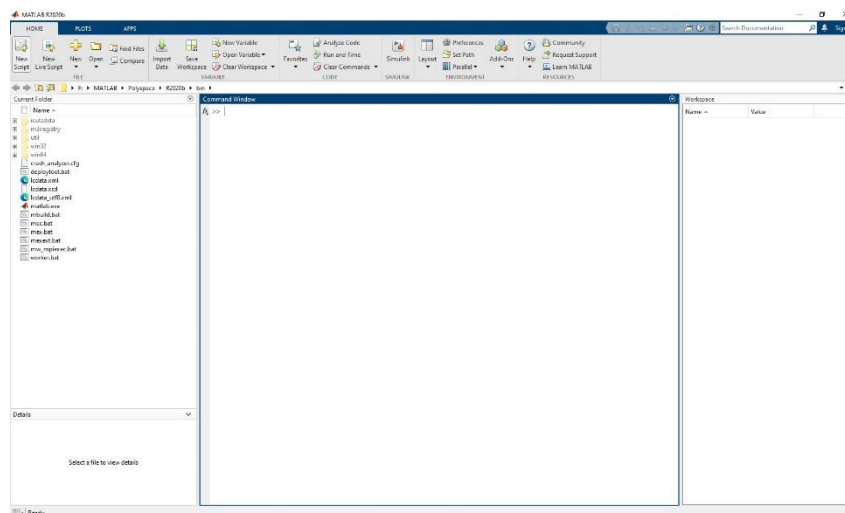


Figura 16. Entorno Matlab R2020.

En este caso se emplea fundamentalmente para obtener las ecuaciones, en dos variables, que definen la fuerza necesaria en cada músculo en función del ángulo de la articulación y la carga aplicada en la muñeca, así como su representación gráfica. El resultado de estos ajustes son superficies en dos variables, como se ven en la Figura 17.

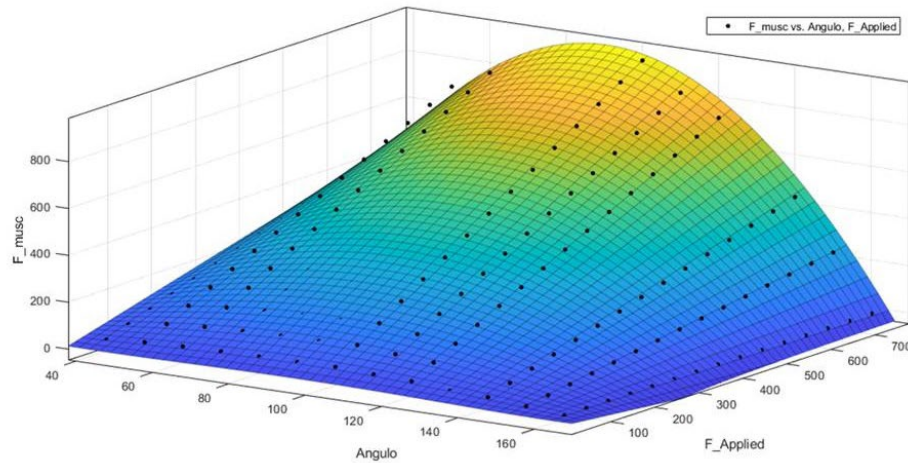


Figura 17. Representación de la fuerza en un músculo, en función de la carga en la muñeca y el ángulo de la articulación.

4.1.3. HERRAMIENTAS Y METODOLOGIA PARA LA EJECUCION DE ENSAYO DE VALIDACION CON PARTICIPANTES

Este apartado ha sido enviado para su publicación a la revista Mathematics, Journal Rank: JCR - Q1 (Mathematics) / CiteScore - Q1 (Mathematics):

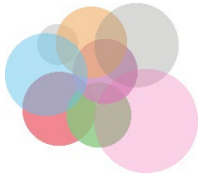
4.1.3.1. Metodología del ensayo

La validación de los resultados teóricos, obtenidos mediante los análisis de modelos de elementos finitos realizados, se lleva a cabo mediante un ensayo con participantes en el que se obtendrán los esfuerzos en los músculos de la articulación del codo para cada uno de ellos en función de la carga aplicada en la muñeca.

El ejercicio realizado con los participantes consiste en medir la fuerza que son capaces de aplicar en la muñeca en ejercicio de carga en flexión o extensión del codo.

La realización del ejercicio, en bíceps o tríceps, se realizará de acuerdo con las siguientes indicaciones.

- El participante mantendrá la espalda en contacto con la pared de modo que no ejerza fuerza curvando la misma.
- El participante aplicará fuerza a través de la eslinga, que se medirá en una célula de carga, en sentido vertical ascendente o descendente, la configuración del ensayo será la adecuada para un ejercicio de carga en bíceps o tríceps.
- La fuerza se aplicará en condición isométrica, no habrá desplazamiento en la articulación ni en el útil de ensayo.
- La articulación del participante se limitará en desplazamiento de modo que se mantenga la condición de carga isométrica, para ello se empleará una órtesis de codo, de modo que se pueda medir el ángulo formado por la articulación al mismo tiempo que se limite la apertura de esta.
- La duración del esfuerzo será lo menor posible para evitar fatigar los músculos.



- Cada participante realizara la prueba de carga (en bíceps o tríceps) en diferentes ángulos de apertura del codo.

Para ello se plantea un ejercicio como el del esquema de la Figura 18.

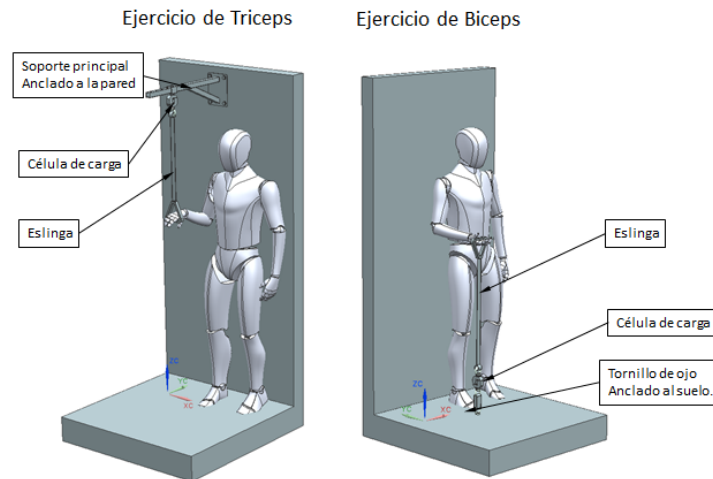


Figura 18. Esquema para la realización del ejercicio de validación del modelo teórico.

Como se observa en la

Figura 18 son necesarias dos configuraciones diferentes, una para el ejercicio de tríceps y otra para el ejercicio de bíceps, diferenciándose fundamentalmente en la posición de la célula de carga (anclada al suelo, o anclada al útil en la parte superior) y la estructura, soporte para el ensayo, necesaria para ello, no siendo necesaria esta para el ejercicio de bíceps, dado que la célula de carga esta fija al suelo, la Figura 19.

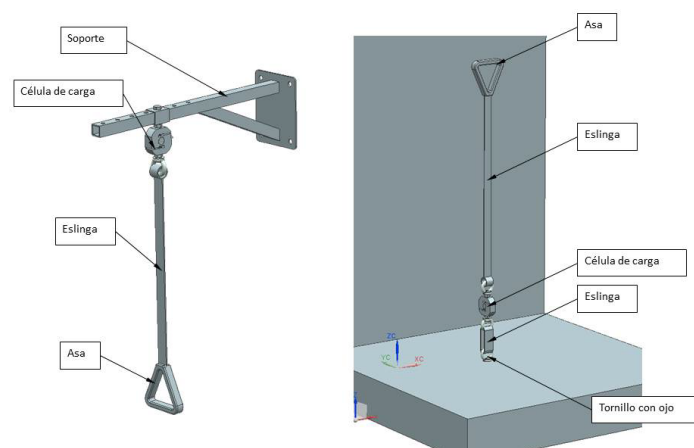
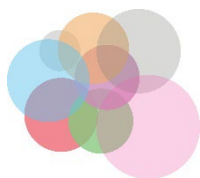


Figura 19. Configuración de útil de medida para ejercicio de tríceps (izquierda) o bíceps (derecha).



4.1.3.2. Tratamiento de los participantes

La participación de estos es, evidentemente, libre. Antes de cualquier acción se ha de llevar a cabo una información de que va a llevarse a cabo. Se ha de obtener de los participantes la aceptación para la participación en el ejercicio.

Los datos que se recopilan de los participantes son anónimos, y los únicos datos recopilados por cada paciente se muestran en Figura 20.

EL consentimiento informado de cada participante se muestra en la Figura 21.

A cada participante se le entrega además una hoja informativa sobre el objeto del estudio y las condiciones de realización de este, Figura 22.

TOMA DE DATOS DEL PARTICIPANTE	
•	NUMERO PARTICIPANTE _____
•	EDAD _____
•	SEXO _____
•	DEPORTE PRATICADO _____
◦	NIVEL DE PRACTICA AFICIONADO _____ COMPETICION _____
◦	HORAS DE DEPORTE PRATICADO ALA SEMANA _____
•	BRAZO DOMINANTE DERECHO _____ IZQUIERDO _____
•	ALTURA (H) _____ (M) 160-170 _____
•	PESO 70-85 _____ 50-65 _____
•	CONTORNO BICEPS BRAZO DERECHO (CM) _____
•	LONGITUD HUMERAL BRAZO DERECHO (CM) _____
•	CONTORNO BICEPS BRAZO IZQUIERDO (CM) _____
•	LONGITUD HUMERAL BRAZO IZQUIERDO (CM) _____

Figura 20: Datos que se recopilan de los participantes

HOJA DE CONSENTIMIENTO INFORMADO	
Yo,.....,	con D.N.I. n°.....,
me declaro mayor de 18 años y declaro que he sido informado/a de manera amplia y satisfactoria, de manera oral y he leído el documento llamado "Hoja de información al participante", he entendido y estoy de acuerdo con las explicaciones del procedimiento, y que esta información ha sido realizada.	
He tenido la oportunidad de hacer todas las preguntas que he deseado sobre el estudio.	
He hablado de ello con: (Nombre del/de la investigador/a que ha dado la Información).....	
Comprendo que mi participación es en todo momento voluntaria.	
Comprendo que puedo retirarme del estudio:	
En el momento en que así lo quiera,	
Sin tener que dar ninguna explicación, y	
Sin que este hecho tenga que repercutir en mi relación con los/las investigadores/as ni promotores del estudio	
Así, pues, presto libremente mi conformidad para participar en este estudio.	
Nombre, apellidos y firma del/de la participante:	
D.N.I.:	
Edad:	
Fecha:	
Firma del participante.	

Figura 21: Hoja consentimiento informado



HOJA DE INFORMACIÓN AL/A LA PARTICIPANTE

Nombre del candidato participante en el estudio:

Título del estudio:

Participación en el estudio

Su participación en este estudio es totalmente voluntaria y si durante el transcurso del estudio usted decide retirarse, puede hacerlo libremente en el momento en que lo considere oportuno, sin ninguna necesidad de dar explicaciones y sin que por este hecho deba verse alterada su relación con el/la investigador/a principal, los/las investigadores/as colaboradores/as, los/las monitores/as o el patrocinador del estudio.

Confidencialidad de los datos

Los resultados de las diversas pruebas realizadas, así como toda la documentación referente a su persona son anónimas y únicamente estarán a disposición del/de la investigador/a principal, los/las colaboradores/as, y.....

Todas las medidas de seguridad necesarias para que los/las participantes en el estudio no sean identificados y las medidas de confidencialidad en todos los casos serán completas, de acuerdo con la Ley Orgánica sobre protección de datos de carácter personal (Ley 15/1999 de 13 de diciembre).

Publicación de los resultados

El promotor del estudio reconoce la importancia y transcendencia del estudio y, por tanto, está dispuesto a publicar los resultados en una revista, publicación o reunión científica a determinar en el momento oportuno y de común acuerdo con los investigadores. Si usted lo desea, el investigador responsable del estudio, podrá informarle de los resultados, así como de cualquier otro dato relevante que se conozca durante el estudio.

Investigador/a responsable del estudio

El Sr/Sra....., en calidad de investigador responsable del estudio o, en su caso un investigador colaborador designado directamente por este, es la persona que le ha informado sobre los diferentes aspectos del estudio. Si usted desea formular cualquier pregunta sobre lo que se le ha expuesto o si desea alguna aclaración de cualquier duda, puede manifestárselo en cualquier momento. Si usted decide participar en este estudio, debe hacerlo otorgando su consentimiento con total libertad. Los promotores del estudio y el/la investigador/a principal le agradecen su inestimable

Figura 22: Hoja informativa sobre el objeto del estudio y las condiciones de realización de este

4.1.3.3. Soporte de ensayo

Para la realización de los ejercicios de tríceps es necesario emplear el útil indicado en las imágenes anteriores.

Dicho útil se diseña específicamente para la realización de estos ejercicios, está fabricado con tubo y placa de acero St37 soldado.

La estructura del soporte es como se ve en la Figura 23.

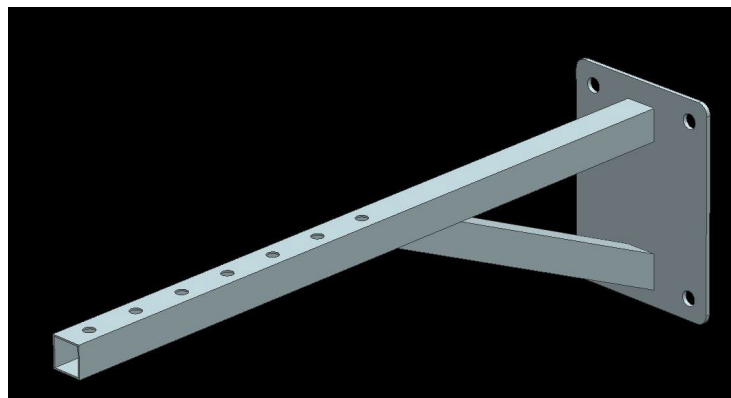
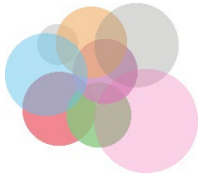


Figura 23. Estructura soldada, útil para la realización de los ejercicios de tríceps.



En él se instala una célula de carga, apta para cargas de tracción de hasta 200 Kg, para poder medir la fuerza desarrollada por cada participante en ejercicio de tríceps. Para asegurar que la carga se aplica en vertical se puede colocar la célula de carga en cualquiera de los taladros incluidos en la estructura, Figura 24. La justificación de la carga admisible en la estructura se lleva a cabo mediante un análisis estructural realizado sobre la propia estructura. La estructura se calcula para una carga máxima aplicada de 100 Kg aplicada en el extremo de ésta, Figura 25.

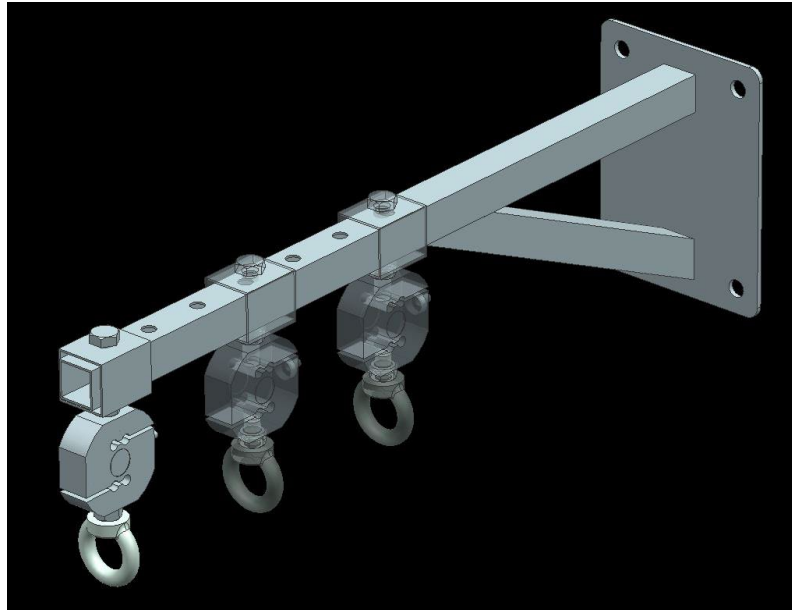


Figura 24. Célula de carga instalada representada en diferentes posiciones.

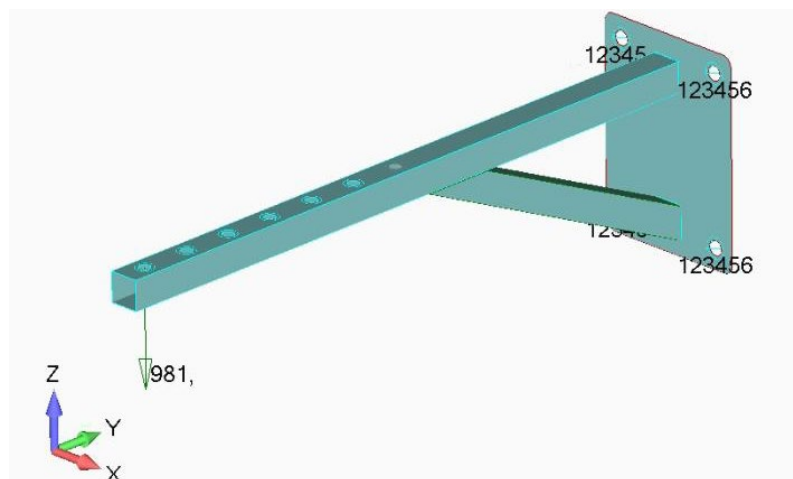


Figura 25. Modelo de carga aplicado al útil del ensayo.

El modelo de carga considerado es el siguiente.

- Los 4 tornillos de la parte posterior se consideran inmovilizados.
- La carga de 981 N aplicada en el extremo del soporte.

Los resultados del análisis estructural se muestran en la Figura 26.

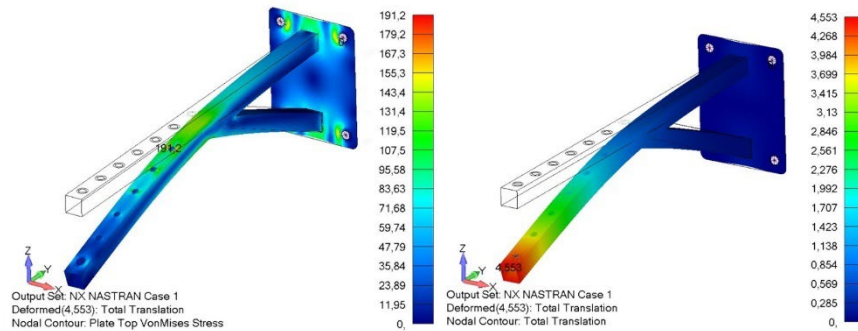


Figura 26. Tensiones y desplazamientos en la estructura.

Como se observa la tensión máxima del análisis es de 191.2 MPa, menor que el límite admisible del acero considerado (220 MPa) y un desplazamiento máximo en el extremo del útil. Dado que el útil empleado no es objeto del trabajo de investigación no es necesario aumentar el detalle de los resultados, se considera que el útil cumple los requisitos necesarios puesto que la tensión máxima de trabajo no supera el límite admisible del material y por lo tanto no se producirá el fallo de la estructura durante la realización de las pruebas de carga.

4.1.3.4. Órtesis de codo

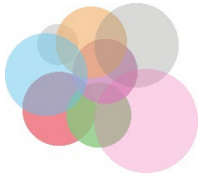
Mediante el empleo de esta órtesis se puede inmovilizar, o limitar, el recorrido en la articulación del codo al mismo tiempo que medimos el ángulo en que ésta se encuentra para así poder asociar los resultados de fuerza ejercida con el ángulo de la articulación. La órtesis es para ser usada en el brazo derecho y en la Figura 27 se observa cómo queda colocada en el participante.



Figura 27. Órtesis de codo. Ángulo de apertura ajustable y bloqueo.

4.1.3.5. Electrónica de adquisición de resultados

Circuito electrónico de fabricación propia, basado en la tecnología de Arduino, a través del cual se recibe la información de la célula de carga y se recopilan de modo sencillo los resultados del ensayo. Igualmente se desarrolla el software específico para recibir correctamente la información de la carga en la célula de carga. Esta se almacena en un archivo de texto que posteriormente se importará en Excel para su tratamiento. Las funciones del software incluyen un procedimiento de calibración de la célula de carga, este proceso de



calibración se debe llevar a cabo una única vez, al comienzo de los ejercicios de cara participante.

4.1.4. REALIZACIÓN DEL EJERCICIO

Las condiciones que cualquier participante debe cumplir son:

- Edad entre 18 y 60 años.
- Altura, entre 165 y 190 cm para hombres y entre 155 y 170 para mujeres
- Peso entre 65 y 95 Kg para hombres y entre 45 y 75 Kg para mujeres.
- El grupo de participantes debe incluir al menos 20 personas.
- Ninguno de los participantes deberá sufrir ningún tipo de dolor en el codo, brazo o antebrazo.
- Ninguno de ellos deberá tener ningún tipo de lesión en la articulación del codo o sus músculos.
- Ninguno de ellos ha tenido cirugía de brazo, hombro o antebrazo.

La prueba se realizará de la siguiente manera:

Los participantes formarán un grupo único, no se necesita grupo de control debido al objetivo de la prueba. El ensayo no tiene por finalidad validar un procedimiento o tecnología específica; si ese fuera el objetivo, se debería necesitar comparar el grupo de control con el grupo instrumentado. La prueba se especifica como un procedimiento de adquisición de datos, donde se obtienen datos para compararlos con los resultados del análisis teórico.

Los participantes realizarán una prueba de carga en bíceps y tríceps, en varios ángulos del codo. Los resultados de esta prueba son la fuerza máxima que cualquier participante puede desarrollar en la muñeca, valor que se puede comparar con la carga máxima en la muñeca que predice nuestro modelo matemático. Para realizar la prueba, utilizando la herramienta mencionada anteriormente, cada participante deberá realizar ejercicios de fuerza isométrica con carga para bíceps y tríceps. La lectura de la fuerza se realizará mediante el uso de una célula de carga y será una lectura directa, la célula de carga dará la fuerza ejercida por el participante en la muñeca. La medición de fuerza se realiza mediante un circuito de adaptación de señal de salida de una célula de carga y un software desarrollado para tal fin. Dicho software será instalado sobre una plataforma con un microprocesador ARM sobre una plataforma Arduino. Para cada uno de los participantes se recopiló la siguiente información, de acuerdo con [28], además de las fuerzas obtenidas en el propio ejercicio: Peso, Altura, Brazo dominante, Edad, Sexo, Deportes practicados, Nivel de práctica deportiva, Longitud del brazo (medida desde el hombro hasta el codo en una posición de flexión en el codo de 90 grados) y, por último, el Diámetro del brazo en la zona central del bíceps, con el brazo relajado en una posición de 150 grados. Dado que cada participante debe ejercer fuerza en diferentes ángulos del codo, hay que medir este ángulo para lo cual se utiliza una prótesis de codo con un goniómetro que permite bloquear el movimiento de la articulación en una de las dos direcciones (flexión y/o extensión) y mida el ángulo. El ejercicio se realiza como se observa en la Figura 28.

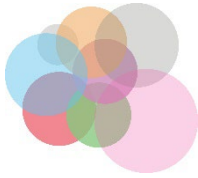


Figura 28. Realización del ejercicio. Carga en tríceps (izquierda) y en bíceps (derecha)

4.1.5. OBTENCION DE LAS FUERZAS MÁXIMAS DISPONIBLES EN LOS MÚSCULOS DE LA ARTICULACIÓN

Puesto que se necesita comparar la fuerza necesaria en el caso de carga analizado, respecto a la situación de carga admisible en los músculos analizados, se ha de conocer la fuerza máxima que cada músculo de la articulación puede desarrollar.

Se consideran los músculos de acuerdo con lo expresado en el modelo de Hill, que es el modelo más aceptado para analizar músculos. Dicho modelo se basa en que la combinación del músculo y el tendón representan lo que se conoce como MTU (unidad músculo tendón) y que ambos participan en el desarrollo de la fuerza en el músculo.

Como se explica en [2], el modelo de Hill explica que la fuerza total en el músculo es la combinación de elementos activos (músculos) y elementos pasivos (tendones), Figura 29. Varios estudios utilizan esta explicación para el comportamiento de los músculos basándose en el modelo de Hill [23], [24], [29]

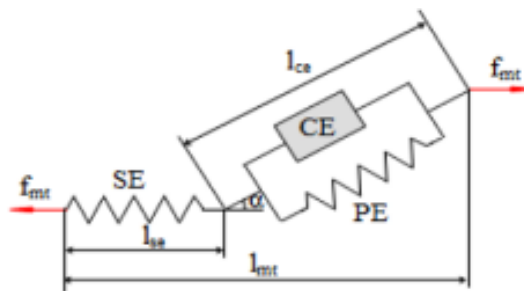
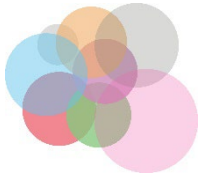


Figura 29. Modelo esquemático de Hill para la representación de la unidad músculotendinosa.



Donde SEE representa los tendones, y CE el elemento contráctil, las fibras musculares. Y PE representa los tejidos circundantes alrededor de las fibras musculares.

La relación entre la fuerza que puede desarrollar un músculo con respecto a su longitud se representa según la Figura 30.

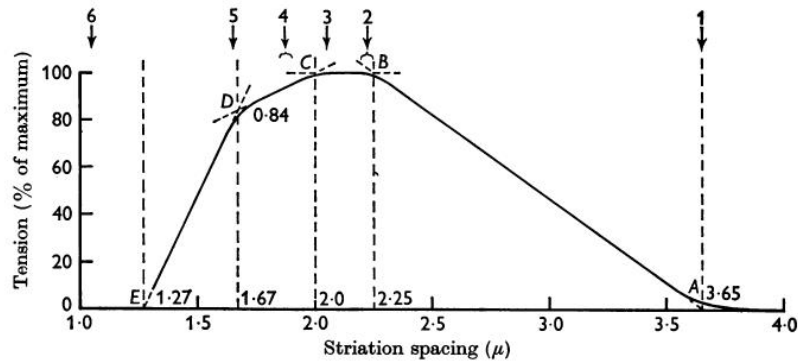


Figura 30. Relación entre la fuerza que puede desarrollar un músculo con respecto a su longitud [24].

Donde el eje X representa la longitud del sarcómero y el eje Y representa la fuerza en el músculo, referida a la máxima fuerza en el músculo – cuando la longitud del sarcómero se encuentra entre 2 y 2.5 μm. Como se expresa en [30], si al elemento contráctil se le suma el efecto del elemento serie en el modelo de Hill, Figura 31:

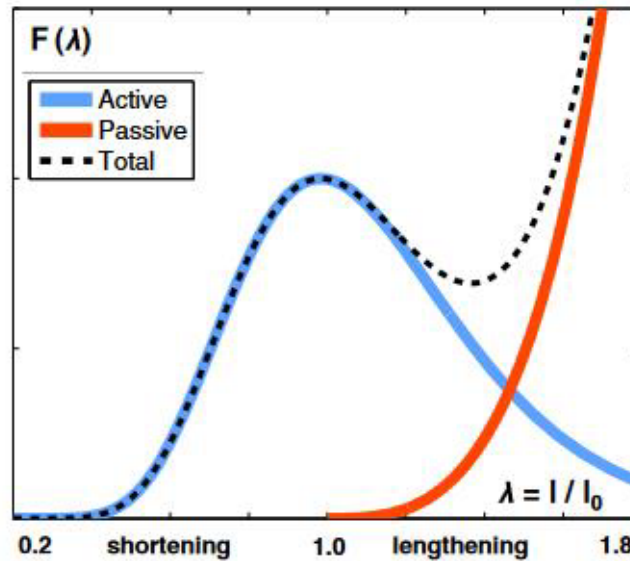


Figura 31. Evolución de la fuerza desarrollada en un músculo, teniendo en cuenta los elementos CE y SEE del modelo de Hill, referidos a su longitud.

Y representando esa gráfica, de forma matemática, aproximándola a segmentos lineales se obtiene la ecuación que se muestra en la Figura 32.

Las ecuaciones en el gráfico corresponden a cada segmento de este. Si se conoce la fuerza máxima de un determinado músculo, se puede conocer la fuerza para cada longitud que adopte el músculo (o ángulo del codo).

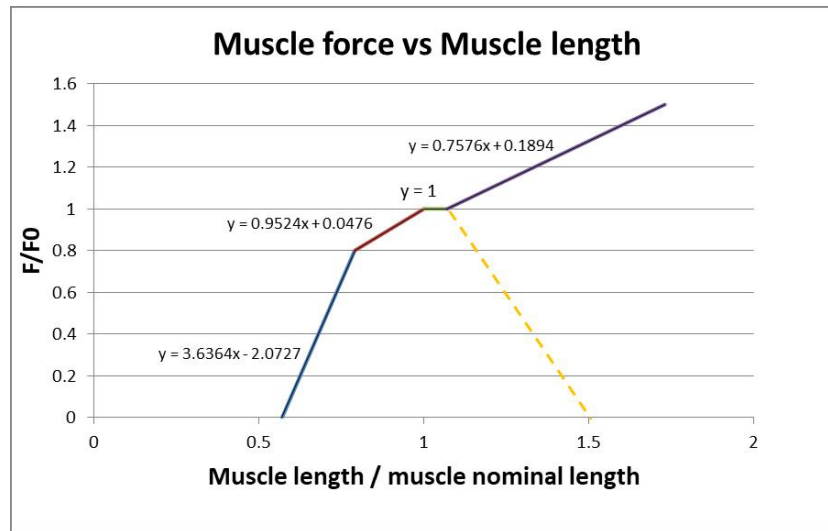


Figura 32. Aproximación matemática, a través de segmentos lineales, de la fuerza desarrollada en un músculo referida a su longitud. El eje X representa la relación entre la longitud del músculo y su longitud óptima, y el eje Y representa la relación entre la fuerza muscular y la fuerza máxima desarrollada por el mismo.

Los parámetros necesarios para caracterizar un músculo en términos de fuerza/longitud han sido estudiados en varias ocasiones. Estos parámetros son principalmente la fuerza máxima del músculo y la longitud óptima. Con esos valores se define el punto máximo del gráfico de la Figura 30 y así se pueden obtener los demás puntos de la curva para cada músculo.

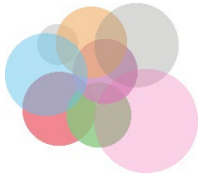
La caracterización se basa en lo dicho en [31] donde se parametrizan varios músculos de esta manera, los valores característicos que definen los músculos involucrados en nuestro estudio se muestran en la Tabla 1.

Tabla 1. Parámetros característicos de los músculos del codo [31].

MÚSCULO	PCAS (cm ²)	FO PICO FUERZA (N)	Long nominal músculo (cm)	Longitud tendón (cm)	Angulo penación (°)
Porción larga del tríceps	5.7	798.5	13.4	14.3	12
Vasto externo ó lateral	4.5	624.3	11.4	9.8	9
Vasto interno ó medial	4.5	624.3	11.4	9.1	9
Porción larga del bíceps braquial	4.5	624.3	11.6	27.2	0
Porción corta del bíceps braquial	3.1	435.6	13.2	19.2	0

Donde:

- **PCAS** es el área de sección transversal fisiológica, calculada dividiendo el volumen muscular por la longitud óptima de fibra normalizada



- **F0 Fuerza Pico** es la fuerza máxima que el músculo es capaz de desarrollar, denominada F0 y asociada a la longitud Nominal del Músculo.
- **Longitud nominal del músculo** es la longitud de las fibras asociadas al valor pico de fuerza.
- **Longitud Tendón** es la longitud del tendón en la que comienza a ofrecer resistencia al cambio de longitud. La longitud del tendón permanece constante en comparación con la longitud de la fibra durante la contracción.
- **Angulo de Penación** es el ángulo entre las fibras musculares y el eje del músculo basado en la línea entre los puntos de inserción.

Esos valores según se obtienen considerando una tensión específica en los músculos de 140 N/cm^2 , [31], sin embargo, debido a que este valor no está explicado en el estudio citado, se han revisado otros para compararlo.

Yasuo Kawakami et Al. En el estudio indicado buscan, precisamente, el valor de tensión específica de los músculos flexores y extensores. Ellos mencionan en su estudio el valor de tensión específica máxima en los músculos bíceps y tríceps de 75 y 65 N/cm^2 respectivamente, [32] como se ve en la Figura 33, considerando la contracción isométrica (0 m/s).

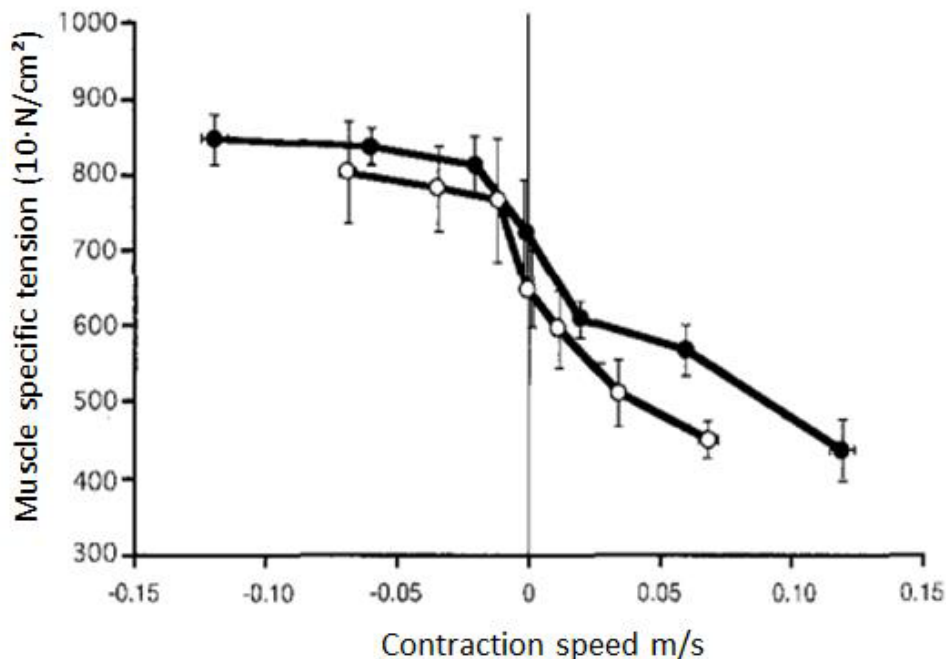


Figura 33. Tensión específica en bíceps y tríceps (eje Y) en función de la velocidad de contracción (eje X)

Modificando los valores de fuerzas máximas en la Tabla 1, manteniendo los valores de PCAS, por lo que la tensión específica se ajusta a los valores anteriores, Tabla 2 para cada unidad músculo + tendón sería:

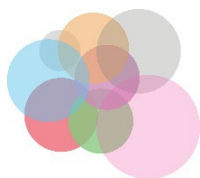


Tabla 2. Parámetros característicos de los músculos del codo. [5]

MÚSCULO	PCAS (cm ²)	F0 PICO FUERZA (N)	Long nominal músculo (cm)	Longitud tendón (cm)	Angulo de penación (°)
Porción larga del tríceps	5.7	370.5	13.4	14.3	12
Vasto externo ó lateral	4.5	292.5	11.4	9.8	9
Vasto interno ó medial	4.5	292.5	11.4	9.1	9
Porción larga del bíceps braquial	4.5	334.4	11.6	27.2	0
Porción corta del bíceps braquial	3.1	233.4	13.2	19.2	0

4.2. INFLUENCIA DEL TAMAÑO DEL MÚSCULO

Resulta intuitivo pensar que la fuerza desarrollada por un músculo es directamente proporcional a la cantidad de fibras musculares que tiene el músculo. Y ese número de fibras debe estar relacionado de alguna manera con el tamaño del músculo, esta puede ser la principal causa de la alta variabilidad que se encuentra en los resultados de las pruebas (Figura 46 y Figura 47).

Una forma sencilla de considerar este parámetro podría ser tener en cuenta el área PCAS declarada en diversos estudios para los músculos de la articulación del codo, sin embargo, para poder caracterizar la fuerza de un individuo específico, medir el PCAS de un músculo requeriría de técnicas de imagen para poder medir el volumen y longitud del músculo, lo que complicaría el estudio personalizado.

En este estudio se normalizarán las ecuaciones obtenidas para cada músculo según las secciones medias que presenta el músculo en reposo, haciendo referencia a un parámetro fácilmente medible como es el contorno del brazo en la posición del bíceps. Si además de estas medidas del contorno, y por tanto el área de la sección transversal del brazo, se aproxima a un cilindro y se considera la distribución de la sección en los diferentes músculos que lo componen, se puede obtener el área aproximada que ocupa cada músculo en la sección y así normalizar la ecuación obtenida para cada músculo por su área estimada.

El motivo de hacer esta aproximación es la de poder caracterizar estos valores en cualquier estudio médico, clínica de fisioterapia, etc.... de modo rápido.

La opción de caracterizar el área de modo más detallado siempre se puede hacer con posterioridad en caso de ser necesario.

4.2.1. Sección teórica para cada músculo

Analizando una sección transversal del brazo, como se ve en la Figura 34 [33].

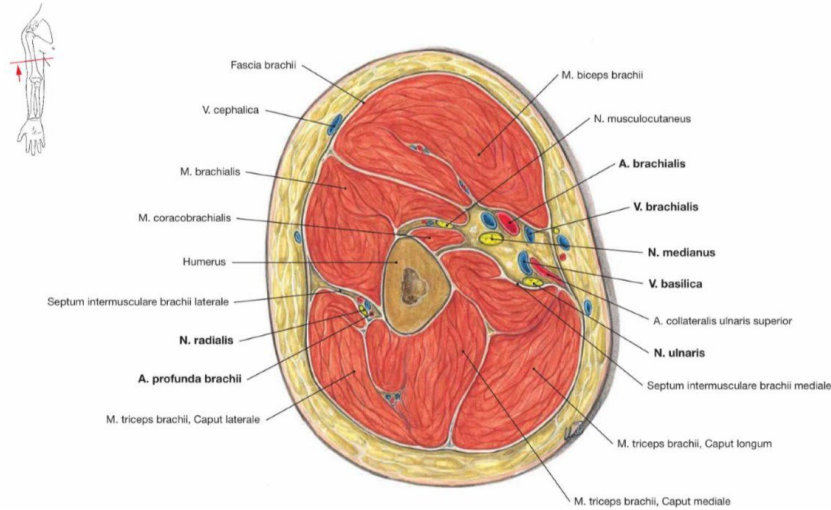
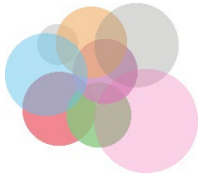


Figura 34. Sección transversal del brazo. Músculos del panorama anterior y posterior.

Se puede medir las áreas de los músculos implicados en el movimiento del codo.

A continuación, se muestra el área relativa de cualquier músculo referida al área del brazo. Para obtener estas áreas se elaboró un modelo 3D, creado a partir de la Figura 34, de la sección del brazo.

Modelando la sección transversal del brazo, e identificando los músculos del brazo, se puede calcular un área aproximada para cada uno de ellos, como se ve en la Figura 35.

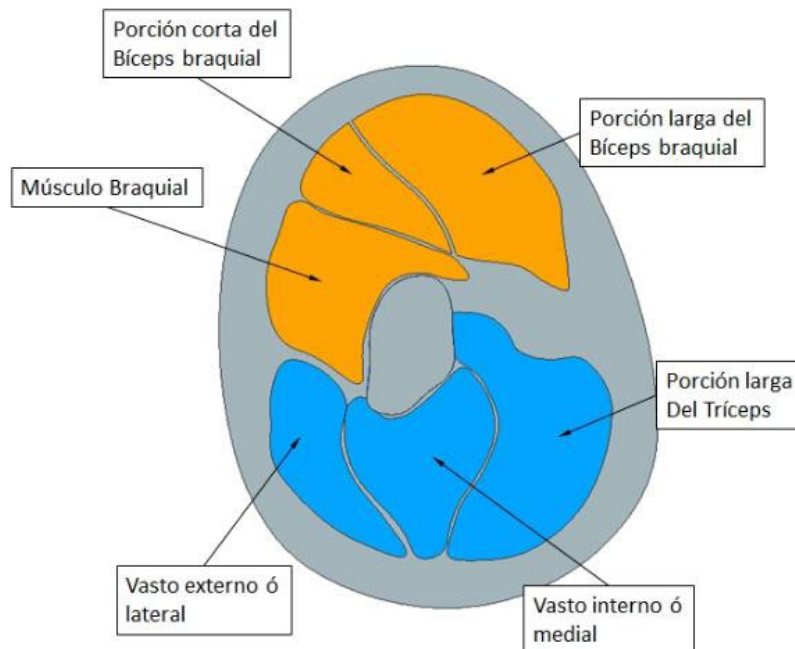


Figura 35. Sección transversal modelada para el brazo. Los músculos están identificados en la propia imagen.

En base en la representación anterior, el % de área asociada a un músculo, referido al área transversal completa de un brazo, es:

- Porción larga del bíceps braquial: 11,7 %



- Porción corta del bíceps braquial: 4,5 %
- Vasto externo o lateral: 6,4 %
- Vasto interno o medial: 8,8 %
- Porción larga del tríceps: 13,6 %
- Músculo braquial: 9,2 %

Así, aproximadamente el 54 % del área transversal de un brazo está ocupada por estos músculos.

El perímetro medio del brazo está entre 34 y 35 cm, [34] y suponiendo una forma cilíndrica en el brazo, el área sería de $97,5 \text{ cm}^2$, de este modo:

- El área ocupada por la porción larga del bíceps braquial sería: $11,4 \text{ cm}^2$
- El área ocupada por la porción corta del bíceps braquial sería: $4,4 \text{ cm}^2$

Siguiendo esta suposición, y midiendo el perímetro de un brazo en posición relajada (aproximadamente 100-110 grados de apertura, no aparece contracción en ningún músculo) se puede obtener un área aproximada de cualquiera de los músculos.

De esta forma, las fuerzas en cualquier músculo se pueden expresar como una relación fuerza/área músculo y expresarse en N/cm^2 , minimizando de esta forma la influencia del tamaño del músculo.

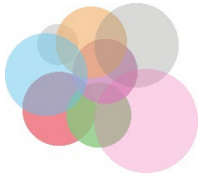
Para normalizar los valores teóricos de las fuerzas de los músculos, se considera un valor teórico de área, o volumen, del brazo y de los músculos considerados. Este cálculo se lleva a cabo según se indica a continuación.

- Se considera la medida media del perímetro del brazo, entre 34 y 35 cm.
- Se considera el área de la sección transversal ocupada por los músculos, A_m , aproximadamente el 54%. Para obtener este valor se utiliza información anatómica, creando un modelo geométrico de la sección transversal del brazo y midiendo sobre él el área de cada músculo.
- Se considera la longitud media del brazo, de la población muestreada en el ensayo, para calcular el volumen ocupado por los músculos, V_m .
- Dividiendo la fuerza teórica de cualquier músculo en cualquier ángulo por el área ocupada por los músculos, se obtiene el F/A_m teórico, expresado en N/cm^2 .
- Dividiendo la fuerza teórica de cualquier músculo en cualquier ángulo por el volumen ocupado por los músculos, se obtiene el F/V_m teórico, expresado en N/cm^3 .

CÁLCULO DEL ÁREA TRANSVERSAL, Y ÁREA DE LOS MÚSCULOS, EN LOS PARTICIPANTES

Dado que para los participantes se han tomado las medidas correspondientes, se conocen los siguientes datos de los participantes en la prueba.

- Altura
- Peso
- Longitud del Brazo
- Contorno del brazo



Se puede obtener para cada uno de ellos un área de sección transversal que representa el área ocupada por las fibras musculares de cualquiera de ellos.

En el caso de los participantes se cuenta con la medición del contorno del brazo en la zona del bíceps. Para calcular el área ocupada por las fibras musculares se realizan los siguientes cálculos.

- Partiendo del contorno del brazo medido se obtiene el área de la sección transversal, aproximándola a una circunferencia.
- En base al peso y altura del participante se puede obtener un IMC (índice de masa corporal).
- De ese IMC se puede obtener un % de grasa corporal, suponiendo una distribución uniforme de la misma en el cuerpo de la persona. Usando la fórmula explicada en [35]
$$\% \text{ grasa} = 1,2 \cdot \text{IMC} + 0,23 \cdot \text{edad (años)} - 10,8 \cdot \text{sexo} \quad (1)$$

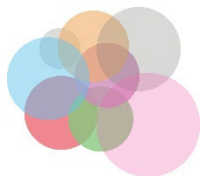
0 si es mujer) - 5,4
- Eliminando el % de grasa calculado y el área ocupada por el húmero en el brazo, el 3,5% del área de la sección transversal del brazo considerando un diámetro medio del húmero en la posición media del brazo de 2 cm [36] con un perímetro medio del brazo de 34-35 cm [34] , se puede obtener lo que se llama sección media, A_{mp} , o lo que es lo mismo, el área ocupada por las fibras musculares.
- Con la medición de la longitud del brazo de cada participante se puede obtener el volumen muscular como el producto de la sección media, A_{mp} , por la longitud medida, considerando una sección transversal uniforme en la zona del brazo, obteniendo de esta forma el volumen muscular, V_{mp} .

De esta manera se normalizan los valores de las fuerzas con respecto a la sección transversal media, A_{mp} o el volumen medio V_{mp} .

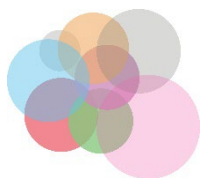
Ha de tenerse en cuenta que lo que se pretende no es obtener la fuerza por unidad de superficie muscular, se busca una forma rápida de predecir la fuerza que puede soportar un individuo en la muñeca en un ejercicio de bíceps o tríceps.

Es por eso por lo que no se normaliza las cargas en la muñeca respecto al área de un determinado músculo, lo que implicaría medir el área muscular, sino respecto a la sección transversal del brazo del individuo para facilitar la medición.

En otras palabras, para predecir la fuerza que desarrollaría una determinada persona en determinadas condiciones, bastaría con medir la longitud y el perímetro del brazo. Y siguiendo los cálculos previos obtendría A_{mp} y V_{mp} para él.



5. RESULTADOS



En este capítulo se exponen los resultados según los métodos expuestos en el capítulo anterior, en la generación de modelos CAD y de elementos finitos, así como los resultados obtenidos de los análisis realizados, que han sido recogidos en las publicaciones y artículos que generan esta Tesis Doctoral. En el estudio “Analysis and Fem Simulation Methodology of Dynamic Behavior of Human Rotator Cuff in Repetitive Routines: Musician Case Study” [37], se ha analizado y simulado el comportamiento dinámico de la articulación glenohumeral en violinistas durante movimientos repetitivos usando el método RULA y software de simulación FEM. Se identificaron zonas críticas de fatiga en el manguito rotador, especialmente en el músculo supraespinoso, con riesgo de lesiones tras ciclos repetitivos de movimientos. En el estudio “Finite Element Simulation and Analysis of the behavior under load of a human shoulder” (Technological Ecosystems for Enhancing Multiculturality, TEEM, Salamanca 2018) [38] y en el artículo “Behavior under Load of A Human Shoulder: Finite Element Simulation and Analysis” publicado en el Journal of Medical Systems en 2019 [39], se confirma que tanto el módulo de elasticidad del tejido muscular como la sección del músculo durante el movimiento de la articulación deben considerarse variables. Este hallazgo subraya la necesidad de investigaciones futuras detalladas sobre estos tejidos. Además, el modelado de los contactos en la zona de los cartílagos y ligamentos ha demostrado ser válido, lo que refuerza la precisión de estos modelos. En el congreso TEEM de 2019 en León [40] y en el artículo “Behavior Under Load of a Human Shoulder: Finite Element Simulation and Analysis” [39], se presentó un modelo de elementos finitos 1D utilizando elementos de barra (CROD) aplicado a la simulación del comportamiento muscular. Los resultados de este análisis fueron correctos al compararlos con estudios previos realizados con otras metodologías [40]. Se observó una reducción significativa en el tiempo de análisis al emplear este tipo de modelado en comparación con el mallado tradicional mediante elementos 3D. Además, la articulación que incluye un músculo modelado con un elemento CROD puede ser reutilizada en diferentes posiciones sin necesidad de modificar el mallado, lo que simplifica el proceso y permite la exportación de este modelo a la simulación de ligamentos. En 2020, en el congreso TEEM [41], se presentó un modelo de elementos finitos de un codo bajo carga, analizando el esfuerzo muscular cuando se modela utilizando elementos de barra 1D. Los resultados mostraron que es posible obtener la fuerza en cada músculo de la articulación por separado y que el modelo FEM puede reutilizarse en diferentes posiciones. Esta simplificación aplicada a los músculos no afecta las distribuciones de tensiones en el modelo en comparación con los métodos habituales mediante elementos 3D. Finalmente, en el artículo “Mechanical Model and FEM Simulations for Efforts on Biceps and Triceps Muscles under Vertical Load: Mathematical Formulation of Results” publicado en Mathematics en 2022 [42], se compararon dos modelos análogos, observando diferencias notables en el tiempo de cálculo. Se adoptó la modelización mediante elementos rígidos en los modelos a resolver, obteniendo ecuaciones específicas para cada músculo. Esto permite predecir la fuerza necesaria en cada músculo afectado en situaciones de carga teórica, como en el caso de una contracción isométrica.

5.1. VALIDACIÓN DEL MODELO TEÓRICO

La validación de los modelos teóricos desarrollados se lleva a cabo mediante un ensayo con participantes a través del cual se recopilan datos sobre las fuerzas desarrolladas por los grupos musculares del tríceps o bíceps para su posterior estudio. La realización del ejercicio se lleva a cabo según lo indicado en el apartado herramientas y metodología. El proceso de validación requiere de un trabajo previo, basado en datos teóricos, que permita definir un patrón contra el que comparar los resultados obtenidos en los ejercicios. Al tratarse de ecuaciones que definen la fuerza en los músculos ha de encontrarse un modo de poder conocer la fuerza real que esos músculos (porciones larga o corta del bíceps braquial, vasto interno o externo y porción larga del tríceps) desarrollan en situaciones reales de carga.

El método de validación elegido se define en las siguientes etapas.

- Determinar la fuerza MAXIMA que puede desarrollar cada uno de los músculos estudiados. Este valor de fuerza MAXIMA está definido mediante estudios teóricos conocidos y publicados en numerosas ocasiones en los que se define la fuerza máxima de uno de los músculos y su variación en función de la longitud de este.
- Con esas fuerzas máximas obtenidas en base a los estudios teóricos y las ecuaciones obtenidas que definen la fuerza en cada músculo en función del ángulo de la articulación (o la longitud del músculo dado que esta es única para cada ángulo) y la carga en la muñeca. Se puede obtener la carga máxima teórica que se podría aplicar en la muñeca a un ángulo dado antes de agotar la capacidad del músculo.
- Mediante un ensayo realizado por un grupo de participantes, se puede obtener la carga máxima que estos pueden desarrollar en la muñeca a diferentes ángulos.
- Una vez se tiene la carga máxima en la muñeca teórica y la carga máxima obtenida en los ensayos se comparan ambos valores y se define, de este modo, la validez de las ecuaciones obtenidas en el artículo anterior.

En este estudio, de cara a procesar los resultados obtenidos en el ensayo, se han de tener en cuenta aspectos de los participantes como son; la altura, el peso, y las dimensiones del brazo. Esta información se recopila durante la ejecución del ensayo para poder llevar a cabo los estudios correspondientes y eliminar las diferencias motivadas por los tamaños de diferentes músculos en cada uno de los participantes. Para determinar las fuerzas máximas que cada músculo puede desarrollar se ha de tener en cuenta la longitud del músculo, medida en el modelo FEM en [11] y los valores de fuerza máxima para cada músculo indicados en la Tabla 2, y aplicando las ecuaciones indicadas en la Figura 32. De este modo se obtiene la fuerza máxima que puede desarrollar un músculo de la articulación del codo, en condición isométrica, para cualquier ángulo que forme. Estos valores se muestran en esta Tabla 3. Las fuerzas isométricas necesarias en cada músculo de la articulación del codo para soportar una fuerza de 150 N aplicada verticalmente en la muñeca se representan en la Tabla 4. Los valores de la Tabla 4 hay que leerlos de forma descendente para el análisis de los músculos bíceps y ascendente para el análisis de los músculos tríceps.

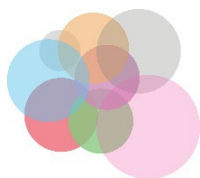


Tabla 3. Fuerzas isométricas máximas para cada músculo de la articulación del codo en función del ángulo que forma.

Angulo codo (°)	Fuerza Porción larga del bíceps braquial (N)	Fuerza Porción corta del bíceps braquial (N)	Fuerza Vasto externo o lateral (N)	Fuerza Vasto interno o medial (N)	Fuerza porción larga del tríceps (N)
44	263.06	179.80	265.00	265.00	265.00
54	267.98	182.68	265.00	265.00	265.00
64	287.69	194.23	265.00	265.00	265.00
74	300.00	200.00	265.00	265.00	265.00
84	305.64	203.17	265.00	265.00	265.00
94	321.32	212.35	265.00	265.00	265.00
104	336.99	220.39	265.00	265.00	265.00
114	350.71	229.57	265.00	265.00	265.00
124	364.42	236.46	262.79	262.79	263.12
134	376.18	243.34	253.93	256.15	257.47
144	384.01	247.93	245.08	247.29	249.93
154	389.89	252.53	236.22	236.22	240.52
164	395.77	254.82	225.15	227.36	232.98

Tabla 4. Fuerzas necesarias en cada músculo de la articulación del codo con una carga de 150 N aplicada en la muñeca en dirección vertical.

Angulo codo (°)	Porción larga del bíceps braquial		Porción corta del bíceps braquial		Vasto externo o lateral		Vasto interno o medial		Porción larga del tríceps	
	Fuerza (N)	Long. (mm)	Fuerza (N)	Long. (mm)	Fuerza (N)	Long. (mm)	Fuerza (N)	Long. (mm)	Fuerza (N)	Long. (mm)
44	244	373	96	310	148	217	70	211	269	284
54	330	375	136	312	128	216	144	209	312	282
64	341	383	152	320	212	215	115	208	340	281
74	341	391	161	328	208	215	119	208	347	281
84	340	399	168	336	235	214	150	207	340	280
94	333	407	170	344	263	214	186	207	347	279
104	302	415	181	351	233	213	164	207	331	279
114	301	422	159	359	233	213	176	206	307	278
124	274	429	143	365	213	211	170	204	271	276
134	261	435	95	371	177	207	150	201	226	273
144	185	439	87	375	131	203	120	197	175	269
154	121	442	46	379	79	199	82	192	119	264
164	52	445	1	381	24	194	40	188	58	260

Si se comparan en una gráfica la fuerza necesaria para equilibrar 150 N aplicados en la muñeca y la fuerza máxima posible en un músculo, se comprueba el nivel de utilización del músculo. En el eje X se representa el ángulo de la articulación y en el eje Y la fuerza expresada en Newtons, Figura 36 y Figura 37.

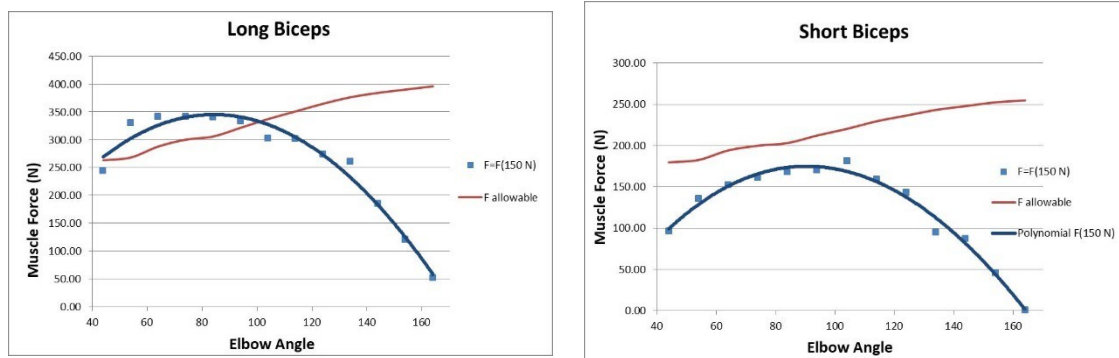


Figura 36. Fuerza de los músculos bíceps (carga de 150 N en la muñeca) versus fuerza permitida. Ángulo del codo (eje X) referido.

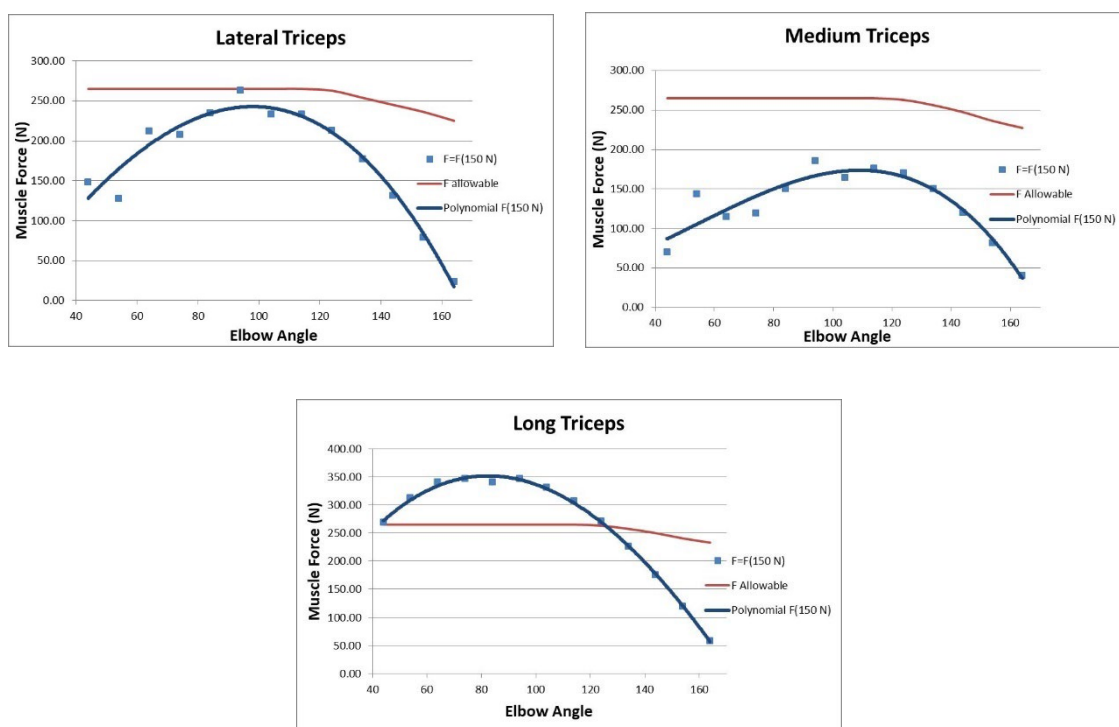


Figura 37. Fuerza de los músculos tríceps (carga de 150 N en la muñeca) versus fuerza permitida. Ángulo del codo (eje X) referido

En el caso bajo estudio se ha elegido una variación de apertura angular en codo de 44° a 164° en pasos de 10° . El codo completamente extendido corresponde a 164° y el codo completamente flexionado corresponde a 44° .

Los músculos del codo son capaces de desarrollar las fuerzas máximas, dependiendo del ángulo del codo, como se muestra en las imágenes a continuación, datos tomados de la **Errore. L'origine riferimento non è stata trovata.** Como se ve en las figuras siguientes, Figura 38 & Figura 39, la fuerza desarrollada por el músculo no es constante, sino que depende de la longitud del músculo.

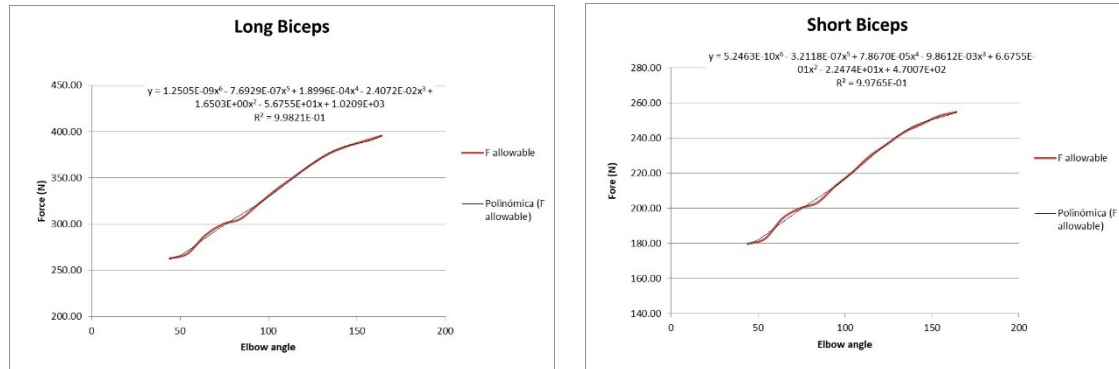


Figura 38. Fuerza máxima permitida en bíceps (corto o largo) dependiendo del ángulo del codo.

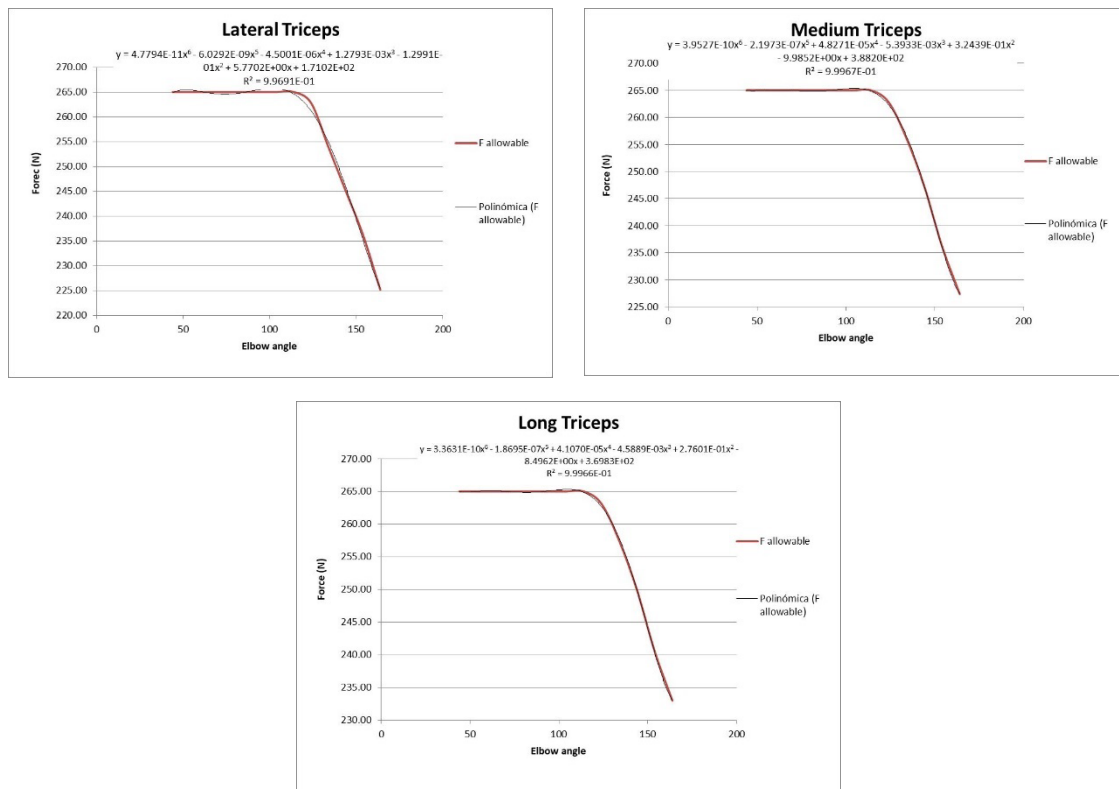


Figura 39. Fuerza máxima permitida en tríceps (Medio, Lateral o Largo) dependiendo del ángulo del codo.

Estos gráficos se han obtenido a partir de los valores de la tabla 3, las ecuaciones que se muestran en los gráficos son en forma polinómica de sexto grado, los coeficientes de las ecuaciones que aparecen en las gráficas se agrupan en la siguiente tabla.

$$f(\alpha) = P_0 + P_1 \cdot \alpha + P_2 \cdot \alpha^2 + P_3 \cdot \alpha^3 + P_4 \cdot \alpha^4 + P_5 \cdot \alpha^5 + P_6 \cdot \alpha^6 \quad (2)$$

Como la fuerza medida en la prueba no es la fuerza muscular, sino la fuerza aplicada en la muñeca, la validación de los resultados teóricos y de los datos experimentales debe realizarse de forma indirecta.

Es decir, considerando que la fuerza en la muñeca medida se corresponde con la fuerza máxima que la persona puede aplicar, se puede considerar que las fuerzas en los músculos involucrados en el ejercicio también son las fuerzas máximas en ellos.

Tabla 5. Coeficientes para los polinomios de grado 6 que predicen la fuerza máxima disponible en un músculo de la articulación del codo en función del ángulo que forma.

Músculo	P ₆	P ₅	P ₄	P ₃	P ₂	P ₁	P ₀	R ²
Porción larga del bíceps braquial / Long bíceps	1.2505e-9	-7.6929e-7	1.8996e-4	-0.024072	1.6503	-56.755	1020.9	0.99821
Porción corta del bíceps braquial / Short bíceps	5.2463e-10	-3.2118e-7	7.8670e-5	-0.0098612	0.66755	-22.474	470.07	0.9976
Porción larga del tríceps / Long Tríceps	3.3631e-10	-1.8695e-7	4.1070e-5	-0.0045889	0.27601	-8.4962	368.83	0.99966
Vasto interno o medial / Medium Tríceps	3.9527e-10	-2.1973e-7	4.8271e-5	-0.0053933	0.32439	-9.9852	388.20	0.99967
Vasto externo o lateral / Lateral Tríceps	4.7794e-11	-6.0292e-9	-4.5001e-6	0.0012793	-12.991	5.7702	171.02	0.9969

Las ecuaciones para predecir fuerzas isométricas en los músculos de la articulación del codo, en cualquier ángulo de 44 a 164°, cuando se carga en la muñeca, son en forma polinómica de cuarto grado, según la expresión siguiente.

$$f(\alpha, w) = p_{00} + p_{10} \cdot \alpha + p_{01} \cdot w + p_{11} \cdot w \cdot \alpha + p_{20} \cdot \alpha^2 + p_{21} \cdot w \cdot \alpha^2 + p_{30} \cdot \alpha^3 + p_{31} \cdot w \cdot \alpha^3 + p_{40} \cdot \alpha^4 \quad (3)$$

Donde α es el ángulo (grados) del codo y w (Newtons) es la carga aplicada a la muñeca. Los coeficientes que dependen del músculo considerado se muestran en la Tabla 6. Durante la prueba, el resultado obtenido será la fuerza máxima que cualquier persona podría aplicar en la muñeca, así se conocerá el valor de w en las ecuaciones anteriores, como también se conoce el ángulo del codo, se podrá predecir la fuerza que cualquiera de los músculos de la articulación necesita generar para soportar la carga w .

Dado que esta predicción corresponde a la fuerza en cualquier músculo, para la carga máxima que la persona puede mantener en la muñeca, se puede comparar la fuerza muscular estimada con la fuerza teórica máxima permitida en el músculo (asumiendo que la carga máxima soportada es la carga soportada cuando los músculos desarrollan la fuerza máxima).

Esta fuerza máxima, permitida, sólo depende de la geometría del músculo, fibras, longitud, sección transversal.... Y para cualquiera de los músculos involucrados en el estudio, estas fuerzas máximas seguirán las gráficas y ecuaciones que se muestran en Figura 38 y Figura 39, en función del ángulo del codo.

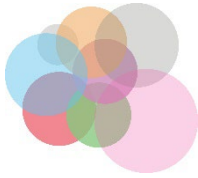


Tabla 6. Coeficientes para los polinomios de las ecuaciones que predicen la fuerza en un músculo de la articulación del codo en función de la carga aplicada en la muñeca (w) y el ángulo que forma el codo (α).

Músculo	P_{00}	P_{10}	P_{01}	P_{20}	P_{11}	P_{30}	P_{21}	P_{40}	P_{31}
Porción larga del bíceps braquial / Long bíceps	-2019	94.81	0.04386	-1.544	0.05606	0.01047	-0.0003463	-2.516e-5	1.082e-7
Porción corta del bíceps braquial / Short bíceps	77.91	-3.618	-0.8343	0.06135	0.04654	-4.245e-4	-2.803e-4	1.022e-6	1.695e-7
Porción larga del tríceps / Long tríceps	-13.56	0.6366	-0.2656	-0.01038	0.06707	7.035e-5	-0.0004743	-1.691e-7	5.439e-7
Vasto interno o medial / Medium tríceps	-118.9	5.576	0.2513	-0.09088	0.0002843	0.000616	0.0002236	-1.481e-6	-1.375e-6
Vasto externo o lateral / Lateral tríceps	1081	-50.7	-0.6042	0.8263	0.03876	-0.005601	-9.62e-5	1.346e-5	-6.915e-7

Según el estudio realizado se puede conocer la fuerza que se necesita desarrollar en cada músculo de la articulación para soportar una carga w aplicada sobre la muñeca en cualquier ángulo α en el que se encuentre la articulación.

La relación fuerza vs carga en la muñeca se considera lineal, siguiendo la ecuación que se muestra a continuación, extraída de [11]:

$$\frac{W_q + W'_s}{F_m} = \frac{F_{ext}}{R_{musc}} = \frac{c \cdot \cos \alpha}{b \cdot \sin \beta} = \text{constante}(\alpha, \beta) = \frac{150 \text{ N}}{F_m(150 \text{ N})} \quad (4)$$

Si se fija el ángulo en la articulación, y se dibuja la relación fuerza vs carga en la muñeca se obtiene una línea recta que daría la fuerza necesaria en el músculo para una carga aplicada en la muñeca.

Para responder a la pregunta “¿qué carga máxima se puede soportar en la muñeca en un determinado ángulo?” se debe buscar el valor en esa recta (fuerza muscular vs. carga aplicada en la muñeca) que coincida con la fuerza máxima teórica que el músculo puede entregar en ese ángulo de apertura articular. Por ejemplo, la carga que debe desarrollar el músculo bíceps largo para equilibrar una carga aplicada a la muñeca con un ángulo de apertura de 64 grados sigue la curva de la Figura 40.

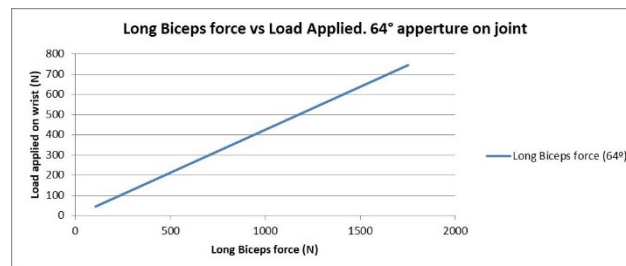


Figura 40. Fuerza que debe desarrollar el músculo bíceps largo (X), con un ángulo de 64° de apertura en el codo, dependiendo de la carga aplicada en la muñeca (Y).

Por otro lado, la fuerza máxima que puede desarrollar el bíceps largo se muestra en la Figura 41. Comprobando como con una apertura de 64° en el codo la fuerza máxima en el músculo sería de 287.89 N

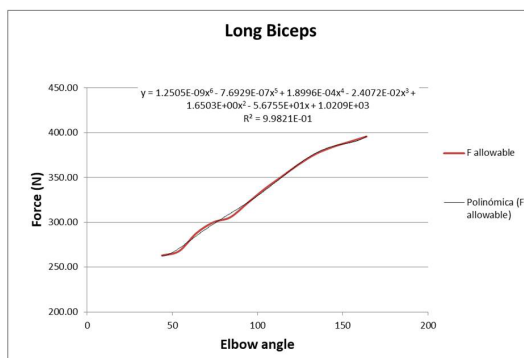


Figura 41. Fuerza máxima en el músculo bíceps largo respecto al ángulo que forma el codo y ecuación característica. Así, con una apertura de 64 grados en la articulación, la fuerza máxima permitida en el músculo bíceps largo sería de 287,89 N, sin depender de la carga aplicada.

Comprobando el punto de corte entre la fuerza máxima permitida en el músculo con la fuerza necesaria en el músculo para una determinada carga, se puede obtener la carga máxima que se podría aplicar en la muñeca se muestra en la Figura 42.

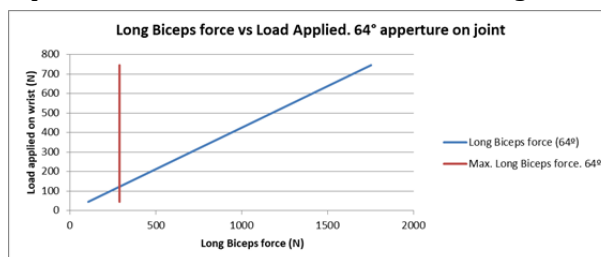


Figura 42. Fuerza máxima en el músculo bíceps largo en un ángulo del codo de 64 grados (rojo) y fuerza en el músculo bíceps largo en función de la carga aplicada en la muñeca (azul). La intersección entre ambas curvas representa el punto de carga máxima permitida en la muñeca para el ángulo de 64 grados.

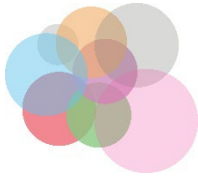
En la gráfica anterior se comprueba como la fuerza en el músculo bíceps largo aumenta de modo lineal a medida que aumenta la carga. Línea azul.

Se observa como la fuerza máxima disponible en el músculo bíceps largo es independiente de la carga aplicada en la muñeca. Línea roja

La carga máxima en la muñeca, limitada por el músculo bíceps largo, será de 105,5 N valor obtenido en la intersección de ambas curvas.

Según el estudio teórico, la fuerza máxima que se puede soportar en la muñeca vendrá determinada por la fuerza máxima que sea capaz de aportar cualquiera de los músculos implicados en la articulación. De esta forma la fuerza máxima, en un determinado ángulo, que soporta la muñeca en un ejercicio de bíceps será la mínima que cualquiera de los dos bíceps, corto o largo, pueda soportar en ese ángulo.

$$F_{\text{máx}} = \text{Min} (f_{\text{máx long biceps}}, f_{\text{máx short biceps}}) \quad (5)$$



Para calcular la carga máxima que puede soportarse en la muñeca se ha de obtener la carga que necesita cualquier músculo para desarrollar su fuerza máxima permitida. Para ello basta con fijar el ángulo del codo e igualar las (2) y (3). El valor obtenido es la carga sobre la muñeca que necesita la máxima fuerza sobre el músculo considerado equilibrado. Y también lo sería la carga máxima permitida en la muñeca. Haciendo esto para cualquier ángulo del codo, y todos los músculos estudiados, se obtienen las gráficas de Figura 43 que muestran la carga máxima en la muñeca en cualquier ángulo del codo.

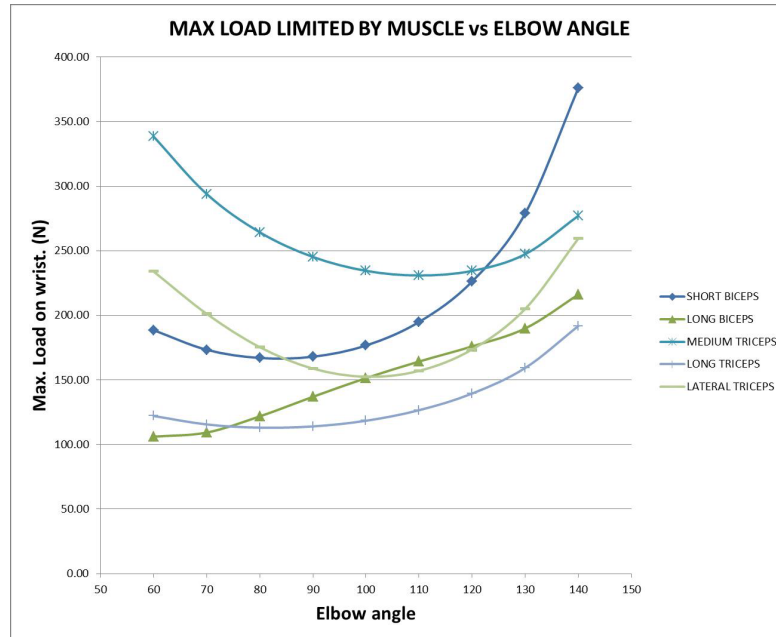


Figura 43. Carga máxima aplicada en la muñeca que cada músculo del codo puede soportar sin exceder su estado de fuerza máxima para el ángulo indicado.

Y las ecuaciones para cada músculo son:

- Porción larga del bíceps braquial:
$$F_m = 1.13E-05\alpha^4 - 4.51E-03\alpha^3 + 6.62E-01\alpha^2 - 4.09E+01\alpha + 1.01E+03$$
$$R^2 = 1.00E+00$$

- Porción corta del bíceps braquial:
$$F_m = 7.34E-04\alpha^3 - 1.54E-01\alpha^2 + 9.92E+00\alpha - 1.40E+01$$
$$R^2 = 9.96E-01$$

- Vasto interno o medial:
$$F_m = -1.31E-05\alpha^3 + 4.99E-02\alpha^2 - 1.03E+01\alpha + 7.80E+02$$
$$R^2 = 9.96E-01$$

- Porción larga del tríceps:
$$F_m = 0.00\alpha^4 - 0.00\alpha^3 + 0.21\alpha^2 - 14.06\alpha + 468.71$$
$$R^2 = 1.00$$

- Vasto externo o lateral:
$$F_m = 0.00\alpha^3 - 0.03\alpha^2 - 2.67\alpha + 449.54$$

$$R^2 = 1.00$$

Como se ve en gráficos anteriores, Figura 43, no ambos músculos se cargan de la misma manera cuando se carga la muñeca. Uno de los músculos implicados en la carga es el que limita la carga máxima que se puede soportar. Agrupando por grupos de músculos bíceps o tríceps, los gráficos se muestran en la Figura 44. Como se observa en Figura 44 y Figura 45, la porción larga del bíceps braquial y la porción larga del tríceps, son los músculos limitantes, esto significa que, aunque los otros músculos no están en estado de máximo esfuerzo, la articulación no es capaz de soportar más carga porque estos dos músculos estarán en un estado de carga mayor que el máximo permitido.

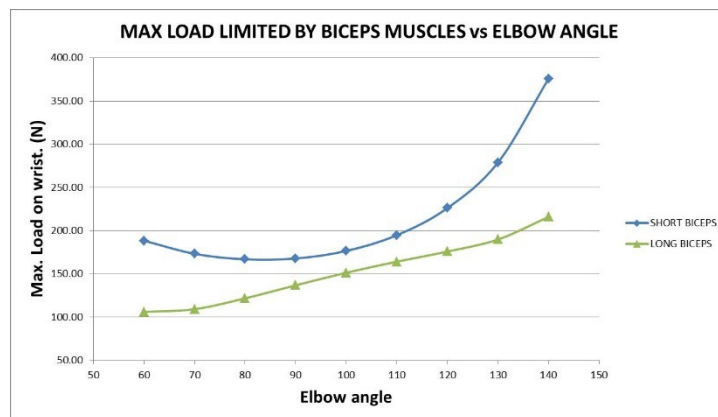


Figura 44. Carga máxima aplicada en la muñeca que los músculos bíceps pueden soportar sin exceder su estado de fuerza máxima para el ángulo indicado.

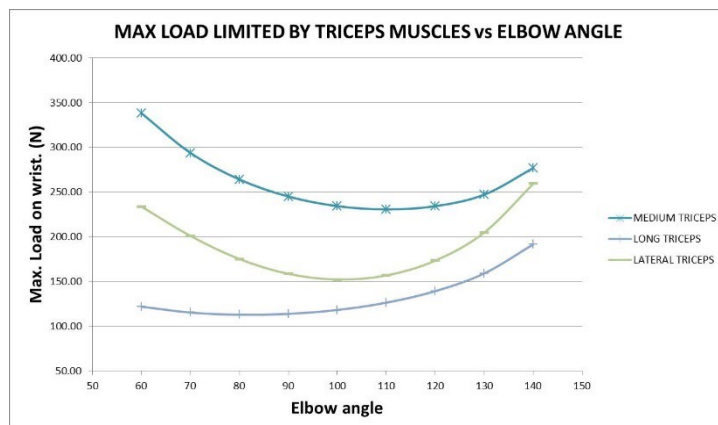


Figura 45. Carga máxima aplicada en la muñeca que los músculos tríceps pueden soportar sin exceder su estado de fuerza máxima para el ángulo indicado.

5.2. ENSAYO PARA VALIDACION

Un total de 43 participantes colaboraron en el desarrollo de la prueba de carga descrita anteriormente. 11 mujeres y 32 hombres realizaron ejercicios de carga en bíceps y/o tríceps. La lectura de la fuerza obtenida en la célula de carga se realiza cada 100 mS, de manera que en cualquier ensayo se puede obtener multitud de resultados (10 valores por segundo). Para obtener el valor máximo y filtrar los posibles picos de fuerza que se producen en la actuación del usuario, se considera como valor máximo de fuerza desarrollada por el



participante el valor medio de los valores medidos que superen el 90% del valor máximo registrado. Es decir:

- Se obtiene el valor máximo leído por el software.
- Se calcula el 90% de ese valor máximo.
- Se eliminan los valores que se encuentran por debajo de ese valor, $0.9 \cdot F_{\text{máx}}$.
- Se calcula el valor medio de los valores considerados.

Representando gráficamente los valores medios obtenidos para cada participante y cada ángulo, se obtienen las curvas de la Figura 46 y Figura 47.

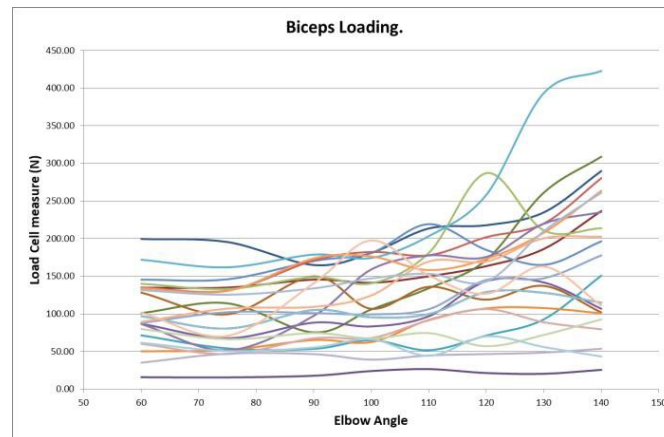


Figura 46. Fuerzas en la muñeca para el ejercicio de carga de bíceps y cualquier participante.

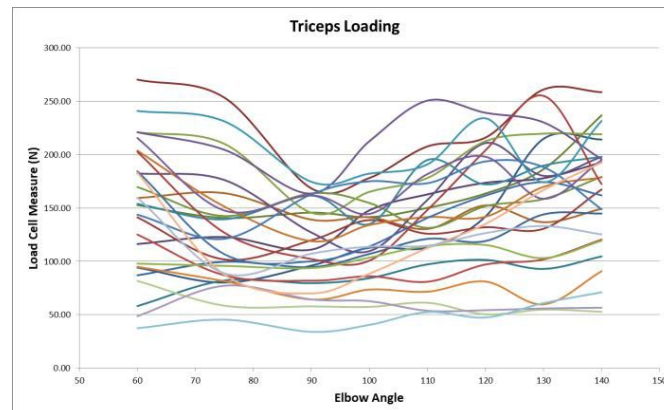


Figura 47. Fuerzas en la muñeca para el ejercicio de carga de tríceps y cualquier participante.

Con estos valores de fuerza isométrica media para cada participante y ángulo en la articulación, se puede obtener el valor medio de fuerza entre todos los participantes para cada ángulo.

Comparación de la fuerza máxima teórica y la fuerza máxima experimental

Representando en el mismo gráfico las curvas correspondientes a los valores medios de fuerzas máximas medidas en la célula de carga y la curva correspondiente a la carga teórica máxima que podría soportar la muñeca (mínima de Figura 44 y Figura 45), se obtienen los siguientes resultados, Figura 48 y Figura 49.

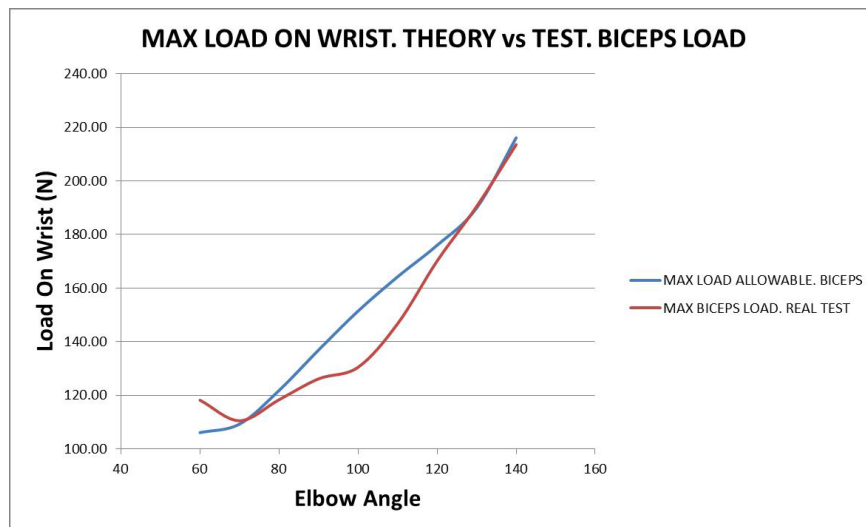


Figura 48. Carga máxima que puede soportar la articulación del codo cargando la muñeca en ejercicio de bíceps.
Resultado teórico (azul vs resultado de la prueba (rojo).

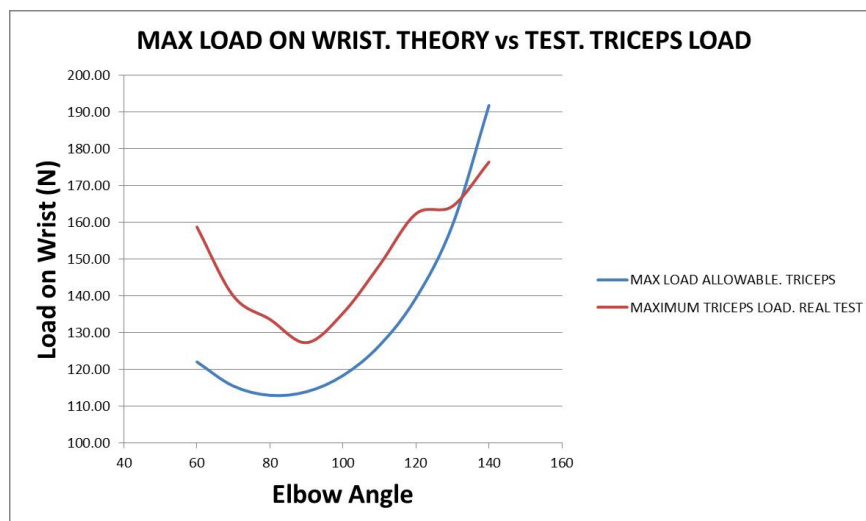


Figura 49. Carga máxima que puede soportar la articulación del codo cargando la muñeca en ejercicio de tríceps.
Resultado teórico (azul) vs resultado de la prueba (rojo).

5.2.2. Efecto de la dimensión del brazo en las fuerzas medidas

Como se ha indicado anteriormente, normalizando los valores de las fuerzas máximas en la muñeca (carga máxima soportada) en relación con el área, o el volumen, del brazo, se minimizan los efectos de variabilidad de carga máxima debidos al tamaño del brazo. Para hacer esto es suficiente dividir los valores de las fuerzas obtenidas entre el área transversal del brazo o el volumen medido de acuerdo con lo indicado en el apartado *INFLUENCIA DEL TAMAÑO DEL MÚSCULO* Representando gráficamente los valores normalizados (al área o volumen), para los test de carga en bíceps y tríceps, se obtienen las curvas de la Figura 50 y Figura 51. Para el caso de carga sobre el bíceps.

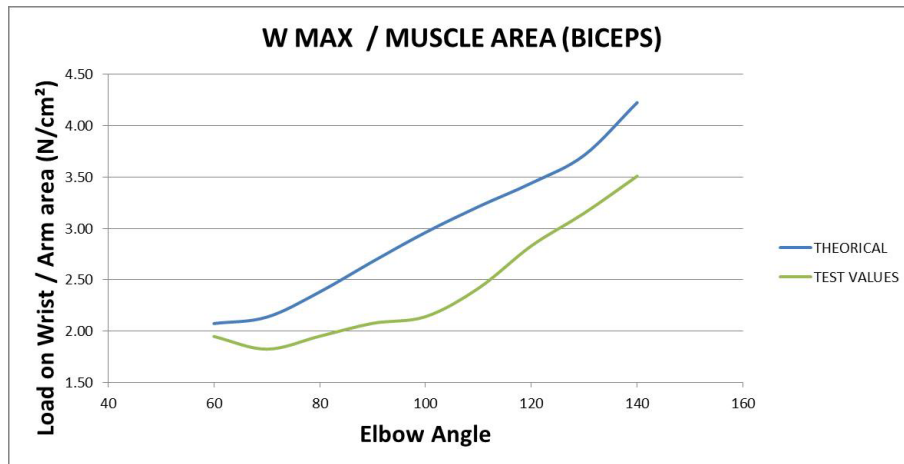


Figura 50. Carga máxima en la muñeca dividida por la sección transversal ocupada por músculo. Ejercicio de carga sobre bíceps. Resultado teórico (azul) vs resultado de la prueba (rojo).

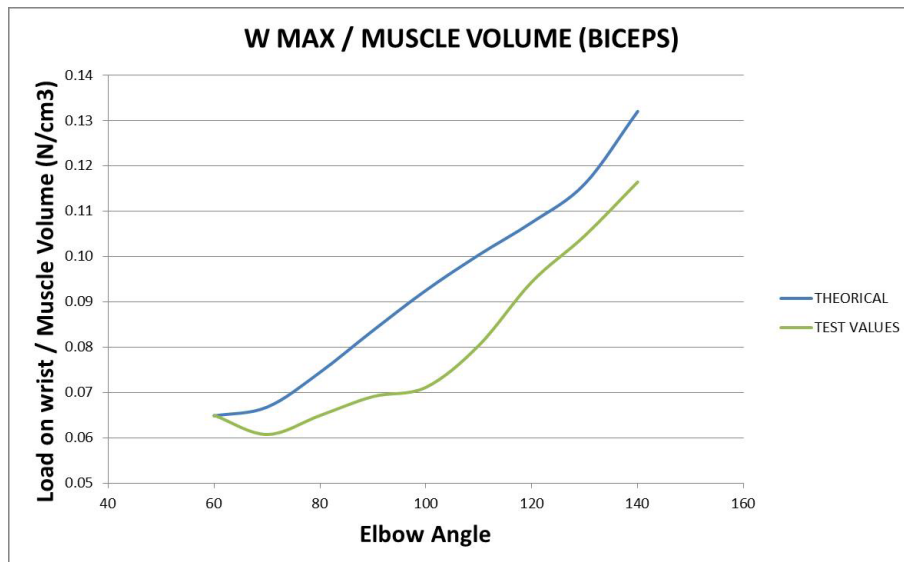


Figura 51. Carga máxima en la muñeca dividida por el volumen ocupado por músculo. Ejercicio de carga sobre bíceps. Resultado teórico (azul) vs resultado de la prueba (rojo).

Con un error máximo de 1 N/cm^2 si se normaliza al área o 0.02 N/cm^3 si se normaliza con respecto al volumen del músculo. La Y para el caso de carga en el tríceps, Figura 52 y Figura 53.

Con un error máximo de 0.5 N/cm^2 normalizando al área transversal y de 0.02 N/cm^3 normalizando al volumen del músculo.

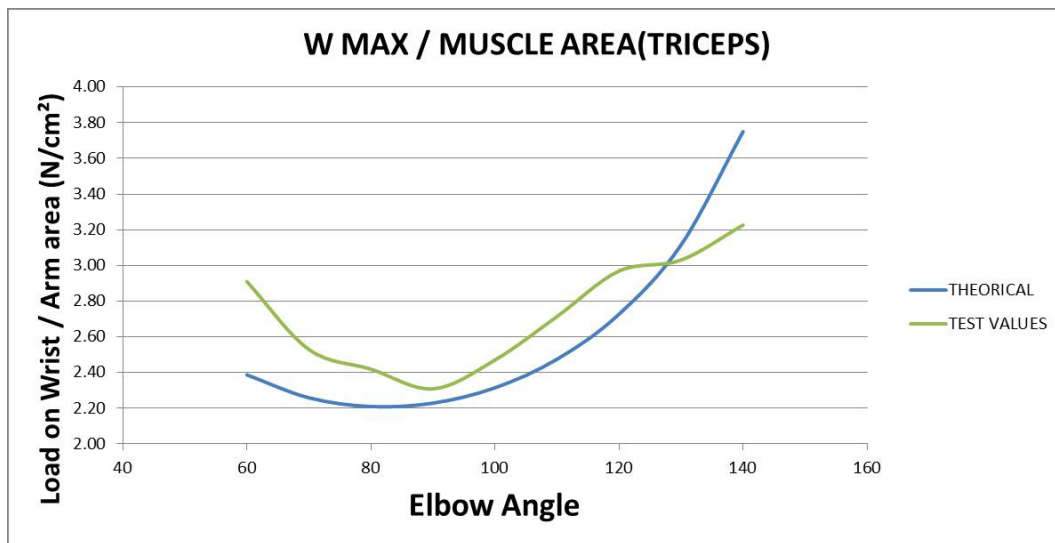


Figura 52. Carga máxima en la muñeca dividida por la sección transversal del músculo del brazo. Ejercicio de carga sobre tríceps. Resultado teórico (azul) vs resultado de la prueba (rojo).

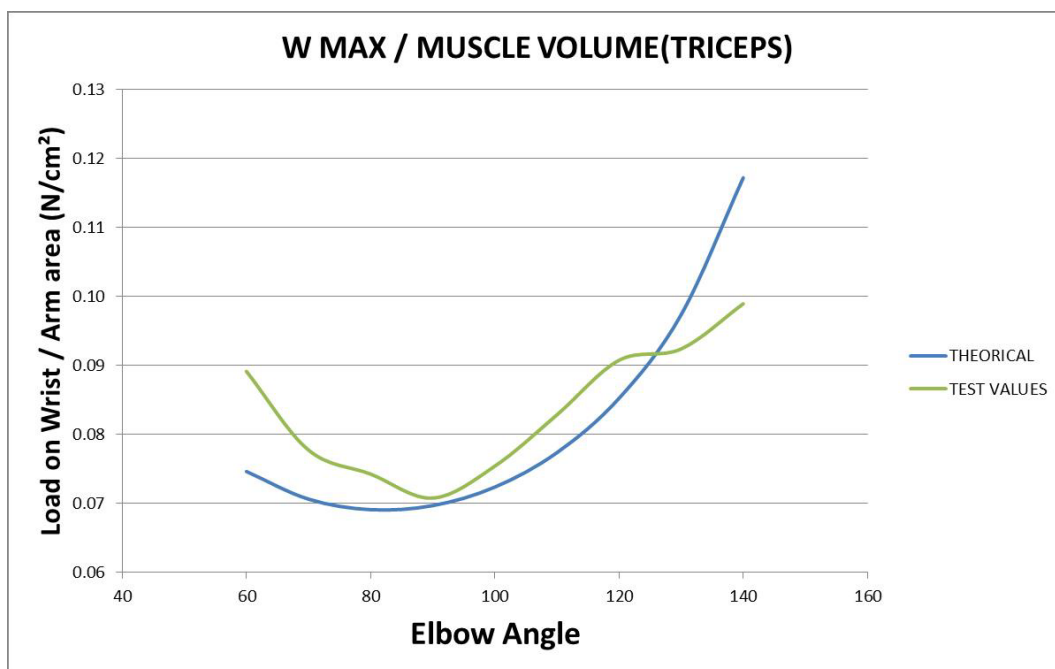
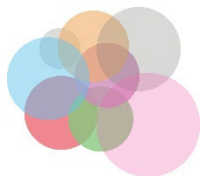
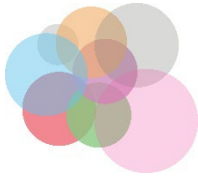


Figura 53. Carga máxima en la muñeca dividida por el volumen muscular del brazo. Ejercicio de carga sobre tríceps. Resultado teórico (azul) vs resultado de la prueba (rojo).



6. DISCUSIÓN



Este estudio se centra en la obtención de unas ecuaciones de comportamiento de los músculos de la articulación del codo, en ejercicio de flexión (bíceps) y extensión (tríceps). Para ello se emplea una combinación de técnicas de modelado 3D, simulación numérica y análisis matemático. Las ecuaciones obtenidas permiten estimar y predecir el comportamiento de los músculos dentro de la articulación de la que forman parte.

Hasta ahora el trabajo de simulación del comportamiento de músculos, o tejido muscular, se ha realizado con fines diferentes al de este estudio. Los estudios realizados hasta la fecha están, en su mayoría, centrados en conocer que le ocurre al músculo en una situación concreta de carga.

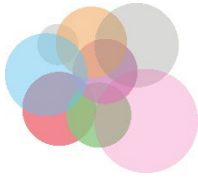
Habitualmente se centran en explicar el estado de carga del músculo en esa situación concreta y no tanto en desarrollar una metodología de estudio que permita prever el estado de carga del músculo en otras situaciones diferentes. Haciendo referencia a los músculos de una articulación, lo normal es que las simulaciones realizadas se centren en una posición concreta de ésta.

En este estudio se propone un método de cálculo analítico para obtener las fuerzas en los músculos flexores (bíceps largo y corto) y extensores (tríceps largo, lateral y medio) en cualquier configuración del ángulo de apertura de la articulación del codo y ante cualquier carga aplicada a la muñeca, [11]. El objetivo final de este estudio es permitir la creación de un nuevo tipo de elementos finitos que pueda incluirse en un modelo musculoesquelético que pueda analizarse en diferentes posiciones y configuraciones sin necesidad de repetir el proceso de mallado del modelo.

Aunque existen numerosos estudios en la misma línea, [4], [21], [26], [43], [44], ninguno aborda el problema desde nuestra perspectiva. El uso del método de los elementos finitos es ampliamente reconocido como herramienta de cálculo para problemas estructurales, estáticos y/o dinámicos. Sin embargo, hasta ahora no se ha abordado la caracterización del músculo, para que sea sencillo incluirlo en un modelo de elementos finitos.

Los principales inconvenientes que presentan los estudios realizados hasta el momento, para poder incluir el músculo en un modelo de elementos finitos válido para analizar el comportamiento de un músculo dentro de una articulación en diversas posiciones de esta son:

- Los estudios se han centrado en una posición de la articulación, obteniendo resultados relacionados con el esfuerzo muscular difíciles de extrapolar a diferentes posiciones [21]
- Las ecuaciones obtenidas para caracterizar el músculo son difíciles de implementar en un modelo FEM, referencias [43], [45]
- Varios de estos estudios utilizan elementos 3D para representar el músculo, lo que permite comprender mejor el comportamiento interno del propio músculo (distribución de



tensiones y tensiones, deformaciones...) pero no permite reutilizar la malla creada en otras posiciones de la articulación, [21], [26], [45], [46].

No existen estudios previos que realicen estudios similares al de esta investigación, por lo que no se puede realizar una comparación directa con lo ya realizado, sin embargo, la tendencia observada en el esfuerzo muscular durante el movimiento articular es similar a lo que otros estudios han publicado, [47], [48]. Aunque estos estudios miden el momento en la articulación, (o lo que es lo mismo, la suma de los momentos generados por cada músculo), y no la fuerza independiente en cada músculo los resultados presentados en estas investigaciones son similares a los presentados en este estudio en términos de la evolución del momento en la articulación, o de la fuerza en el músculo en nuestro caso, a medida que varía el ángulo de la articulación.

Dichas ecuaciones además han sido validadas mediante la realización de un ejercicio con participantes. Para ello se ha realizado un ensayo/ejercicio siguiendo la metodología de ensayo simple ciego con la autorización de los participantes mediante consentimiento informado.

En estudios similares a la presente investigación, se han desarrollado ensayos con participantes, [49], [50], [51], [52], [53], [54], aunque sus objetivos no estaban enfocados como en este trabajo. J.S. Lee et al. [49] obtienen la relación entre las fuerzas musculares y los ángulos del codo mediante el uso de EMG y un total de 8 sujetos realizando una prueba con un banco de ejercicios hecho a medida.

Way Tong Virginia Chu et al. [50] analizan niños con distonía secundaria durante la flexión del codo buscando la variabilidad de las medidas.

A.L. Hof y otros [51] comparan las mediciones de un proceso de tecnología EMG y una técnica de medición de calímetro realizando una prueba con 4 sujetos. Miden el torque y el ángulo de cada participante, también se considera en el estudio la velocidad angular.

Elsa Sarabia et al. [52], [53] realizaron un ensayo doble ciego en el que buscaron el efecto del uso de una máquina isoinercial en el tratamiento de nadadores. Comprobación de parámetros como percepción del dolor, mejora de la fuerza y rango de movimiento.

Hanjabam Barun Sharma et al. [53] buscan la relación entre la fuerza muscular y la longitud del músculo mediante la medición del torque del codo con una prueba realizada por una persona, y Hyung Yun Choi et al. [54] realizan una prueba con varios participantes para medir las fuerzas en los brazos y antebrazos cuando ocurre un choque y el paciente está sosteniendo el volante.

Benjamín Goislard de Monsabert et al. en su estudio Modelando las relaciones fuerza-longitud-activación de los músculos extensores de la muñeca y los dedos, mide los torques y establece una relación longitud-fuerza en las articulaciones de la muñeca y los dedos cuando diez participantes realizaron contracciones de extensión isométrica con variación de fuerza en diferentes posturas en un ergómetro, almacenando en un registro el momento resultante.

Todos ellos utilizan el ensayo y los diferentes participantes como herramienta preferida para comprobar y validar modelos teóricos, aunque sólo unos pocos siguen nuestros mismos objetivos; Loss JF, Candotti [47] obtienen los torques en la articulación y las



fuerzas en los músculos del codo y sus resultados y los nuestros están alineados mostrando la misma tendencia en los esfuerzos en los músculos.

Charles Pontonnier et al. [55] utilizan métodos dinámicos y técnicas de optimización para estimar las fuerzas de los músculos en el movimiento del codo, nuevamente, sus resultados, aunque recorren un camino diferente para obtenerlos, están alineados con los nuestros, en tendencia de esfuerzos en diferentes músculos, aunque no tan similares en cuanto al valor de estimación de fuerza para diferentes músculos

La mayoría de las pruebas se centran en conocer los esfuerzos musculares en uno o varios músculos en una situación de carga definida, pero no es común que busquen la extrapolación de estos resultados a otros estados de carga. El único estudio encontrado en el mismo campo en el que se trabaja es el realizado por Christian Kleinbach et al. [29] Implementación y validación del modelo muscular extendido tipo Hill con capacidades robustas de enrutamiento en LS-DYNA para modelos de cuerpo humano activo, donde modifican el material LS-DYNA para simulación muscular en condiciones dinámicas obteniendo los parámetros para ejecutar simulaciones con LS -DYNA.

Otros estudios validan los resultados teóricos mediante EMG [47], [56], Alejandro Zar et al. [56] calculan las fuerzas en los músculos basándose en los resultados de EMG obtenidos y Loss JF et al. [47] realizaron el mismo estudio basado en EMG y obtuvieron el valor resultante de las fuerzas musculares en la mano.

No se ha encontrado ningún otro estudio previo que busque ecuaciones de comportamiento para predecir el valor de la fuerza de los músculos y, por supuesto, ninguna ecuación característica que pueda exportarse a un nuevo modelo de elementos finitos como se ha hecho en este trabajo.

Darío Panarello et al. en su estudio [57] Análisis biomecánico de la parte superior del cuerpo durante tareas industriales aéreas utilizando electromiografía y captura de movimiento integradas con modelos humanos digitales desarrollan un análisis biomecánico de la parte superior del cuerpo, incluyendo las extremidades superiores, el cuello y el tronco durante la ejecución de tareas industriales aéreas (trabajo debajo de un automóvil por ejemplo) que apoyan su estudio en la adquisición de imágenes del sujeto durante las tareas y también sensores EMG durante la actividad. El resultado de su estudio es un resumen de los ángulos, velocidades y fuerzas en los músculos mientras se realizan las tareas mencionadas.

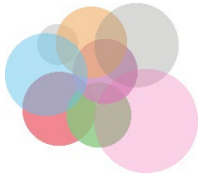
Una vez validadas las ecuaciones de comportamiento de los músculos estudiados, se hace posible exportar el estudio a otras situaciones de trabajo diferentes de las estudiadas aquí, por ejemplo;

Como parte de un grupo musculoesquelético más amplio que la articulación del codo.

Como base para estudios de comportamiento muscular relacionado con la velocidad de la contracción.

Como base para estudios diferentes de fatiga o sobrecarga muscular.

Además, se ha desarrollado un método de análisis que no requiere de información anatómica compleja de adquirir, básicamente con las medidas antropométricas del



participante objeto de estudio es suficiente para llevar a cabo un estudio de su comportamiento.

Con el desarrollo de esta técnica de análisis, personal sanitario involucrado en el diagnóstico de afecciones de tipo muscular (médicos traumatólogos, fisioterapeutas, entrenadores deportivos, etc....) pueden llevar a cabo estudios complementarios a los habituales, encaminados a predecir dolencias o traumas que puedan producirse en el paciente.

A pesar de que el estudio está centrado en contracciones isométricas, constituye el primer paso de otros que puedan darse con posterioridad; el estudio de comportamiento de los músculos teniendo en cuenta la velocidad de contracción puede llevarse a cabo de modo relativamente sencillo, ampliar el número de músculos analizados con este método, una vez validado, requiere de la repetición de la misma metodología aplicada en esta investigación (modelo CAD, modelo FEM, análisis FEM y comparación con datos teóricos referidos al músculo considerado). Es por ello, que una librería de músculos caracterizados según este método puede ser incrementada en un espacio de tiempo relativamente pequeño.

Con dicha librería elaborada, el análisis de comportamiento de grupos musculares más complejos se llevaría a cabo rápidamente conjuntando la potencia de cálculo y simulación de los métodos de elementos finitos y la simplicidad de la expresión matemática que caracteriza al músculo.

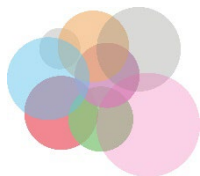
Comparando este trabajo de investigación con los realizados anteriormente, se observa que el enfoque de éste difiere de aquellos.

Los estudios realizados hasta el momento no permiten incluir sus conclusiones, de modo directo, en modelos de cálculo posteriores, los enfoques que se han dado en numerosos estudios sobre comportamiento muscular coinciden con lo indicado a continuación

Los estudios realizados hasta la fecha se han centrado en una posición articular, obteniendo resultados relacionados sobre el esfuerzo muscular que son difíciles de extrapolar a otras posiciones. [21], obtiene los resultados para el músculo tibial anterior bajo condiciones de estudio, pero trasladar las conclusiones del estudio a otras configuraciones es complicado y requiere repetir determinadas tareas para establecer el modelo matemático.

Las ecuaciones matemáticas que rigen el comportamiento del músculo son difíciles de implementar en herramientas de cálculo, [4] usa elementos 1D para simular músculos, pero no hay ecuaciones que puedan implementarse en otro análisis. Lo mismo sucede con Alonso et al. [43], donde el resultado se basa en una optimización estática y fisiológica de cualquier unidad MTU según el modelo de Hill.

Varios de estos estudios utilizan elementos 3D para representar el músculo, como ya se ha comentado anteriormente, el empleo de elementos 3D en el mallado de músculos permite una mejor comprensión del comportamiento interno del propio músculo (distribución de tensiones, deformaciones, etc.) pero no permite que el modelo creado sea reutilizado en otras posiciones de la unión. Ejemplos de esto son los trabajos realizados por Martínez [12] en el que se analiza el músculo tibial anterior en una determinada posición, debiendo generarse una nueva malla si la posición de la articulación cambia.



La presente investigación soluciona ambas limitaciones.

Las ecuaciones de comportamiento que rigen los músculos son lo suficientemente simples como para ser implementadas en un modelo FEM de la articulación.

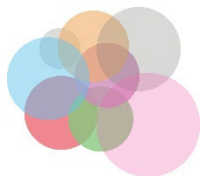
El modelo FEM generado puede reutilizarse en diferentes posiciones.

Los resultados no están vinculados a una posición concreta de la articulación, la misma puede modificarse en función de las necesidades y los resultados seguirán siendo válidos.

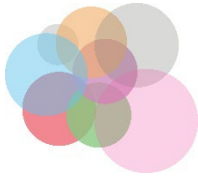
Los objetivos de la investigación pasan por obtener unas ecuaciones que puedan predecir la fuerza muscular, y más aún, obtener ecuaciones que puedan implementarse en un nuevo tipo de elementos finitos que caracteriza el músculo considerado para realizar análisis de la articulación del codo con mayor precisión, sin necesidad de ejecutar pruebas complejas con personas y con un tiempo de procesamiento reducido para obtener los resultados.

Las aplicaciones de este estudio son varias, análisis de esfuerzos musculares en tareas deportivas, análisis de esfuerzos musculares en actividades laborales, estudio de sobrecargas musculares, etc...

El estudio presentado en este artículo tiene la limitación de ser válido sólo para el análisis de esfuerzos isométricos en los músculos, así como la de que sólo es válido para la articulación del codo. Sin embargo, con la validación del método utilizado es posible realizar estudios similares en diferentes grupos articulares y así ampliar el número de músculos caracterizados, pudiendo, a medida que se realicen estos trabajos, generar un modelo FEM apto para analizar estructuras musculoesqueléticas más complejas.

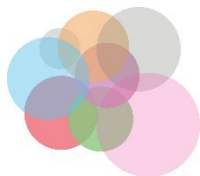


7. CONCLUSIONES



Una vez realizado el trabajo de investigación en el que se desarrolla la metodología necesaria para predecir el comportamiento de los músculos estudiados en la articulación del codo, y según los resultados obtenidos en la misma, validados a través de una prueba realizado con participantes, concluimos que:

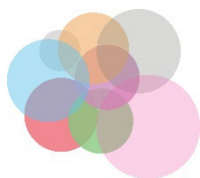
- **Primera.** Se ha logrado un modelo de elementos finitos de la articulación del codo reutilizable en diferentes posiciones mediante el empleo de elementos 1D, tipo CROD, para simular los músculos de una articulación.
- **Segunda.** Se han planteado las ecuaciones de comportamiento de los músculos flexores y extensores de la articulación, validos en cualquier ángulo que esta forme y para cualquier carga que puede aplicarse en la muñeca. Se puede predecir la fuerza necesaria en cada uno de los músculos en función de la posición que forme la articulación y la carga aplicada en la misma.
- **Tercera.** Se han obtenido los valores teóricos de fuerza máxima que los músculos flexores y extensores del codo pueden desarrollar en cualquier ángulo que forme la articulación. Se definen a través de ecuaciones características que permiten predecir la fuerza máxima que un músculo del grupo flexor y/o extensor es capaz de entregar en una posición concreta de la articulación.
- **Cuarta.** Se ha comprobado, mediante un test realizado con participantes, los valores de carga máxima que puede soportar una persona en la muñeca variando el ángulo que forme la articulación del codo.
- **Quinta.** Se han obtenido las expresiones matemáticas que definen la carga máxima en la muñeca que pueden soportar los músculos flexores o extensores de la articulación del codo.
- **Sexta.** Se ha correlado el comportamiento teórico con los resultados del ensayo realizado, comprobando cómo existe similitud de comportamiento entre el estudio y el ensayo.
- **Séptima.** Se ha desarrollado una metodología que permite extrapolar los resultados del estudio a diferentes personas, con diferente antropometría, mediante la toma de unas sencillas medidas sobre el paciente.
- **Octava.** Dado que se han obtenido expresiones que indican la fuerza máxima admisible por un músculo de la articulación del codo, y la fuerza necesaria en un músculo para un estado de carga, se puede definir un factor de utilización en los músculos anticipándonos al agotamiento del mismo.
- **Novena y última.** Las ecuaciones obtenidas para expresar la fuerza máxima admisible en cada músculo y la fuerza necesaria en función del estado de carga pueden incluirse en la programación de un nuevo tipo de elementos finitos que responda al comportamiento del músculo y permita resolver situaciones de carga en la articulación comprobando los márgenes de seguridad con los que la misma se encontraría.



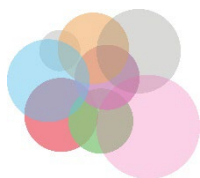
8. BIBLIOGRAFIA



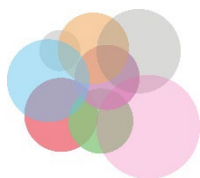
- [1] J. A. C. Martins, M. P. M. Pato, and E. B. Pires, "A finite element model of skeletal muscles," *Virtual Phys Prototyp*, vol. 1, no. 3, pp. 159–170, 2006, doi: 10.1080/17452750601040626.
- [2] D. Hodgson, "Modelling muscles in vitro: a finite element analysis study," 2015.
- [3] B. Scholarsarchive and M. S. Hortin, "Ligament Model Fidelity in Finite Element Analysis of the Human Lumbar Spine," 2015.
- [4] E. C. Teo, Q. H. Zhang, and T. X. Qiu, "Finite Element Analysis of Head-neck Kinematics Under Rear-end Impact Conditions."
- [5] W. A. Siswanto, C. C. Yoon, S. M. Salleh, M. Z. Ngali, and E. M. Yusup, "Computational Modelling and Movement Analysis of Hip Joint with Muscles," in *IOP Conference Series: Materials Science and Engineering*, Institute of Physics Publishing, Jan. 2017. doi: 10.1088/1757-899X/165/1/012030.
- [6] J. N. Parekh, "Using Finite Element Methods to study Anterior Cruciate Ligament Injuries: Understanding the role of ACL modulus and Tibial Surface Geometry on ACL loading."
- [7] J. A. Weiss, J. C. Gardiner, B. J. Ellis, T. J. Lujan, and N. S. Phatak, "Three-dimensional finite element modeling of ligaments: Technical aspects," *Med Eng Phys*, vol. 27, no. 10, pp. 845–861, Dec. 2005, doi: 10.1016/j.medengphy.2005.05.006.
- [8] L. A. Spyrou and N. Aravas, "Muscle-driven finite element simulation of human foot movements," *Comput Methods Biomech Biomed Engin*, vol. 15, no. 9, pp. 925–934, Sep. 2012, doi: 10.1080/10255842.2011.566564.
- [9] C. Y. Tang, C. P. Tsui, B. Stojanovic, and M. Kojic, "Finite element modelling of skeletal muscles coupled with fatigue," *Int J Mech Sci*, vol. 49, no. 10, pp. 1179–1191, Oct. 2007, doi: 10.1016/j.ijmecsci.2007.02.002.
- [10] E. C. Teo, Q. H. Zhang, and T. X. Qiu, "Finite Element Analysis of Head-neck Kinematics Under Rear-end Impact Conditions," 2006.
- [11] E. Lechosa Urquijo, F. Blaya Haro, J. D. Cano-Moreno, R. D'Amato, and J. A. Juanes Méndez, "Mechanical Model and FEM Simulations for Efforts on Biceps and Triceps Muscles under Vertical Load: Mathematical Formulation of Results," *Mathematics*, vol. 10, no. 14, Jul. 2022, doi: 10.3390/math10142441.
- [12] A. Ruggiero, S. Affatato, M. Merola, and M. C. De Simone, "FEM analysis of metal on UHMWPE total hip prosthesis during normal walking cycle," in *AIMETA 2017 - Proceedings of the 23rd Conference of the Italian Association of Theoretical and Applied Mechanics*, 2017.
- [13] M. Zheng, Z. Zou, P. jorge D. silva Bartolo, C. Peach, and L. Ren, "Finite element models of the human shoulder complex: a review of their clinical implications and modelling techniques," 2017. doi: 10.1002/cnm.2777.
- [14] N. Seki *et al.*, "Mechanical environment of the supraspinatus tendon: Three-dimensional finite element model analysis," *Journal of Orthopaedic Science*, vol. 13, no. 4, 2008, doi: 10.1007/s00776-008-1240-8.



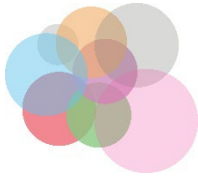
- [15] B. Seral Garcia, J. Cegoñino Banzo, J. M. Garcia Aznar, M. Doblaré Castellano, and F. Seral Iñigo, "Simulación en 3D con elementos finitos de un modelo de prótesis de rodilla," in *Revista de Ortopedia y Traumatologia*, Jan. 2003, pp. 64–72. doi: 10.1016/s1888-4415(03)76072-x.
- [16] J. C. Carr, X. Ye, and H. M. Tharp, "Biceps Brachii Muscle Length Affects Force Steadiness with and Without Fatigue," *Journal of Science in Sport and Exercise*, vol. 3, no. 4, pp. 364–373, Nov. 2021, doi: 10.1007/s42978-020-00102-0.
- [17] N. A. Z. Abidin, M. Rafiq Abdul Kadir, and M. H. Ramlee, "Three Dimensional Finite Element Modelling and Analysis of Human Knee Joint-Model Verification," in *Journal of Physics: Conference Series*, Institute of Physics Publishing, Nov. 2019. doi: 10.1088/1742-6596/1372/1/012068.
- [18] E. C. Teo, Q. H. Zhang, and T. X. Qiu, "Finite Element Analysis of Head-neck Kinematics Under Rear-end Impact Conditions."
- [19] M. I. Marcos, R. D'Amato, E. L. Urquijo, J. A. Juanes, F. B. Haro, and E. S. Heras, "Finite Element Simulation and Analysis of the behavior under load of a human shoulder," in *ACM International Conference Proceeding Series*, Association for Computing Machinery, Oct. 2018, pp. 454–461. doi: 10.1145/3284179.3284256.
- [20] T. L. H. Donahue, M. L. Hull, M. M. Rashid, and C. R. Jacobs, "A finite element model of the human knee joint for the study of tibio-femoral contact," *J Biomech Eng*, vol. 124, no. 3, pp. 273–280, 2002, doi: 10.1115/1.1470171.
- [21] Ramírez Martínez Angélica María, "Modelado y simulación del tejido músculo-esquelético. Validación experimental con el músculo tibial anterior de rata," Zaragoza, May 2011.
- [22] A. Samani, J. Bishop, M. J. Yaffe, and D. B. Plewes, "Biomechanical 3-D Finite Element Modeling of the Human Breast Using MRI Data," 2001.
- [23] J. M. Winters, "Hill-Based Muscle Models: A Systems Engineering Perspective," in *Multiple Muscle Systems*, Springer New York, 1990, pp. 69–93. doi: 10.1007/978-1-4613-9030-5_5.
- [24] "The variation in isometric tension with sarcomere length in vertebrate muscle fibres".
- [25] I.A.Kapandji, *Fisiología articular: esquemas comentados de mecánica humana*. 2009.
- [26] M. Islan, R. D'Amato, J. Carvajal, J. A. Juanes, P. S. Pedro, and E. Soriano, "Linear approximation of the behavior of the rotator cuff under fatigue conditions. violinist case study," in *ACM International Conference Proceeding Series*, Association for Computing Machinery, Oct. 2017. doi: 10.1145/3144826.3145408.
- [27] J. Grasa, B. Hernández-Gascón, A. Ramírez, J. F. Rodríguez, and B. Calvo, "Modelado numérico del comportamiento del tejido músculo- esquelético," *Revista Internacional de Metodos Numericos para Calculo y Diseno en Ingenieria*, vol. 28, no. 3, pp. 177–186, 2012, doi: 10.1016/j.rimni.2012.03.005.
- [28] *NATIONAL HEALTH AND NUTRITION EXAMINATION SURVEY* . Westat, Inc. 1650 Research Boulevard Rockville, 1988.



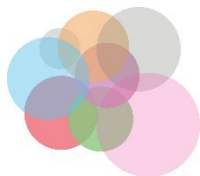
- [29] C. Kleinbach, O. Martynenko, J. Promies, D. F. B. Haeufle, J. Fehr, and S. Schmitt, "Implementation and validation of the extended Hill-type muscle model with robust routing capabilities in LS-DYNA for active human body models," *Biomed Eng Online*, vol. 16, no. 1, Sep. 2017, doi: 10.1186/s12938-017-0399-7.
- [30] K. M. Wisdom, S. L. Delp, and E. Kuhl, "Use it or lose it: multiscale skeletal muscle adaptation to mechanical stimuli," Apr. 01, 2015, *Springer Verlag*. doi: 10.1007/s10237-014-0607-3.
- [31] K. R. S. Holzbaur, W. M. Murray, and S. L. Delp, "A model of the upper extremity for simulating músculoskeletal surgery and analyzing neuromuscular control," *Ann Biomed Eng*, vol. 33, no. 6, pp. 829–840, Jun. 2005, doi: 10.1007/s10439-005-3320-7.
- [32] Y. Kawakami, K. Nakazawa, T. Fujimoto, D. Nozaki, M. Miyashita, and T. Fukunaga, "Specific tension of elbow flexor and extensor muscles based on magnetic resonance imaging," *Eur J Appl Physiol Occup Physiol*, vol. 68, no. 2, pp. 139–147, Feb. 1994, doi: 10.1007/BF00244027.
- [33] Sobotta, *Atlas of Human Anatomy*, 15th ed., vol. 1. 2013.
- [34] Cheryl D. Fryar, Margaret D. Carroll, M. S. Joseph Afful, Cynthia L. Ogden, and Qiuping Gu, "Anthropometric Reference Data for Children and Adults," Jan. 2021.
- [35] Á. Carbajal Azcona, "Manual de Nutrición y Dietética," *Facultad de farmacia. Universidad Complutense de Madrid*.
- [36] A. D. Aydin Kabakci, M. Buyukmumcu, M. T. Yilmaz, A. E. Cicekcibasi, D. Akin, and E. Cihan, "Estudio Osteométrico del Húmero," *International Journal of Morphology*, vol. 35, no. 1, pp. 219–226, 2017, doi: 10.4067/S0717-95022017000100036.
- [37] M. Islan, F. Blaya, P. S. Pedro, R. D'Amato, E. L. Urquijo, and J. A. Juanes, "Analysis and Fem Simulation Methodology of Dynamic Behavior of Human Rotator Cuff in Repetitive Routines: Musician Case Study," *J Med Syst*, vol. 42, no. 3, pp. 1–10, Mar. 2018, doi: 10.1007/S10916-018-0908-7/FIGURES/14.
- [38] M. I. Marcos, R. D'Amato, E. L. Urquijo, J. A. Juanes, F. B. Haro, and E. S. Heras, "Finite Element Simulation and Analysis of the behavior under load of a human shoulder," *ACM International Conference Proceeding Series*, pp. 454–461, Oct. 2018, doi: 10.1145/3284179.3284256.
- [39] M. Islán Marcos, E. Lechosa Urquijo, F. Blaya Haro, R. D'Amato, E. Soriano Heras, and J. A. Juanes, "Behavior under Load of A Human Shoulder: Finite Element Simulation and Analysis," *J Med Syst*, vol. 43, no. 5, pp. 1–18, May 2019, doi: 10.1007/S10916-019-1248-Y/FIGURES/31.
- [40] E. Lechosa, R. D'Amato, F. B. Haro, and J. A. Juanes, "1D Finite Element model using rod element applied to muscle behavior simulation," *ACM International Conference Proceeding Series*, pp. 411–415, Oct. 2019, doi: 10.1145/3362789.3362868.
- [41] E. Lechosa Urquijo, F. Blaya Haro, R. D'Amato, and J. A. Juanes Méndez, "Finite Element model of an elbow under load, muscle effort analysis when modeled using 1D rod



- element," *ACM International Conference Proceeding Series*, vol. 8, no. 20, pp. 475–482, Oct. 2020, doi: 10.1145/3434780.3436612.
- [42] L. Urquijo *et al.*, "Mechanical Model and FEM Simulations for Efforts on Biceps and Triceps Muscles under Vertical Load: Mathematical Formulation of Results," *Mathematics 2022*, Vol. 10, Page 2441, vol. 10, no. 14, p. 2441, Jul. 2022, doi: 10.3390/MATH10142441.
- [43] F. J. Alonso, G. Galan-Marin, D. R. Salgado, and R. Pàmies-Vilà, "Cálculo de esfuerzos musculares en la marcha humana mediante optimización estática-fisiológica." [Online]. Available: <https://www.researchgate.net/publication/267258732>
- [44] W. Il Park, H. D. Lee, and J. Kim, "Estimation of isometric joint torque from muscle activation and length in intrinsic hand muscle," in *2008 International Conference on Control, Automation and Systems, ICCAS 2008*, 2008, pp. 2489–2493. doi: 10.1109/ICCAS.2008.4694273.
- [45] O. A. Sachenkov, R. F. Hasanov, P. S. Andreev, and Y. G. Konoplev, "Numerical study of stress-strain state of pelvis at the proximal femur rotation osteotomy," *Russian Journal of Biomechanics*, vol. 20, no. 3, pp. 220–232, 2016, doi: 10.15593/RJBiomech/2016.3.06.
- [46] F. A. Syomin and A. K. Tsaturyan, "Mechanical model of the left ventricle of the heart approximated by axisymmetric geometry," *Russian Journal of Numerical Analysis and Mathematical Modelling*, vol. 32, no. 5, pp. 327–337, Oct. 2017, doi: 10.1515/rnam-2017-0031.
- [47] Jefferson. Fangudes Loss and C. Tarragó Candoti, "Comparative study between two elbow flexion exercises using the estimated resultant muscle orce.," *Revista Brasileira de Fisioterapia*, 2008, doi: 10.1590/S1413-35552008000600012.
- [48] W. M. Murray, S. L. Delp, and T. S. Buchanan, "VARIATION OF MUSCLE MOMENT ARMS WITH ELBOW AND FOREARM POSITION," 1995.
- [49] J. S. Leedham and J. J. Dowling, "Force-length, torque-angle and EMG-joint angle relationships of the human in vivo biceps brachii," 1995.
- [50] W. T. V. Chu and T. D. Sanger, "Force variability during isometric biceps contraction in children with secondary dystonia dueto cerebral palsy," *Movement Disorders*, vol. 24, no. 9, pp. 1299–1305, Jul. 2009, doi: 10.1002/mds.22573.
- [51] A. L. Hof and J. Van den Berg, "EMG to force processing II: Estimation of parameters of the Hill muscle model for the human triceps surae by means of a calfergometer," *J Biomech*, vol. 14, no. 11, 1981, doi: 10.1016/0021-9290(81)90032-4.
- [52] E. Sarabia Marañón, "Efectos de la inclusión de máquina isoinercial en el tratamiento del hombro del nadador con gomas elásticas. Ensayo aleatorizado controlado doble ciego," 2014.
- [53] H. B. Sharma, A. Das, P. Tayade, and K. K. Deepak, "Recording of length-tension relationship of elbow flexors and extensors by varying elbow angle in human," *Indian J Physiol Pharmacol*, vol. 64, no. 5, pp. S46–S50, 2020, doi: 10.25259/IJPP_285_2020.



-
- [54] H. Y. Choi, "Experimental and numerical studies of muscular activations of bracing occupant," 2005. [Online]. Available: <https://www.researchgate.net/publication/238729776>
- [55] C. Pontonnier and G. Dumont, "Inverse dynamics method using optimization techniques for the estimation of muscles forces involved in the elbow motion," *International Journal on Interactive Design and Manufacturing*, vol. 3, no. 4, pp. 227–236, 2009, doi: 10.1007/s12008-009-0078-4.
- [56] Alejandro Zar Casás, "ESTIMACIÓN DE FUERZAS MUSCULARES DURANTE EL MOVIMIENTO DEL BRAZO HUMANO MEDIANTE OPTIMIZACIÓN ESTÁTICO-FISIOLÓGICA," 2016.
- [57] D. Panariello, S. Grazioso, T. Caporaso, A. Palomba, G. Di Gironimo, and A. Lanzotti, "Biomechanical analysis of the upper body during overhead industrial tasks using electromyography and motion capture integrated with digital human models," *International Journal on Interactive Design and Manufacturing*, vol. 16, no. 2, pp. 733–752, Jun. 2022, doi: 10.1007/s12008-022-00862-9.



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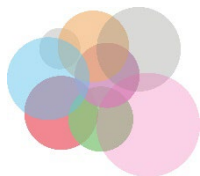


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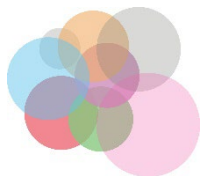
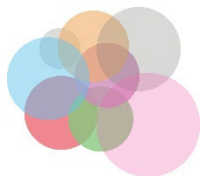
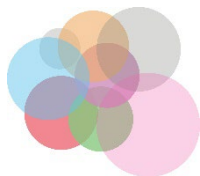


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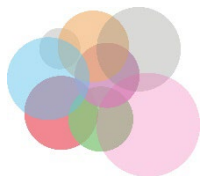


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11. ANEXOS



ANEXO I:

Lechosa, E., D'Amato, R., Haro, F. B., & Juanes, J. A. (2019). 1D Finite Element model using rod element applied to muscle behavior simulation. *ACM International Conference Proceeding Series*, 411–415. <https://doi.org/10.1145/3362789.3362868>

ANEXO II:

Lechosa Urquijo, E., Blaya Haro, F., D'Amato, R., & Juanes Méndez, J. A. (2020). Finite Element model of an elbow under load, muscle effort analysis when modeled using 1D rod element. *ACM International Conference Proceeding Series*, 8(20), 475–482. <https://doi.org/10.1145/3434780.3436612>

ANEXO III:

Marcos, M. I., D'Amato, R., Urquijo, E. L., Juanes, J. A., Haro, F. B., & Heras, E. S. (2018). Finite Element Simulation and Analysis of the behavior under load of a human shoulder. *ACM International Conference Proceeding Series*, 454–461. <https://doi.org/10.1145/3284179.3284256>

ANEXO IV:

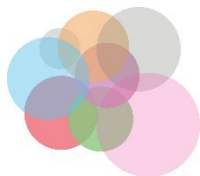
Islan, M., Blaya, F., Pedro, P. S., D'Amato, R., Urquijo, E. L., & Juanes, J. A. (2018). Analysis and FEM Simulation Methodology of Dynamic Behavior of Human Rotator Cuff in Repetitive Routines: Musician Case Study. *Journal of Medical Systems*, 42(3), 55. <https://doi.org/10.1007/s10916-018-0908-7>

ANEXO V:

Islán Marcos, M., Lechosa Urquijo, E., Blaya Haro, F., D'Amato, R., Soriano Heras, E., & Juanes, J. A. (2019). Behavior under Load of A Human Shoulder: Finite Element Simulation and Analysis. *Journal of Medical Systems*, 43(5), 132. <https://doi.org/10.1007/s10916-019-1248-y>

ANEXO VI:

Lechosa Urquijo, E., Blaya Haro, F., David Cano-Moreno, J., D'Amato, R., & Antonio Juanes Méndez, J. (2022b). Mechanical Model and FEM Simulations for Efforts on Biceps and Triceps Muscles under Vertical Load: Mathematical Formulation of Results. *Mathematics* 2022, Vol. 10, Page 2441, 10(14), 2441. <https://doi.org/10.3390/MATH10142441>



11.1.ANEXO I

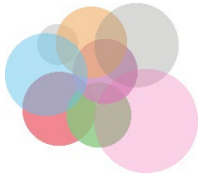
FINITE ELEMENT SIMULATION AND ANALYSIS OF THE BEHAVIOR UNDER LOAD OF A HUMAN SHOULDER

La mayoría de las lesiones musculoesqueléticas localizadas en el hombro suelen deberse a movimientos repetitivos o sostenidos que ocurren en las rutinas de trabajo en diferentes áreas.

En el caso de los trabajadores del sector de la construcción, la rutina repetitiva y los movimientos bajo carga producen desplazamientos y posturas antinaturales de la articulación del hombro. Por este motivo, el objetivo de este estudio es modelar en 3D los elementos biológicos que forman la articulación del hombro para su posterior Análisis de Elementos Finitos (FEA).

En este primer estudio se han considerado tres casos de cargas diferentes. Con el análisis desarrollado, y gracias al abordaje articular lineal e isotrópico, se han podido evaluar las tensiones en los elementos: músculos, tendones y ligamentos.

La metodología utilizada permitió obtener una malla mejorada de la articulación del hombro para analizar situaciones reales con análisis FEM con aplicaciones en el campo de la medicina deportiva, laboral, etc. Además, la simplificación adoptada para el modelado de los músculos articulares, como elementos 1D en FEM, ha permitido establecer diferentes posiciones de la articulación humana sin tener que volver a mallar cada una de las posiciones que se pueden estudiar. Los resultados son consistentes con los estados de cargas aplicadas.



Finite Element Simulation and Analysis of the behavior under load of a human shoulder*

Extended Abstract[†]

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ABSTRACT

Majority¹ of musculoskeletal injuries located in the shoulder are often due to repetitive or sustained movements which occur in work routines in different areas. In the case of workers in the construction sector, repetitive routine and movements under load produce displacements and unnatural postures of the shoulder joint. For this reason, the objective of this study is to model in 3D the biological elements that form the shoulder joint for subsequent Finite Element Analysis (FEA). Three cases of different loads have been considered in this first study. With the analysis developed, and thanks to the linear and isotropic joint approach has been possible to evaluate the stresses in the elements: muscles, tendons and ligaments. The methodology used allowed to obtain an improved mesh of the shoulder joint to analyze real situations with FEM analysis with applications in the field of sports medicine, work, etc. Furthermore, the simplification adopted for the modeling of the joint muscles, as 1D elements in FEM, has made it possible to establish different positions of the human joint without having mesh again each one of the positions that can be studied. The results are consistent with the states of applied loads.

KEYWORDS: FEM analysis, Shoulder, Humerus, Muscles, Tendons, Ligaments.

1 INTRODUCTION

Due to its high mobility, the shoulder is the Human joint that has the most injuries in its tissues [1].

The rotator cuff is formed by the tendons of the subscapular, supraspinatus, infraspinatus, and minor rotator cuff muscles that meet at the shoulder joint [2], [3]. It is located in the arch called coracoacromial, constituting the subacromial space with a serous pouch that lubricates the cuff. This pathology is characterized by progressive pain as a result of an accumulated overload or acute after an overexertion, which is located on the upper or lateral side of the shoulder and increases with movement, mainly with the elevation of the arm (ex. When combing) and with internal rotation (e.g. when putting on a jacket) [4]. The pain may be more acute at night when leaning on the affected side. The chronic injury can cause a rupture of the tendons of the cuff. Chronic tendinitis of the rotator cuff may be related to shoulder overload, joint instability and rotator cuff degeneration with age. It is relatively frequent, reaching up to 18% in workers in heavy manual tasks with great efforts on the shoulders, associated or not with repetitive movements and forced postures. 75% of people over 50 years have rotator cuff injuries and most of them are asymptomatic [5]. Many jobs in different trades in the construction sector involve important musculoskeletal efforts. This pathology is associated with work that involves stabilizing the shoulders to maintain or move the elbows in an elevated position by force, lifting and catching continuously, or working the arm in flexion or abduction. In the construction sector, this occurs in the assembly of metallic or prefabricated structures, also of heavy prefabricated elements that must be manipulated to unload. In order to understand the behavior of human joints, in recent years, several authors have proposed scientific studies with Finite Element Model (FEM) [6] as well as the studies conducted to simulate the dynamic behavior of human

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joints [7]–[9]. Many authors, in recent years, have focused their studies on the description of the dynamic behavior of the shoulder joint. For example, in the 1998 Luo et al. present for the first time a simplified 2D-FEM in order to describe the stress environment in the supraspinatus tendon [10]. Wakabayashi et al. in the 2003 performed 2D FEM analysis in order to estimate the mechanical environment of the supraspinatus tendon. In the 2008 Seki et al. [11] presented the first 3D FE model of rotator cuff tears in order to study the stress distribution in the supraspinatus tendon. Most of these studies have a substantial limitation. In fact, in these studies only the scapula and the bones of the Humerus were considered. This means that in presenting their results both in the 2D and 3D geometry, the structure of the muscles and other soft tissues have been neglected [12]. For this reason, in this study a FEM of the shoulder joint, glenohumeral, classified as a diarthrosis, of the genus enarthrosis or spheroid is presented. Based on the FEM made for the case of a violinist [13] they are carried out modifications and improvements that allow us increasing the accuracy of the study and improving the predictive performance of it. This is due to include the ligaments, glenohumeral and coracohumeral, and the glenohumeral-capsule in the anatomical model. In the same way the meshing process is improved, by working on the characterization of the tendon-bone insertion, achieving a transfer of efforts between both more precise and in accordance with reality.

The study focuses on a unique position shoulder, flexion 53°. The scope of work aims the following aspects:

- Improving the mesh used in the finite element model.
- Improving the characterization of the contact tendon-bone insertion area. To do this, meshes between both elements have been made congruent, with nodes coinciding on the interface surface.
- Including the main ligaments involved in the joint, glenohumeral ligaments (lower, middle and upper) and coracohumeral ligament.
- Obtaining a qualitative model of joint behavior.

2 METHODOLOGY

Starting from the cloud of points (Fig. 1) obtained through a resonance. A 3D Computer Aided Design (CAD) model of the elements involved in the articulation, scapula, Humerus, tendons and muscles is generated.

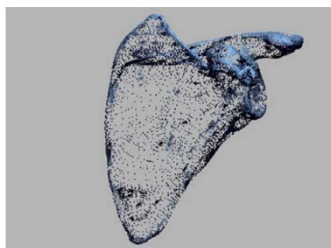


Figure 1: Cloud of points corresponding to the scapula

The reconstruction of the models of the bones and tendons involved in the joint is carried out using Geomagic Design X software.

This software is capable of generating a solid model, or surfaces, based on a cloud of points (Fig. 2).

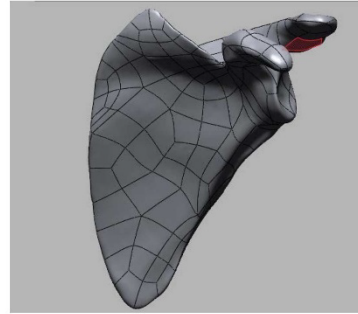


Figure 2: Cloud of points corresponding to the scapula

To verify that the interfaces between tendons and bones are perfectly matched, a 3D CAD, NX tool from Siemens has been used.

The basis of the meshing process in these areas is the coincidence of the meshes between both surfaces in contact. To do this, a modeling work on the tendons that ends with a Boolean operation has been carried out.

This operation produces the subtraction of volume between the tendon and the bone, thus achieving a matching surface between both elements (Figs. 3 and 4).

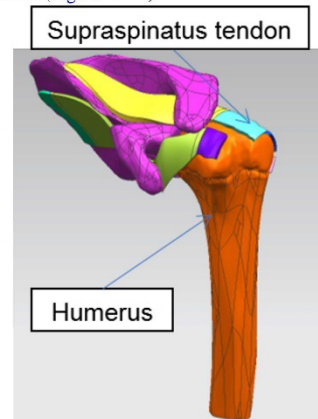
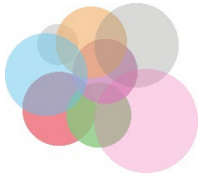


Figure 3: Location of the supraspinatus tendon and the head of the humerus.



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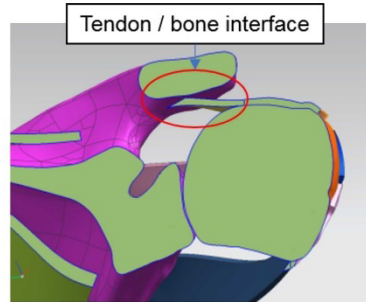


Figure 4: Sectional view of the humerus + supraspinatus tendon

With the solid models of the articulation elements, a file is generated "*.stp" or "*.igs". These files are imported into a FEM software (siemens FEMAP). As in the case described before, it must be ensured matching surfaces between tendon and bone (Fig. 5). In this way both surfaces will be coincident in geometry (achieved with the Boolean operation in the CAD 3D software) and also in the contour of them (Fig. 6). Once verified that the geometries between bone and tendon are coincident, the model mesh is generated. Table 1 summarizes the elements used to model the parts which the human shoulder joint consists of.

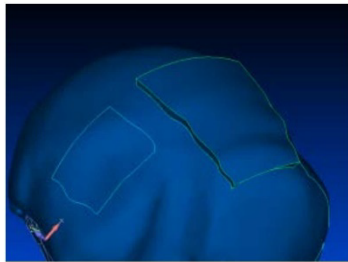


Figure 5: View of the Humerus + supraspinatus tendon.

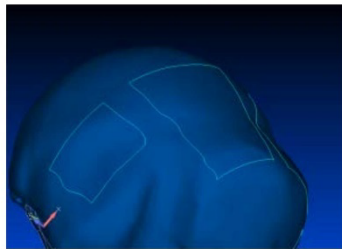


Figure 6: Outline of the tendon projected on the Humerus.

Table 1: Frequency of Special Characters

N ^o	Element type	Parameters	Application
1	tetrahedral 10 nodes	Material, Young modulus and Poisson coef.	Meshing bones, muscles and tendons
2	beam element 2 nodes	Material, Young modulus and Poisson coef. and transverse section	Mesh of the ligaments and the glenohumeral capsule.
3	rigid element 2 nodes	none	Simulation insertion of the ligament in the bone.
4	gap element 2 nodes	Gap and stiffness of the contact.	Simulation of the cartilage contact between the head of the Humerus and the scapula.

Once the mesh is generated separately from the muscles, tendons, bones and ligaments, the next step is the fusion of the matching nodes to generate the complete model. As an example, the case of the interface between bone and tendon is of special importance (Figs. 7 and 8). In fact, with this fusion it is possible to transfer the forces between both elements through matching nodes between both surfaces (Fig. 9). This avoids modeling contacts between them.

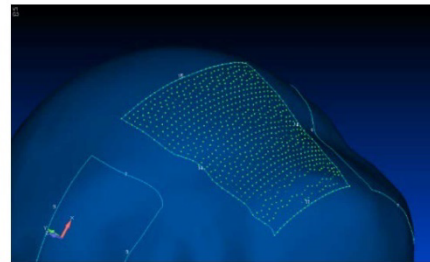


Figure 7: Nodes located on the tendon-bone interface surface corresponding to the Humerus.

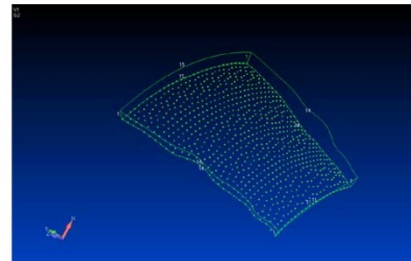


Figure 8: Nodes located on the tendon-bone interface surface corresponding to the tendon.



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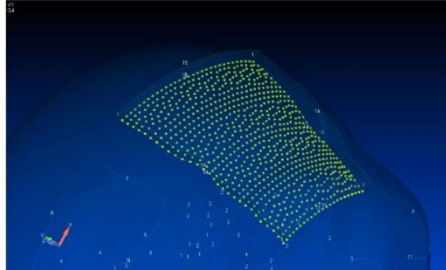


Figure 9: Nodes fused between the two coinciding surfaces of the tendon and the Humerus.

To improve the response of the FEM proposed, the ligaments that form part of the human shoulder joint have been considered. These ligaments are:

- Superior glenohumeral ligament: medially inserted into the top of the glenoid labrum and neck of the scapula and extends to the base of the coracoid process.
- Middle glenohumeral ligament: it is inserted medially in the glenoid rim and in the neighboring part of the neck of the scapula, to end in the lower part of the minor tubercle, along the insertion of the subscapularis tendon.
- Lower glenohumeral ligament: reinforces the anteroinferior part of the joint capsule. It is inserted into the glenoid rim and reaches to the lower part of the lower tubercle of the humerus.
- Coracohumeral ligament: thick fibrous lamina, which connects the humeral head with the coracoid process of the scapula, which generates tension and stability when the arm is on the side of the body. It merges with the capsule of the shoulder joint.

Similarly, it has been included in the model the glenohumeral capsule wrapping the joint formed between the humerus and the scapula.

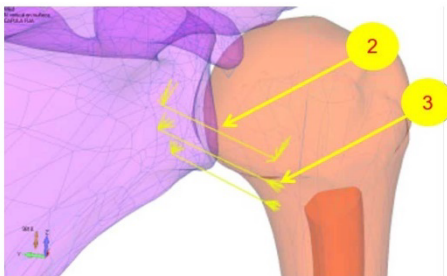


Figure 10: Rigid element (3) and element ID type Beam (2) to simulate ligaments of the joint (Table 1).

For the ligaments, 1D beam elements have been used, where the section of the element is that of the real ligament (Fig. 10). The use of rigid elements to simulate the insertion of the ligament in the bone allows to eliminate the concentration of tensions that would occur if the insertion were made in a single node (Fig. 10).

Table 2: Materials used for the model.

	Young's Modulus (Mpa)	Poisson	Density (Kg/m3)
Bone	17200	0,41	1790
Muscle	0,3	0,43	1350
Tendon	2000	0,43	1350
Cartilage	8	0,49	1150
Ligament	300	0,43	1350

The materials used for the model are summarized in Table 2. In this study three load cases have been considered (Table 3).

Table 3: Load cases considered for FEA.

Load Case	Load Conditions	Constrain Conditions
1	Gravity applied negative Z axis (9.841 m/s ²)	Fixed scapula on areas shown in Fig. 13
2	Gravity applied along negative Z axis (9.841 m/s ²) + 50 N applied along positive Y axis.	Fixed scapula on areas shown in Fig. 11
3	Gravity applied along negative Z axis (9.841 m/s ²) + 50 N applied along negative Z axis.	Fixed scapula on areas shown in Fig. 12

For the proposed FEM has considered the effect of gravity. This has been applied as shown in Fig. 13.

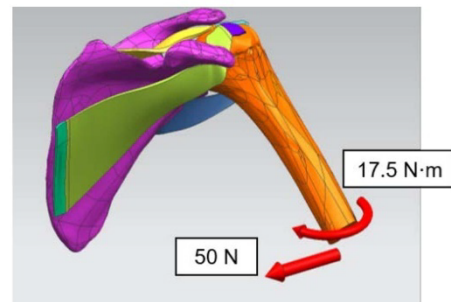
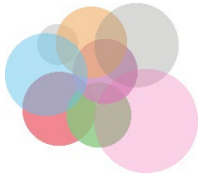


Figure 11: Load case 1 used in FEM.



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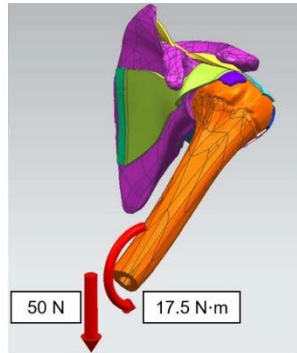


Figure 12: Load case 2 used in FEM.

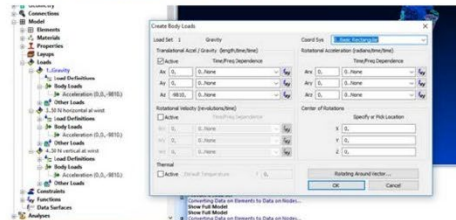


Figure 13: Effect of gravity applied to FEM.

For the boundary conditions, the scapula was considered fixed in the area where the muscles of the scapula would be inserted. These muscles are (Fig.14):

1. Levator scapula muscle.
2. Rhomboid minor muscle.
3. Rhomboid major muscle.

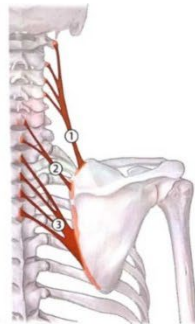


Figure 14: Muscles attached to the scapula fixed.

Figs. 15 and 16 show the areas of the scapula without displacements applied to the developed FEM. Fig. 15 shows the scapula in the front view and figure 16 in the rear view.

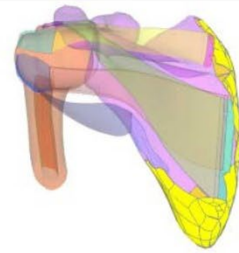


Figure 15: Areas of the scapula without displacement, in yellow

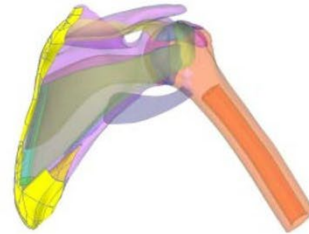


Figure 16: Areas of the scapula without displacement, in yellow

3 RESULTS AND DISCUSSION

This section shows the results of the meshed geometric model and the FEA implemented in this study.

3.1 Computer Aided Design (CAD) Model

Figs. 17 and 18 show the CAD model obtained from the clouds of points.

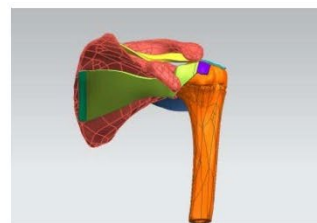


Figure 17: CAD model obtained from the clouds of points (frontal view).



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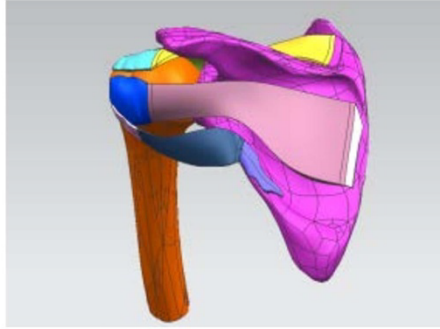


Figure 18: CAD model obtained from the clouds of points (rear view).

From this CAD model, the corresponding mesh has been generated for the FEA, in which all the elements mentioned in Table 1 are included.

As shown in Fig. 19, the generated mesh is suitable for carrying out the FEA. The combination of 3D elements and 1D elements used meets the objectives of the study. In fact, the main objective is the adequacy of the mesh so that the FEM simulation reproduces the behavior of the ligaments in the model. The stresses that appear in bones, muscles, cartilage area and insertions of the ligaments are consistent with the expected result.

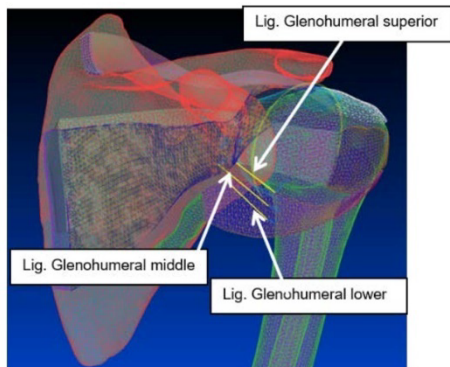


Figure 19: Detail of glenohumeral ligaments modeled as beam elements, in yellow.

3.2 Finite Element Analysis (FEA)

The results shown in Figs. 20 to 22 represent the stresses and displacements in the articulation under study by color coding. The colors vary from the maximum value represented in red to the minimum value represented in blue.

3.2.1 Stresses in bones. As shown in the figures of the three load cases studied (Figs. 20 to 22, section a and b), the maximum stresses in bones are in the insertion areas of the ligaments. This makes physical sense since ligaments are responsible for keeping the joint in position. Therefore, the ligaments transfer the necessary efforts to the area of insertion of the ligaments in the bones, causing these points to be the most stressed.

3.2.2 Stresses in muscles. Due to the static positions of the joint under study, the muscles do not support high stresses (Fig. 20-22 sections c). According to the stresses distribution, the maximum values are in the zones of tendons (Figs. 20 to 22, section d).

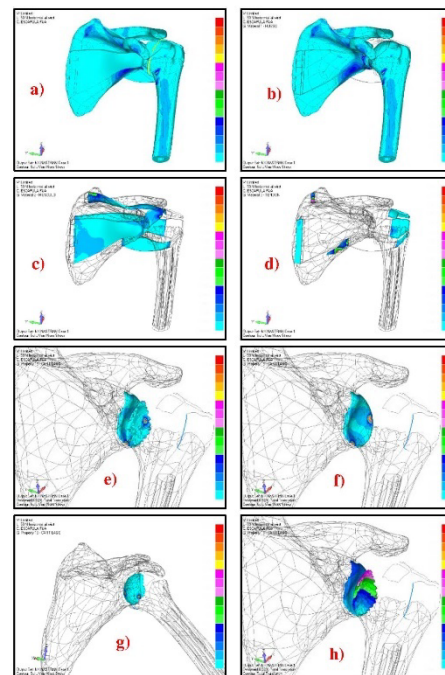


Figure 20: Load case 1, FEM results: a) Stresses in complete model; b) Maximum stresses in bones; c) Stresses in muscles; d) Stresses in tendons; e) Stresses in cartilage zone, scapula-Humerus; f) Stresses in cartilage, only Humerus; g) Stresses in cartilage zone, only scapula; h) Displacements in cartilage zone.

This situation has great physical sense since tendons are responsible for transmitting efforts of muscles to the bone structure while maintaining the position of the joint.



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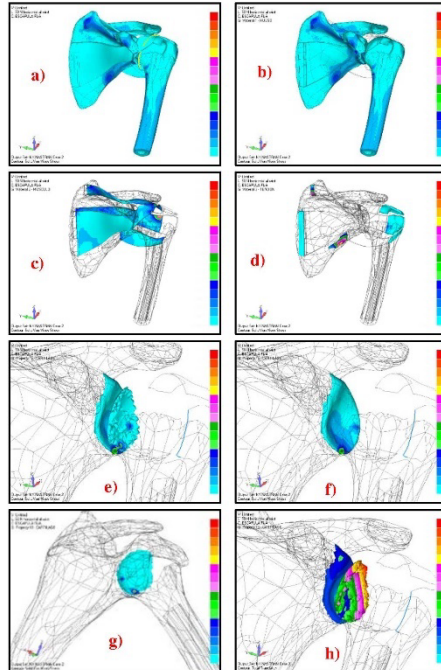


Figure 21: Load case 2, FEM results: a) Stresses in complete model; b) Maximum stresses in bones; c) Stresses in muscles; d) Stresses in tendons; e) Stresses in cartilage zone, scapula-Humerus; f) Stresses in cartilage, only Humerus; g) Stresses in cartilage zone, only scapula; h) Displacements in cartilage zone.

3.2.2 Stresses in cartilage. In the cartilage zone (Fig. 20-22, section e, f and g) a uniform and low value stresses distribution is observed. However, points with high stress concentration does not have a great physical sense, they are due to the alignment of the gap element. As the gap element, behaves like a compression spring, the stresses that appear in it are proportional to the displacement that occurs in this kind of element (Figs. 20 to 22, section h). Observing the images of the displacements that occur around the cartilage (Figs. 20 to 22, section h) it is verified how they adjust to the expected, producing the displacement of the head of the Humerus. This displacement moves the head away from the Humerus of the scapula, which in comparison remains with very low values. However, if the gap element alignment, in relation to the displacement that occurs, is not correct, high stresses may appear that do not correspond to reality. Despite this, it is observed how the stresses distribution in the cartilage area maintains

coherence with reality since the maximum stresses appear in the lower half of the cartilage. In this area in which the Humerus-scapula contact is greater, the compression tensions are greater.

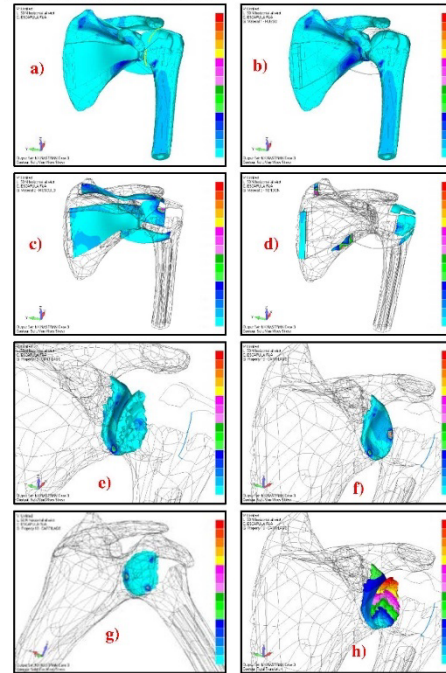


Figure 22: Load case 3, FEM results: a) Stresses in complete model; b) Maximum stresses in bones; c) Stresses in muscles; d) Stresses in tendons; e) Stresses in cartilage zone, scapula-Humerus; f) Stresses in cartilage, only Humerus; g) Stresses in cartilage zone, only scapula; h) Displacements in cartilage zone.

4 CONCLUSIONS

This study validates the FEM developed presenting a first qualitative study of the performances of the joint of the human shoulder. The proposed analysis meets the stated objective:

- Obtaining an improved mesh of the shoulder joint to analyze real situations with application in the field of sports medicine, work, etc. For these reason the follow improvements have been carried out:
 - Meshing of the joint, bone, tendons and muscles as 3D elements.
 - Improved mesh in the insertion of tendon-bone.



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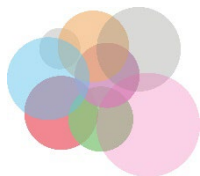
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- Congruence between the matching meshes of the tendon insertion area with the bone.
 - Including ligaments as 1D elements, node to node.
 - b) Obtaining congruent and logical results of the state of load of the articulation in the three proposed cases.
 - Simplifying the modeling of muscles. It is of great interest model muscles simplified to 1D elements. It makes possible to establish different positions in the articulation without having to mesh again each one of the positions that can be studied.
- This improvement must consider:
- Young's modulus of muscle is not linear. This depends on the load to which the muscle is subjected and its length
 - Muscle geometry (cross-section) varies depending on the position in which it is located.
 - Modifying the model used to simulate the ligaments considering that the strain-stress curve of them is not linear along its entire length. This involves using 1D elements to adapt its Young's modulus to the length having the ligament at each time.
 - Improving simulation at the contact zone cartilage-bones of the joint. It must be investigated about the possible improvement of the alignment of the gap elements.

The final objective of the study is to obtain an FEM to implement different positions of the same joint without modifying the initial mesh. This requires use 1D elements for the anatomical elements that deform, ligaments and muscles, and 3D elements for bones and tendons involved in the joint.

REFERENCES

- [1] M. P. Brunelli and T. J. Gill, "Fractures and tendon injuries of the athletic shoulder," *Orthop. Clin. North Am.*, vol. 33, no. 3, pp. 497-508, Jul. 2002.
- [2] M. Islán, J. Carvajal, P. S. Pedro, R. D'Amato, J. A. Juanes, and E. Soriano, "Linear approximation of the behavior of the rotator cuff under fatigue conditions. Violinist case study," in *Proceedings of the 5th International Conference on Technological Ecosystems for Enhancing Multiculturality - TEEM 2017*, 2017, pp. 1-8.
- [3] M. Islán, F. Blaya, P. San Pedro, R. D'Amato, E. Lechosa Urquijo, and J. A. Juanes, "Analysis and Fem Simulation Methodology of Dynamic Behavior of Human Rotator Cuff in Repetitive Routines: Musician Case Study," *J. Med. Syst.*, vol. 42, no. 3, 2018.
- [4] CDC - NIOSH Publications and Products, "Cumulative Trauma Disorders in the Workplace (95-119)," *DHHS Publ. Number 95-119*, 1995.
- [5] International Labour Office - Geneva, *LIST OF OCCUPATIONAL DISEASES (revised 2010) Identification and recognition of occupational diseases: Criteria for incorporating diseases in the ILO list of occupational diseases*. Geneva: INTERNATIONAL LABOUR OFFICE • GENEVA, 2010.
- [6] A. Ruggiero, S. Alfaiato, M. Merola, and M. C. De, "FEM analysis of metal on UHMWPE total hip prosthesis during normal walking cycle," in *Proceedings of the XXIII Conference of the Italian Association of Theoretical and Applied Mechanics (AIMETA 2017)*, 2017, pp. 4-7.
- [7] A. Ruggiero, E. Gómez, and R. D'Amato, "Approximate analytical model for the squeeze-film lubrication of the human ankle joint with synovial fluid filtrated by articular cartilage," *Tribol. Lett.*, vol. 41, no. 2, pp. 337-343, Feb. 2011.
- [8] R. D'Amato, R. Calvo, and E. Gómez, "Sensitivity study of the morphometric fitting on the pressure field inside ankle joints," *Case Stud. Mech. Syst. Signal Process.*, vol. 1, pp. 8-14, 2015.
- [9] A. Ruggiero, E. Gómez, and R. D'Amato, "Approximate closed-form solution of the synovial fluid film force in the human ankle joint with non-Newtonian lubricant," *Tribology International*, vol. 57, pp. 156-161, 2013.
- [10] Z. P. Luo, H. C. Hsu, J. J. Grabowski, B. F. Morrey, and K. N. An, "Mechanical environment associated with rotator cuff tears," *J. shoulder Elb. Surg.*, vol. 7, no. 6, pp. 616-20, 1998.
- [11] N. Seki et al., "Mechanical environment of the supraspinatus tendon: Three-dimensional finite element model analysis," *J. Orthop. Sci.*, vol. 13, no. 4, pp. 348-353, Jul. 2008.
- [12] M. Zheng, Z. Zou, P. J. D. S. Bartolo, C. Peach, and L. Ren, "Finite element models of the human shoulder complex: a review of their clinical implications and modelling techniques," *International Journal for Numerical Methods in Biomedical Engineering*, vol. 33, no. 2, Wiley-Blackwell, 2017.
- [13] M. Islán, F. Blaya, P. S. Pedro, R. D'Amato, E. L. Urquijo, and J. A. Juanes, "Analysis and Fem Simulation Methodology of Dynamic Behavior of Human Rotator Cuff in Repetitive Routines: Musician Case Study," *J. Med. Syst.*, vol. 42, no. 3, p. 55, Mar. 2018.



11.2.ANEXO II

1D FINITE ELEMENT MODEL USING ROD ELEMENT APPLIED TO MUSCLE BEHAVIOR SIMULATION

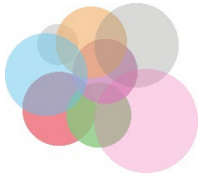
Las técnicas de elementos finitos se aplican en varios estudios para analizar el comportamiento de los músculos bajo carga. La mayoría de ellos han utilizado elementos 3D (hexaédricos o tetraédricos) para mallar la geometría del modelo y normalmente han analizado casos de carga estática, contracciones isométricas.

El objetivo de este estudio es implementar un nuevo modelo simplificado de un músculo, basado en un elemento 1D, tipo CROD, utilizado en los análisis más comunes, en lugar de una malla 3D de la geometría del músculo.

La principal ventaja de este nuevo método para modelar músculos es que se pueden analizar varias posiciones de una articulación en la que está involucrado el músculo sin ningún esfuerzo añadido al mallado original del modelo.

Sólo se necesita desplazar los huesos implicados en la articulación a su nueva posición, y así los músculos se moverán a su posición correcta. Los modelos 3D permiten estudiar de forma más detallada los mapas de tensión muscular, y las deformaciones, pero el principal problema que hay de afrontar es la necesidad de volver a mallar la geometría 3D cada vez que la articulación cambia de posición. Con este método simplificado, sólo se necesitan sus puntos de inserción, la sección y la longitud para definir el músculo, en el FEA.

De este modo, se puede emplear el mismo músculo en diferentes posiciones sin modificar la malla de la articulación completa.



1D Finite Element model using rod element applied to muscle behavior simulation*

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ABSTRACT

Finite element techniques are applied in several studies to analyze muscles behavior under load. Most of them have used 3D element (hexahedral or tetrahedral) to mesh the geometry of the model [1], [2], [3], [7] and usually they have analyzed static load cases, isometric contractions.

The objective of this study is to implement a new simplified model of a muscle, based on a 1D element, rod type, used in most common analysis, instead of a 3D mesh of the muscle geometry. The main advantages of this new method for modeling muscles is that several position of a joint, where the muscle is involved, could be analyzed without any added effort to the original meshing of the model. We only need to locate the bones involved in the joint in their new position, and so the muscles will move to their right position. 3D models allows us to study in a more detailed manner the muscle stress maps, an deformations, but the main issue we must work on is the need of re-meshing the 3D geometry any time the joint changes in position. With this simplified method, only its insertion points, the section and the length, define the muscle, in the FEA. And none of the changes when the joint changes in position.

KEYWORDS: FEM analysis, Joints, Muscles, Tendons, Ligaments.

1 INTRODUCTION

Muscles are the most complex element involved in a joint. Muscles, ligaments, tendons and bones mainly, form a joint. Bones can be considered as rigid bodies in the joint, comparing them with muscles and ligaments, and so it is not a problem meshing them using high rigidity material and 3D elements. Tendons correspond to the end of the muscles, where they are inserted into the bones, and can be considered high rigidity elements as well. Ligaments can be simplified as cables, not working under compression, made on isotropic material with

nonlinear behavior, and so Young's modulus should vary as the length of the ligament does.

Muscles can be classified in several types, depending on the fibers alignment, the geometry of the muscle itself, its function... Hill's model of a muscle determines that the force on a muscle is obtained adding the passive effects and the active effects on the muscle, this means, the reaction of the muscle to be deformed and the action of the muscle contracting itself when the stimulation appears. [9]

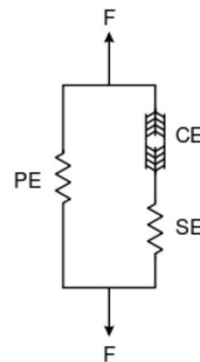


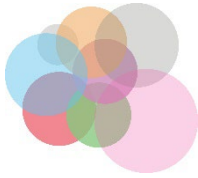
Figure 1: Hill's muscle model schema.

Where:

PE is the parallel element, the fascia surrounding the fibers.

CE is the contractile element, the fibers of the muscle.

SE is the serial element the tendons at the end of the muscle.



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According to their geometry, mechanical muscles as part of joints of the human body, can be classified as:

- Plain muscles, where two dimensions are bigger than the third one.
- Fusiform muscles, where fibers are aligned with the tendons.
- Biceps muscles, where only one tendon is at one end, but two are in the other.
- Single angled muscles, where fibers form an angle with action line of the muscles
- Doubled angled muscles, where fibers form two different angles with the action line of the muscle.

According to the type of fibers in the muscle, these can be classified as, [6], [8]

Type I, slow fibers, high resistance.

Type II A, fast oxidating - glucolitic fibers

Type II B, fast glucolitic fibers.

This study is based on fusiform muscles, mainly composed by fibers aligned with the tendons, and uniaxial efforts applied on the muscle in a static load case.

There are studies where a muscle has been tested in a uniaxial load case and the muscle has been characterized [6] In this studies the Young's modulus of a muscle has been defined as a nonlinear parameter depending on the length. This nonlinear Young's will be used in our study with a 1D element.

2 METHODOLOGY

The relation on a rat front tibia muscle between strain and stress is as the following tables shows.

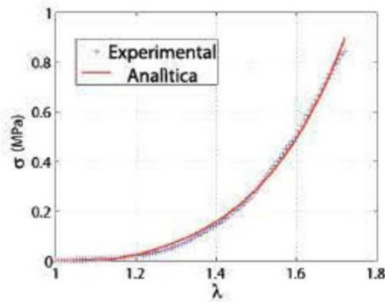


Figure 2: Stress vs strain on a front tibia muscle of a rat

That can be expressed as a table

Stress (σ) Mpa	0,1	0,2	0,4	0,6	0,8	0,9
Strain (λ)	1,3	1,42	1,55	1,62	1,7	1,72

These values are obtained through a uniaxial load test done on a rat muscle specimen. [6], we will use them once we define the muscle material of the 1D element used to simulate the muscle. As the muscle does not work in compression, the table, and the graph, must be edited in order to set to 0 the stress below strain value of 1.

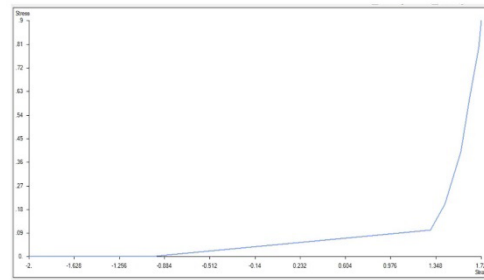


Figure 3: Stress vs strain on a muscle tissue.

The 1D element planned for model the muscle is a ROD, this element only transmit effort along the line that join the start and the end on the element. Axial efforts, no bending moments are transferred through the tendons to the bones.

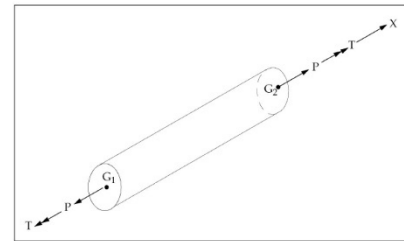
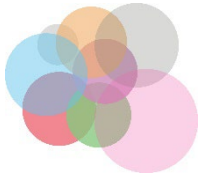


Figure 4: CROD element, only axial efforts are transferred

Using that material characterization and a rod element, a FEA model is developed to obtain the result of the simulations done using 3D elements for a front tibia rat muscle and a femoral biceps. [6]



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The results for these analysis, using 3D elements, and the characterization for muscle tissue above mentioned are shown below [6]

Front tibia rat muscle.

- Muscle length: 22.3 m
- Transverse area: 148.09 mm²
- Elongation: 10 mm
- F max: 1.9 N (estimated from curve)

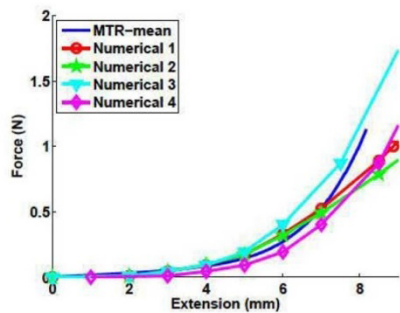


Figure 5: Force vs extension, front tibia rat muscle.

For human femoral biceps muscle.

- Muscle length: 240 m
- Transverse area: 2827.4 mm²
- Elongation: 35.37 mm
- Max. stress: 4.189 MPa

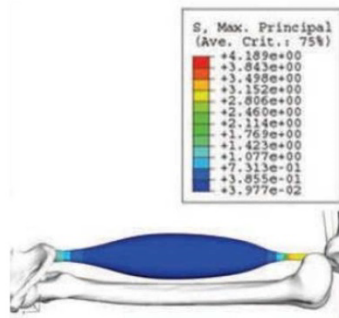


Figure 6: stress under 35 mm elongation, femoral biceps.

To develop our FEA model, Feamp+Nastran software has been used.

The characterization of the rod needs a material definition and a transverse area. The material is defined according to Fig 3 curve for stress/strain and the transverse area is the one mentioned above.

The use of the stress/strain curve obtained for the rat muscle in the human muscle I supported in studies proving both of them produce similar maximum stress [Peterson and Bronzino, 2008]

Especie	Tipo de Músculo	Preparación	Tensión activa máxima [KPa]	Referencia
Various	-	Síntesis	300	[Josephson and Edman, 1988]
Rata	Tipo I	Fibra	134	[Fitts et al., 1991]
Humano	Tipo IIB	Fibra	133	[Fitts et al., 1991]
Rata	Tipo I	Fibra	108	[Fitts et al., 1991]
Gato	Tipo I	Unidad motora	172	[Bodine et al., 1987]
Gato	Tipo IIA	Unidad motora	211	[Bodine et al., 1987]
Gato	Tipo IIB	Unidad motora	249	[Bodine et al., 1987]
Humano	Codo	Músculo completo	230 - 420	[Edgerton et al., 1996]
Humano	Tobillo	Músculo completo	45 - 250	[Palomaga et al., 1996]
Rata	Tibial Anterior	Músculo completo	272	[Wells, 1965]
Rata	Sóleo	Músculo completo	319	[Wells, 1965]
Cerdo de Guinea	Sóleo	Músculo completo	154	[Powell et al., 1984]

The ROD element needs two nodes to define its length and position in space.

The transverse area is assumed as constant, and circular, in all the element except for the tendons insertion points where we reduce it till 20% of nominal section.

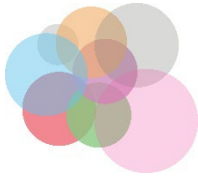
The element transverse area is defined as:

- Rat muscle: circle 6.86 mm radius.
- Femoral biceps: circle, 30 mm radius.

Both elements are constraint in one of their ends, one of the nodes of the element is fixed in all DOF, the other one is forced to move the distances shown above.

- Rat muscle, load condition. Displacement of 10 mm at one end along the muscle line.
- Femoral biceps load condition. Displacement of 35.37 mm at one end along the muscle line.

The element modeled is below. Top node of the element, in the image, is constrained in all DOF, bottom node is forced to move along Y axis, 10 mm for the rat muscle simulation, and 35.37 mm for the femoral biceps muscle.



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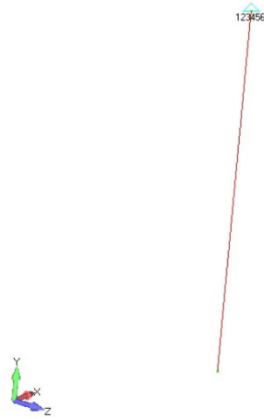


Figure 7: CROD element, Y axis aligned.

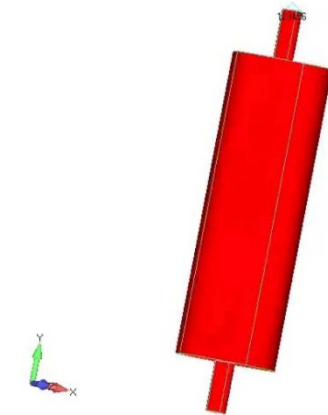


Figure 8: CROD element, Y axis aligned. Transverse section shown.

The analysis are solved as nonlinear analysis with the following parameters.

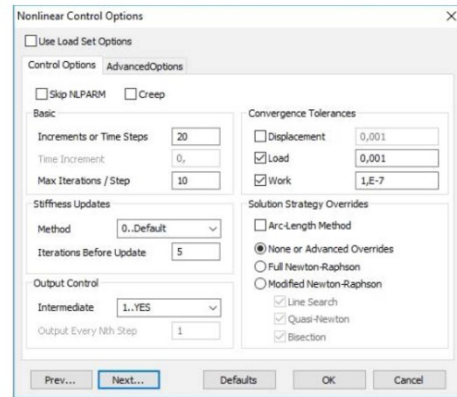


Figure 9: Nonlinear analysis parameters definition.

20 steps per analysis and 10 iterations per step to converge.

The results of the analysis are

- **Front tibia rat muscle.**

Results on the analysis are:

Maximum displacement, 10 mm. same displacement applied in 3D element model.

Maximum force, 1.897 N vs

Maximum axial stress, 0.127 MPa

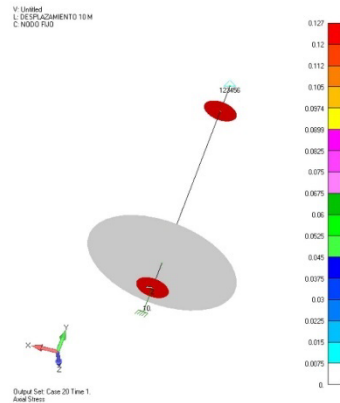


Figure 10: axial stress, front tibia rat muscle.



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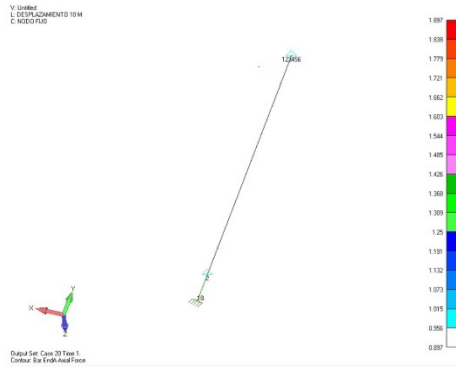


Figure 11: axial force, front tibia rat muscle.

As expected, maximum stress appears on reduced area parts of the muscle, tendon insertions.

- **Femoral biceps muscle**

Results on the analysis are:
Maximum displacement, 35.37 mm. same displacement applied in 3D element model.
Maximum force, 1.897 N vs
Maximum axial stress, 0.127 MPa

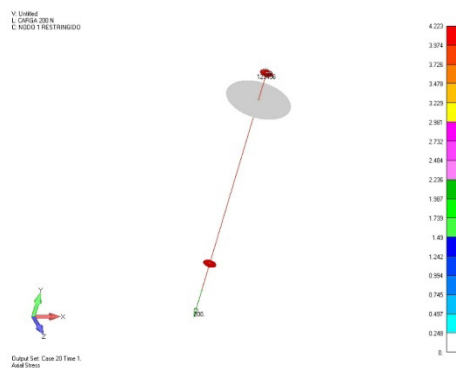


Figure 12: axial stress, femoral biceps muscle.

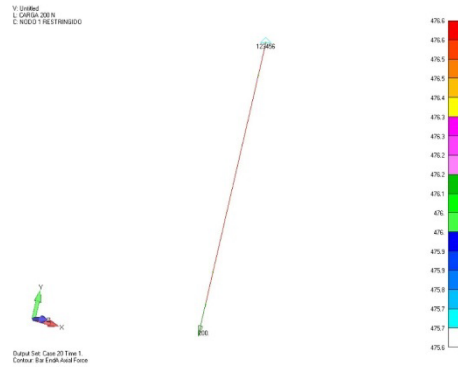


Figure 13: axial force, femoral biceps muscle.

As seen on images above, results are close to the values obtained in the muscle model done with 3D elements.

4 CONCLUSIONS

This study validates the FEM developed presenting the first model in which a muscle is simplified using a 1D element Rod type.

The main advantages of this simplification are:

- The time of analysis is highly reduced, nonlinear analysis using 3D elements to model a muscle should require about 10 minutes while the same muscle modeled using 1D element can be analyzed in about 30 seconds. Intel i7 station, 64 GB ram and SSH drive has been used.
- The joint where the muscle is inserted can be modified in position when using a 1D element type for muscle simulation. If the muscle is modeled using 3D elements, the joint must be re-meshed once the new position of its members is defined.

The main advantage of this study is the improvement of the relation results / time processing. But it's limited to the study of the joint in a general manner, if we need to study detailed effects on the muscle this simplification will not be valid as the muscle is considered uniform in dimensions.

The proposed analysis meets the stated objective:

- The results of the analysis run using 3D elements to model a load muscle are close to the results obtained using 1D element, rod type
- The 1D element type allow us to change geometry of the joint in a simple and fast way just moving the nodes that define the element



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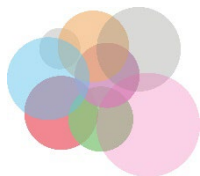
- The 1D element can be easily modified to model different muscles, fusiform aligned with its fibers aligned with tendons. The parameter that has to be changed is the transverse section considered
- The muscle tissue behavior can be simulated with the stress strain relation shown in this study.

This study is the first step in order to get a full joint modeled in a FEA where muscles can be modified in position in a simple way.

Following studies will work in the same simplification applied to ligaments, once this is done a full model of a joint can be developed using muscles and ligaments as 1D elements.

REFERENCES

- [1] M. Islan, J. Carvajal, P. S. Pedro, R. D'Amato, J. A. Juanes, and E. Soriano, "Linear approximation of the behavior of the rotator cuff under fatigue conditions. Violinist case study," in Proceedings of the 5th International Conference on Technological Ecosystems for Enhancing Multiculturality - TEEM 2017, 2017, pp. 1-8.
- [2] M. Islan, F. Blaya, P. San Pedro, R. D'Amato, E. Lechosa Urquijo, and J. A. Juanes, "Analysis and Fem Simulation Methodology of Dynamic Behavior of Human Rotator Cuff in Repetitive Routines: Musician Case Study," *J. Med. Syst.*, vol. 42, no. 3, 2018.
- [3] A. Ruggiero, S. Affatato, M. Merola, and M. C. De, "FEM analysis of metal on UHMWPE total hip prosthesis during normal walking cycle," in Proceedings of the XXIII Conference of the Italian Association of Theoretical and Applied Mechanics (AIMETA 2017), 2017, pp. 4-7.
- [4] N. Seki et al., "Mechanical environment of the supraspinatus tendon: Three-dimensional finite element model analysis," *J. Orthop. Sci.*, vol. 13, no. 4, pp. 348-353, Jul. 2008.
- [5] M. Zheng, Z. Zou, P. J. D. S. Bartolo, C. Peach, and L. Ren, "Finite element models of the human shoulder complex: a review of their clinical implications and modelling techniques," *International Journal for Numerical Methods in Biomedical Engineering*, vol. 33, no. 2, Wiley-Blackwell, 2017.
- [6] A. Maria Ramírez, Begoña Calvo, Jorge Grasa., "Modelado y simulacion del tejido musculo-esquelético. Validacion experimental con el musculo tibial anterior de rata" Tesis doctoral, Univ. de Zaragoza, 2011
- [7] Belen Seral, Jose Manuel Garcia Aznar, Manuel Doblaré, "Simulacion en 3D con elementos finitos de un modelo de protesis de rodilla" *Rev Ortop Traumatol* 2003;47:64-72.
- [8] Kigo. Lk: MS Francisco Guede R. "Biomecanica del músculo". Universidad de las Americas.
- [9] F.J.Alonso, Gloria Galan-Marin, D.R.Salgado, Rosa Pàmies-Vila "Calculo de esfuerzos musculares en la marcha humana mediante optimización elastica-fisiologica" XVIII Congreso Nacional de Ingeniería Mecánica, 2010



11.3.ANEXO III

FINITE ELEMENT MODEL OF AN ELBOW UNDER LOAD, MUSCLE EFFORT ANALYSIS WHEN MODELED USING 1D ROD ELEMENT

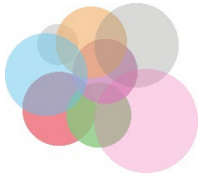
Las contracciones isométricas de los músculos se han analizado varias veces aplicando el análisis del modelo de elementos finitos. La mayoría de estos análisis han utilizado elementos 3D (hexaédricos o tetraédricos) para mallar la geometría del músculo.

Este estudio es un paso adelante en la búsqueda de una técnica de modelado simplificada para estos músculos. Se han realizado estudios previos sobre el mismo tema y los resultados muestran que es posible simplificar un músculo utilizando un elemento 1D, tipo CROD, para modelar el músculo.

El objetivo final es obtener una metodología en la que se pueda predecir el comportamiento muscular bajo carga, y durante el movimiento articular, para poder estudiar los efectos de la fatiga, el esfuerzo máximo obtenido, las mejoras en los movimientos de la articulación para minimizar el esfuerzo en los músculos, y muchos más.

Una vez que se pueda predecir el comportamiento muscular en movimiento se pueden desarrollar varias aplicaciones.

Este estudio utiliza el poder de las herramientas de modelado CAD y análisis de elementos finitos existentes, además de técnicas para la reconstrucción de articulaciones humanas basadas en imágenes de RMN.



Finite Element model of an elbow under load, muscle effort analysis when modeled using 1D rod element. *

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ABSTRACT

Isometric contractions on muscles have been analyzed several times applying Finite Elements Model analysis. Most of these analysis have used 3D elements (hexahedral or tetrahedral) to mesh the geometry of the muscle [1], [2], [4],[7] .

This study is one step ahead on the seek for a simplified modeling technique for this muscles. Previous studies on the same topic have been done and the results show that is possible simplify a muscle using a 1D element, rod type, for muscle modeling.

The final aim is obtaining a methodology in which we can predict the muscle behavior under load, and during joint movement, so we can study fatigue effects, maximum effort obtained, improvements in movements in the joint in order to minimize effort on muscles, and many others. Once we can predict the muscle behavior in movement several applications can be developed.

This study uses the power of existing CAD modeling and finite elements analysis tools, plus techniques for reconstruction human joints based on NMR images.

KEYWORDS: FEM analysis, Joints, Muscles, Tendons, Ligaments, NMR reconstruction, reverse engineering, elbow, cartilages.

1 INTRODUCTION

Muscles are the most complex elements in a human joint to be modeled. Muscles, tendons, bones, cartilages and ligaments, commonly form joints.

- Bones are commonly approached to rigid bodies (considering proper Young modulus, density, etc..) under linear behavior, [1], [2], [4],[7].

- Ligaments are used to be considered as springs, considering their stiffness, they can be quite well approximated as a reactive element that reacts to the effort applied, [10].
- Cartilages are rigid bodies (considering proper Young modulus) with a linear behavior.
- Tendon a muscles form what is commonly defined as Muscle-tendon element, their behavior is complex to simulate and this is the main aim of this study [6].

As an introduction to this couple tendon-joint, [6].

- They cannot be approached as a spring, they are not reactive elements.
- The force it can generate depends on the length they have.
- The force it can generate depends on the speed of the contraction.
- The force it can generate depends on the type of stimulation is applied (frequency and voltage).

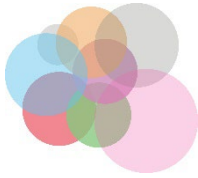
Based on this premises, the muscle modeling is a complex task that cannot be done in a direct way, using one of the common 1D elements that exist in FEM libraries.

This study is focused on one muscle type.

There are several types of muscles, classified in different types depending on different characteristics.

According to their geometry, mechanical muscles as part of joints of the human body, can be classified as:

- Plain muscles, where two dimensions are bigger that the third one.
- Fusiform muscles, where fibers are aligned with the tendons.
- Biceps muscles, where only one tendon is at one end, but two are in the other.



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- Single angled muscles, where fibers form an angle with action line of the muscles
- Doubled angled muscles, where fibers form two different angles with the action line of the muscle.

According to the type of fibers in the muscle, these can be classified as, [6], [8]

Type I, slow fibers, high resistance.

Type II A, fast oxidating - glucolitic fibers

Type II B, fast glucolitic fibers.

This study is based on fusiform muscles, mainly composed by fibers aligned with the tendons, and uniaxial efforts applied on the muscle under a isometric load case.

The joint chosen for the study is the elbow, moved by biceps and triceps muscles (both of them can be considered fusiform muscles). The scope of work includes the following aspects:

- Improving the mesh used in the finite element model.
- Improving the mesh of the muscles using 1D element type.
- Improving the mesh of the complete joint so it can be used in several positions (different angles in elbow joint)
- Obtaining reaction forces for muscles in different positions of the joint.

2 METHODOLOGY

Starting from the cloud of points (Fig 1) obtained through a resonance. A 3D Computer Aided Design (CAD) model of the elements involved in the articulation, (scapula, humerus, radius and ulna, -biceps and triceps- muscles and tendons) is generated.

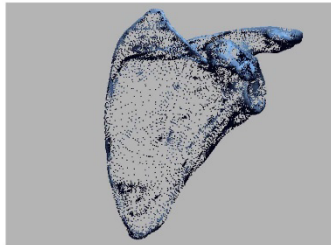


Fig 1: Cloud of points corresponding to the scapula

The reconstruction of the models of the bones and tendons involved in the joint is carried out using Geomagic Design X software.

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This software is capable of generating a solid model (fig 2), or surfaces, based on a cloud of points.

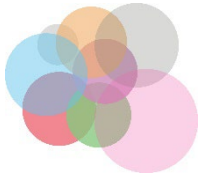


Fig 2: 3d model auto-generated from cloud of point

Once the 3D model is generated it has to be exported to a FEA software (FEMAP) to be meshed and solved.

To do this, we need the bodies of the joint (bones, muscles) to be generated as independent solid bodies. In our case, muscles and tendons must be reconstructed thus they are surfaces (open and not sewed surfaces) bodies.

After this reconstruction, a complete CAD 3D (fig 3) model of the joint is obtained.



1D Finite Element model using rod element applied to muscle behavior simulation

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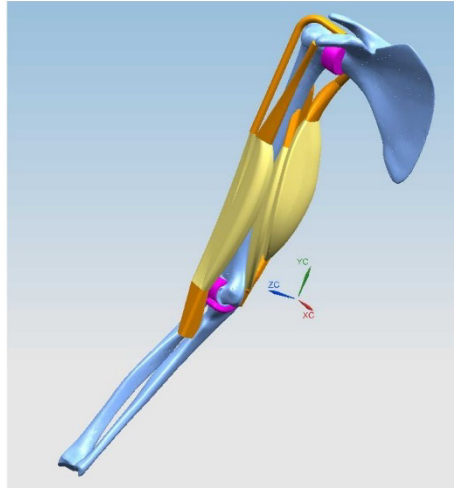


Fig 3: 3d model reconstructed into solid bodies.

The complete model is composed by tendons, muscles, cartilages, and bones, can be seen in following images, element by element.

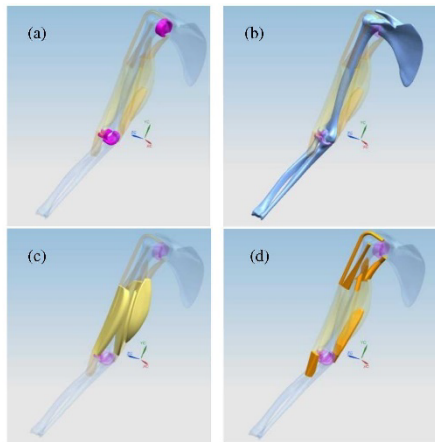


Fig 4: 3d model of different elements in joint

- | | |
|---------------|-------------|
| a. Cartilages | c. Muscles |
| b. Bones | d. Tendons. |

Once the model is converted into solid bodies, it must be exported (via step or parasolid file) to the FEA software.

The meshing is done in this FEA software (Siemens FEMAP) and some hints must be considered prior this task is run.

Mesh type.

The mesh type chosen depends on the type of element is going to be meshed.

- Bones: they are meshed using 3D elements (tetrahedral type, 2nd order, 8 nodes). Proper material (bone tissue) is considered for these elements.
- Cartilages: they are meshed using 3D elements (tetrahedral type, 2nd order, 8 nodes). Proper material (cartilage tissue) is considered for these elements.
- Ligaments. They are meshed as 1D elements, spring damper elements, with a proper stiffness (30 N/mm)
- Muscles. These are meshed using 1D elements (type is discussed in this study).
- Tendons. These are meshed in combination with muscles, as a unique element muscle+tendon. Same 1D element is considered for tendons.

MUSCLE ELEMENT TYPE.

Muscles must be modeled using 1D element, to nodes, that could allow us to re-use same meshed model in several positions just moving the elements on the mesh, without re meshing new parts. For this purpose, classical FEM software offer some 1D element type, their principal characteristics are:

Rod type

- Only transmit axial forces
- Based on Young modulus behavior
- Linear or non linear
- Constant area along the element
- Transverse area remains unchanged, small deformations considered.

Beam type

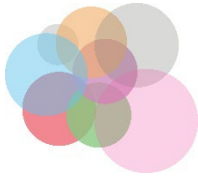
- Transmit axial and shear forces
- Based on Young modulus behavior
- Linear or non linear
- Constant area along the element
- Transverse area remains unchanged, small deformations considered.

Spring-damper type

- Spring characteristics can be defined independently
- No transverse area definition is needed
- No material behavior, stiffness is numerically defined
- Linear or non- linear spring can be defined.

MUSCLE BEHAVIOR.

Muscles have been modeled and studied several times.



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Their behavior has been described several times, they are based mainly in Hill's model (fig 3), [3], [5], [9].

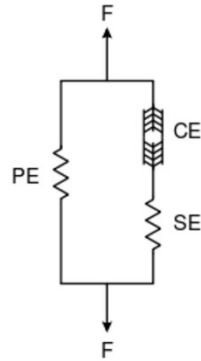


Fig 5: Muscle Hill's model.

Where:

PE is the parallel element, the fascia surrounding the fibers.
CE is the contractile element, the fibers of the muscle.
SE is the serial element the tendons at the end of the muscle.

According to this model, the muscle can be considered as the combination of the three elements on the image, fascia + muscle fibers+tendons to obtain the force of the muscle.

In order to apply this model into a FEM model, we may know the stiffness of SE and CE elements, assuming PE (fascia) do not have effect compared to PE+SE contribution, and this is the main issue thus the stiffness of a muscle is not a constant value and depends on several parameters.

Muscle force depends on following parameters:

- Muscle length
- Speed of contraction
- Voltage of exciting of electrical signal
- Contraction duration.

Following curves show this dependency, [6].

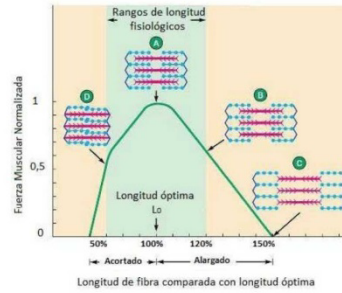


Fig 6: Muscle force vs muscle length

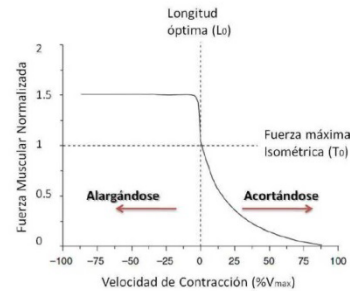


Fig 7: Muscle force vs contraction speed

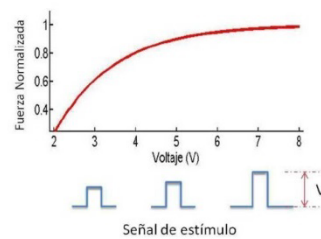


Fig 8: Muscle force vs excitation



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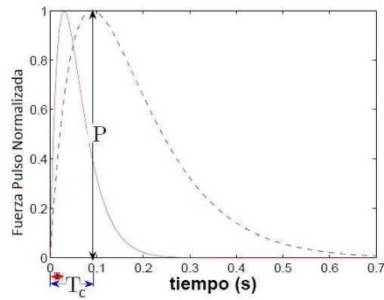


Fig 9: Muscle force vs excitation duration

As seen on previous images, there is not a unique stiffness value for a muscle, it depends on several factors.

Considering only isometric contractions, no length variation on muscle, force vs length is the only relation that should be considered, but taking this information into a FEM model is not possible due to the type of elements that can be used in such mathematical model.

The second condition for this muscle element is that must be used in the FEM model at any configuration of the model, joint position, without any new configuration, so we can automate analysis process in any situation.

The most suitable element that could be used for this purpose is the spring-damper element with a variable stiffness vs element length parameter.

At a first glance could be considered a non linear stiffness parameter in the element, but this non linear stiffness can be defined only dependant on one parameter, for example:

- Stiffness vs length
- Stiffness vs extension (positive or negative)

It is not possible defining this stiffness as a relation depending on two parameters simultaneously, thus if we define the stiffness dependant on length, the element will work even in compression, on the other hand, if we define this stiffness dependant on extension, we can not consider the relation between stiffness vs length.

One more issue associated to this element type is that, considering the muscle as a spring, and considering an extension of X mm, the force obtained from the FEM will be a constant value following the equation

$$F=k \cdot x$$

But this does not apply to a muscle, that force would be the maximum force the muscle could develop, but not the real force the muscle develop in such load configuration.

In other words, a certain muscle at a certain length, can develop a maximum force of X Newtons, but that muscle under a certain load condition may need to develop just 40% of that maximum force, simulating this issue with a spring-damper element is not possible. The force a spring develops is directly referred to the length variation it suffers.

MUSCLE MODELING

The conclusion after this point is that there is no possibility of modeling a muscle in a FE model in such a way that automatic analysis on different configurations of the joint can be solved.

The alternative is modeling a muscle in such a way that a unique FEM model can be analyzed in different positions of the joint in order to obtain in a direct manner the force developed by a certain muscle, under certain load conditions.

If we analyze the model in several configurations we can obtain the force in the muscle in different configurations, and so, compare that needed force with the force that muscle could develop.

Is a non-direct analysis method, with a previous study of the dynamic condition of the joint (forces balance) and a second step comparing this results with the muscle itself.

The decision adopted is modeling muscles using rod elements, with a high stiffness (greater than real stiffness of muscles) in order to find the force that a muscle should develop to withstand the load applied with minimum displacements on the joint to maintain the isometric condition.

The mesh has been done using the following element types:

N ^o	Element type	Parameters	Application
1	tetrahedral 10 nodes	Material, Young modulus and Poisson coef.	Meshing bones
2	Spring damper	Stiffness definition	Mesh of the ligaments
3	rigid element spider	Independent node and dependent nodes	Simulation insertion of the tendon in the bone.
4	rod element 2 nodes	Material, Young modulus and Poisson coef.	Muscles modeling

As we work with the elbow joint, two different models are used, one for loading biceps muscles and other one for triceps loading, depending on loading direction.



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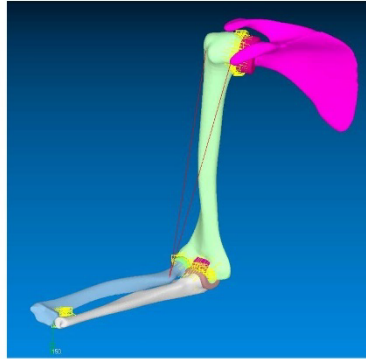


Fig 10: Biceps loading load case

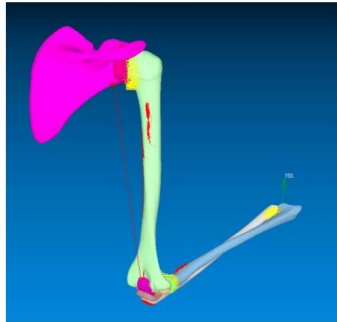


Fig 11: Triceps loading load case

The characteristics of the muscle+tendon elements considered are:

- Transverse section: 100 mm²
- Material:
 - Young modulus: 2000 N/mm²
 - Poisson coef: 0.43

As we are not studying the element itself but the force that muscle develop under load, we are not concern about proper material (muscle tissue) and section (according to real muscle). Our objective is to maintain isometric condition on joint, and so, small displacements on wrist, that is why muscle stiffness, in our model, is higher than real one.

CARTILAGE CONTACT CONDITION

At a real joint, cartilage are not joined together, the real condition is a contact between the surfaces of both cartilages involved in the movement.

This can be simulated in FEM analysis using contacts between parts, transmitting only compression but not tractions. But this technique implies high effort in computing (memory and processing resources).

The alternative is connecting both cartilages involved in joint by rigid elements, in this case the stress distribution in the contact area is affected due to the traction efforts transmission between surfaces.

Doing a preliminary test on the elbow analyzed, analyzing it with and without contact on cartilages, can find the difference in processing resources needed, and the difference in the result when using contact vs using rigid elements.

Both analysis run on same computer, analysis parameters are:

System configuration:

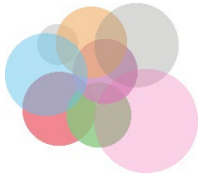
- Hostname: INTEL-I5
- Architecture: cm64int
- Platform: Intel64 Family 6 Model 158 Stepping 9 Windows
- Model: Intel(R) Core(TM) i5-7400 CPU @ 3.00GHz
- Clock freq.: 3000 MHz
- Number of CPUs: 4
- Executable: standard
- Raw model ID: 8664
- Config number: 8664
- Physical memory: 32707 MB
- Virtual memory: 51064 MB
- Numeric format: 64-bit little-endian IEEE.
- Bytes per word: 8
- Disk block size: 512 bytes (64 words)

The time to solve the model using contact modeling in cartilages rises up till 67 hours, 45 minutes

The time to solve the model without contacts is 26 minutes.

As seen on previous results, the use of contacts in the cartilages areas in the joint makes sense only if the alternative (use of rigid elements joining cartilages on involved bones) gives us results highly different.

Following images show the results of the same model using contacts and rigid elements.



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Contact modeled in the cartilages.

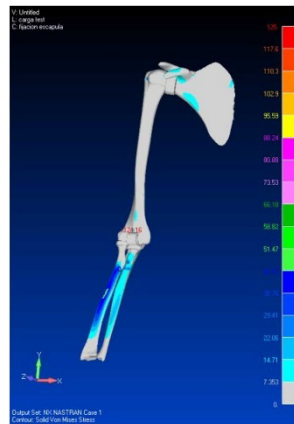


Fig 12: Stresses map using contacts

Rigid elements joining cartilages

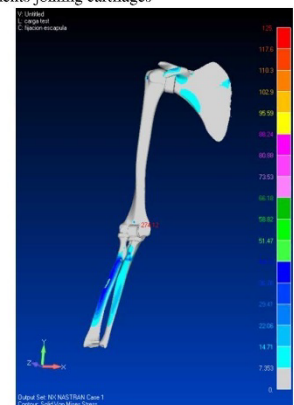


Fig 13: Stresses map using rigid elements on cartilages

Although when using rigid elements, maximum stress is higher, this maximum correspond to the stress concentration due to the rigid elements, discarding these stress concentrations the stress mapping in the joint is similar in both cases, and so, the use of rigid elements vs contact modeling can be accepted as a valid technique.

MUSCLE INSERTION IN BONES

Due to one of the aims of the FEA model is that it must be used in several positions of the joint (aperture angle in elbow), the insertion of the muscle in the bones must be done in such way that the insertion condition is maintained when moving the joint elements from one position to another

The muscle + elbow has been modeled using the rod element mentioned before, two nodes element. The condition needed is that nodes of these elements must belong to the bones involved (for example, for one of the biceps couple of muscles, one of the nodes must belong to the humerus, and the other one must belong to the ulna or the radius, depending on the muscle).

Doing this, and only this, the stress at the insertion points on the bones will be much higher than real ones. To obtain a more real stress map at these areas, a rigid element is included in the model, distributing the effects on the independent node to a group of dependent nodes located on the bones, as well.

In the image can be found this modeling.

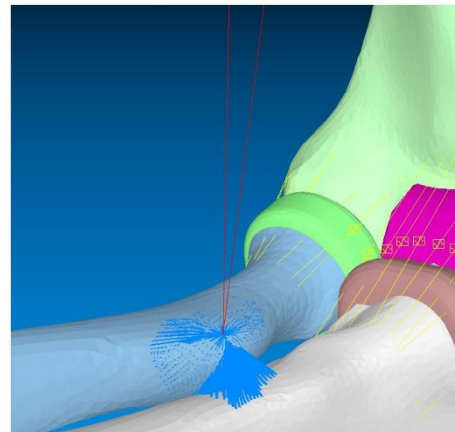


Fig 14: Rigid elements at muscle-tendon insertion

Muscle+tendon elements, red color, ends in the center node of a rigid element, blue one, that distributes the effect at this central node to the surrounding nodes, all of them located on the bones).



Finite Element model of an elbow under load, muscle effort analysis when modeled using 1D rod element.

TEEM2020, October 2020, Salamanca, SPAIN
J.A. Juanes et al.

3 ANALYSIS

In order to validate model proposal for muscles in a human joint, two different models have been modeled, one for biceps loading and the other one for triceps loading.

The objective in these analysis are:

- Check the force that appear in muscles when loading the joint
- Check the stress map that appear in model when loading.
- Check deflections in model.

The load case considered is an elbow, at a 74° opening (angle between upper arm and forearm), and 150 N applied in vertical in the wrist (like when holding a dumbbell).

The movement restrictions are applied in the scapula, area constrained has been fixed in all 6 DOF, the elbow is not supported, and so the complete arm will be loaded.

The load case can be seen in the following image.

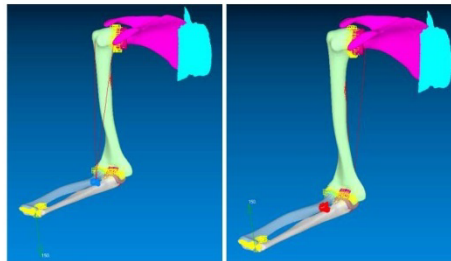


Fig 15: Biceps and triceps loadcases (150 N vertical)

RESULTS

Following table show the effort (N) that appear on each muscle depending on the load case considered.

	Biceps loaded	Triceps loaded
Biceps brachii, long head	500 N	-
Biceps brachii short head	218 N	-
Triceps brachii lateral	-	495 N
Triceps brachii longus	-	346 N
Triceps brachii medium	-	94 N

And the stress maps on the joint can be seen in the following images.

Biceps loading.

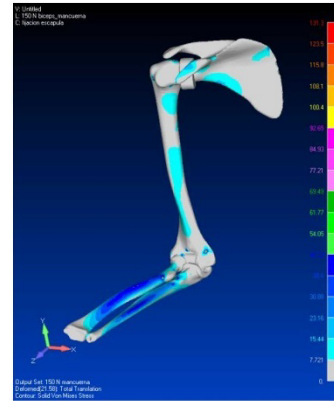


Fig 16: Stresses map biceps loading

Triceps loading

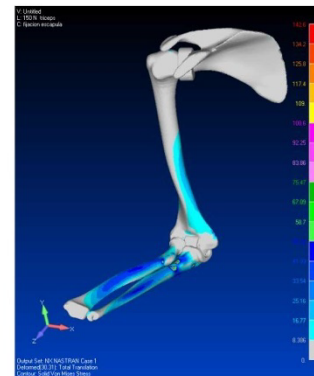
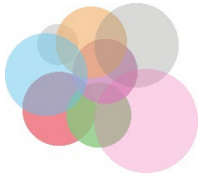


Fig 17: Stresses map triceps loading

In both cases, maximum stress (discarding stress concentrations on muscle insertions) appears on radius and ulna bones. Humerus also present some stress due to the fixation on scapula.

Maximum values for stress (discarding stress concentrations due to rigid element use) are:

- 131.3 MPa in biceps loading



1D Finite Element model using rod element applied to muscle behavior simulation

TEEM2019, October 2019, León, SPAIN

- 142.6 MPa in triceps loading.

4 CONCLUSIONS

The use of a rod element to model muscles allow us to reach the objectives:

- A complete elbow can be meshed and analyzed in several positions without remesh the geometry every time.
- The results converge showing a proper stress map on bones.
- There is no possibility to mesh a muscle that can work as is expected from muscle tissue, so the analysis of the joint must be done in two steps, one of the looking for the effort that appear on the muscle and a second one comparing that effort with the force a muscle can develop in that situation (type of contraction, speed of contraction, length of the muscle...)
- Modeling the contact in the cartilages area the resources needed to run the analysis are increased dramatically.

This analysis and modeling technique represent the preliminary step to develop new analysis, on same mesh, obtaining results for the joint in other positions and different load so we can create a database with results so we can check them with real specimen.

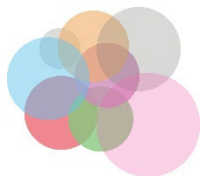
Next step on this investigation are:

- Running analysis on elbow, under same load (150 N on biceps and triceps) and get effort on muscles for all of them
- Search for a common parameter in all of the that could describe the behavior of s fusiform muscle.
- Compare theoretical results with real specimen, different sizes of muscles, different ages,....

REFERENCES

- [1] M. Islan, J. Carvajal, P. S. Pedro, R. D'Amato, J. A. Juanes, and E. Soriano, "Linear approximation of the behavior of the rotator cuff under fatigue conditions. Violinist case study," in Proceedings of the 5th International Conference on Technological Ecosystems for Enhancing Multiculturality - TEEM 2017, 2017, pp. 1-8.
- [2] M. Islan, F. Blaya, P. San Pedro, R. D'Amato, E. Lechosa Urquijo, and J. A. Juanes, "Analysis and Fem Simulation Methodology of Dynamic Behavior of Human Rotator Cuff in Repetitive Routines: Musician Case Study," J. Med. Syst., vol. 42, no. 3, 2018.
- [3] G.A. Lichtwark and A.M. Wilson, "A modified Hill muscle model that predicts muscle power output and efficiency during sinusoidal length changes," Journal of experimental biology, 2005.
- [4] N. Seki et al., "Mechanical environment of the supraspinatus tendon: Three-dimensional finite element model analysis," J. Orthop. Sci., vol. 13, no. 4, pp. 348-353, Jul. 2008.
- [5] Eric J. Perreault, Thomas G. Sandercock and Charles J. Heckman "Hill muscle model performance during natural activation and electrical stimulation" 23rd annual conference - IEEE/EMBS, Turkey, 2001.

- [6] A.María Ramírez, Begoña Calvo, Jorge Grasa, "Modelado y simulación del tejido muscular-esquelético. Validación experimental con el músculo tibial anterior de rata" Tesis doctoral, Univ. de Zaragoza, 2011
- [7] Belen Seral, Jose Manuel Garcia Aznar, Manuel Doblaré, "Simulación en 3D con elementos finitos de un modelo de prótesis de rodilla" Rev Ortop Traumatol 2003;47:64-72.
- [8] Klgo. Lic: MS Francisco Guede R. "Biomecánica del músculo". Universidad de las Américas.
- [9] Hatherine R.S. Holzbaur, Wendy M.Murray and Scott L. "A model of the upper extremity for simulating musculoskeletal surgery and analyzing neuromuscular control", Annals of Biomedical Engineering Vol 33, 2005
- [10] Jeffrey A. Weiss, John C Gardiner, Benjamin J. Ellis, Trevor J. Lujan and Nikhil S. Phatak, "Three-dimensional finite element modeling of ligaments, technical aspects", Medical engineering and physics, 2005.



11.4.ANEXO IV

ANALYSIS AND FEM SIMULATION METHODOLOGY OF DYNAMIC BEHAVIOR OF HUMAN ROTATOR CUFF IN REPETITIVE ROUTINES: MUSICIAN CASE STUDY

La mayoría de las lesiones musculoesqueléticas localizadas en el hombro suelen deberse a movimientos repetitivos o sostenidos que se producen en rutinas de trabajo en diferentes áreas. En el caso de músicos, como los violinistas, que tienen rutinas de entrenamiento largas y diarias, los movimientos repetitivos que realizan son forzados y en ocasiones las posturas no son naturales.

Este artículo tiene como objetivo estudiar y simular el comportamiento dinámico de la articulación glenohumeral bajo condiciones repetitivas que representan las diferentes posturas asumidas por un violinista durante su entrenamiento diario. Para ello se utilizan los criterios proporcionados por el método RULA (evaluación rápida de miembros superiores) ha sido usado.

Posteriormente, utilizando como geometría de referencia la de la articulación en estudio generada y modelada en CATIA® [VERSIÓN 5R21], se ha propuesto un análisis FEM con el software ANSYS® [VERSIÓN 17.1] simulando los movimientos cortos y cíclicos del Húmero de los violinistas.

Con el análisis realizado, gracias a las simplificaciones realizadas, consideraciones lineales y materiales isotrópicos de la articulación, se ha podido conocer el comportamiento dinámico aproximado de tejidos, músculos y tendones, así como la respuesta de la articulación en términos de fatiga.



Analysis and Fem Simulation Methodology of Dynamic Behavior of Human Rotator Cuff in Repetitive Routines: Musician Case Study

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Abstract

The majority of musculoskeletal injuries located in the shoulder are often due to repetitive or sustained movements that occur in work routines in different areas. In the case of musicians, such as violinists, who have long and daily training routines, the repetitive movements they perform are forced and sometimes the postures are not natural. Therefore, this article aims to study and simulate the dynamic behavior of the glenohumeral joint under repetitive conditions that represent the different postures assumed by a violinist during his daily training. For this purpose, the criteria provided by the RULA (rapid upper limb assessment) method have been used. Subsequently, by using as a reference geometry that of the articulation under study generated and modeled in CATIA[*VERSIÓN 5R21*], a FEM analysis has been proposed with the software ANSYS[*VERSIÓN 17.1*] simulating the short and cyclic movements of the Humerus of the violinists. With the analysis carried out, thanks to linear and isotropic approximations of the joint, it has been possible to know the approximate dynamic behavior of tissues, muscles and tendons, and the response of the joint in terms of fatigue.

Keywords Musician · FEM analysis · Humerus · Rotator cuff · RULA criteria

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Introduction

The shoulder represents the most unstable and susceptible to injury joint in the human body for its high mobility [1]. In most cases this is due to excessive use of the joint and bad postures in repetitive routines [2–5]. Thanks to the action of the muscles, ligaments and tendons of the rotator cuff (Fig. 2) this type of instability is compensated allowing all kinds of movements [6, 7]. The functional unit of the shoulder includes the proximal humerus, the clavicle and the scapula, and their connections to each other, in addition to the sternum and the rib cage (Fig. 1). Together these elements form the following joints:

- The scapulohumeral or glenohumeral joint (Diarthrosis enarthrosis or spheroidica). It joins the Humerus with the scapula.
- The sternocostoclavicular joint (Diarthrosis). It joins the sternum and first rib with the clavicle.
- The acromioclavicular joint (Diarthrosis arthrodes). It links the acromion to the clavicle.

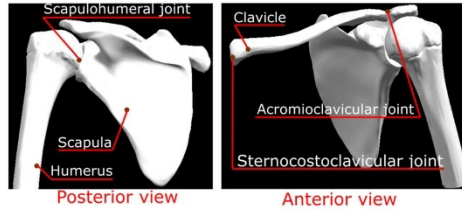
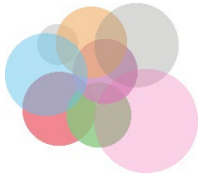


Fig. 1 Shoulder joint

- The omoserrathoracic joint, whose articular surfaces are muscles.

Together these joints allow the upper extremity to rotate up to 180° in three different planes. This allows the arm to perform a variety of versatile activities.

The correct knowledge of this articulation is important for those people who carry out repetitive movements. For example, musicians such as violinists who train the shoulder daily and who with excessive use of the instrument leave the shoulder joint susceptible to muscle and nervous lesions, muscular pains, contractures and also structural disorders in the back, neck, arms and hands [8, 9]. The musculoskeletal problems most frequently observed in musicians are: overuse (50%), nerve compression or thoracic outlet syndrome (20%), focal dystonia (10%) [10]. The initial symptoms that often appear are pain, in about 85% of cases, burning sensation, tiredness or heavy limb in one or more parts of the body, and may arise even after a short period of activity [11, 12]. Heming [13] in his study of 107 musicians aged between 16 and 72 years has found that 70% had suffered an injury related to the instrument. Women (72%) and stringed instrument musicians

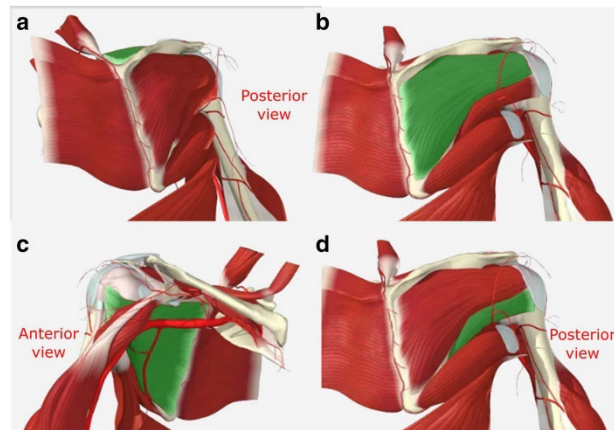
(77%) showed a higher prevalence, together with professional teachers (57%) who harbored the majority of injuries. The absence of pathologies in beginner musicians, in his study, suggests that the most influential risk factor is strongly related to the increase in the use time of the instrument and to the frequency of training. According to recent studies, 25% of orchestra musicians suffer to some pathology as pain in the neck and shoulder. The 35.3% are violinists. It is due to the complex posture of the arms and neck and who usually practice more than 3 h per day [14]. The glenohumeral joint of the shoulder is composed mainly of bones such as scapula (glenoid cavity) and Humerus (humerus head), articular cartilage, and the muscle complex called the rotator cuff. It is formed by the follow muscles:

- Supraspinatus. It is located in the upper area of the shoulder joint; it protects the deltoid muscle from the movement of separation (abduction) of the arm (Fig. 2a).
- Infraspinatus. Its function is to allow the external rotation of the arm (Fig. 2b).
- Subscapular. Its function is to allow the internal rotation of the arm (Fig. 2c).
- Teres minor. Its function is to allow backward stretching, external rotation, bring the arms towards the body (adduction) (Fig. 2d).

The function of the rotator cuff is to provide stability at the moment of generating combined movements of: abduction-adduction or flexion-extension with internal and external rotation.

Tendinitis, tendinosis, entrapment syndrome or impingement on the supraspinatus muscle are the most frequently present disorders in the shoulder in the case of the right

Fig. 2 Rotator cuff muscles in green color: a supraspinatus; b infraspinatus; c subscapular; d teres minor



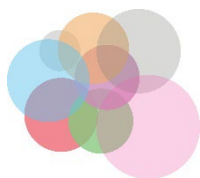
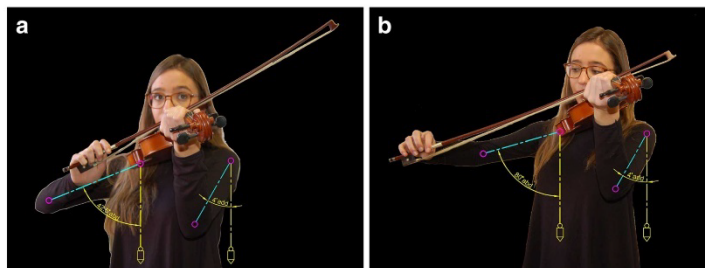


Fig. 3 **a** Right arm with abduction angles of 42.5 degrees. **b** Right arm with abduction angle of 80 degrees



shoulder of the violin and flute players, when there are prolonged positions of abduction or static flexion of the forearm [15]. In the left shoulder there is flexion and sometimes adduction resulting in a rotator cuff friction phenomenon produced for poor blood supply and due to the compression effect of the supraspinatus tendon. For this reason, the strengthening of external rotators and stabilization of the tendon deserves additional attention, such as treatment or prevention [16].

In recent years, the use of FEM simulations has made it possible to improve the characteristics of industrial production processes [17, 18] and the materials used, especially in the medical field [19–23], as well as the studies conducted to simulate the dynamic behavior of human joints [24–26]. For these reasons, in this work models based on usual shoulder positions held by violinists during their practice will be developed and analyzed, considering the role of the rotator cuff in the glenohumeral joint. The representative positions have been obtained from the observation of the position assumed by the arms throughout a routine [27–29]. It has been possible to recreate the selected positions and measure their risk index under the RULA criterion with the help of CATIA®[VERSION 5R21] design software. The RULA method was developed in 1993 by McAtamney and Corlett, (Nottingham

University) in order to evaluate individual postures of the upper limbs of the body [30]. The next step was to generate the 3D-models of the articulation maintaining the representative positions of the case under study. Finally, the behavior of the glenohumeral joint and the rotator cuff was simulated considering the effects of gravity and assuming the properties of the materials as linear and isotropic. The developed FEM simulations show the results of the safety factor in a different number of cycles. The observation of their behavior under fatigue and the increase in the number of cycles have allowed to know where the first faults can occur and also how many cycles of repetitive movements are required. The used software provides simulations in order to evaluate the fatigue of the materials, by defining a short cycle internal rotation of the humerus from the positions described above. It is estimated that there is a type of short movement of the rotating humerus maintained by the violinists when performing a long rotation of the elbow and that occurs for each bow or “bow stroke”.

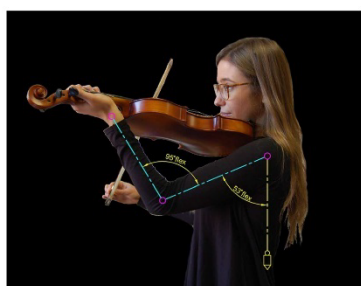


Fig. 4 Flexion angle at 53 degrees of the left arm of the violinist

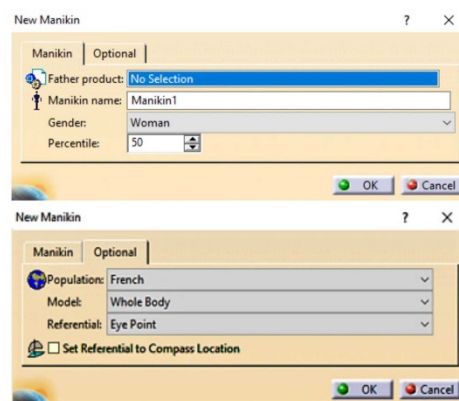
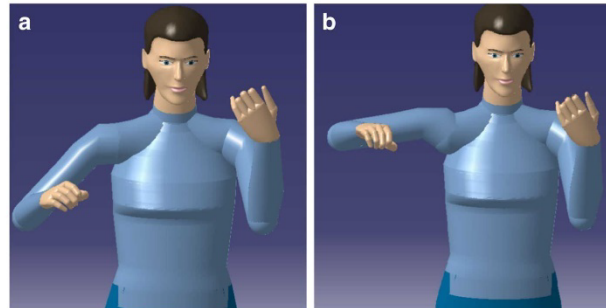


Fig. 5 Ergonomics Tool window of CATIA®



Fig. 6 Flexion angle position of 53 degrees for the left arm and Abduction angle position for the right arm of: **a** 42.5 degrees; **b** 80 degrees



Materials and Methods

RULA Analysis

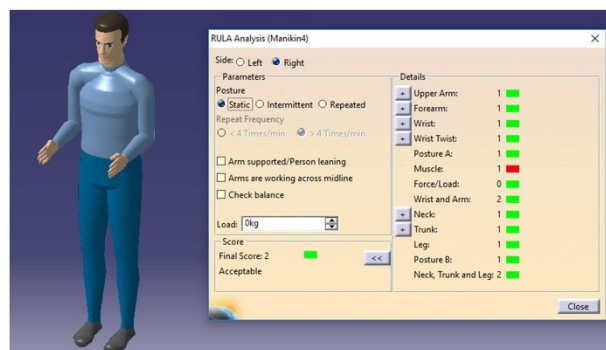
The most representative postural gestures have been extracted from the sonata performed by a professional violinist woman of 24 years hold. The movements, considered extreme in terms of flexion and abduction, that represent the position of the violinist have been measured in CATIA® in terms of angles. An observational, indirect and digital measurement method is functional and is used for the ergonomic evaluation of the postures [27]. Two positions have been obtained for the “mobile” right arm and one for the “fixed” left arm. The violinist’s right arm is in abduction, while his left arm is in flexion. Measuring the minimum and maximum space between a vertical line that assumes the vertical position of the body and the line parallel to the posture of the arms in both situations, the angle of abduction is obtained. In this study, the result of the measurement for the right arm, when the violinist assumes the two positions of abduction shown in Fig. 3, has been 42.5° and 80°. The positions shown in Fig. 3 are the

upper and lower limits of the range of abduction movement performed by the arm during this particular practice.

An almost constant position for the left arm can be assumed, with the exception of the movements made with the fingers and the wrist (Fig. 4). The same method is also used to size the flexion angle assumed by the violinist’s left arm during execution. It is considered a vertical line that represents the upright posture of the body. This line, together with the lines parallel to the profile of the left arm, helps to define the flexion angle of the shoulder and of the elbow. As a result, the left arm has a flexion position of 53° with respect to the body (Fig. 4), with an elbow flexion of the 95.8° with respect to the forearm and an adduction of 4° of the elbow with respect to the shoulder (Fig. 3).

Two possible complete postures, once obtained the abduction limits of the right arm and the fixed position of flexion of the left arm, have been analyzed. These postures are generated in CATIA® by using “Ergonomics Design and Analysis” tool. This tool offers the possibility of creating mannequins with human morphology and interacting with them to simulate daily human activities. As a reference avatar,

Fig. 7 Result of RULA analysis on a resting position



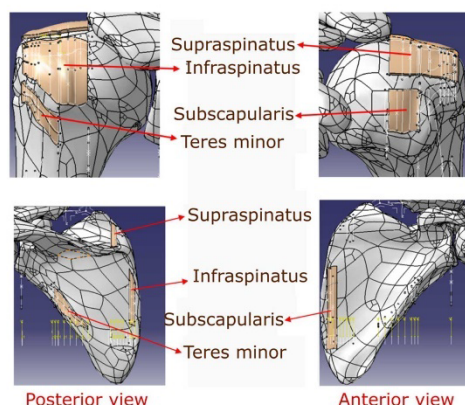


Fig. 8 Models of tendons

a human body of a Caucasian woman belonging to the 50th percentile of the population was created (Fig. 5).

Figure 6 shows the two obtained positions.

The left arm is placed with a flexion of 53° and 4° of adduction of the shoulder and with a flexion of 95° for the elbow, as obtained from Fig. 4. In Fig. 6a, the right arm is positioned with an abduction of 42.5° , and in Fig. 6b with an

abduction of 80° . Once the postures have been created it has been possible to perform a RULA analysis for each one of them. Assuming that the right arm of the body performs the described abduction postures more than 4 times per minute (common routine for a violinist). The RULA analysis results show by means of a numerical category the levels of action that must be taken on said position. Scores between 1 and 2 indicate that the risk of the task is acceptable and that no changes are necessary, between 3 and 4 indicate that a deeper study of the position is necessary because changes may be required, between 5 and 6 indicate that the changes are necessary and finally 7 indicates that the changes are urgent. Figure 7 shows the result of the RULA analysis on a resting position where the values are all between 1 and 2.

3D Model

The glenohumeral joint is composed of different sets of tissues such as: bones, ligaments, cartilages, synovial fluid bags, tendons and muscles. In this study the dynamic behavior of those tissues directly related to the support, at the level of efforts, of the joint (bones, tendons and muscles) have been simulated. Using 3D-Builder, a platform that allows to soften textures, to scale, to measure and to cut with some ease, the file with

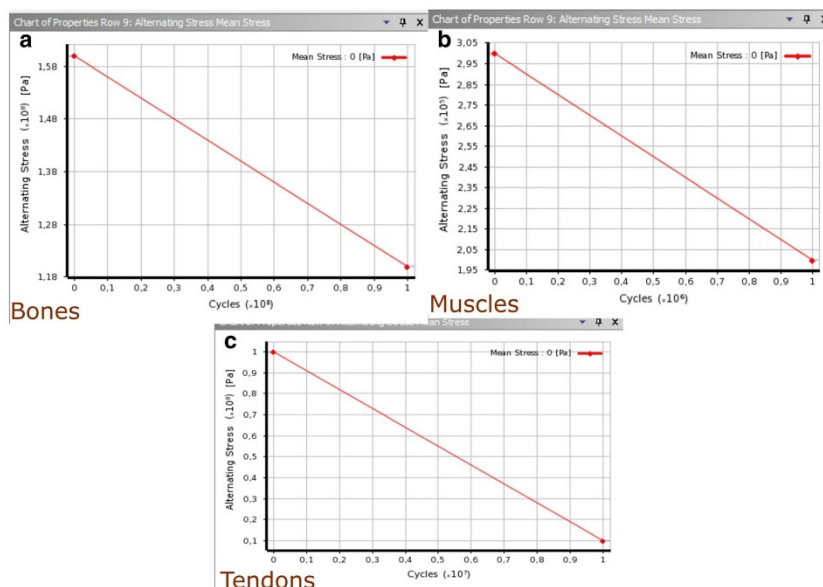


Fig. 9 Material Properties and material behaviors for: a Bones, b Muscles, c Tendons



Table 1 Material properties for the bones, muscles and tendons

	Bones	Muscles	Tendons
Density [kg mm^{-3}]	1,79E-06	1,35E-06	1,35E-06
Isotropic Elasticity			
Derive from	Young's modulus	Young's modulus	Young's modulus
Young's Modulus [MPa]	17200	0,16	2000
Poisson's Ratio	0,41	0,43	0,43
Bulk Modulus [MPa]	31852	0,38095	4761,9
Shear Modulus [MPa]	6099,3	0,055944	699,3
Tensile Yield Strength [MPa]	120	0,1	50
Tensile Ultimate Strength [MPa]	160	0,3	100

extension *.stl has been imported from 3D-scanning of the bony surfaces that make up the shoulder. The first step was to adjust the size of reconstructed bones taking into account the size ratio of the humeral head that depend on sex, age and race. For the case under study, the work of G. R. Milner has been taken into account, which has estimated the average value of the humeral head diameter, in the case of a Caucasian woman from Crete, of 41.2 mm [31]. Once the bones have been obtained in real scale and through a texture smoothing process, the three *.stl files are exported in the CATIA® software. Under the recommendations of Rockwood [32], it has been possible to develop and to locate the tendons in the 3D-model. Once the tendons have been obtained it has been possible to generate the muscles. The result has been a 3D-model of joints consisting of two bones, 4 muscles and 8 tendons (Fig. 8).

In order to guarantee the actual position for the Humerus in the 3D-model, an auxiliary sketch and feature restrictions have

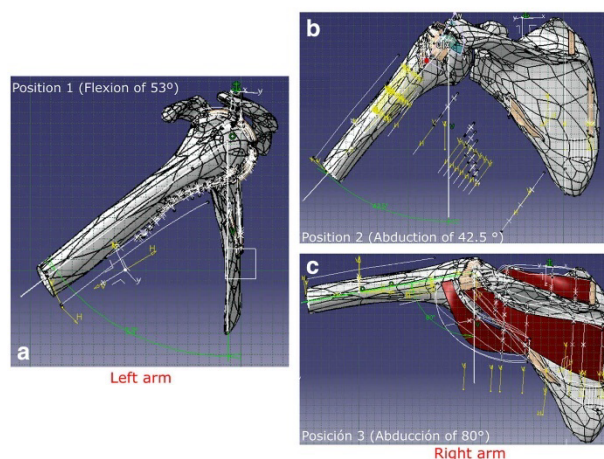
been used. These restrictions represent the abduction degree or amplitude of flexion between the humerus and the body. All models have been generated in the required positions and imported in ANSYS®.

Materials

Taking into account the CES-Edupack database® [33] and recent works [6–8, 34] it has been possible to evaluate the properties of the tissues (Fig. 9 and Table 1).

The fatigue behavior of the tendons has been evaluated considering the value in percentage of Ultimate Tensile Strength for different amounts of cycles, by performing tests in vitro on human tendons [35]. Subsequently, it has been possible to generate in ANSYS the material behavior under fatigue conditions according to a linear approximation.

Fig. 10 Position of the left arm with a flexion angle of 53° and with a rotation angle of 4° of the elbow with respect to the shoulder (a). Position of the right arm with abduction of 42.5° (b) and 80° (c). CATIA®



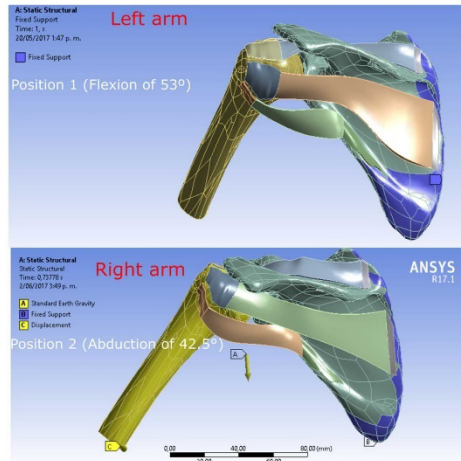
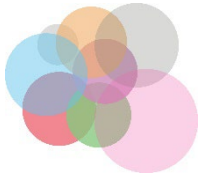


Fig. 11 Models for the ANSYS® simulation of the left arm

Results

From the RULA analysis it can be deduced that:

- The left hemisphere of the body suffers more and therefore is more at risk in its posture than the right in the two analyzed postures.
- In the position of Fig. 6b, when the maximum possible abduction is reached using only the arm without using the clavicle, it is when it is less exposed to injuries compared to the two evaluated positions.
- The results suggest that for the right hemisphere, attention must be paid because it may be that some change in technique is required to prevent injuries.

The results show that the highest risk is taken by the left arm in a position of sustained flexion with a score of 5. Follow the right arm in the case of a repetitive movement to reach the abduction position of 42.5° and of abduction of 80° with a score of 4. The analysis corroborates that violinists will have

injuries when adopting this type of position. However, the analysis cannot locate what area or tissue of the shoulder in which it will occur. For this reason, in this study the dynamic behavior of the joint has been simulated, by using a 3D model of the general shoulder morphology.

Figure 10 shows the positions obtained with the reconstruction in CATIA® of the shoulder joint for the left arm with a flexion of 53°(a) and for the right arm with an abduction angle of 42.5°(b) and 80°(c). It is worth mentioning that all the images of the right shoulder in CATIA® and ANSYS® are made on a left shoulder model, but with the data of the right shoulder.

Figure 11 shows the models for the ANSYS® simulation of the left arm with a flexion of 53°(a) and for the right arm with an abduction angle of 42.5°(b).

The way in which the simulation software evaluates the fatigue in the different models is based on the evaluation of the behavior of the S-N material curves [35–37] and the Stress-Life criterion, by using the following equation:

$$\frac{\Delta\sigma}{2} = \sigma_f (2N_f)^b \quad (1)$$

σ_f = Fatigue strength coefficient (Y intercept in SN curve)
 b = Fatigue strength exponent (slope of data in SN curve)

Once the geometries and the starting positions have been generated, the properties of the aforementioned materials are assigned and the dynamic behavior of the shoulder is simulated at 10, 10800 and 50000 cycles. As the arc strike occurs in a time equal to 1 s, 10 cycles represent 10s, this means that 10800 cycles represent 3 h. This represents the estimated time of daily and continuous practice of a violinist.

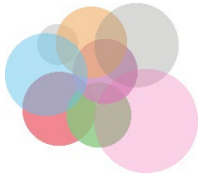
Table 2 shows the results that relate the position to the critical risk zone.

Discussions

The results represent the time in which the area under study begins to show problems in order to continue resisting the conditions of cyclic loading. The approximations used in the

Table 2 Results of ANSYS simulation for right arm and for the left arm

Position (degrees)	Critical zone (Connection)	Minimum life cycle safety factor 1 (Cycles)	Time (Hours)
Flexion 53°	Supraspinatus Muscle - Tendons Subscapular muscle-tendon Humerus	6218	1,7
Abduction 42.5° with a rotation of 4° of the elbow with respect to the shoulder	Supraspinatus Muscle-Tendons Infraspinatus Muscle-Tendons	7617	2,1
Abduction 80°	Teres Minor Muscle- Tendons	14903	4,14



simulation have not considered the variations in the abduction angle, in the cycle time and in the depth of the rotation that depend on the order of the musical notes that will be executed. These variables are difficult to quantify and probably will dilate the fatigue effect. These variations are important for the arm and elbow, but the shoulder will continue to have short and cyclical movements. The supraspinatus muscle is affected by both movements of flexion and abduction. Even when the movement has a great amplitude, (80° or 90°), the musician can suffer the syndrome of entrapment (compression syndrome). Due to these results, it could be said that this connection with their respective tendons is the critical zone. Fatigue can manifest itself in a matter of hours due to unnatural, highly frequent or sustained positions.

The critical zones of the FEM simulation have been found in the flexion movements of 53° of the left shoulder, and in abduction of the right shoulder of 42.5° and 80° (Table 2).

The detected critical points are the origin of the supraspinatus, the origin and insertion of the infraspinatus and the insertion, origin and muscular belly of the teres minor for abduction of 42.5° . This can be explained considering that the supraspinatus is the starter or initiator of abduction. It is fundamental in the first grades of abduction, but reaches its maximum efficiency and electromyographic activity around 90° . The deltoid is its main synergistic muscle in the first time of abduction (up to 90°), and it offers its effectiveness in the last stages of the abduction movement of this first time, so it

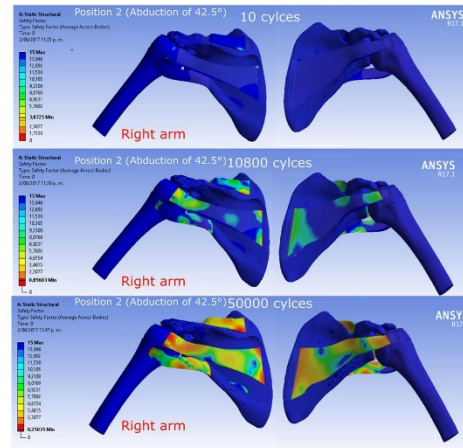


Fig. 13 Results of the fatigue simulation for 10, 10800 and 50000 cycles for the right arm with an Abduction angle of 42.5 degrees

collaborates when approaching 80° , not at the beginning [38]. Something similar happens with the long biceps [38].

The least fatigue was detected for an abduction of 80° . In this case the critical points are the origin of the supraspinatus and the origin and insertion of the infraspinatus. This can be explained considering that the abduction of the arm to execute the technical movement of the violin is not a pure movement. It automatically carries with it a progressive flexion of the arm in which the biceps and the trapezius collaborate, and a

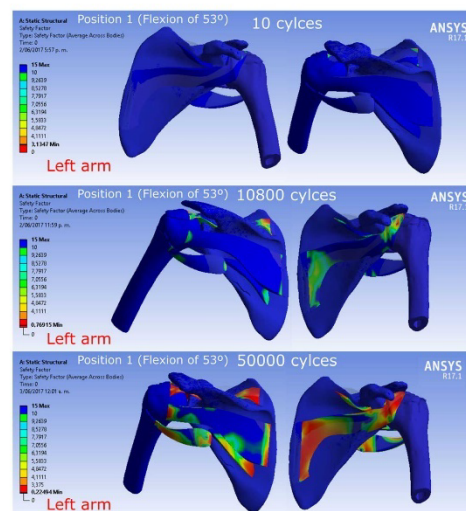


Fig. 12 Results of the fatigue simulation for 10, 10800 and 50000 cycles for the left arm with a flexion angle of 53 degrees and with a rotation of 4° of the elbow with respect to the shoulder

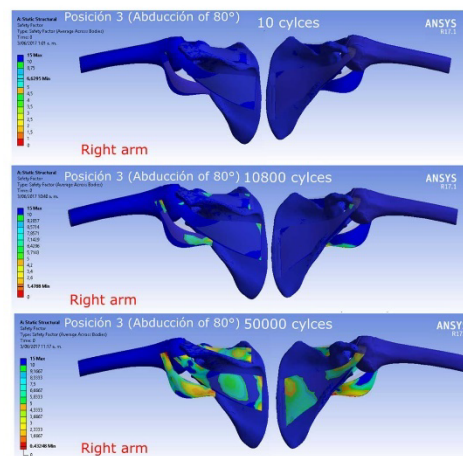
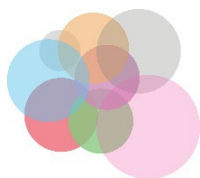


Fig. 14 Results of the fatigue simulation for 10, 10800 and 50000 cycles for the right arm with an Abduction angle of 80 degrees



simultaneous external rotation of the arm on its longitudinal axis (Codman Paradox) [38–40]. In addition to this, when the arm abduction approaches 90°, the supraspinatus tendon reaches the horizontal and achieves an anatomically favorable position (Figs. 12, 13 and 14). This is due to the insertion, the origin of the supraspinatus muscle and the start of the supraspinatus tendon are contained in a horizontal plane that favors its passage through the supraspinatus canal [38].

Conclusions

In this study the level of risk was evaluated, observing and measuring the representative positions of the routine of a musician playing the violin by performing a RULA analysis. Subsequently, thanks to the simulation in ANSYS it has been possible to predict, approximately, the number of cycles associated with a time of activity related to the routine for which a possible injury will be generated. This work has allowed to identify and locate the first joint fault. The results imply that this is the limit that the violinist can afford in order to practice continuously and that after this limit she will feel an increase in fatigue and pain. Future studies can be useful to determine an approximation of the life cycles of tissues by assuming a continuous rhythm for movement. However, they are living tissues that have associated a recovery rate that prolongs their useful life. The result should be taken as the pain threshold that the violinist will suffer when his safety factor is less than 1. Pain is the main signal that the body must feel before a possible injury. Also with the developed methodology of analysis and simulation it will be possible to extend the study to other joints and/or case of study where the subject performs repetitive movements or routines.

Compliance with Ethical Standards

Conflict of Interest Manuel Islan declares that she has no conflict of interest. Fernando Blaya declares that he has no conflict of interest. Pilar San Pedro declares that he has no conflict of interest. Roberto D'Amato declares that he has no conflict of interest. Emilio Lechosa Urquijo declares that he has no conflict of interest. Juan Antonio Juanes declares that he has no conflict of interest.

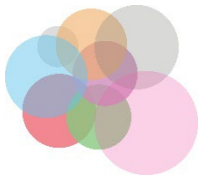
Ethical Approval This article does not contain any studies with human participants or animals performed by any of the authors.

References

1. Brunelli, M.P., and Gill, T.J., Fractures and tendon injuries of the athletic shoulder. *Orthop. Clin. North Am.* 33(3):497–508, 2002.
2. Calle, Y., and Hincapié, S., *CES Movimiento y Salud.*, vol. 2, no. 1, 2014.
3. Allman, F.L., Fractures and ligamentous injuries of the clavicle and its articulation. *J. Bone Joint Surg. Am.* 49(4):774–784, 1967.
4. McKoy, B.E., Bensen, C.V., and Hartsock, L.A., Fractures about the shoulder: conservative management. *Orthop. Clin. North Am.* 31(2):205–216, 2000.
5. Fiff, M. P., Hatch, R., and Higgins, M. K., Fracture management for primary care. Saunders/Elsevier, 2012.
6. Ellenbecker, T. S., Clinical examination of the shoulder. W.B. Saunders, 2004.
7. Moore, K. L., Agur, A. M. R., Dalley, A. F., and K. L. D. of (work) Moore, Essential clinical anatomy.
8. Quillen, D.M., Wuchner, M., and Hatch, R.L., Acute shoulder injuries. *Am. Fam. Physician.* 70(10):1947–1954, 2004.
9. Zaza, C., and Farewell, V.T., Musicians' playing-related musculoskeletal disorders: An examination of risk factors. *Am. J. Ind. Med.* 32(3):292–300, 1997.
10. de Moraes, G.F., and Antunes, A.P., Musculoskeletal disorders in professional violinists and violists. Systematic review. *Acta Ortop. Bras.* 20(1):43–47, 2012.
11. Lederman, R.J., Neuromuscular and musculoskeletal problems in instrumental musicians. *Muscle Nerve.* 27(5):549–561, 2003.
12. Schuele, S.U., and Lederman, R.J., Occupational disorders in instrumental musicians. *Med. Probl. Perform. Art.* 19(3):123–129, 2004.
13. Heming, M.J.E., Occupational injuries suffered by classical musicians through overuse. *Clin. Chiropr.* 7(2):55–66, 2004.
14. Nyman, T., Wiktorin, C., Mulder, M., and Johansson, Y.L., Work postures and neck-shoulder pain among orchestra musicians. *Am. J. Ind. Med.* 50(5):370–376, 2007.
15. Marqués, F., Directrices para la Decisión Clínica en Enfermedades Profesionales. Instituto nacional de seguridad e higiene en el trabajo, Madrid. Instituto nacional de seguridad e higiene en el trabajo, Madrid, 2010.
16. Rietveld, A.B.M., Dancers' and musicians' injuries. *Clin. Rheumatol.* 32(4):425–434, 2013.
17. Wojciechowski, S., Maruda, R.W., Barrans, S., Nieslony, P., and Krolczyk, G.M., Optimisation of machining parameters during ball end milling of hardened steel with various surface inclinations. *Measurement.* 111:18–28, 2017.
18. Jarosz, K., Löschner, P., Nieslony, P., and Krolczyk, G., Optimization of CNC face milling process of Al-6061-T6 aluminum alloy. *J. Mach. Eng.* 17(1):69–77, 2017.
19. Ruggiero, A., Merola, M., and Affatato, S., On the biotribology of total knee replacement: A new roughness measurements protocol on in vivo condyles considering the dynamic loading from musculoskeletal multibody model. *Measurement.* 112:22–28, 2017.
20. Ruggiero, A., Affatato, S., Merola, M., and De, M. C., FEM analysis of metal on UHMWPE total hip prosthesis during normal walking cycle. In *Proceedings of the XXIII conference of the Italian Association of Theoretical and Applied Mechanics (AIMETA 2017)*, pp. 4–7, 2017.
21. Merola, M., Ruggiero, A., De Mattia, J.S., and Affatato, S., On the tribological behavior of retrieved hip femoral heads affected by metallic debris. A comparative investigation by stylus and optical profilometer for a new roughness measurement protocol. *Measurement.* 90:365–371, 2016.
22. Ruggiero, A., D'Amato, R., and Gómez, E., Experimental analysis of tribological behavior of UHMWPE against AISI420C and against TiAl6V4 alloy under dry and lubricated conditions. *Tribol. Int.* 92:154–161, 2015.
23. Ruggiero, A., Amato, R.D., Gómez, E., and Merola, M., Experimental comparison on tribological pairs UHMWPE/TiAl6V4 alloy, UHMWPE/AISI316L austenitic stainless and UHMWPE/AL2O3 ceramic, under dry and lubricated conditions. *Tribol. Int.* 96:349–360, 2016.
24. Ruggiero, A., Gómez, E., and D'Amato, R., Approximate analytical model for the squeeze-film lubrication of the human ankle joint



- with synovial fluid filtrated by articular cartilage. *Tribol. Lett.* 41(2): 337–343, 2011.
25. D'Amato, R., Calvo, R., and Gómez, E., Sensitivity study of the morphometric fitting on the pressure field inside ankle joints. *Case Stud. Mech. Syst. Signal Process.* 1:8–14, 2015.
26. Ruggiero, A., Gómez, E., and D'Amato, R., Approximate closed-form solution of the synovial fluid film force in the human ankle joint with non-Newtonian lubricant. *Tribology International.* 57: 156–161, 2013.
27. Diego-Mas, J.-A., Poveda-Bautista, R., and Garzon-Leal, D.-C., Influences on the use of observational methods by practitioners when identifying risk factors in physical work. *Ergonomics.* 58(10):1660–1670, 2015.
28. David G. C., “Ergonomic methods for assessing exposure to risk factors for work-related musculoskeletal disorders.”
29. LI, G., and BUCKLE, P., Current techniques for assessing physical exposure to work-related musculoskeletal risks, with emphasis on posture-based methods. *Ergonomics.* 42(5):674–695, 1999.
30. Mcatamney, L., and Corlett E. N., RULA: a survey method for the investigation of world-related upper limb disorders. *Appl. Ergon.* 19(2):91–99.
31. Milner, G.R., and Boldsen, J.L., Humeral and femoral head diameters in recent white American skeletons. *J. Forensic Sci.* 57(1):35–40, 2012.
32. Rockwood, C.A., Matsen, Wirth, and Lippitt. Hombro, Márban, 2013.
33. CES Edu Pack. Granta Design, Cambridge, 2017.
34. Holzapfel, G., Biomechanics of soft tissue. *Handb. Mater. Behav.* 3(1):1049–1063, 2000.
35. Schechtman, H., and Bader, D. L., In vitro fatigue of human tendons. *J. Biomech.* 30(8):829–835, 1997.
36. Chen, E. J., Novakofski, J., Jenkins, W. K., and O'Brien, W. D., Young's modulus measurements of soft tissues with application to elasticity imaging. *IEEE Trans. Ultrason. Ferroelectr. Freq. Control.* 43(1):191–194, 1996.
37. Kent-Braun, J. A., Fitts, R. H., and Christie A., “Skeletal muscle fatigue,” In: *Comprehensive Physiology*, vol. 2, no. 2, Hoboken, NJ: John Wiley & Sons, Inc., pp. 997–1044, 2012.
38. I. A. Ibrahim A. Kapandji, E. Martínez, and A. Jornet Cases, *Cuadernos de fisiología articular: esquemas comentados de mecánica articular*. Barcelona: Toray-Masson, 1982.
39. Cheng, P. L., Simulation of Codman's paradox reveals a general law of motion. *J. Biomech.* 39(7):1201–1207, 2006.
40. Pearl, M. L., Sidles, J. A., Lippitt, S. B., Harryman, D. T., and Matsen, F. A., Codman's paradox: Sixty years later. *J. Shoulder Elb. Surg.* 1(4):219–225, 1992.



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ANEXO V:

BEHAVIOR UNDER LOAD OF A HUMAN SHOULDER: FINITE ELEMENT SIMULATION AND ANALYSIS

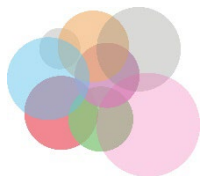
La mayoría de las lesiones músculoesqueléticas ocurren durante las rutinas de trabajo en diferentes áreas, debido a movimientos repetitivos y sostenidos, que a menudo se encuentran en el hombro. Para los trabajadores del sector de la construcción, los movimientos y desplazamientos repetitivos se producen bajo carga y en posturas antinaturales de la articulación del hombro.

Este estudio pretende modelar en 3D los componentes biológicos que forman la articulación del hombro para el posterior análisis de elementos finitos. Para este estudio se han considerado tres casos con diferentes cargas.

Gracias a un abordaje articular lineal e isotrópico se ha podido evaluar las tensiones en los principales componentes de los hombros:

Músculos, tendones y ligamentos. La metodología utilizada permitió obtener una malla mejorada de la articulación del hombro para analizar situaciones con análisis del método de elementos finitos con aplicaciones en el campo de la medicina deportiva, laboral, etc. Además, la simplificación adoptada para modelar los músculos de las articulaciones, ya que los elementos 1D en el modelo de elementos finitos han hecho posible establecer diferentes posiciones de articulación humana sin mallar nuevamente cada una de las posiciones estudiadas. Los resultados son consistentes con los estados de cargas aplicadas.

Las tensiones máximas en los huesos se encuentran en las zonas de inserción de los ligamentos. Debido a la estática en las posiciones de la articulación en estudio, los músculos no soportan tensiones elevadas. Según la distribución de tensiones, los valores están localizados en las zonas de los tendones. Del análisis de resultados se observa cómo la distribución de tensiones en la zona del cartílago mantiene coherencia con la realidad ya que las tensiones máximas aparecen en la mitad inferior del cartílago. En esta zona en la que el contacto húmero-escápula es mayor, las tensiones de compresión son mayores.



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EDUCATION & TRAINING



Behavior under Load of A Human Shoulder: Finite Element Simulation and Analysis

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Abstract

Most musculoskeletal injuries occur during the work routines in different areas, due to repetitive and sustained movements, they are often located in the shoulder. For workers in the building sector, the repetitive movements and displacements occur under load and unnatural postures of the shoulder joint. For this reason, this study aims to model in 3D the biological components which form the shoulder joint for the later finite element analysis. Three cases with different loads have been considered for this study. Due to a linear and isotropic joint approach it has been possible to evaluate the tensions in the main components of the shoulders: muscles, tendons and ligaments. The methodology used allowed obtaining an improved mesh of the shoulder joint to analyse real situations with finite element method analysis with applications in the field of sports medicine, work, etc. Furthermore, the simplification adopted for modelling the joint muscles, as 1D elements in the finite element model has made it possible to establish different positions of human joint without mesh again each of the studied positions. The results are consistent with the states of applied loads. In fact, the maximum stresses in bones are in the insertion areas of the ligaments. Due to the static positions of the joint under study, the muscles do not support high stresses. According to the stresses distribution, the maximum values are in the zones of tendons. From the result analysis, it is observed how the stresses distribution in the cartilage area maintains coherence with reality since the maximum stresses appear in the lower half of the cartilage. In this area in which the Humerus-scapula contact is greater, the compression tensions are greater.

Keywords Finite element analysis · Shoulder · Humerus · Muscles · Tendons · Ligaments

Introduction

Due to its high mobility, the shoulder is the Human joint that has the most injuries in its tissues [1]. The

rotator cuff is formed by the tendons of the subscapular, supraspinatus, infraspinatus, and minor rotator cuff muscles that meet at the shoulder joint [2, 3]. It is placed in the arch called coracoacromial, constituting the

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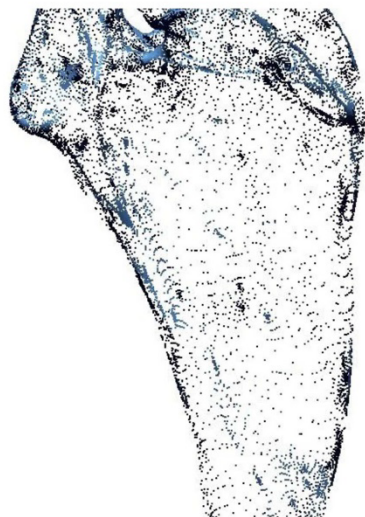


Fig. 1 Cloud of points obtained through a resonance

subacromial space with a serous pouch that lubricates the cuff. This pathology is characterized by progressive pain as a result of an accumulated overload or acute after an overexertion, which is located on the upper or lateral side of the shoulder and increases with movement, mainly with the elevation of the arm (ex. when

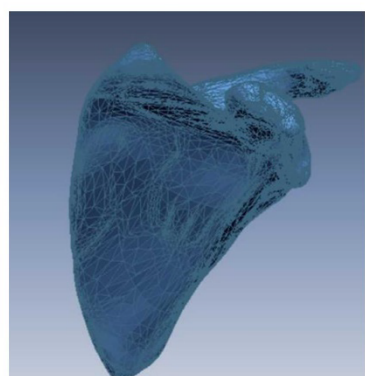


Fig. 2 Mesh corresponding to the scapula

combining) and with internal rotation (ex. when putting on a jacket) [4]. The pain may be more acute at night when leaning on the affected side. The chronic injury can cause a rupture of the tendons of the cuff. Chronic tendinitis of the rotator cuff may be related to shoulder overload, joint instability and rotator cuff degeneration with age. It is relatively frequent, reaching up to 18% in workers in heavy manual tasks with great efforts on the shoulders, associated or not with repetitive movements and forced postures. 75% of people over 50 years have rotator cuff injuries and most of them are asymptomatic [5]. Many jobs in different trades in the construction sector involve important musculoskeletal efforts. This pathology is associated with work that involves stabilizing the shoulders to maintain or move the elbows in an elevated position by force, lifting and catching continuously, or working the arm in flexion or abduction. In the construction sector, this occurs in the assembly of metallic or prefabricated structures, also of heavy prefabricated elements that must be manipulated to unload. In order to understand the behaviour of human joints, in recent years, several authors have proposed scientific studies with finite element model [6] as well as the studies conducted to simulate the dynamic behaviour of human joints. For example, Ruggiero et al. in their study present a new mathematical model in order to understand the tribological performances of the human ankle joint with synovial fluid filtrated by articular cartilage in the case of Newtonian fluid and of no Newtonian fluid [7, 8]. D'Amato et al. in their study investigate the importance of the morphometric parameter of the human joint for the mathematical modelling of the human ankle joint [9].- Many authors, in recent years, have focused their studies on the description of the dynamic behavior of the shoulder joint. For example, in the 1998 Luo et al. present for the first time a simplified 2D-finite element model in order to described the stress environment in the supraspinatus tendon [10]. Wakabayashi et al. in the 2003 performed 2D-finite element model analysis in order to estimate the mechanical environment of the supraspinatus tendon. In the 2008 Seki et al. [11] presented the first 3D-finite element model of rotator cuff tears in order to study the stress distribution in the supraspinatus tendon. Most of these studies have a substantial limitation. In fact, in these studies only the scapula and the bones of the Humerus were considered. This means that in presenting their results both in the 2D and 3D geometry, the structure of the muscles and other soft tissues have been neglected [12]. For this reason, in this study a finite

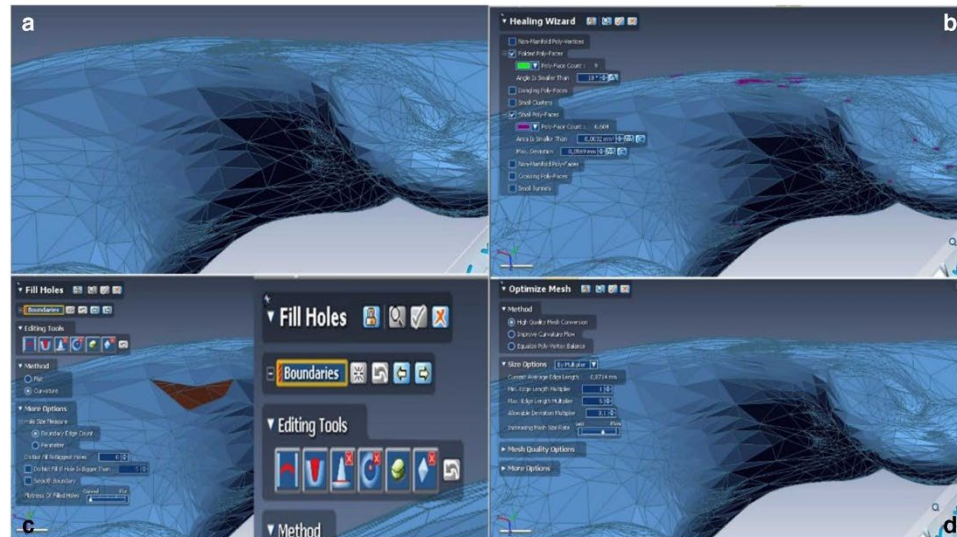
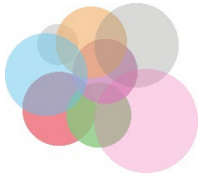


Fig. 3 Mesh and cloud of points operations

element model of the shoulder joint, glenohumeral, classified as a diarthrosis, of the genus enarthrosis or spheroid is presented. Based on the finite element model made for the case of a violinist [3] they are carried out modifications and improvements that allow us

increasing the accuracy of the study and improving the predictive performance of it. This is due to include the ligaments, glenohumeral and coracohumeral, and the glenohumeral-capsule in the anatomical model. In the same way the meshing process is improved, by working

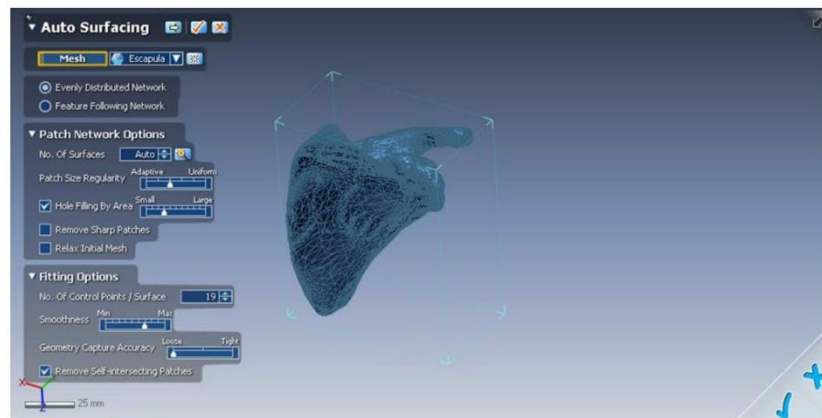


Fig. 4 Reconstruction of surfaces from the cloud of points

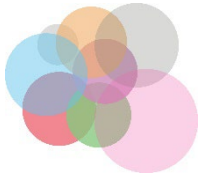


Fig. 5 Surface generated by Geomagic Design X associated with the scapula

on the characterization of the tendon-bone insertion, achieving a transfer of efforts between both more precise and in accordance with reality.

The study focuses on a unique position shoulder, flexion 53°. The scope of work aims the next aspects:

- Improving the mesh used in the finite element model.
- Improving the characterization of the contact tendon-bone insertion area. To do this, meshes between both elements have been made congruent, with nodes coinciding on the interface surface.
- Including the main ligaments involved in the joint, glenohumeral ligaments (lower, middle and upper) and coracohumeral ligament.

Fig. 6 Process to model tendons

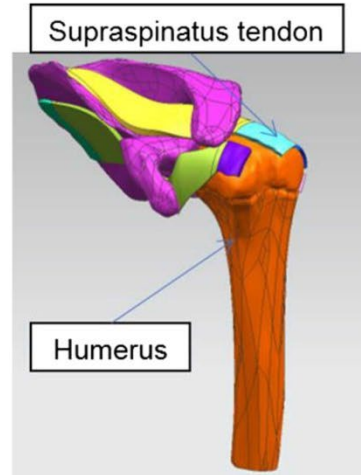
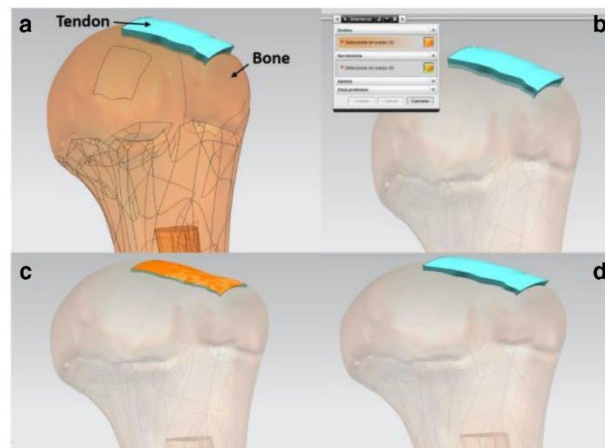


Fig. 7 Location of the supraspinatus tendon and the head of the humerus

- Obtaining a qualitative model of joint behavior.

Methodology

The process to generate a 3D model from a cloud of points is described in the following steps:

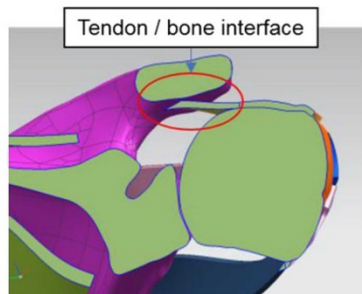
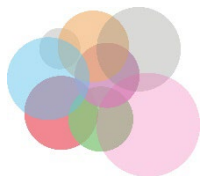


Fig. 8 Sectional view of the humerus + supraspinatus tendon

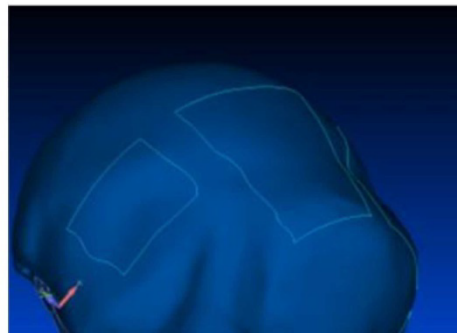


Fig. 10 Outline of the tendon projected on the Humerus

Starting from the cloud of points (Fig. 1) obtained through a resonance. A 3D Computer Aided Design (CAD) model of the elements involved in the scapula.

Using the software Geomagic Design X, it is possible to create a mesh supported in the obtained cloud of points (Fig. 2). The mesh is a triangulation of the surface of the object joining the points of the cloud.

Among the various operations that can be performed using the software Geomagic Design X, all of them are aimed at improving the cloud of points and its associated mesh, are the following (Fig. 3):

- Repairing the mesh, by creating points in the cloud, based on the existing points in it (Fig. 3a).
- Filling holes in the mesh. The mesh is completed creating the triangles and points in the cloud that are missing to close the outer surface of it (Fig. 3b). The filling of holes can be done in different ways (Fig. 3c).
- Optimizing the mesh. It is possible to simplify the number of points in the cloud, eliminating both duplicated points and points which do not provide information, decreasing the size of the cloud (Fig. 3d).

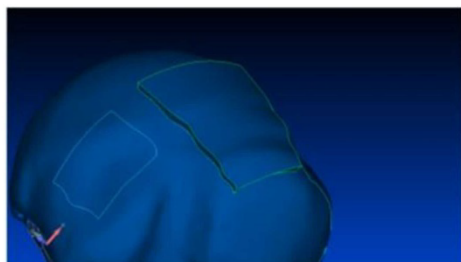


Fig. 9 View of the Humerus + supraspinatus tendon

In addition to these operations, there are many other operations aimed at modifying the cloud of points such as symmetry operations, copy points, scaling of the piece, etc. These operations are also included in commercial Computer Aided Design (CAD) systems.

The most important operation that Geomagic Design X offers us, is the reconstruction of surfaces based on the cloud of points. The result of this operation is an entity that can be treated in any commercial CAD system. The importance of this operation lies in the possibility of using the reconstructed entity for other purposes such as modification of the geometry in a commercial CAD system, integration of the entity in a set of pieces, finite element analysis operations.

Table 1 Types of elements for each body that intervenes in the joint

Nº	Element type	Parameters	Application
1	Tetrahedral 10 nodes	Material, Young modulus and Poisson coef.	Meshing bones, muscles and tendons
2	Beam element 2 nodes	Material, Young modulus and Poisson coef. and transverse section	Mesh of the ligaments and the glenohumeral capsule.
3	Rigid element 2 nodes	none	Simulation insertion of the ligament in the bone.
4	Gap element 2 nodes	Gap and stiffness of the contact.	Simulation of the cartilage contact between the head of the Humerus and the scapula.

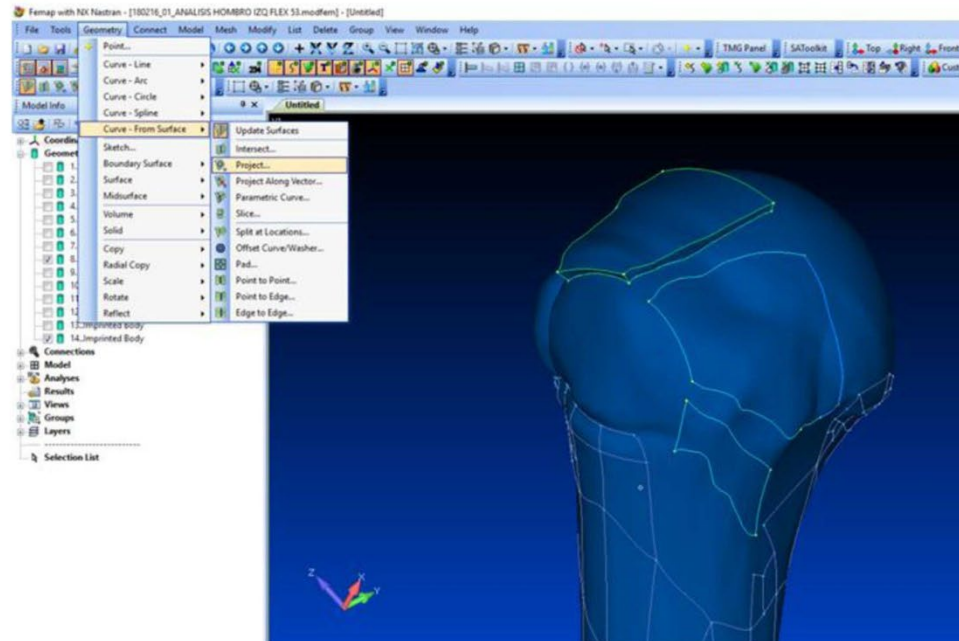


Fig. 11 Contour of the tendons in Siemens FEMAP

The reconstruction of the cloud of points can be done in two different ways:

- Based on a uniformly distributed network. Geomagic Design X is responsible for generating “patches” based

Fig. 12 Projection of the contour

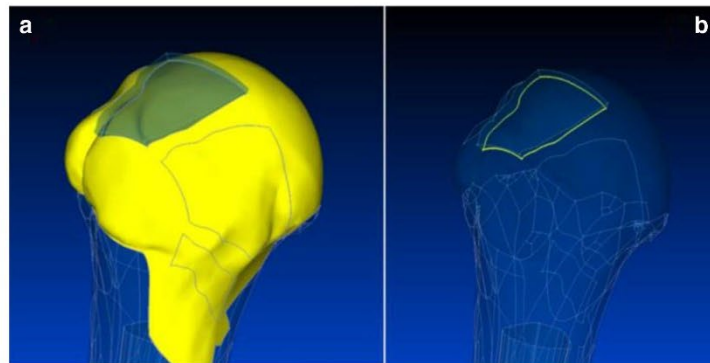
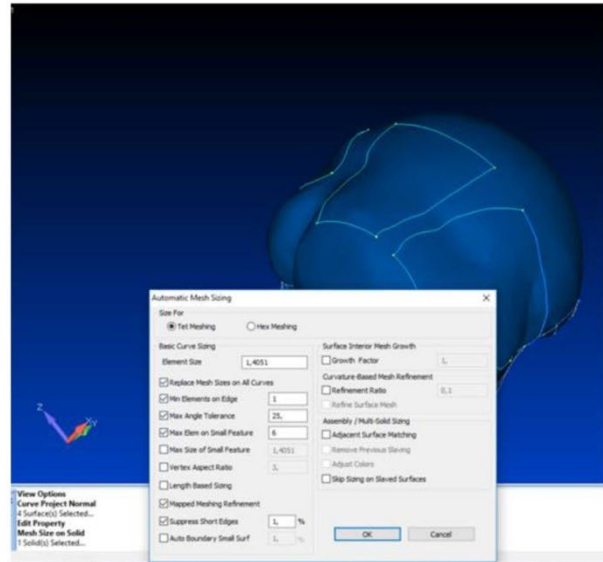




Fig. 13 Mesh Sizing



on the cloud of points to build the outer surface of the piece.

- Based on geometrical operations. Geomagic Design X looks for geometrical surfaces in the point cloud (cylinders, flat faces, etc. ...) to reconstruct the piece.

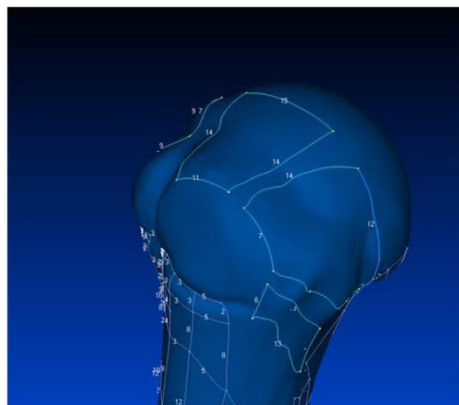


Fig. 14 Number of nodes in each curve

In the case of complicated geometrical pieces, it is recommended to use patch reconstruction. Locating known geometries in a piece as the scapula is complicated so the patches are a suitable solution to obtain the surface of the piece (Fig. 4).

The result of the reconstruction is shown in Fig. 5. The surface of the piece is composed of a series of patches, surfaces of different sizes, which adapt to the

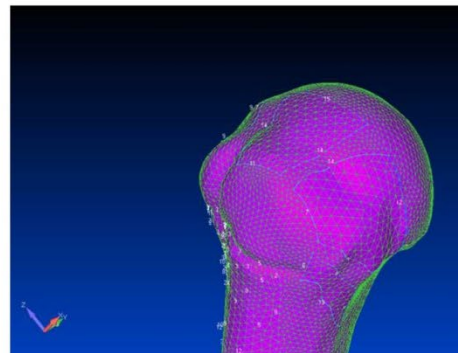
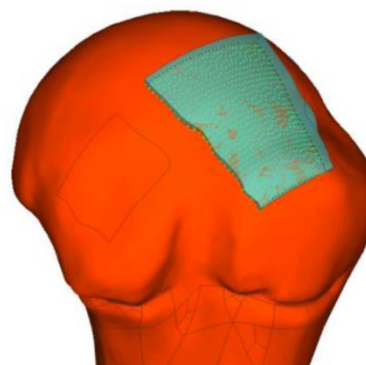


Fig. 15 Obtained mesh



Fig. 16 Nodes located on the tendon-bone interface surface corresponding to the Humerus

V: Unlabeled
L: 50 N horizontal al wrist
C: ESCAPULA FLA
G: 26x27



cloud of points, thus creating the outer surface of the piece and, therefore, its volume.

To verify that the interfaces between tendons and bones are perfectly matched, a 3D CAD, NX tool from Siemens has been used. The basis of the meshing process in these areas is the coincidence of the meshes between both surfaces in contact. To do this, a modeling work on the tendons that ends with a Boolean operation has been carried out.

Taking as an example the insertion of the supraspinatus tendon in the head of the humerus, it is possible to explain the process of modeling the tendon in order to achieve a perfect match. Figure 6a shows that the tendon has a thickness greater than the real one, so that between the tendon and the bone there is an interference between both pieces. by means of a Boolean subtraction operation it is possible to subtract the interference to the tendon volume. As shown in

Fig. 17 Nodes located on the tendon-bone interface surface corresponding to the tendon

V: Unlabeled
L: 50 N horizontal al wrist
C: ESCAPULA FLA
G: Tendon Supraspinoso - Humero

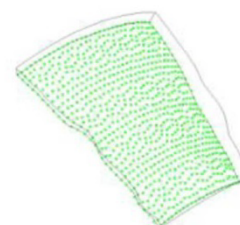




Fig. 18 Nodes fused between the two coinciding surfaces of the tendon and the Humerus

V: Untitled
L: 50° N Horizontal al nivel
C: ESCAPULA.FUA
G: HUMERO

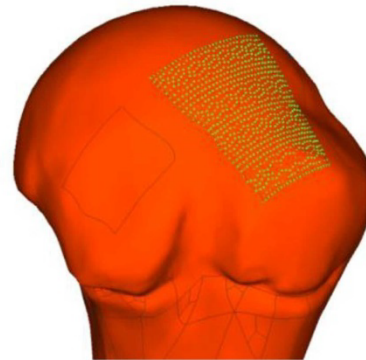


Fig. 6b, it is enough select the two volumes so that Geomagic Design X automatically will calculate the volume of the intersection. The volume of the intersection, the interference between both bodies, is observed in Fig. 6c. subtracting the volume from the tendon, a modeled tendon is obtained with the support surface between the head of the humerus and the coincident tendon between both bodies (Fig. 6d).

This Boolean subtraction operation must be carried out in all the tendons involved in the articulation so that the coincidence between them and the surface in which they are inserted in the bones is perfect. This coincidence is necessary so that during the process of meshing the finite elements a mesh is obtained that shares nodes between the tendons and the bones so that the insertion of the same is modelled correctly.

After performing the modelling operations in which the match between tendons and bones is checked, the obtained model is shown in Figs. 7 and 8.

With the solid models of the articulation elements, a file is generated “*.stp” or “*.igs”. These files are imported into a FEM software (Siemens FEMAP). As in the case described before, it must be ensured matching surfaces between tendon and bone (Fig. 9). In this way both surfaces will be coincident in geometry (achieved with the Boolean operation in the CAD 3D software) and also in the contour of them (Fig. 10). Once verified that the geometries between bone and tendon are coincident, the model mesh is generated. Table 1 summarizes the

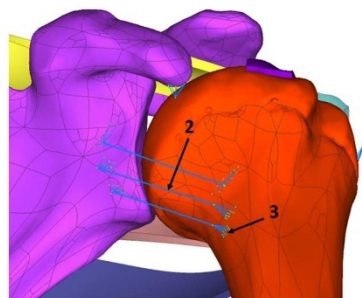


Fig. 19 Rigid element (3) and element 1D type Beam (2) to simulate ligaments of the joint (Table 1)

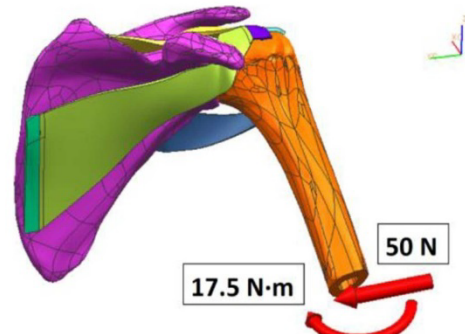


Fig. 20 Load case 1 used in finite element model



Table 2 Materials used for the model

	Young's Modulus (MPa)	Poisson	Density (Kg/m ³)
Bone	17,200	0,41	1790
Muscle	0,3	0,43	1350
Tendon	2000	0,43	1350
Cartilage	8	0,49	1150
Ligament	300	0,43	1350

elements used to model the parts which the human shoulder joint consists of.

This projection of the contour of the tendons on the surface of the bones is achieved in Siemens FEMAP by projecting the curves that make up the contour of the tendon on the insertion surface in the bone (Fig. 11). Then select the surfaces on which the curves are going to be projected, in our case shown in yellow in the Fig. 12a. Finally, the curves wish to project on the surface are selected, in our case, the curves that define the contour of the tendon (Fig. 12b).

The projection is produced according to the normal to the surface at each point, it is also possible to project the curves according to a defined vector. The result of the projection is a partition on the surface (Fig. 13).

Once this operation of projection of curves on the surfaces is done, the solids are obtained in situation of starting the process of generating the finite element mesh to proceed to the calculation of the different load cases. The meshing process is carried out using the types of elements for each body that intervenes in the joint, see Table 1. To generate the mesh must follow the following steps:

1. Generation of the physical properties of bodies.
Consisting of defining the physical characteristics of the finite elements used:
 - a) Characteristics of the material that defines the element, bone, tendon, muscle, etc.

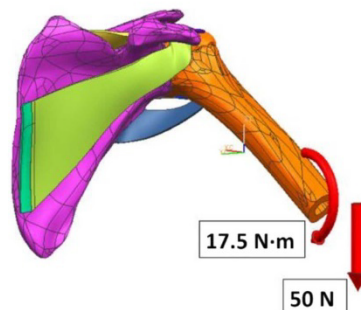


Fig. 21 Load case 2 used in finite element model

- b) Geometrical characteristics of the element, thickness (in the case of using elements 2d), section (in the case of using elements 1d, elastic constant (in the case of using spring elements)
2. A different physical property is assigned to each body involved in the articulation, in this way it is necessary to create as many physical properties as bodies participate in the model.

Once the physical properties of the bodies involved in the articulation have been defined, meshing is carried out. The first step is to define the size of the element to be used, the smaller element is the more accurate the analysis, it is worth mentioning that the calculation process by means of the finite elements method obtains results in the defined nodes, the rest of the model, the spaces without nodes, are calculated by extrapolation between nodes depending on the type of element, and its characteristics, that unites them with each other. However, the smaller the selected element size, the greater the time and resources needed for the calculation, that is why the number of nodes to be used should be minimized, keeping the minimum size necessary to

Table 3 Load cases considered for FEA

Load Case	Load Conditions	Constrain Conditions
1	Gravity applied negative Z axis (9.841 m/s ²)	Fixed scapula on areas shown in Fig. 20
2	Gravity applied along negative Z axis (9.841 m/s ²) + 50 N applied along positive Y axis.	Fixed scapula on areas shown in Fig. 21
3	Gravity applied along negative Z axis (9.841 m/s ²) + 50 N applied along negative Z axis.	Fixed scapula on areas shown in Fig. 22

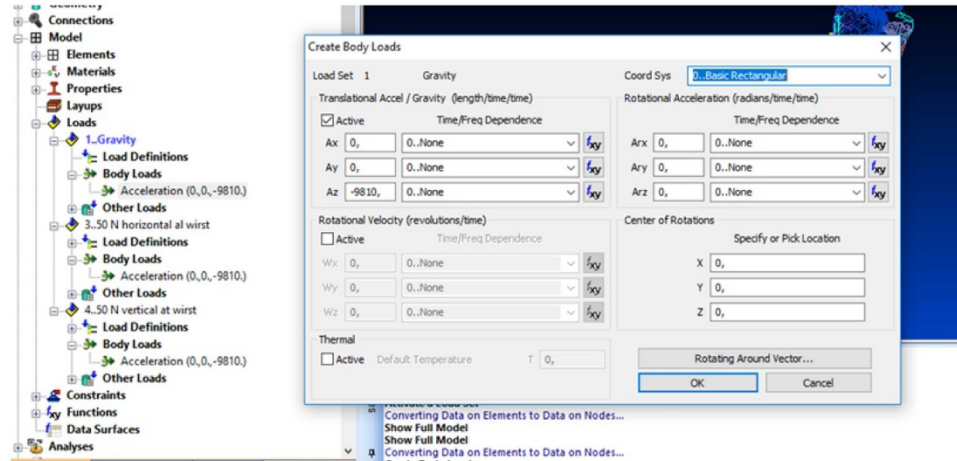


Fig. 22 Effect of gravity applied to finite element model

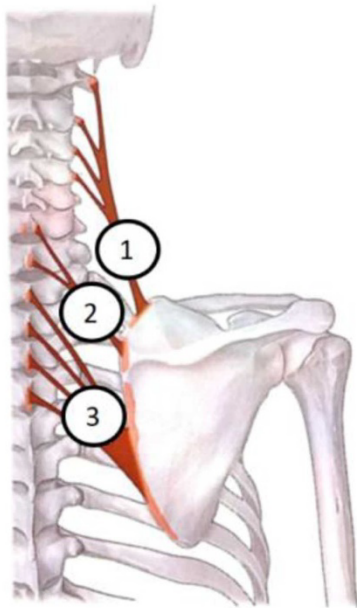


Fig. 23 Muscles attached to the scapula fixed

reproduce with precision the geometry of a discrete mode and a finite number of nodes. To define the element size, select the desired volume and assign it a size. Siemens FEMAP is responsible for locating the nodes in that volume in the optimal way (Fig. 13).

Once the number of nodes that make up the piece have been defined, the mesh of elements is generated. Figure 14 shows the number of nodes that have been created in each curve that defines the body, Siemens FEMAP is responsible for generating the nodes inside the surfaces by interpolation and triangulation of them. After the meshing process of the body the result obtained is a mesh, composed of nodes and elements, which define the geometry of the body (Fig. 15). The process must be repeated in all the bodies that make up the joint.

Once the mesh is generated separately from the muscles, tendons, bones and ligaments, the next step is the fusion of the matching nodes to generate the complete model. As an example, the case of the interface between bone and tendon is of special importance (Figs. 16 and 17). In fact, with this fusion it is possible to transfer the forces between both elements through matching nodes between both surfaces (Fig. 18). This avoids modelling contacts between them.

To improve the response of the FEM proposed, the ligaments that form part of the human shoulder joint have been considered. These ligaments are:

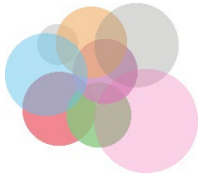
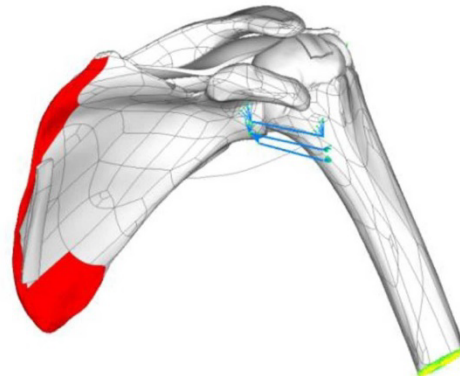


Fig. 24 Areas of the scapula without displacement, in red

V: Unfixed
L: 50 N horizontal at wrist
C: ESCAPULA FUA



1. Superior glenohumeral ligament: medially inserted into the top of the glenoid labrum and neck of the scapula and extends to the base of the coracoid process.
2. Middle glenohumeral ligament: it is inserted medially in the glenoid rim and in the neighbouring part of the neck of the scapula, to end in the lower part of the minor tubercle, along the insertion of the subscapularis tendon.
3. Lower glenohumeral ligament: reinforces the anteroinferior part of the joint capsule. It is inserted into the glenoid rim and reaches to the lower part of the lower tubercle of the humerus.
4. Coracohumeral ligament: thick fibrous lamina, which connects the humeral head with the coracoid process of

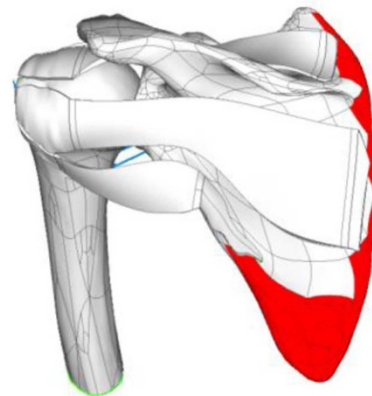
the scapula, which generates tension and stability when the arm is on the side of the body. It merges with the capsule of the shoulder joint.

Similarly, it has been included in the model the glenohumeral capsule wrapping the joint formed between the humerus and the scapula.

For the ligaments, 1D beam elements have been used, where the section of the element is that of the real ligament. The use of rigid elements to simulate the insertion of the ligament in the bone allows to eliminate the concentration of tensions that would occur if the insertion were made in a single node (Fig. 19).

Fig. 25 Areas of the scapula without displacement, in red

V: Unfixed
L: 50 N horizontal at wrist
C: ESCAPULA FUA



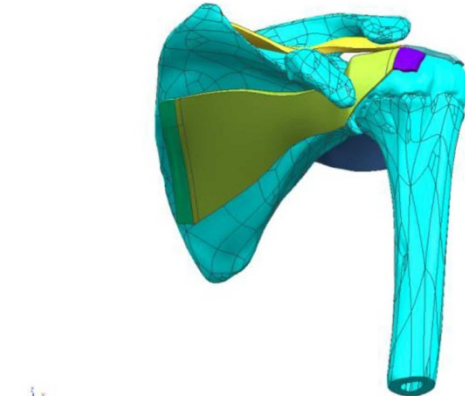


Fig. 26 CAD model from the cloud of points (frontal view)

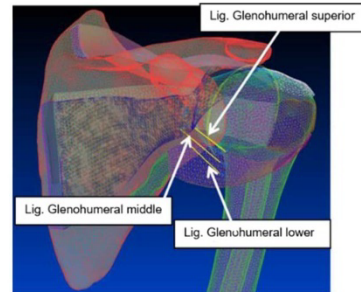


Fig. 28 Detail of glenohumeral ligaments modeled as beam elements, in yellow

Figs. 24 and 25 show the areas of the scapula without displacements applied to the developed FEM. Fig. 24 shows the scapula in the front view and Fig. 25 in the rear view.

The materials used for the model are summarized in Table 2. In this study three load cases have been considered (Table 3).

For the proposed finite element model has considered the effect of gravity. This has been applied as shown in Fig. 22.

For the boundary conditions, the scapula was considered fixed in the area where the muscles of the scapula would be inserted. These muscles are (Fig. 23):

1. Levator scapula muscle.
2. Rhomboid minor muscle.
3. Rhomboid major muscle.

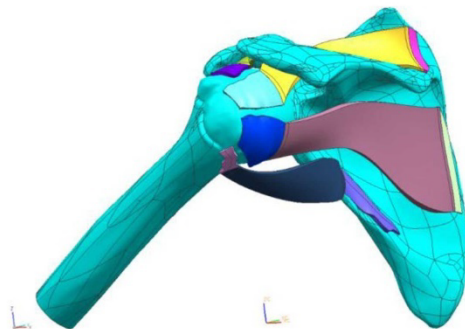


Fig. 27 CAD model from the cloud of points (rear view)

Results and discussion

This section shows the results of the meshed geometric model and the finite element analysis implemented in this study.

Computer Aided Design (CAD) model

Figures 26 and 27 show the CAD model obtained from the clouds of points.

From this CAD model, the corresponding mesh has been generated for the FEA, in which all the elements mentioned in Table 1 are included.

As shown in Fig. 28, the generated mesh is suitable for carrying out the finite element analysis. The combination of 3D elements and 1D elements used meets the objectives of the study. In fact, the main objective is the adequacy of the mesh so that the finite element method simulation reproduces the behaviour of the ligaments in the model. The stresses that appear in bones, muscles, cartilage area and insertions of the ligaments are consistent with the expected result.

Finite Element Analysis (FEA)

The results shown in Figs. 29, 30 and 31 represent the stresses and displacements in the articulation under study by colour coding. The colours vary from the maximum value represented in red to the minimum value represented in blue.

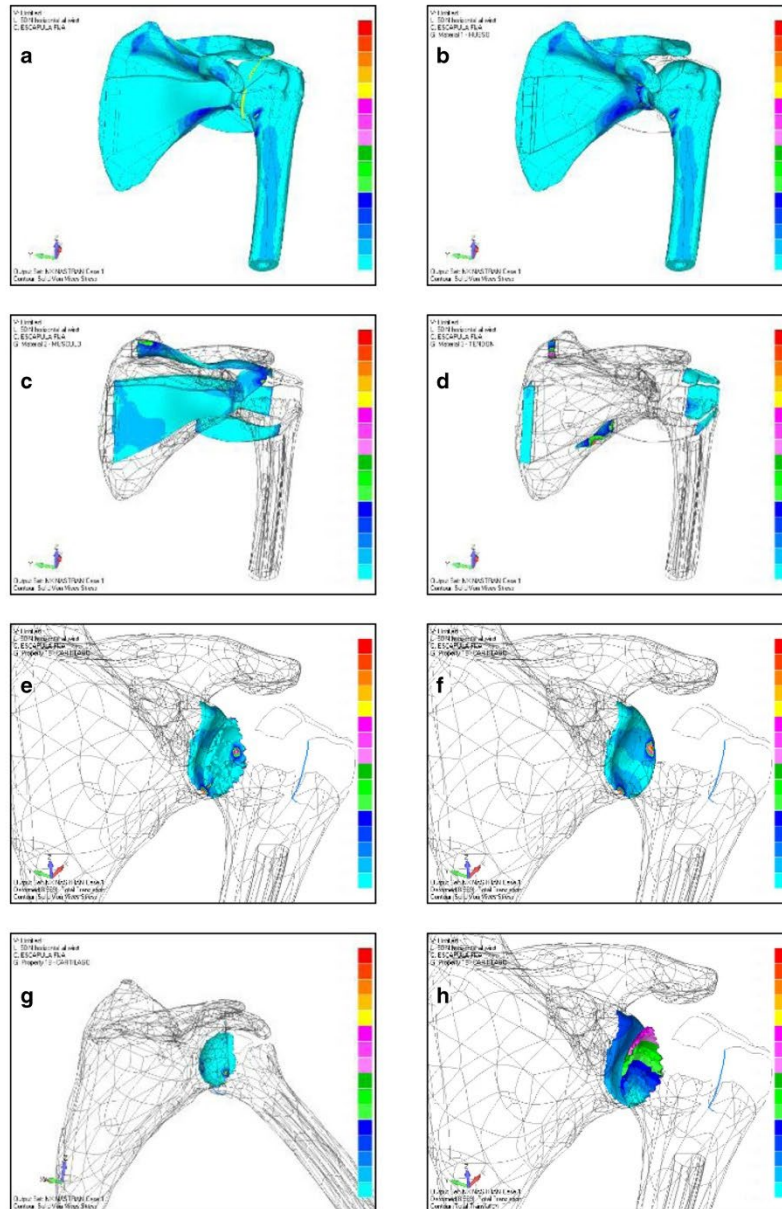


Fig. 29 Load case 1, FEM results: **a)** Stresses in complete model; **b)** Maximum stresses in bones; **c)** Stresses in muscles; **d)** Stresses in tendons; **e)** Stresses in cartilage zone, scapula-Humerus; **f)** Stresses in cartilage zone, only Humerus; **g)** Stresses in cartilage zone, only scapula; **h)** Displacements in cartilage zone

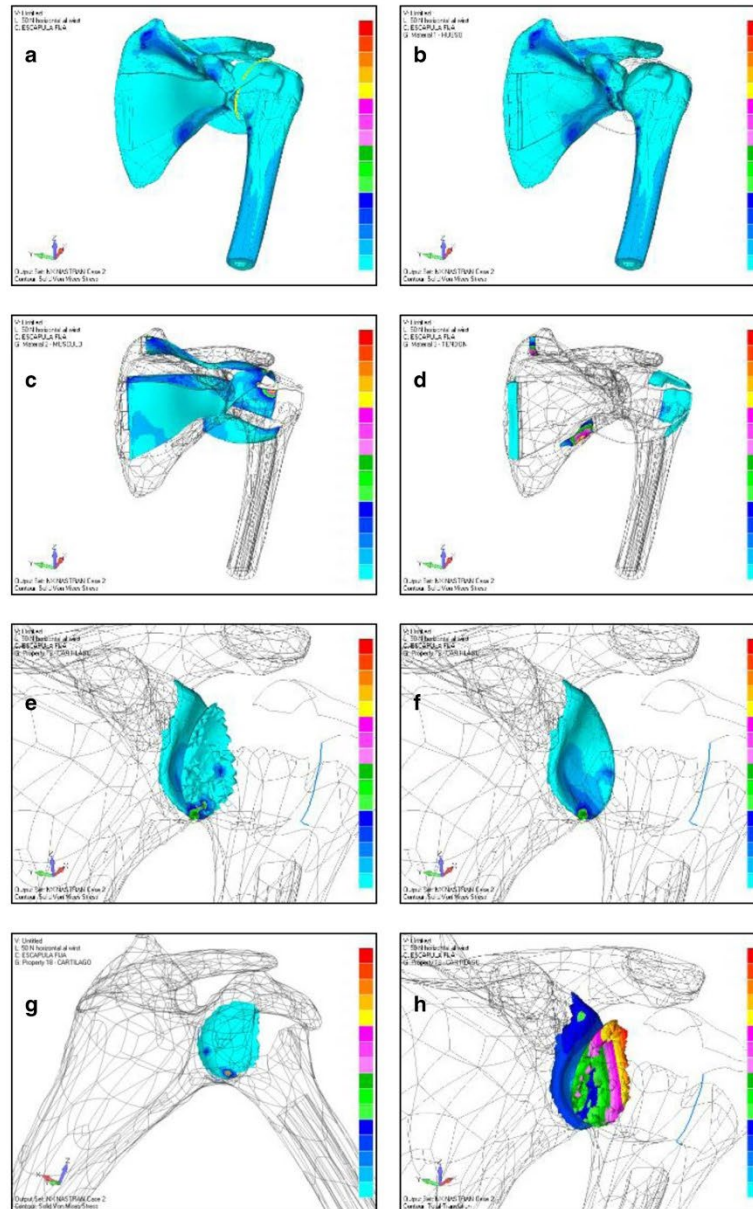
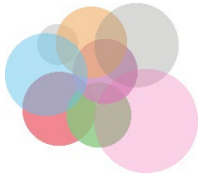


Fig. 30 Load case 2, FEM results: **a)** Stresses in complete model; **b)** Maximum stresses in bones; **c)** Stresses in muscles; **d)** Stresses in tendons; **e)** Stresses in cartilage zone, scapula-Humerus; **f)** Stresses in cartilage, only Humerus; **g)** Stresses in cartilage zone, only scapula; **h)** Displacements in cartilage zone

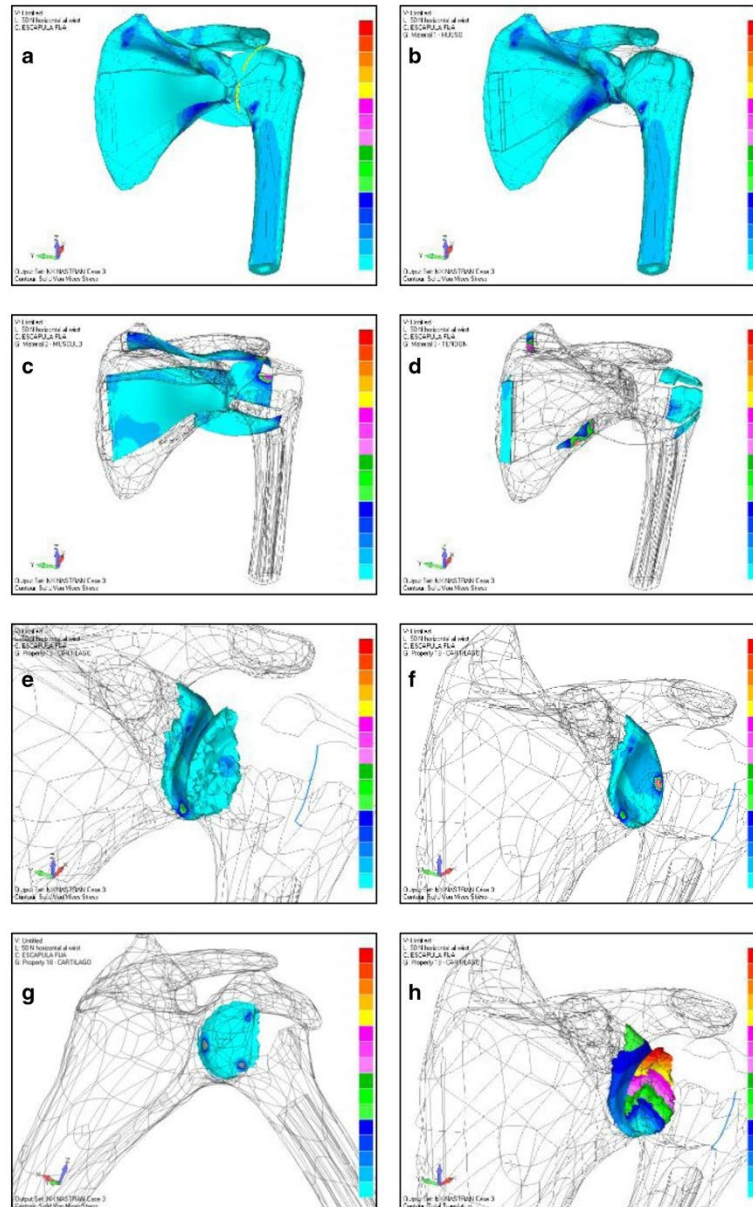
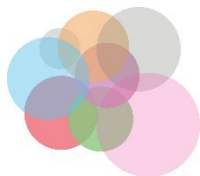


Fig. 31 Load case 3, FEM results: **a)** Stresses in complete model; **b)** Maximum stresses in bones; **c)** Stresses in muscles; **d)** Stresses in tendons; **e)** Stresses in cartilage zone, scapula-Humerus; **f)** Stresses in cartilage zone, only Humerus; **g)** Stresses in cartilage zone, only scapula; **h)** Displacements in cartilage zone



Stresses in bones

As shown in the figures of the three load cases studied (Figs. 29, 30 and 31, section a to b), the maximum stresses in bones are in the insertion areas of the ligaments. This makes physical sense since ligaments are responsible for keeping the joint in position. Therefore, the ligaments transfer the necessary efforts to the area of insertion of the ligaments in the bones, causing these points to be the most stressed.

Stresses in muscles

Due to the static positions of the joint under study, the muscles do not support high stresses (Figs. 29, 30 and 31, section c). According to the stresses distribution, the maximum values are in the zones of tendons (Figs. 29, 30 and 31, section d).

This situation has great physical sense since tendons are responsible for transmitting efforts of muscles to the bone structure while maintaining the position of the joint.

Stresses in cartilage

In the cartilage zone (Figs. 29, 30 and 31, section e, f and g) a uniform and low value stresses distribution is observed. However, points with high stress concentration does not have a great physical sense, they are due to the alignment of the gap element. As the gap element, behaves like a compression spring, the stresses that appear in it are proportional to the displacement that occurs in this kind of element (Figs. 29, 30 and 31, section h). Observing the images of the displacements that occur around the cartilage (Figs. 29, 30 and 31, section h) it is verified how they adjust to the expected, producing the displacement of the head of the Humerus. This displacement moves the head away from the Humerus of the scapula, which in comparison remains with very low values. However, if the gap element alignment, in relation to the displacement that occurs, is not correct, high stresses may appear that do not correspond to reality. Despite this, it is observed how the stresses distribution in the cartilage area maintains coherence with reality since the maximum stresses appear in the lower half of the cartilage. In this area in which the Humerus-scapula contact is greater, the compression tensions are greater.

Conclusions

This study validates the FEM developed presenting a first qualitative study of the performances of the joint of the human shoulder. The proposed analysis meets the stated objective:

a) Obtaining an improved mesh of the shoulder joint to analyse real situations with application in the field of sports medicine, work, etc. For this reason, the follow improvements have been carried out:

- Meshing of the joint, bone, tendons and muscles as 3D elements.
- Improved mesh in the insertion of tendon-bone.
- Congruence between the matching meshes of the tendon insertion area with the bone.
- Including ligaments as 1D elements, node to node.

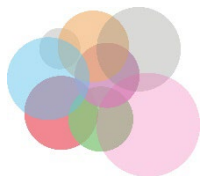
b) Obtaining congruent and logical results of the state of load of the articulation in the three proposed cases.

- Simplifying the modelling of muscles. It is of great interest model muscles simplified to 1D elements. It makes possible to establish different positions in the articulation without having to mesh again each one of the positions that can be studied.

c) This improvement must consider:

- Young's modulus of muscle is not linear. This depends on the load to which the muscle is subjected and its length
- Muscle geometry (cross-section) varies depending on the position in which it is located.
- Modifying the model used to simulate the ligaments considering that the strain-stress curve of them is not linear along its entire length. This involves using 1D elements to adapt its Young's modulus to the length having the ligament at each time.
- Improving simulation at the contact zone cartilage-bones of the joint. It must be investigated about the possible improvement of the alignment of the gap elements.

The final objective of the study is to obtain an FEM to implement different positions of the same joint without modifying the initial mesh. This requires use 1D elements for the anatomical elements that deform, ligaments and muscles, and 3D elements for bones and tendons involved in the joint. The FEM model developed demonstrate that the results are consistent with the states of applied loads in the human joint under real work conditions. In fact, the maximum stresses in bones are in the insertion areas of the ligaments. Due to the static positions of the joint under study, the muscles do not support high stresses. According to the stresses distribution, the maximum values of these one are in the zones of tendons. From the result analysis, it is observed how the stresses distribution in the cartilage area maintains coherence with reality since the maximum stresses appear in the lower half of the cartilage. In this area in which the Humerus-scapula contact is greater, the compression tensions are greater.



Compliance with ethical standards

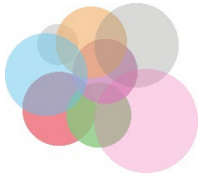
Conflict of interest Manuel Islán Marcos declares that he has no conflict of interest. Emilio Lechosa Urquijo declares that he has no conflict of interest. Fernando Blaya Haro declares that he has no conflict of interest. Roberto D'Amato declares that he has no conflict of interest. Enrique Soriano Heras declares that he has no conflict of interest. Juan Antonio Juanes declares that he has no conflict of interest.

Ethical approval This article does not contain any studies with human participants performed by any of the authors. This article does not contain any studies with animals performed by any of the authors.

References

1. Brunelli, M. P., and Gill, T. J., Fractures and tendon injuries of the athletic shoulder. *Orthop. Clin. North Am.* 33(3):497–508, 2002.
2. Islán, M., Carvajal, J., Pedro, P. S., D'Amato, R., Juanes, J. A., and Soriano, E., Linear approximation of the behavior of the rotator cuff under fatigue conditions. Violinist case study, in *Proceedings of the 5th International Conference on Technological Ecosystems for Enhancing Multiculturality - TEEM 2017*, pp. 1–8, 2017.
3. M. Islán, F. Blaya, P. San Pedro, R. D'Amato, E. Lechosa Urquijo, and J. A. Juanes, "Analysis and fem simulation methodology of dynamic behavior of human rotator cuff in repetitive routines: Musician case study," *J. Med. Syst.*, vol. 42, no. 3, 55, 2018.
4. CDC - NIOSH Publications and Products, Cumulative trauma disorders in the workplace (95-119), *DHHS Publ. Number 95-119*, 1995.
5. International Labour Office - Geneva, *LIST OF OCCUPATIONAL DISEASES (revised 2010) Identification and recognition of occupational diseases: Criteria for incorporating diseases in the ILO list of occupational diseases*. Geneva: INTERNATIONAL LABOUR OFFICE GENEVA, 2010.
6. Ruggiero, A., Affatato, S., Merola, M., and De, M. C., FEM analysis of metal on UHMWPE total hip prosthesis during normal walking cycle, in *Proceedings of the XXIII Conference of the Italian Association of Theoretical and Applied Mechanics (AIMETA 2017)*, pp. 4–7, 2017.
7. Ruggiero, A., Gómez, E., and D'Amato, R., Approximate analytical model for the squeeze-film lubrication of the human ankle joint with synovial fluid filtrated by articular cartilage. *Tribol. Lett.* 41(2): 337–343, 2011.
8. Ruggiero, A., Gómez, E., and D'Amato, R., Approximate closed-form solution of the synovial fluid film force in the human ankle joint with non-Newtonian lubricant. *Tribology International* 57: 156–161, 2013.
9. D'Amato, R., Calvo, R., and Gómez, E., Sensitivity study of the morphometric fitting on the pressure field inside ankle joints. *Case Stud. Mech. Syst. Signal Process.* 1:8–14, 2015.
10. Luo, Z. P., Hsu, H. C., Grabowski, J. J., Morrey, B. F., and An, K. N., Mechanical environment associated with rotator cuff tears. *J. shoulder Elb. Surg.* 7(6):616–620, 1998.
11. Seki, N. et al., Mechanical environment of the supraspinatus tendon: Three-dimensional finite element model analysis. *J. Orthop. Sci.* 13(4):348–353, 2008.
12. Zheng, M., Zou, Z., Bartolo, P. J. D. S., Peach, C., and Ren, L., Finite element models of the human shoulder complex: a review of their clinical implications and modelling techniques, *International Journal for Numerical Methods in Biomedical Engineering*, vol. 33, no. 2. Wiley-Blackwell, 2017.

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11.5.ANEXO V

BEHAVIOR UNDER LOAD OF A HUMAN SHOULDER: FINITE ELEMENT SIMULATION AND ANALYSIS

Este artículo tiene como objetivo Modelar en 3D los componentes biológicos del hombro humano para el análisis mediante elementos finitos, enfocado en lesiones músculoesqueléticas comunes. Para ello Se generó un modelo 3D a partir de una nube de puntos obtenida por resonancia. Se mejoró la malla del modelo y se incluyeron ligamentos principales y cápsula glenohumeral.

Posteriormente, utilizando como geometría de referencia la de la articulación en estudio generada y modelada se ha propuesto un análisis FEM con el software ANSYS® [VERSIÓN 17.1].

Con el análisis realizado, gracias a las simplificaciones realizadas se pudieron evaluar las tensiones máximas en los huesos que se encuentran en las áreas de inserción de los ligamentos.

El modelo FEM desarrollado es coherente con las cargas aplicadas y puede ser utilizado para analizar situaciones reales en medicina deportiva y laboral.



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EDUCATION & TRAINING



Behavior under Load of A Human Shoulder: Finite Element Simulation and Analysis

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Abstract

Most musculoskeletal injuries occur during the work routines in different areas, due to repetitive and sustained movements, they are often located in the shoulder. For workers in the building sector, the repetitive movements and displacements occur under load and unnatural postures of the shoulder joint. For this reason, this study aims to model in 3D the biological components which form the shoulder joint for the later finite element analysis. Three cases with different loads have been considered for this study. Due to a linear and isotropic joint approach it has been possible to evaluate the tensions in the main components of the shoulders: muscles, tendons and ligaments. The methodology used allowed obtaining an improved mesh of the shoulder joint to analyse real situations with finite element method analysis with applications in the field of sports medicine, work, etc. Furthermore, the simplification adopted for modelling the joint muscles, as 1D elements in the finite element model has made it possible to establish different positions of human joint without mesh again each of the studied positions. The results are consistent with the states of applied loads. In fact, the maximum stresses in bones are in the insertion areas of the ligaments. Due to the static positions of the joint under study, the muscles do not support high stresses. According to the stresses distribution, the maximum values are in the zones of tendons. From the result analysis, it is observed how the stresses distribution in the cartilage area maintains coherence with reality since the maximum stresses appear in the lower half of the cartilage. In this area in which the Humerus-scapula contact is greater, the compression tensions are greater.

Keywords Finite element analysis · Shoulder · Humerus · Muscles · Tendons · Ligaments

Introduction

Due to its high mobility, the shoulder is the Human joint that has the most injuries in its tissues [1]. The

rotator cuff is formed by the tendons of the subscapular, supraspinatus, infraspinatus, and minor rotator cuff muscles that meet at the shoulder joint [2, 3]. It is placed in the arch called coracoacromial, constituting the

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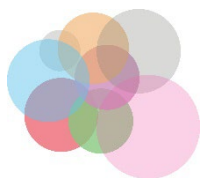


Fig. 1 Cloud of points obtained through a resonance

subacromial space with a serous pouch that lubricates the cuff. This pathology is characterized by progressive pain as a result of an accumulated overload or acute after an overexertion, which is located on the upper or lateral side of the shoulder and increases with movement, mainly with the elevation of the arm (ex. when

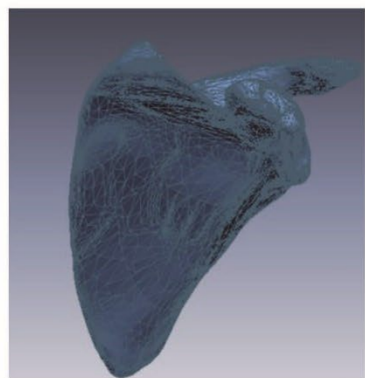


Fig. 2 Mesh corresponding to the scapula

combining) and with internal rotation (ex. when putting on a jacket) [4]. The pain may be more acute at night when leaning on the affected side. The chronic injury can cause a rupture of the tendons of the cuff. Chronic tendinitis of the rotator cuff may be related to shoulder overload, joint instability and rotator cuff degeneration with age. It is relatively frequent, reaching up to 18% in workers in heavy manual tasks with great efforts on the shoulders, associated or not with repetitive movements and forced postures. 75% of people over 50 years have rotator cuff injuries and most of them are asymptomatic [5]. Many jobs in different trades in the construction sector involve important musculoskeletal efforts. This pathology is associated with work that involves stabilizing the shoulders to maintain or move the elbows in an elevated position by force, lifting and catching continuously, or working the arm in flexion or abduction. In the construction sector, this occurs in the assembly of metallic or prefabricated structures, also of heavy prefabricated elements that must be manipulated to unload. In order to understand the behaviour of human joints, in recent years, several authors have proposed scientific studies with finite element model [6] as well as the studies conducted to simulate the dynamic behaviour of human joints. For example, Ruggiero et al. in their study present a new mathematical model in order to understand the tribological performances of the human ankle joint with synovial fluid filtrated by articular cartilage in the case of Newtonian fluid and of no Newtonian fluid [7, 8]. D'Amato et al. in their study investigate the importance of the morphometric parameter of the human joint for the mathematical modelling of the human ankle joint [9]. Many authors, in recent years, have focused their studies on the description of the dynamic behavior of the shoulder joint. For example, in the 1998 Luo et al. present for the first time a simplified 2D-finite element model in order to described the stress environment in the supraspinatus tendon [10]. Wakabayashi et al. in the 2003 performed 2D-finite element model analysis in order to estimate the mechanical environment of the supraspinatus tendon. In the 2008 Seki et al. [11] presented the first 3D-finite element model of rotator cuff tears in order to study the stress distribution in the supraspinatus tendon. Most of these studies have a substantial limitation. In fact, in these studies only the scapula and the bones of the Humerus were considered. This means that in presenting their results both in the 2D and 3D geometry, the structure of the muscles and other soft tissues have been neglected [12]. For this reason, in this study a finite

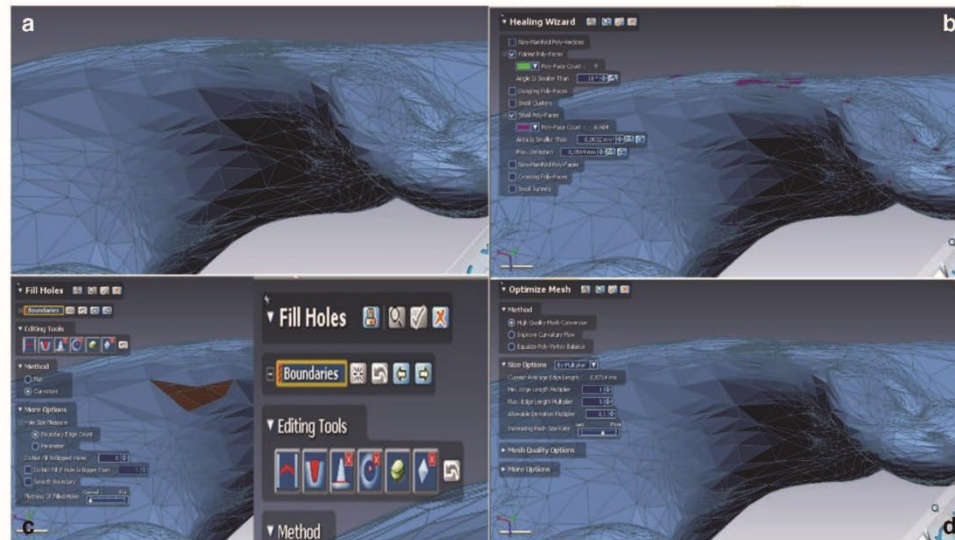


Fig. 3 Mesh and cloud of points operations

element model of the shoulder joint, glenohumeral, classified as a diarthrosis, of the genus enarthrosis or spheroid is presented. Based on the finite element model made for the case of a violinist [3] they are carried out modifications and improvements that allow us

increasing the accuracy of the study and improving the predictive performance of it. This is due to include the ligaments, glenohumeral and coracohumeral, and the glenohumeral-capsule in the anatomical model. In the same way the meshing process is improved, by working

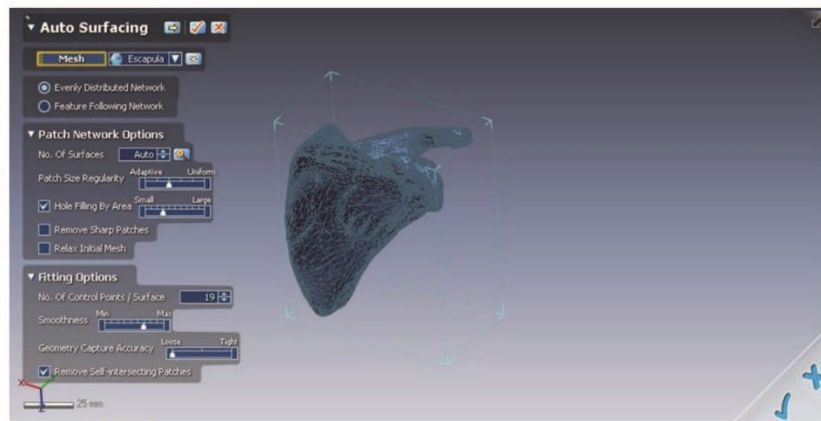


Fig. 4 Reconstruction of surfaces from the cloud of points

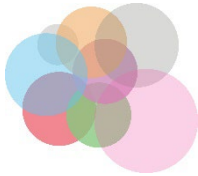


Fig. 5 Surface generated by Geomagic Design X associated with the scapula

on the characterization of the tendon-bone insertion, achieving a transfer of efforts between both more precise and in accordance with reality.

The study focuses on a unique position shoulder, flexion 53°. The scope of work aims the next aspects:

- Improving the mesh used in the finite element model.
- Improving the characterization of the contact tendon-bone insertion area. To do this, meshes between both elements have been made congruent, with nodes coinciding on the interface surface.
- Including the main ligaments involved in the joint, glenohumeral ligaments (lower, middle and upper) and coracohumeral ligament.

Fig. 6 Process to model tendons

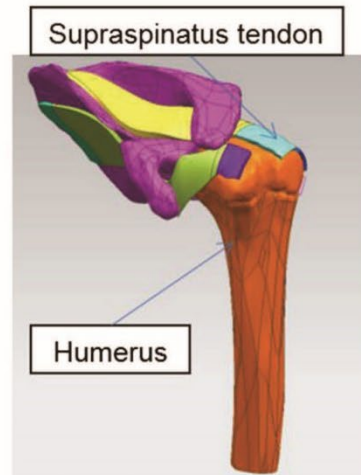
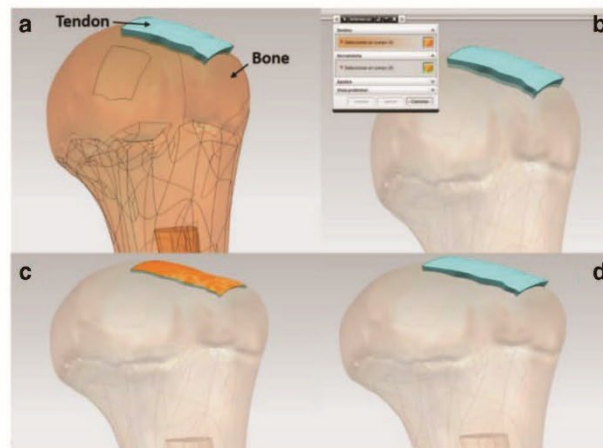


Fig. 7 Location of the supraspinatus tendon and the head of the humerus

- Obtaining a qualitative model of joint behavior.

Methodology

The process to generate a 3D model from a cloud of points is described in the following steps:

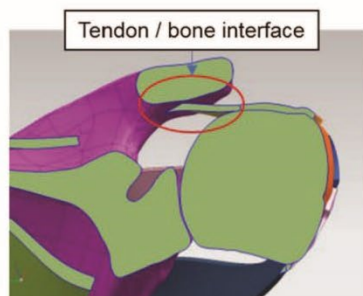


Fig. 8 Sectional view of the humerus + supraspinatus tendon

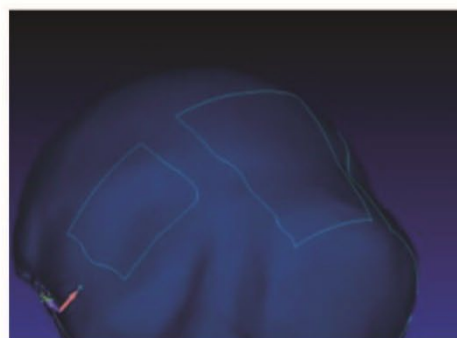


Fig. 10 Outline of the tendon projected on the Humerus

Starting from the cloud of points (Fig. 1) obtained through a resonance. A 3D Computer Aided Design (CAD) model of the elements involved in the scapula.

Using the software Geomagic Design X, it is possible to create a mesh supported in the obtained cloud of points (Fig. 2). The mesh is a triangulation of the surface of the object joining the points of the cloud.

Among the various operations that can be performed using the software Geomagic Design X, all of them are aimed at improving the cloud of points and its associated mesh, are the following (Fig. 3):

- Repairing the mesh, by creating points in the cloud, based on the existing points in it (Fig. 3a).
- Filling holes in the mesh. The mesh is completed creating the triangles and points in the cloud that are missing to close the outer surface of it (Fig. 3b). The filling of holes can be done in different ways (Fig. 3c).
- Optimizing the mesh. It is possible to simplify the number of points in the cloud, eliminating both duplicated points and points which do not provide information, decreasing the size of the cloud (Fig. 3d).

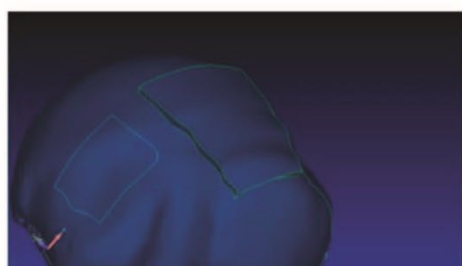


Fig. 9 View of the Humerus + supraspinatus tendon

In addition to these operations, there are many other operations aimed at modifying the cloud of points such as symmetry operations, copy points, scaling of the piece, etc. These operations are also included in commercial Computer Aided Design (CAD) systems,

The most important operation that Geomagic Design X offers us, is the reconstruction of surfaces based on the cloud of points. The result of this operation is an entity that can be treated in any commercial CAD system. The importance of this operation lies in the possibility of using the reconstructed entity for other purposes such as modification of the geometry in a commercial CAD system, integration of the entity in a set of pieces, finite element analysis operations.

Table 1 Types of elements for each body that intervenes in the joint

Nº	Element type	Parameters	Application
1	Tetrahedral 10 nodes	Material, Young modulus and Poisson coef.	Meshing bones, muscles and tendons
2	Beam element 2 nodes	Material, Young modulus and Poisson coef. and transverse section	Mesh of the ligaments and the glenohumeral capsule.
3	Rigid element 2 nodes	none	Simulation insertion of the ligament in the bone.
4	Gap element 2 nodes	Gap and stiffness of the contact.	Simulation of the cartilage contact between the head of the Humerus and the scapula.

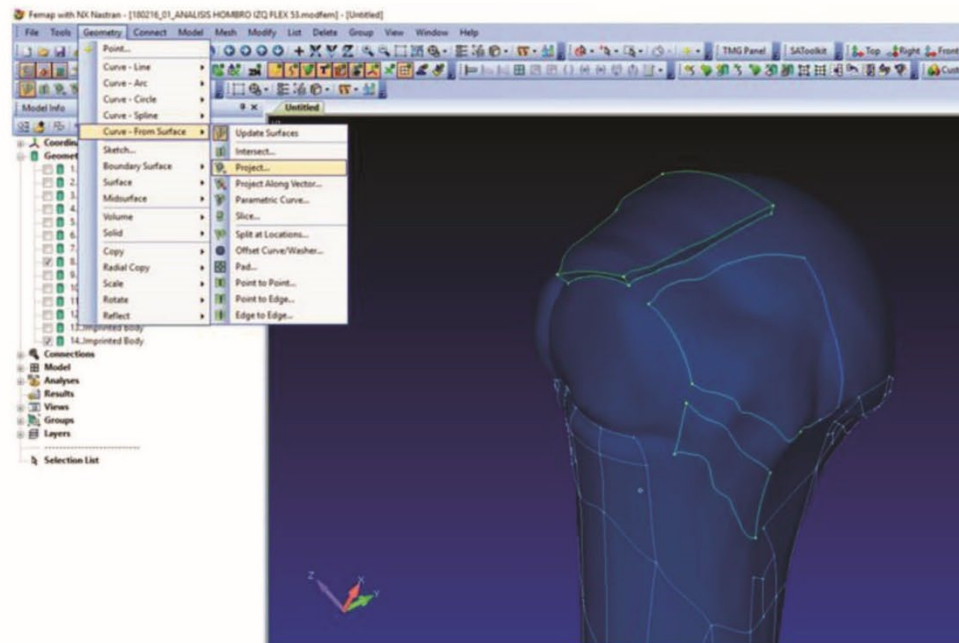


Fig. 11 Contour of the tendons in Siemens FEMAP

The reconstruction of the cloud of points can be done in two different ways:

- Based on a uniformly distributed network. Geomagic Design X is responsible for generating “patches” based

Fig. 12 Projection of the contour

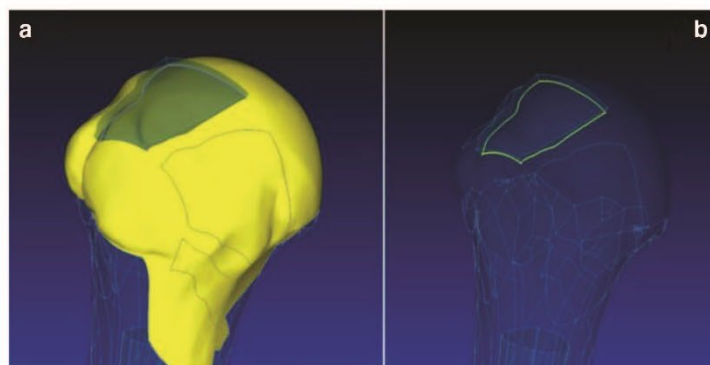
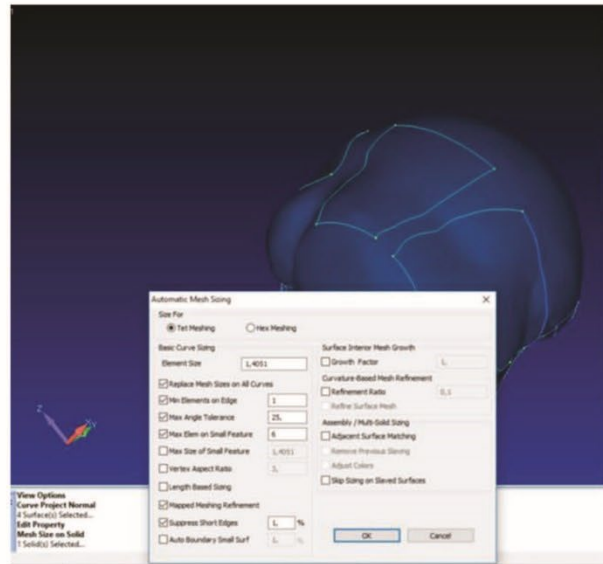




Fig. 13 Mesh Sizing



on the cloud of points to build the outer surface of the piece.

- Based on geometrical operations. Geomagic Design X looks for geometrical surfaces in the point cloud (cylinders, flat faces, etc. ...) to reconstruct the piece.

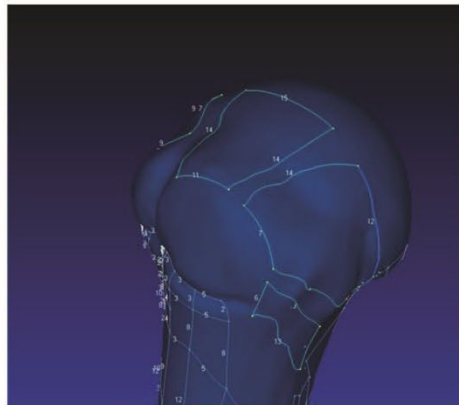


Fig. 14 Number of nodes in each curve

In the case of complicated geometrical pieces, it is recommended to use patch reconstruction. Locating known geometries in a piece as the scapula is complicated so the patches are a suitable solution to obtain the surface of the piece (Fig. 4).

The result of the reconstruction is shown in Fig. 5. The surface of the piece is composed of a series of patches, surfaces of different sizes, which adapt to the

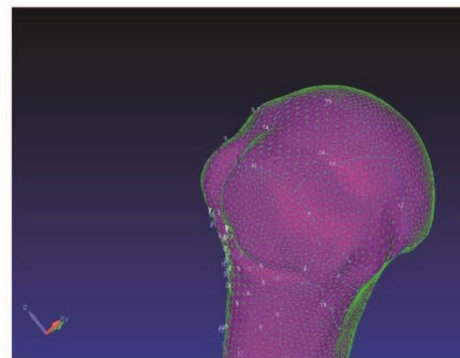


Fig. 15 Obtained mesh

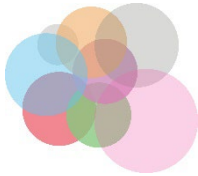
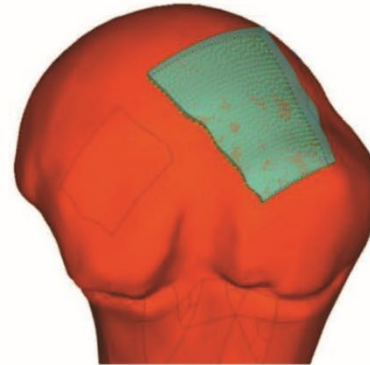


Fig. 16 Nodes located on the tendon-bone interface surface corresponding to the Humerus

V: Unlabeled
L: 50 N horizontal at wrist
C: ESCAPULA FLA
S: 26x27



cloud of points, thus creating the outer surface of the piece and, therefore, its volume.

To verify that the interfaces between tendons and bones are perfectly matched, a 3D CAD, NX tool from Siemens has been used. The basis of the meshing process in these areas is the coincidence of the meshes between both surfaces in contact. To do this, a modeling work on the tendons that ends with a Boolean operation has been carried out.

Taking as an example the insertion of the supraspinatus tendon in the head of the humerus, it is possible to explain the process of modeling the tendon in order to achieve a perfect match. Figure 6a shows that the tendon has a thickness greater than the real one, so that between the tendon and the bone there is an interference between both pieces. by means of a Boolean subtraction operation it is possible to subtract the interference to the tendon volume. As shown in

Fig. 17 Nodes located on the tendon-bone interface surface corresponding to the tendon

V: Unlabeled
L: 50 N horizontal at wrist
C: ESCAPULA FLA
G: Tendon Supraspinoso - Humerus





Fig. 18 Nodes fused between the two coinciding surfaces of the tendon and the Humerus

V: Vómito
L: 50 N horizontal al nivel
C: ESCAPULA/FUJA
S: HUMERO

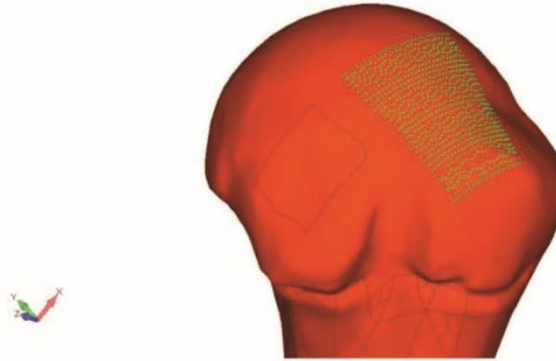


Fig. 6b, it is enough select the two volumes so that Geomagic Design X automatically will calculate the volume of the intersection. The volume of the intersection, the interference between both bodies, is observed in Fig. 6c. subtracting the volume from the tendon, a modeled tendon is obtained with the support surface between the head of the humerus and the coincident tendon between both bodies (Fig. 6d).

This Boolean subtraction operation must be carried out in all the tendons involved in the articulation so that the coincidence between them and the surface in which they are inserted in the bones is perfect. This coincidence is necessary so that during the process of meshing the finite elements a mesh is obtained that shares nodes between the tendons and the bones so that the insertion of the same is modelled correctly.

After performing the modelling operations in which the match between tendons and bones is checked, the obtained model is shown in Figs. 7 and 8.

With the solid models of the articulation elements, a file is generated “*.stp” or “*.igs”. These files are imported into a FEM software (Siemens FEMAP). As in the case described before, it must be ensured matching surfaces between tendon and bone (Fig. 9). In this way both surfaces will be coincident in geometry (achieved with the Boolean operation in the CAD 3D software) and also in the contour of them (Fig. 10). Once verified that the geometries between bone and tendon are coincident, the model mesh is generated. Table 1 summarizes the

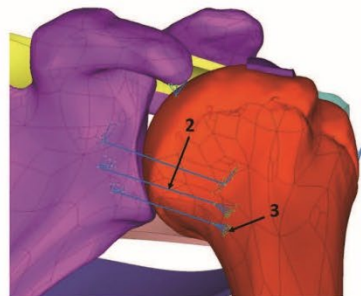


Fig. 19 Rigid element (3) and element 1D type Beam (2) to simulate ligaments of the joint (Table 1)

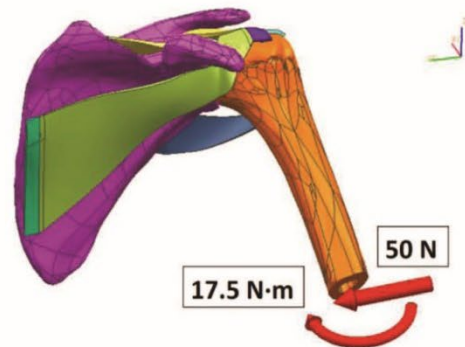


Fig. 20 Load case 1 used in finite element model



Table 2 Materials used for the model

	Young's Modulus (MPa)	Poisson	Density (Kg/m ³)
Bone	17,200	0,41	1790
Muscle	0,3	0,43	1350
Tendon	2000	0,43	1350
Cartilage	8	0,49	1150
Ligament	300	0,43	1350

elements used to model the parts which the human shoulder joint consists of.

This projection of the contour of the tendons on the surface of the bones is achieved in Siemens FEMAP by projecting the curves that make up the contour of the tendon on the insertion surface in the bone (Fig. 11). Then select the surfaces on which the curves are going to be projected, in our case shown in yellow in the Fig. 12a. Finally, the curves wish to project on the surface are selected, in our case, the curves that define the contour of the tendon (Fig. 12b).

The projection is produced according to the normal to the surface at each point, it is also possible to project the curves according to a defined vector. The result of the projection is a partition on the surface (Fig. 13).

Once this operation of projection of curves on the surfaces is done, the solids are obtained in situation of starting the process of generating the finite element mesh to proceed to the calculation of the different load cases. The meshing process is carried out using the types of elements for each body that intervenes in the joint, see Table 1. To generate the mesh must follow the following steps:

1. Generation of the physical properties of bodies. Consisting of defining the physical characteristics of the finite elements used:
 - a) Characteristics of the material that defines the element, bone, tendon, muscle, etc.

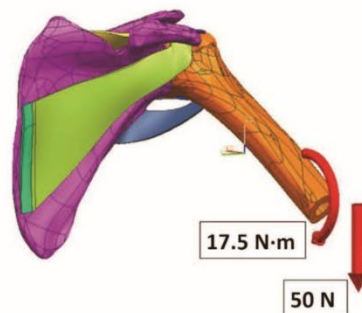


Fig. 21 Load case 2 used in finite element model

- b) Geometrical characteristics of the element, thickness (in the case of using elements 2d), section (in the case of using elements 1d, elastic constant (in the case of using spring elements)
2. A different physical property is assigned to each body involved in the articulation, in this way it is necessary to create as many physical properties as bodies participate in the model.

Once the physical properties of the bodies involved in the articulation have been defined, meshing is carried out. The first step is to define the size of the element to be used, the smaller element is the more accurate the analysis, it is worth mentioning that the calculation process by means of the finite elements method obtains results in the defined nodes, the rest of the model, the spaces without nodes, are calculated by extrapolation between nodes depending on the type of element, and its characteristics, that unites them with each other. However, the smaller the selected element size, the greater the time and resources needed for the calculation, that is why the number of nodes to be used should be minimized, keeping the minimum size necessary to

Table 3 Load cases considered for FEA

Load Case	Load Conditions	Constrain Conditions
1	Gravity applied negative Z axis (9.841 m/s ²)	Fixed scapula on areas shown in Fig. 20
2	Gravity applied along negative Z axis (9.841 m/s ²) + 50 N applied along positive Y axis.	Fixed scapula on areas shown in Fig. 21
3	Gravity applied along negative Z axis (9.841 m/s ²) + 50 N applied along negative Z axis.	Fixed scapula on areas shown in Fig. 22

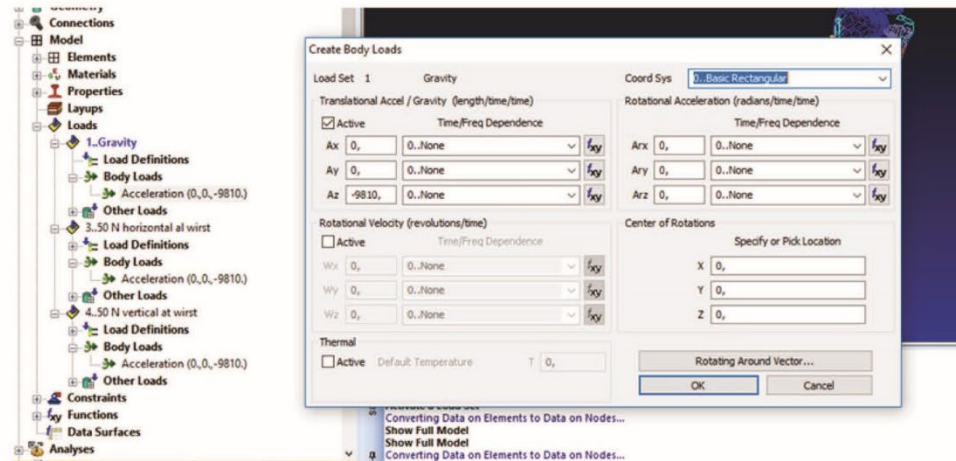
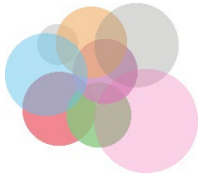


Fig. 22 Effect of gravity applied to finite element model

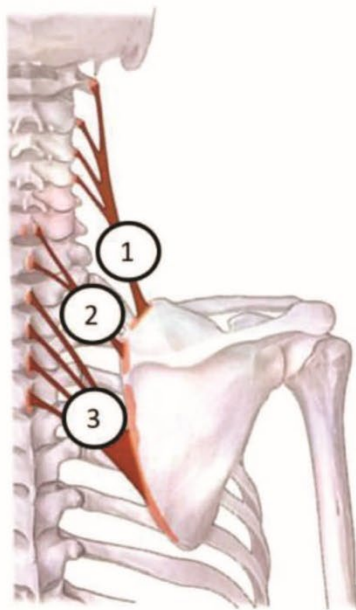


Fig. 23 Muscles attached to the scapula fixed

reproduce with precision the geometry of a discrete mode and a finite number of nodes. To define the element size, select the desired volume and assign it a size. Siemens FEMAP is responsible for locating the nodes in that volume in the optimal way (Fig. 13).

Once the number of nodes that make up the piece have been defined, the mesh of elements is generated. Figure 14 shows the number of nodes that have been created in each curve that defines the body, Siemens FEMAP is responsible for generating the nodes inside the surfaces by interpolation and triangulation of them. After the meshing process of the body the result obtained is a mesh, composed of nodes and elements, which define the geometry of the body (Fig. 15). The process must be repeated in all the bodies that make up the joint.

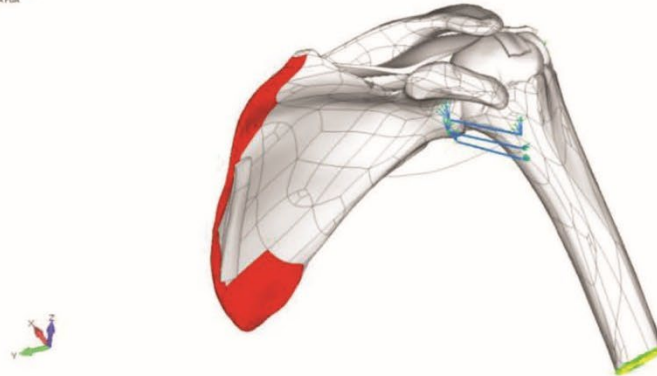
Once the mesh is generated separately from the muscles, tendons, bones and ligaments, the next step is the fusion of the matching nodes to generate the complete model. As an example, the case of the interface between bone and tendon is of special importance (Figs. 16 and 17). In fact, with this fusion it is possible to transfer the forces between both elements through matching nodes between both surfaces (Fig. 18). This avoids modelling contacts between them.

To improve the response of the FEM proposed, the ligaments that form part of the human shoulder joint have been considered. These ligaments are:



Fig. 24 Areas of the scapula without displacement, in red

V: Unfixed
L: 50 N horizontal at wrist
C: ESCAPULA FLUX



1. Superior glenohumeral ligament: medially inserted into the top of the glenoid labrum and neck of the scapula and extends to the base of the coracoid process.
2. Middle glenohumeral ligament: it is inserted medially in the glenoid rim and in the neighbouring part of the neck of the scapula, to end in the lower part of the minor tubercle, along the insertion of the subscapularis tendon.
3. Lower glenohumeral ligament: reinforces the anteroinferior part of the joint capsule. It is inserted into the glenoid rim and reaches to the lower part of the lower tubercle of the humerus.
4. Coracohumeral ligament: thick fibrous lamina, which connects the humeral head with the coracoid process of

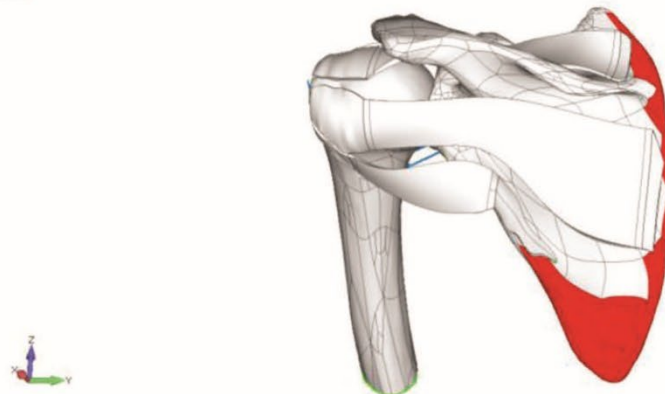
the scapula, which generates tension and stability when the arm is on the side of the body. It merges with the capsule of the shoulder joint.

Similarly, it has been included in the model the glenohumeral capsule wrapping the joint formed between the humerus and the scapula.

For the ligaments, 1D beam elements have been used, where the section of the element is that of the real ligament. The use of rigid elements to simulate the insertion of the ligament in the bone allows to eliminate the concentration of tensions that would occur if the insertion were made in a single node (Fig. 19).

Fig. 25 Areas of the scapula without displacement, in red

V: Unfixed
L: 50 N horizontal at wrist
C: ESCAPULA FLUX



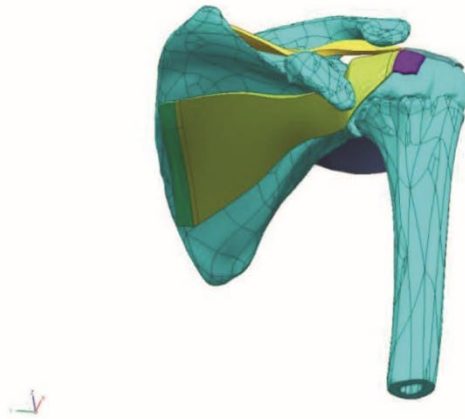


Fig. 26 CAD model from the cloud of points (frontal view)

The materials used for the model are summarized in Table 2. In this study three load cases have been considered (Table 3).

For the proposed finite element model has considered the effect of gravity. This has been applied as shown in Fig. 22.

For the boundary conditions, the scapula was considered fixed in the area where the muscles of the scapula would be inserted. These muscles are (Fig. 23):

1. Levator scapula muscle.
2. Rhomboid minor muscle.
3. Rhomboid major muscle.

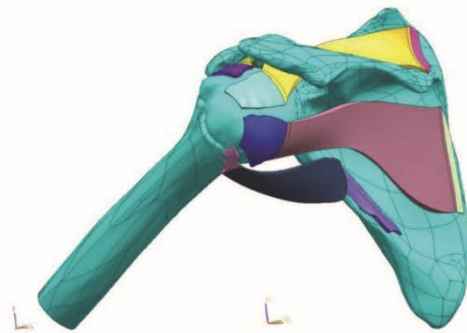


Fig. 27 CAD model from the cloud of points (rear view)

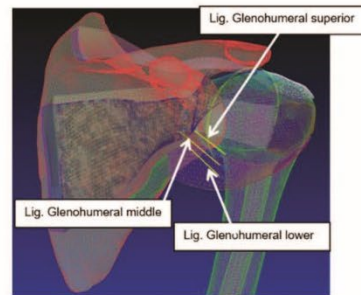


Fig. 28 Detail of glenohumeral ligaments modeled as beam elements, in yellow

Figs. 24 and 25 show the areas of the scapula without displacements applied to the developed FEM. Fig. 24 shows the scapula in the front view and Fig. 25 in the rear view.

Results and discussion

This section shows the results of the meshed geometric model and the finite element analysis implemented in this study.

Computer Aided Design (CAD) model

Figures 26 and 27 show the CAD model obtained from the clouds of points.

From this CAD model, the corresponding mesh has been generated for the FEA, in which all the elements mentioned in Table 1 are included.

As shown in Fig. 28, the generated mesh is suitable for carrying out the finite element analysis. The combination of 3D elements and 1D elements used meets the objectives of the study. In fact, the main objective is the adequacy of the mesh so that the finite element method simulation reproduces the behaviour of the ligaments in the model. The stresses that appear in bones, muscles, cartilage area and insertions of the ligaments are consistent with the expected result.

Finite Element Analysis (FEA)

The results shown in Figs. 29, 30 and 31 represent the stresses and displacements in the articulation under study by colour coding. The colours vary from the maximum value represented in red to the minimum value represented in blue.

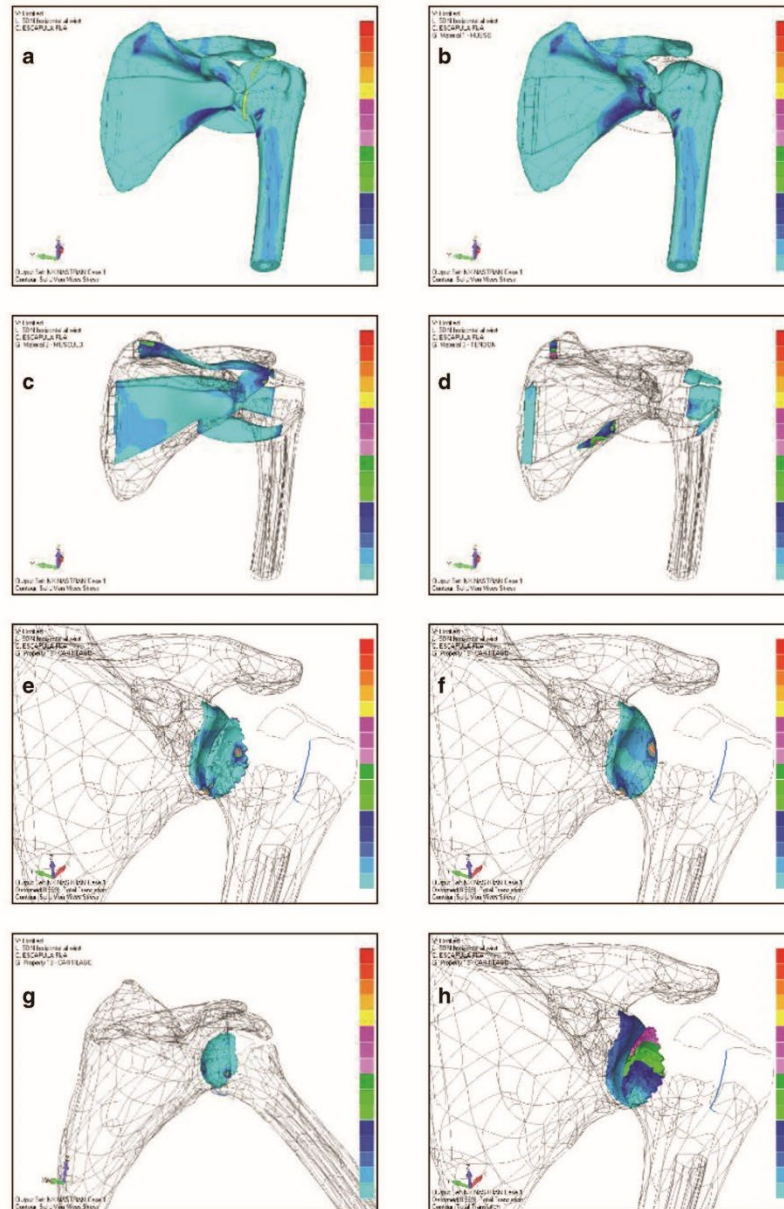
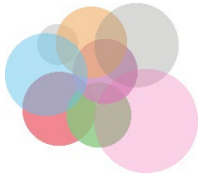


Fig. 29 Load case 1, FEM results: **a)** Stresses in complete model; **b)** Maximum stresses in bones; **c)** Stresses in muscles; **d)** Stresses in tendons; **e)** Stresses in cartilage zone, scapula-Humerus; **f)** Stresses in cartilage zone, only Humerus; **g)** Stresses in cartilage zone, only scapula; **h)** Displacements in cartilage zone

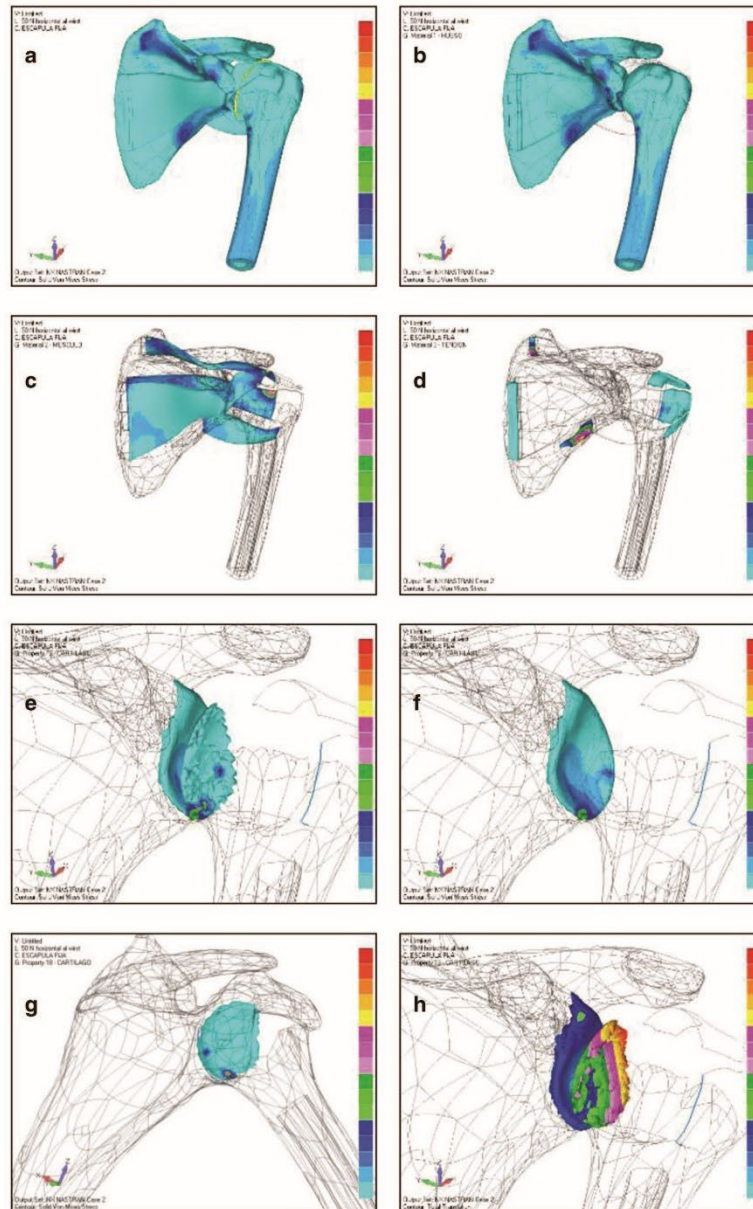


Fig. 30 Load case 2. FEM results: a) Stresses in complete model; b) Maximum stresses in bones; c) Stresses in muscles; d) Stresses in tendons; e) Stresses in cartilage zone, scapula-Humerus; f) Stresses in cartilage, only Humerus; g) Stresses in cartilage zone, only scapula; h) Displacements in cartilage zone

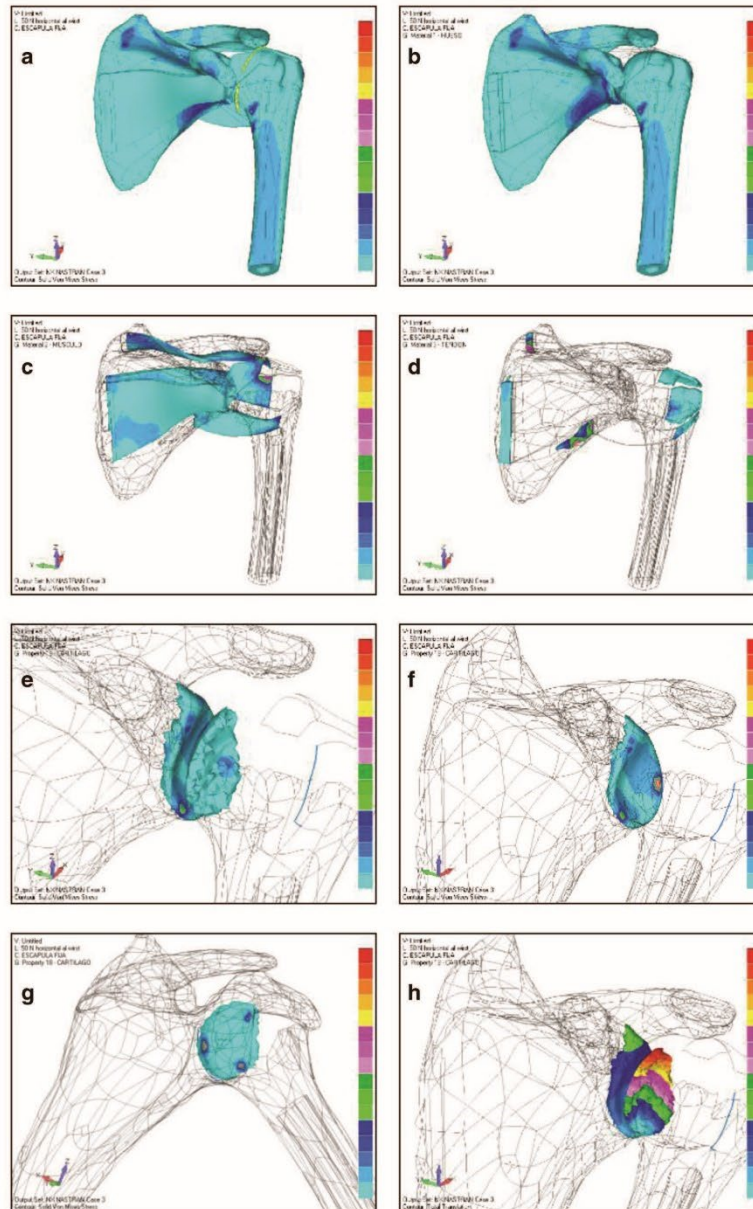
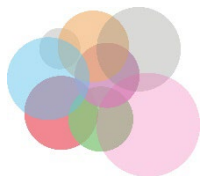


Fig. 31 Load case 3. FEM results: a) Stresses in complete model; b) Maximum stresses in bones; c) Stresses in muscles; d) Stresses in tendons; e) Stresses in cartilage zone, scapula-Humerus; f) Stresses in cartilage zone, only Humerus; g) Stresses in cartilage zone, only scapula; h) Displacements in cartilage zone



Stresses in bones

As shown in the figures of the three load cases studied (Figs. 29, 30 and 31, section a to b), the maximum stresses in bones are in the insertion areas of the ligaments. This makes physical sense since ligaments are responsible for keeping the joint in position. Therefore, the ligaments transfer the necessary efforts to the area of insertion of the ligaments in the bones, causing these points to be the most stressed.

Stresses in muscles

Due to the static positions of the joint under study, the muscles do not support high stresses (Figs. 29, 30 and 31, section c). According to the stresses distribution, the maximum values are in the zones of tendons (Figs. 29, 30 and 31, section d).

This situation has great physical sense since tendons are responsible for transmitting efforts of muscles to the bone structure while maintaining the position of the joint.

Stresses in cartilage

In the cartilage zone (Figs. 29, 30 and 31, section e, f and g) a uniform and low value stresses distribution is observed. However, points with high stress concentration does not have a great physical sense, they are due to the alignment of the gap element. As the gap element, behaves like a compression spring, the stresses that appear in it are proportional to the displacement that occurs in this kind of element (Figs. 29, 30 and 31, section h). Observing the images of the displacements that occur around the cartilage (Figs. 29, 30 and 31, section h) it is verified how they adjust to the expected, producing the displacement of the head of the Humerus. This displacement moves the head away from the Humerus of the scapula, which in comparison remains with very low values. However, if the gap element alignment, in relation to the displacement that occurs, is not correct, high stresses may appear that do not correspond to reality. Despite this, it is observed how the stresses distribution in the cartilage area maintains coherence with reality since the maximum stresses appear in the lower half of the cartilage. In this area in which the Humerus-scapula contact is greater, the compression tensions are greater.

Conclusions

This study validates the FEM developed presenting a first qualitative study of the performances of the joint of the human shoulder. The proposed analysis meets the stated objective:

a) Obtaining an improved mesh of the shoulder joint to analyse real situations with application in the field of sports medicine, work, etc. For this reason, the follow improvements have been carried out:

- Meshing of the joint, bone, tendons and muscles as 3D elements.
- Improved mesh in the insertion of tendon-bone.
- Congruence between the matching meshes of the tendon insertion area with the bone.
- Including ligaments as 1D elements, node to node.

b) Obtaining congruent and logical results of the state of load of the articulation in the three proposed cases.

- Simplifying the modelling of muscles. It is of great interest model muscles simplified to 1D elements. It makes possible to establish different positions in the articulation without having to mesh again each one of the positions that can be studied.

c) This improvement must consider:

- Young's modulus of muscle is not linear. This depends on the load to which the muscle is subjected and its length
- Muscle geometry (cross-section) varies depending on the position in which it is located.
- Modifying the model used to simulate the ligaments considering that the strain-stress curve of them is not linear along its entire length. This involves using 1D elements to adapt its Young's modulus to the length having the ligament at each time.
- Improving simulation at the contact zone cartilage-bones of the joint. It must be investigated about the possible improvement of the alignment of the gap elements.

The final objective of the study is to obtain an FEM to implement different positions of the same joint without modifying the initial mesh. This requires use 1D elements for the anatomical elements that deform, ligaments and muscles, and 3D elements for bones and tendons involved in the joint. The FEM model developed demonstrate that the results are consistent with the states of applied loads in the human joint under real work conditions. In fact, the maximum stresses in bones are in the insertion areas of the ligaments. Due to the static positions of the joint under study, the muscles do not support high stresses. According to the stresses distribution, the maximum values of these one are in the zones of tendons. From the result analysis, it is observed how the stresses distribution in the cartilage area maintains coherence with reality since the maximum stresses appear in the lower half of the cartilage. In this area in which the Humerus-scapula contact is greater, the compression tensions are greater.



Compliance with ethical standards

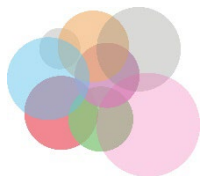
Conflict of interest Manuel Islán Marcos declares that he has no conflict of interest. Emilio Lechosa Urquijo declares that he has no conflict of interest. Fernando Blaya Haro declares that he has no conflict of interest. Roberto D'Amato declares that he has no conflict of interest. Enrique Soriano Heras declares that he has no conflict of interest. Juan Antonio Juanes declares that he has no conflict of interest.

Ethical approval This article does not contain any studies with human participants performed by any of the authors. This article does not contain any studies with animals performed by any of the authors.

References

1. Brunelli, M. P., and Gill, T. J., Fractures and tendon injuries of the athletic shoulder. *Orthop. Clin. North Am.* 33(3):497–508, 2002.
2. Islán, M., Carvajal, J., Pedro, P. S., D'Amato, R., Juanes, J. A., and Soriano, E., Linear approximation of the behavior of the rotator cuff under fatigue conditions. Violinist case study, in *Proceedings of the 5th International Conference on Technological Ecosystems for Enhancing Multiculturality - TEEM 2017*, pp. 1–8, 2017.
3. M. Islán, F. Blaya, P. San Pedro, R. D'Amato, E. Lechosa Urquijo, and J. A. Juanes, "Analysis and fem simulation methodology of dynamic behavior of human rotator cuff in repetitive routines: Musician case study," *J. Med. Syst.*, vol. 42, no. 3, 55, 2018.
4. CDC - NIOSH Publications and Products, Cumulative trauma disorders in the workplace (95-119), *DHHS Publ. Number* 95-119, 1995.
5. International Labour Office - Geneva, *LIST OF OCCUPATIONAL DISEASES (revised 2010) Identification and recognition of occupational diseases: Criteria for incorporating diseases in the ILO list of occupational diseases*. Geneva: INTERNATIONAL LABOUR OFFICE GENEVA, 2010.
6. Ruggiero, A., Affatato, S., Merola, M., and De, M. C., FEM analysis of metal on UHMWPE total hip prosthesis during normal walking cycle, in *Proceedings of the XXIII Conference of the Italian Association of Theoretical and Applied Mechanics (AIMETA 2017)*, pp. 4–7, 2017.
7. Ruggiero, A., Gómez, E., and D'Amato, R., Approximate analytical model for the squeeze-film lubrication of the human ankle joint with synovial fluid filtrated by articular cartilage. *Tribol. Lett.* 41(2): 337–343, 2011.
8. Ruggiero, A., Gómez, E., and D'Amato, R., Approximate closed-form solution of the synovial fluid film force in the human ankle joint with non-Newtonian lubricant. *Tribology International* 57: 156–161, 2013.
9. D'Amato, R., Calvo, R., and Gómez, E., Sensitivity study of the morphometric fitting on the pressure field inside ankle joints. *Case Stud. Mech. Syst. Signal Process.* 1:8–14, 2015.
10. Luo, Z. P., Hsu, H. C., Grabowski, J. J., Morrey, B. F., and An, K. N., Mechanical environment associated with rotator cuff tears. *J. shoulder Elb. Surg.* 7(6):616–620, 1998.
11. Seki, N. et al., Mechanical environment of the supraspinatus tendon: Three-dimensional finite element model analysis. *J. Orthop. Sci.* 13(4):348–353, 2008.
12. Zheng, M., Zou, Z., Bartolo, P. J. D. S., Peach, C., and Ren, L., Finite element models of the human shoulder complex: a review of their clinical implications and modelling techniques, *International Journal for Numerical Methods in Biomedical Engineering*, vol. 33, no. 2. Wiley-Blackwell, 2017.

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11.6.ANEXO VI

MECHANICAL MODEL AND FEM SIMULATIONS FOR EFFORTS ON BICEPS AND TRICEPS MUSCLES UNDER VERTICAL LOAD: MATHEMATICAL FORMULATION OF RESULTS

Aunque las contracciones isométricas en los músculos humanos se han analizado varias veces, no existen modelos FEA que permitan utilizar la misma articulación modelada (el codo en nuestro caso) en diferentes condiciones.

La mayoría de las simulaciones realizadas de la articulación del codo utilizan elementos 3D para el mallado. Representar los músculos de la articulación con elementos 3D es de gran utilidad cuando el estudio se centra en el propio músculo, conociendo la distribución de tensiones en el mismo y comprobar el daño que se produce en el tejido de manera detallada (inserción tendón-músculo, por ejemplo).

Sin embargo, esta técnica no es útil para estudiar el comportamiento de los músculos en diferentes posiciones de la articulación.

Este estudio, basado en el modelo mecánico de la articulación del codo, propone una metodología para modelar músculos que se estudiarán en diferentes posiciones mallándolos con elementos 1D.

Además, la metodología permite calcular los esfuerzos de bíceps y tríceps bajo carga para diferentes ángulos de apertura de la articulación del codo. Los resultados de la simulación han sido modelados matemáticamente para obtener informaciones generales para estos esfuerzos, dependiendo de la carga y del ángulo de apertura



Article

Mechanical Model and FEM Simulations for Efforts on Biceps and Triceps Muscles under Vertical Load: Mathematical Formulation of Results

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Abstract: Although isometric contractions in human muscles have been analyzed several times, there are no FEA models that allow us to use the same modeled joint (the elbow under our case) in different conditions. Most elbow joints use 3D elements for meshing. Representing the muscles in the joint is quite useful when the study is focused on the muscle itself, knowing stress distribution on muscle, and checking damage in muscle in a detailed manner (tendon–muscle insertion, for example). However, this technique is not useful for studying muscle behavior at different positions of the joint. This study, based on the mechanical model of the elbow joint, proposes a methodology for modelling muscles that will be studied in different positions by meshing them with 1D elements. Furthermore, the methodology allows us to calculate biceps and triceps efforts under load for different angles of elbow joint aperture. The simulation results have been mathematically modelled to obtain general formulations for these efforts, depending on the load and the aperture angle.

Keywords: FEM analysis; biomechanics; effort mathematical models; NMR reconstruction; reverse engineering; elbow joint

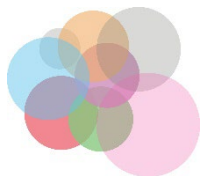
MSC: 37M05

1. Introduction

Over the years, the finite element method (FEM or FEA model) has become the preferred tool [1,2] by researchers for the study of analyzes of practically any type. The main reason for this preference is the accuracy [3,4] in the solution and the flexibility to simulate almost any physical event that may be explained by mathematical equations.

One of the fields in which the finite element method has been used on numerous occasions has been biomechanics [5–11], with applications to specific studies such as joint analysis, prosthesis design and muscle behavior analysis.

Li et al. [5] used the FEM method to run a simulation of fluid structure interactions coupling the lattice Boltzmann method and the FEM, and this new study can be used on blood flow and the heart, or urine interaction with bladder. Della Rosa et al. used the FEM method to validate the new fixator with application in fractures [6]. Zhang et al. developed a FEM analysis of a lumbar spine implanted with a lordotic cage [7] at L3–L4, and the results were validated by in vitro testing. The stress distribution on the cage and the range of motion (ROM) of L3–L4 were used to assess the stability of the implant. Denozière and Ku, in their study, proposed a three-dimensional model of a two-level ligamentous lumbar segment [8], simulated by static analyses with the FEM software to predict that mobility after arthrodesis at the upper level was reduced in all rotational degrees of freedom by an



average of approximately 44%, relative to healthy normal discs. Samani et al., proposed a breast biomechanical model using a FEM formulation [9], specially focused on the modeling of breast tissue deformation which takes place in breast imaging procedures.

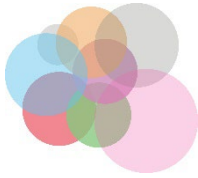
The study proposed by Martinez [12] is focused on the characterization of the muscle and its response under different conditions. Muscle length changes and the internal stress distribution for different electrical stimulations were studied. Weiss et al. [13] described strategies for addressing technical aspects of the 3D computational modelling of ligaments with FEM analysis. Islan et al. [14], studied the behavior of the glenohumeral joint under different postures that represent the routine of a violinist from an ergonomic point of view (RULA method) and using a 3D FEM model to study this movement for a high number of cycles. Sachenkov et al. [15] studied the movement of the femur in the hip, using 1D models for the muscles. Martins et al. [16] presented a study with FEM techniques applied to the pelvic floor, using the Hill model for muscles as a 1D element, extrapolating it to 2D and 3D models, working with isometric and isotonic contractions. Tang et al. [17] studied muscle fatigue with a 3D muscle model in different activation situations. Syomin et al. [18] performed a numerical simulation of contraction of the left ventricle approximated by the axisymmetric body.

Perreault and Heckman evaluated the ability of the Hill model for muscles to describe muscle force responses for naturally and electrically stimulated muscles [19]. Alonso et al. [20] proposed a static physiological optimization model instead of the dynamic optimization model. For this purpose, they used a modeling of muscles by means of 1D elements along with the line of action of the same. Holzbaur et al. proposed a different biomechanical model representing the upper extremity, including shoulder, forearm and arm [21]. The muscles are replaced by a 1D element with a force generation parameter based on experimental data. Park et al. [22] proposed a torque estimation method at the joint of an index finger in the human hand while pinching, the study is supported by electromyography and Hill muscle model. Soechting and Flanders used a simplified muscle model [23] to predict torques during arm movement, comparing their prediction with the EMG results measured in 20 subjects. The muscle model is also based on the Hill muscle model. Zajac developed a mathematical model to analyze the muscle tendon, ‘musculotendon’ actuator [24] based only on the relation between tendon length and muscle fiber length, both at rest.

In the study and FEM analysis of human joints, bones are considered rigid elements, their geometry does not vary when the joint moves, the same happens with cartilages, and they can be approximated to rigid, nondeformable parts [25]. On the other hand, when meshing a muscle in 3D elements into a joint, this mesh is only valid for a certain position, and if there is need to study a different position in the joint, the muscles should be remeshed once again. This problem disappears by using a 1D element, rod-type, to mesh the muscles in the joint. Ligaments can be modeled as springs, 1D elements with two nodes. According to this, a model meshed with 3D elements for rigid bodies and 1D element for muscles elements, can adopt different positions just by moving the elements of the rigid bodies to the new positions.

In the case of 1D elements with just two nodes, if those nodes move in the space, the element will move as well. As ligaments are defined from two different nodes of two different bones, if the bones move to another position, the ligament will adopt the new position. This allows us to reuse the same meshed model in several configurations. For this reason, the final objective of this study is discovering the equation that would allow one to create a new finite element type that could be used in biomechanics analysis where muscles forces are involved in “isometric contractions”. This study focuses on the analysis of human muscle behavior in order to define a new equation that could predict the muscle force that would appear in a certain muscle under known conditions. The human joint under study will be the elbow. The proposed FEM analysis is focused only on the biceps and triceps muscles and the muscle elbow joint considered in this study are (Figure 1):

- Long biceps
- Short biceps



- Brachii
- Long supinator
- Long triceps
- Medium triceps
- Lateral triceps

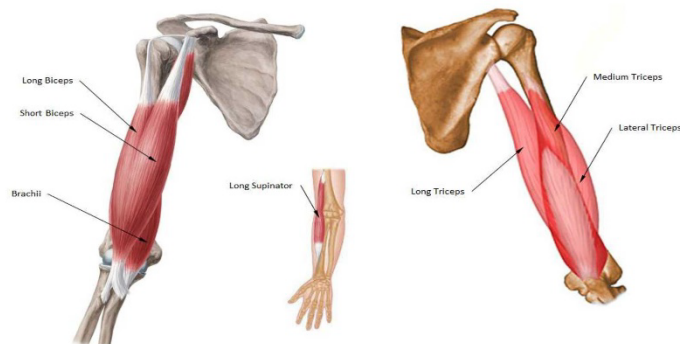


Figure 1. Flexors (left) and extensors (right) elbow muscles identification.

Since muscles can only work under traction (they do not transmit compression loads), different models must be proposed for a separate study of the triceps and biceps muscles. Several analyses, on different aperture angles of the elbow, will be solved looking for the effort needed on referred muscles, biceps or triceps, to balance 150 N applied on the wrist in an “isometric contraction”. The scope of the study includes the following aspects:

- Analyzing the elbow under load at different angles;
- Obtaining efforts in different muscles involved in the joint for different positions;
- Defining mathematical models to predict efforts in muscles depending on:
 - Joint angle or muscle length;
 - Applied load on the wrist.

2. Materials and Methods

The proposed methodology will allow one to calculate elbow muscle effort. Furthermore, a post-processing of the results is proposed in order to obtain the formulation of efforts in the studied muscles when the load applied changes. These efforts are referred to isometric load cases. An isometric load involves muscle contraction against resistance in which the length of the muscle remains constant. The research can be divided into two main parts:

1. Finite element models of the mechanism of elbow joint. The 3D anatomical model developed is based on two diagnostic imaging techniques. CT for the bones and MRI for the muscles. From this three-dimensional model, different FEM models were generated, for biceps and triceps muscles studies.
2. Mathematical models predicting efforts in muscles studied. From FEM models, isometric muscle efforts will be obtained for a vertical load applied on wrist, 150 N including forearm mass, at different angles of aperture in the joint, from 44 to 164 degrees. These efforts will be extrapolated for any load applied on the wrist.

The first part is based on previous studies [25]. This model is shown in Figure 2, and it will be described in the more detail in next section.

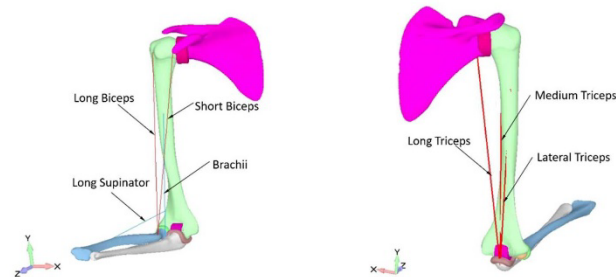


Figure 2. FEA model, muscles identification, flexors (left) and extensors (right).

2.1. Finite Element Model

For the creation of the FEM model, a 3D CAD model of the elbow joint is necessary, from which the mesh of the FEM model will be created.

Said CAD model is obtained starting from the images obtained by means of CT and MRI, of the elements of the joint (bones and muscles). Thanks to the use of software for generating 3D CAD models from point clouds, Geomagic Desing X, we can generate the CAD model of solid bodies.

The FEA model was implemented in Siemens FEMAP 10.2 as preprocessor and post-processor, and NX Nastran 7.1, as solver. FEA software was described in detail in previous research [25]. Figure 3 shows the two developed FEA models for the biceps and triceps FEM analysis.

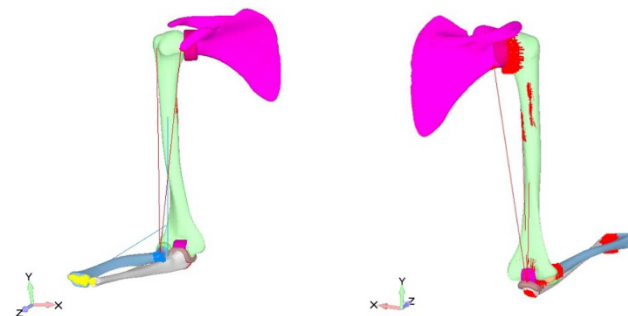
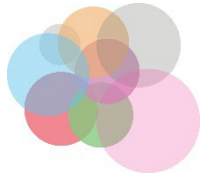


Figure 3. FEA model, biceps study (left) and triceps study (right), 74° aperture angle.

2.1.1. FEA Model Creation Meshing Process, Element Types

Different types of elements have been used for different components:

- **Bones.** They are meshed using 3D elements (tetrahedral type, 2nd order, 10 nodes). Teo et al. [26] use 3D elements for meshing neck and head parts of the joints in their study. In the same way, Donahue et al. [27] use 3D elements with axisymmetric condition, to model knee parts involved in the study.
- **Cartilages.** They are meshed using 3D elements (tetrahedral type, 2nd order, 10 nodes). Donahue et al. [27] and also Abidin et al. [28] model cartilages using 3D elements.
- **Bone cartilage junction.** Since the cartilage and bone form a solidary union between both parts, during the meshing process, a coincidence of the nodes of the bone and the cartilage in the contact area has been carried out. By ensuring that the bone and cartilage nodes coincide in that area, it is possible to ensure that the transmission of efforts between both parts is correct.



- **Contact between cartilages.** It has been modeled using rigid elements that join the nodes of both cartilage meshed parts involved in the joint, as explained in previous studies [25], using an RBE2 element. In the RBE2 elements used, the degrees of rotation freedom have been released so that only forces are transmitted in the X, Y or Z axes.
- **Muscles.** These are meshed using 1D elements (the type is discussed in this study). Previous studies have considered muscles in this manner, as 1D element acting along the imaginary axis of the muscle. Alonso et al. [20] model muscles in the leg to study human march in the same manner, 1D elements aligned with the axis of the muscles; Sackenvik et al. [29] use the same approximation for muscles, Parekh [30] approximates muscles to a 1D FEM element aligned with the muscle axis. This is a simplification, but it is used by several authors because fusiform muscles only have two insertion points and low pennation angles. Thus, fibers are aligned so with this imaginary axis which is formed by joining both insertion points.
- **Tendons.** These are meshed in combination with muscles, as a unique element musculotendon (MTU). The same 1D element is considered for tendons.
- **Tendon insertion in bone.** It has been modeled by adding a rigid element (RBE2), joining the end node of the 1D MTU element with several nodes in the bone, and thus distributing the reaction force in the area of the insertion of the tendons, as explained in previous research [25]. Of the restrictions that an element with these characteristics can apply, only the displacements in the X, Y or Z axes have been applied, freeing the turns around them. In this way, and since the MTU only transmits axial forces, we eliminate any possible bending moments that may affect the insertion of the tendon in the bone.

The RBE2 elements constitute constraint equations between the nodes to which the element is connected, they are characterized by having a node, called independent, and whose degrees of freedom are independent. The rest of the nodes connected to the RBE2 element are called dependent, and in them it must be fulfilled that the displacements of the independent node must be the same as in the dependent nodes.

That is why they are commonly known as rigid elements, since the distance between the component nodes remains unchanged.

In Figure 4, an example of an RBE2 element can be found, with an independent node (i) and four dependent nodes (j1 ... j4).

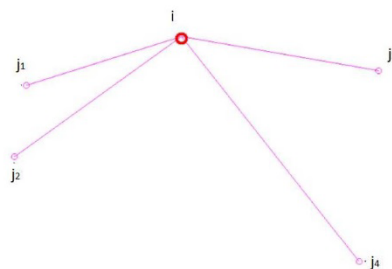


Figure 4. RBE2 element example: 1 Independent node and 4 dependent nodes.

The force balancing equation on an RBE2 is such that the reactions that appear on dependent nodes of the element must balance the force applied in the independent node:

$$\sum_{dependent} \{Reactions\ forces\} = F_{independent} \quad (1)$$

In this way, using an element of this type to simulate tendon insertion in the bone, the stress that appears in the 1D element that simulates MTU in the insertion zone is distributed, eliminating stress concentrations.

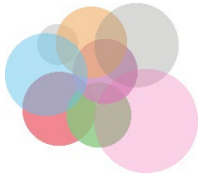


Figure 5 shows the application of this insert model.

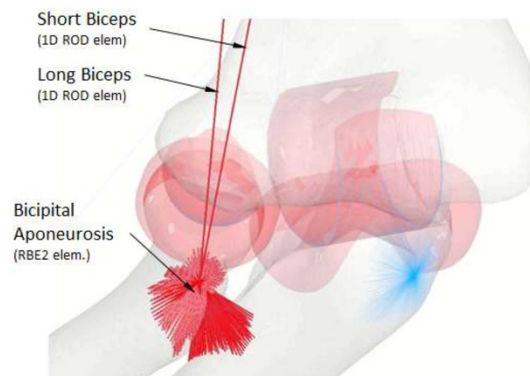


Figure 5. Biceps tendon insertion using RBE2 element.

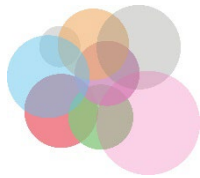
2.1.2. FEA Model Creation Muscles Modeling

Given that the proposed study is focused on knowing the forces developed by each of the muscles of the elbow joint, in different load situations, the modeling of the muscle must be such that it allows one to obtain these values.

As mentioned above, the objective is to be able to use a 1D element that allows us to reuse the mesh created to solve the FEM simulation without the need to re-mesh the entire model for each position of the joint.

The behavior of the element used in our study to simulate the muscle in the joint must be such that:

- It works only in traction.
- The force of the element used must be generated along the axis that joins the two nodes that define it; it must not have transverse forces.
- It must not present deformations or displacements that alter the isometric character of the contraction studied. Therefore, the deformations have to be low.
- The 1D-type elements existing in FEM analysis tools are described below, also indicating the reasons that lead us to discard them as valid for our purpose.
- Constant spring stiffness. In this case, the results are not valid, since a spring delivers the force along a curve according to the equation $F = K \cdot x$, where K is its stiffness (N/mm) and x is its elongation (mm). The use of this element would mean that a muscle would deliver a single force value for a given length, which is not true. A muscle at a given length is capable of delivering a force between 0 and the maximum force that muscle fibers are capable of developing.
- Variable spring stiffness. For the same reason as before, although in this case the maximum force could adjust according to the length of the muscle, it does not serve the purpose of the proposed study. The muscle can deliver a force between 0 and the maximum force for a given length. In the case of using a spring element, there is also the drawback of not always being able to balance the joint and the present study would no longer be isometric.
- Beam element. With this element, it is possible to obtain the necessary force in the joint to balance it. Beam elements are characterized in that they present axial, shear, and bending forces, and the muscles only transmit axial tensile forces.
- Rod element. The use of this element meets the study requirements; it only transmits axial forces, and we can decide on a rigidity in it such that the joint does not present large displacements and therefore it is no longer considered an isometric contraction.



In search of the element that best reflects the behavior of the muscle within the joint, the ROD element, given that it only presents axial forces, is the one that best suits the needs of the proposed mathematical study. However, the said element can work under tension and/or compression, which would alter the results. That is why two different FEM models are generated for the study of the biceps or triceps muscles. One in which the triceps muscles are re-moved, and another in which the biceps muscles are removed. In this way, and considering the direction of the applied load, it can be ensured that the analyzed muscles work only under traction.

Several tests have been carried out to define the characteristics of the ROD element that will be used, maintaining the objective of minimizing its stiffness and keeping the element deformations at low values so as not to lose the isometric load condition in the analyzes.

For these reasons, the elements used in the FEA models are defined as:

- Rod element (1D element);
- Constant cross section: 100 mm²;
- Material parameters:
 - Young modulus: 2000 MPa;
 - Poisson coefficient: 0.43.

This type of element transmits only axial forces. Its constitutive equation is shown below.

$$\begin{bmatrix} F \\ M \end{bmatrix} = \begin{bmatrix} K_x & 0 \\ 0 & K_T \end{bmatrix} \cdot \begin{bmatrix} \varepsilon_x \\ \alpha_x \end{bmatrix} \quad (2)$$

where

- F is the axial force in the element;
- M is the moment along the axis of the element;
- K_x is the axial stiffness of the element;
- K_T is the rotational stiffness along the axis of the element;
- ε_x is the length variation of the element;
- α_x is the angle rotated along the element axis;

To ensure that no torsional stresses are produced in the element, it is sufficient to assign a value of $K_T = 0$.

2.1.3. Material Properties

The meshing process of the 3D bodies that comprise the model, bones and cartilage, has been carried out automatically using the meshing tool that includes Femap. The geometric characteristics of the elements used are common to both bones and cartilage; they are:

- Second-order tetrahedral elements, 10 nodes;
- Nominal element size 1.5 mm.

The materials considered for each body have the following characteristics, taken from the CES Edu Pack [31].

The use of an isotropic and linear material for cartilage has previously been used in those cases in which the duration of the applied load is less than the time constant of the viscoelastic properties of cartilage, around 1500 s. In this case, given that it is considered an isometric contraction, in which there is no displacement in the joint and the application time is considered well below the 1500 s indicated, it is possible to consider it as isotropic, linear, and elastic with the properties indicated in Table 1. Donahue et al. [27] considered isotropic, linear and elastic tissue for cartilages in the knee joint in their study of tibio-femoral contact in the human knee joint, based on the time required to apply the load compared to the time constant for viscoelastic properties of cartilages, for the same reason. Abidin et al. [28] consider cartilage as an isotropic, linear and elastic material in their study on a human knee joint model and verification.

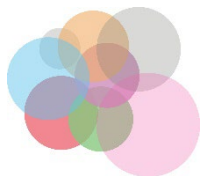


Table 1. Tissue properties for bones and cartilages.

	Young's Modulus (Mpa)	Poisson	Density (kg/m ³)
Bone	17,200	0.41	1790
Cartilage	8	0.49	1150

The use of an isotropic material for bone modelling is known and has previously been used, Teo et al. [26], meshed head and spine using 3D elements with bone tissue as isotropic, linear, and elastic.

2.1.4. Mesh Resume

The summary of the FEM mesh is as follows:

- Number of elements: 548,139;
- Number of nodes: 821,755;
- Number of second-order tetrahedral elements: 546,035;
- Number of 1D ROD elements: 4 (for biceps analysis) 3 (for triceps analysis);
- Number of rigid elements RBE2: 2100.

A study of the quality of the mesh, based on the Jacobian value of the elements, shows us how good its quality is for our objective. Figure 6 shows the elements that are below the limit value of 0.6 (good quality for the considered analysis).

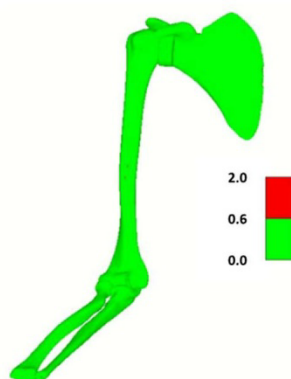


Figure 6. Mesh quality. Jacobian criteria for evaluating mesh quality. Value represented is (1-measured value on element/ideal value on element). Maximum threshold defined as 0.6.

2.1.5. Boundary Conditions

For the boundary conditions, the scapula is considered fixed in the area (pink color on the below image) where the muscles of the scapula would be inserted. These muscles (see Figure 7) are:

1. Levator scapula muscle;
2. Rhomboid minor muscle;
3. Rhomboid major muscle.

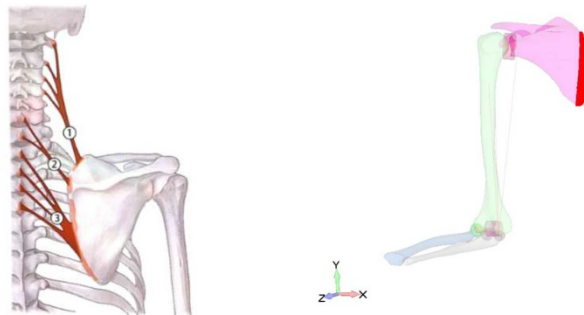
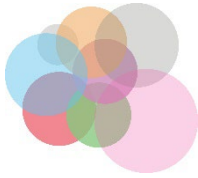


Figure 7. Boundary condition, muscles fixing scapula (left), constrain in FEA model, (right), scapula fixed.

2.1.6. Load Application

In order to be able to validate results of the analysis, the load considered must be easy to replicate in a controlled environment; that is why we have defined the load as the mass of a dumbbell of 15 kg (approximately, 150 N) for the biceps analysis. In the case of triceps loading, the simulation is more similar to the exercise done using a guided load machine.

The application point of this load is the wrist (the hand is not modeled) and although this can affect slightly to the results, due to the difference in distance of the load applied in the joint, this can be solved varying the load as needed when applied in the hand.

All models have the same load applied, at the same node, and in the same vertical direction (see Figure 8).

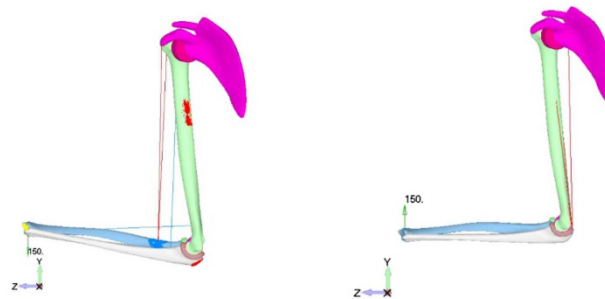


Figure 8. Applied load (150 N) in wrist, vertical direction, biceps analysis (left) and triceps analysis (right).

The load is applied in vertical direction, positive axis (ascending) when analyzing the triceps and vertical direction, negative axis (descending) when analyzing the biceps. All load cases are considered as static load cases, isometric contraction of the muscle, and, to avoid compression loads on muscles, two different models have been developed; one of them includes only biceps muscles, with descending load, and the other one includes only triceps muscles, with ascending load.

In the model, gravity force has also been considered. Thus, the mass of the arm is considered in the analysis.

To do so, the density of elements has been modified so that the values of the upper arm and forearm mass are [32]:

- Forearm mass: 2.5 kg;
- Upper arm mass: 4.3 kg.



All load cases are considered as static load cases; due to the isometric contraction of the muscle, there must be no displacement in the joint.

2.1.7. Design of Experiment

The FEM model is considered linear and static. No non-linearity has been considered, neither in the materials nor in the restrictions of the model. That is why the general equation of the calculation is so important and can be written as follows:

$$[R] = [K] \cdot [U] \quad (3)$$

where

$[R]$ = global loading matrix;

$[K]$ = global stiffness matrix;

$[U]$ = global displacement matrix.

Equation (3) can be solved directly using the Gauss elimination method, there is no iterative process, and the solution is unique.

The resolution of the problem is carried out through several linear static analyses, each one of them representing a different load condition in the joint, and the result of them is the necessary effort in each muscle of the joint to balance the applied force on the wrist.

As mentioned above, there are two different models, thus guaranteeing that the muscles work only in traction, to study the biceps and the triceps muscles.

In each of the models, the configuration of the joint will be modified by varying its angle between 44 and 164°, at intervals of 10°.

The applied load will always be 150 N.

As a reference for the reading of the results, the fully extended arm corresponds to 164 degrees and the fully flexed arm corresponds to 44 degrees (see Figure 9).

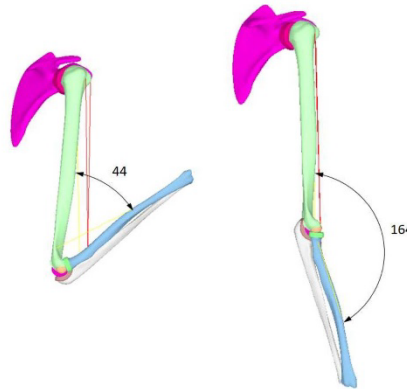


Figure 9. FEA model, minimum (left) and maximum (right) studied angles.

2.1.8. FEA model Solution

The FEA model has been solved on a workstation with the following characteristics.

- Windows 10, 64-bit, operating system;
- Intel(R) Core (TM) i9-10900X CPU @ 3.70 GHz Processor (20 cores);
- 128 GB RAM memory;
- 1 TB SSD disk drive.

The solution time for any configuration is approximately 20 min.

As contacts in cartilages have been replaced by rigid elements, RBE2 type with free rotation on both nodes, the solution time has been reduced from 67 h to 20 min.



As described in a previous study [23] the results, comparing the model with contacts and the model with rigid elements instead, are similar in both cases, and therefore the use of rigid elements improves the efficiency of the analysis.

2.2. Mathematical Approach

In this work, several mathematical models are derived from the results of the FEM simulation. First, there are grade 6 polynomial curves, which represents the efforts of the five muscles involved for the simulated cases, described in previous section, for a vertical force of 150 N applied on wrist. Muscle effort is defined as the total force that appeared in the 1D element used for each muscle in FEM models. These models make it possible to obtain this effort for each muscle by following an equation for each muscle, with the following general formulation:

$$Muscle_{effort} = \sum_{i=0}^{i=6} a_i \cdot \alpha^i \quad (4)$$

Once the forces needed in muscles have been calculated to balance the joint, with different angles and with a load of 150 N applied at the wrist, a mathematical approach will be proposed for obtaining these efforts for other loads, due to the mechanism type considered in the elbow joint, the applied forces and torque analysis done in the joint. The value of the force exponent is 1 in all cases, so there is a linear relation between the force applied and the muscle reaction force [33]. According to this, the greater the force applied to the wrist is, the greater the muscle force must be, and both values are proportional. Figure 10 and the following equations justify this simplification in the analysis, where:

- W_s means the weight of the forearm;
- W_s' means the equivalent force for W_s applied at the wrist. The moment created by W_s or W_s' is the same referred to O point;
- W_q means the load applied in the wrist;
- α means the angle between biceps axis and the normal to the radius longitudinal axis;
- β means the angle between the vertical and the longitudinal axis of the radius;
- a , b and c are geometrical dimensions.

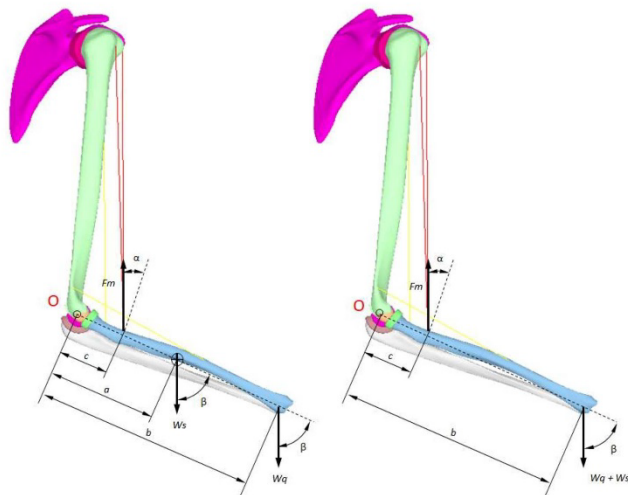
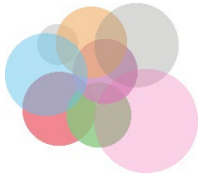


Figure 10. Forces scheme in elbow joint.



The left image considers the proper mass of the arm located in its center of gravity. Balancing equation on that forces scheme, referred to O point scheme, is:

$$F_m = \frac{(W_s \cdot a + W_q \cdot b) \cdot \sin \beta}{c \cdot \cos \alpha} \quad (5)$$

Instead of using W_s , we can use W_s' , applied on the wrist (right image). This W_s' should comply:

$$W_s \cdot a \cdot \sin \beta = W_s' \cdot b \cdot \sin \beta \quad (6)$$

So, the balancing equation would be the following:

$$F_m = (W_q + W_s') \frac{b \cdot \sin \beta}{c \cdot \cos \alpha} \quad (7)$$

As seen in Equation (7), the relation between F_m and $W_q + W_s'$ is linear, and constant for a certain angle only depending on b and c and α and β angles.

According to this, the forces in the muscles when the load is different from the 150 N applied in this study (including the W_s' in this 150 N) can be obtained by the relation between the external load applied and the reaction in the muscle.

$$\frac{W_q + W_s'}{F_m} = \frac{F_{ext}}{R_{musc}} = \frac{c \cdot \cos \alpha}{b \cdot \sin \beta} = constant(\alpha, \beta) = \frac{150 \text{ N}}{F_m(150 \text{ N})} \quad (8)$$

In summarizing, forces in muscles depend on:

- Muscle length (or elbow angle);
- Force applied to wrist.

By applying these mathematical approaches, several polynomial equations have been obtained, one per muscle, which establishes the relation between force in muscle as:

$$F_m = F_m(W, \alpha) \quad (9)$$

where:

- F_m is the force in the muscle (N);
- α is the angle of the elbow;
- $W = W_q + W_s'$, is the load applied on the wrist (N).

In all mathematical models of muscle efforts, the effort could be obtained for any angle, α , and for any vertical load, f_v . If the forearm mass, m_{fa} , is considered and g is the gravity acceleration (9.81 m/s^2) W_s' can be expressed as follows:

$$W_s' = W_s \cdot \frac{a}{b} = m_{fa} \cdot \left(\frac{a}{b} \cdot g \right) \quad (10)$$

Thus, for obtaining the effort:

$$Effort(f_v) = F_m \left(f_v + m_{fa} \cdot \left(\frac{a}{b} \cdot g \right), \alpha \right) \quad (11)$$

General equations will be obtained describing the effort in each muscle depending on the angle of the joint and the load applied. These equations and their surface representations will be the results of this research. Both of them, the equations and surfaces, have been obtained using the curve fitting tool from MATLAB. It has been noted that these effort models for the five involve muscles are only used to study isometric muscular contractions, when muscle length does not vary.



3. Results

The model analyzed includes all muscles in the elbow joint, but the post-processing results are only applied to the biceps and triceps muscles.

3.1. FEM Simulations, 3D Results

Figure 11 shows the results of running the design of the experiment described hereinbefore for the 13 cases of elbow angle for the biceps model. The figure represents the Von Mises stress in the involved bones due to the action of the 150 N as a vertical force (descendent) on the wrist. Each of the images represents the stress map resulting from the study of an isometric contraction in the elbow joint at the angle indicated in the image for a load of 150 N applied to the wrist, vertically and downward.

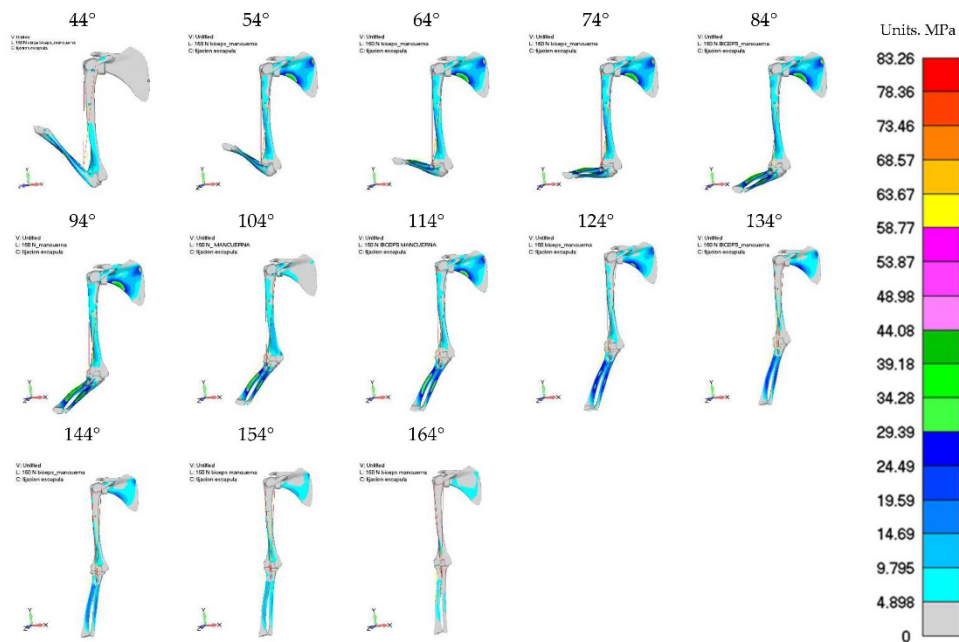


Figure 11. Different elbow angle configuration: bone Von Mises stress distribution under biceps loading (150 N on wrist) cases. Figures are included in more quality in Appendix A.

Similarly, Figure 12 shows the Von Mises stresses on the bones due to the FEM simulation of 150 N of vertical load (ascendant) on the wrist. Each of the images represents the stress map resulting from the study of an isometric contraction in the elbow joint at the angle indicated in the image for a load of 150 N applied to the wrist, vertically and upward.

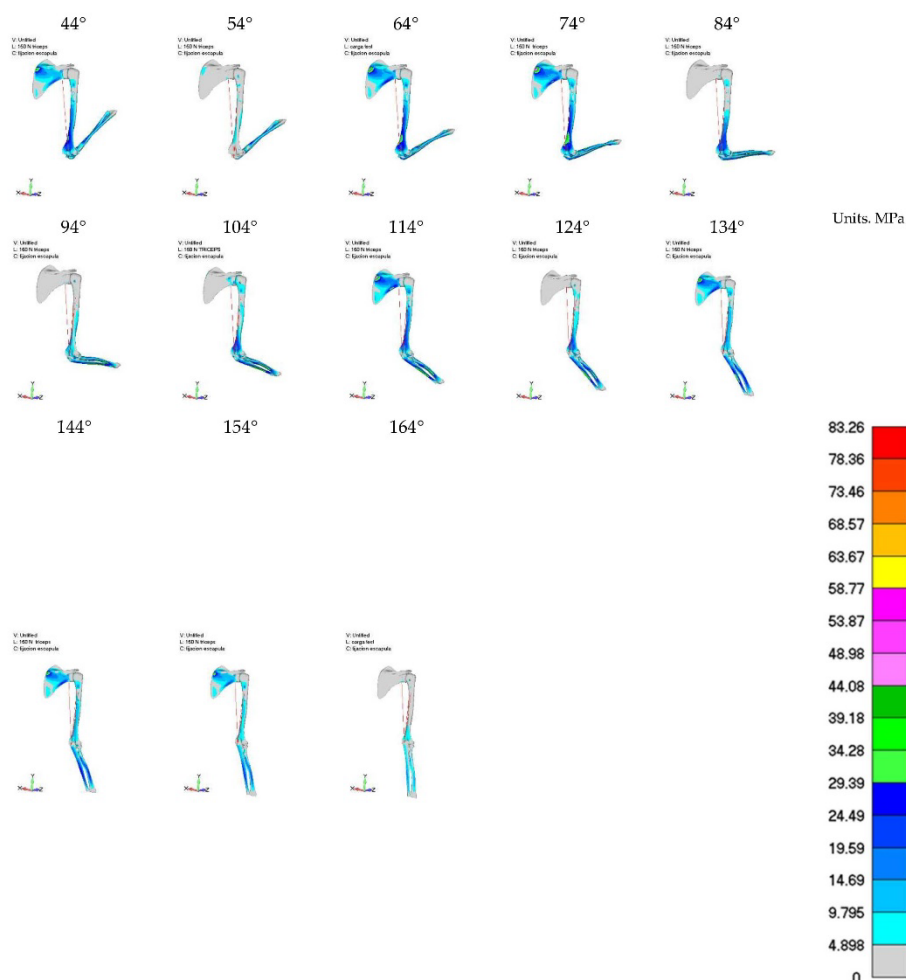
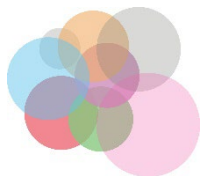


Figure 12. Different elbow angle configuration: bone Von Mises stress distribution under triceps loading (150 N on wrist) cases. Figures are included in more quality in Appendix A.

3.2. Numerical Axial Efforts on Main Different Muscles

This subsection collects axial efforts on biceps and triceps involved muscles for the 26 FEM simulations (13 for biceps and 13 for triceps). These efforts correspond to the reaction obtained in the muscle itself, and length corresponds to the length of the muscle (the whole muscle length including tendons) measured in the FEA model, in that certain elbow angle. Table 2 has these results if the form of force and the corresponding length for this applying force for all muscles studied:

- Biceps muscles (long and short biceps);
- Triceps muscles (lateral, medial and long triceps).

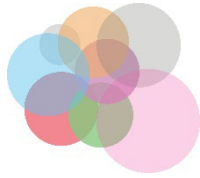


Table 2. Axial efforts (N) in main simulated muscles, 150 N applied on wrist. Isometric contractions at each angle value considered.

Elbow Angle (°)	Long Biceps		Short Biceps		Lateral Triceps		Medial Triceps		Long Triceps	
	Force (N)	Length (mm)	Force (N)	Length (mm)	Force (N)	Length (mm)	Force (N)	Length (mm)	Force (N)	Length (mm)
44	244	373	96	310	148	217	70	211	269	284
54	330	375	136	312	128	216	144	209	312	282
64	341	383	152	320	212	215	115	208	340	281
74	341	391	161	328	208	215	119	208	347	281
84	340	399	168	336	235	214	150	207	340	280
94	333	407	170	344	263	214	186	207	347	279
104	302	415	181	351	233	213	164	207	331	279
114	301	422	159	359	233	213	176	206	307	278
124	274	429	143	365	213	211	170	204	271	276
134	261	435	95	371	177	207	150	201	226	273
144	185	439	87	375	131	203	120	197	175	269
154	121	442	46	379	79	199	82	192	119	264
164	52	445	1	381	24	194	40	188	58	260

3.3. Mathematical Models for Muscle Efforts

3.3.1. Efforts for 150 N of Vertical Load

The obtained R-square values are 99.3%, 98.2%, 99.9%, 93.6%, and 97.2%, respectively (see Table 3). Showing results in graph form, one graph for biceps muscles and the other for triceps muscles, a tendency can be observed in both of them. Figure 13 shows the efforts of biceps muscles at each angle (dots) and the mathematical curves (dash point line) that approximate them. The effort in muscle (y -axis on the graph) is related to the elbow angle (x -axis on the graph). The red color is associated with long biceps and the blue color is associated with short biceps. These curves represent the total effort of each muscle involved. Thus, long biceps, for a constant vertical load on the wrist, including forearm mass, reach the maximum effort (340 N) between 60–90°, while for short biceps the maximum effort (170 N) is achieved for the range of 90–110°.

Table 3. The equations developed as polynomial curves of grade 6th, based on the general formulation of Equation (4), has the following constant values for each muscle involved.

Muscle	a_0	a_1	a_2	a_3	a_4	a_5	a_6	SSE	R ²	RMSE
Long biceps	−2323 (−3730, −916.1)	136.6 (35.39, 217.8)	−2.7 (−4.477, −0.9232)	0.02579 (0.007278, 0.04431)	−0.0001193 (−0.0002116, −2.692 × 10 ^{−5})	2.116 × 10 ^{−7} (3.433 × 10 ^{−8} , 3.889 × 10 ^{−7})	0	694.5	0.993	10.76
Short biceps	−409 (−1729, 911.5)	24.12 (−52.09, 100.3)	−0.4361 (−2.104, 1.232)	0.004284 (−0.01309, 0.02166)	−2.182 × 10 ^{−5} (−0.0001085, 6.486 × 10 ^{−5})	4.28 × 10 ^{−8} (−1.236 × 10 ^{−7} , 2.092 × 10 ^{−7})	0	367.3	0.982	7.82
Long Triceps	−1550 (−3477, 377.3)	110.4 (−23.88, 244.7)	−2.702 (−6.433, 1.028)	0.03508 (−0.01795, 0.0881)	−0.0002503 (−0.0006584, 0.0001577)	9.18 × 10 ^{−7} (−7 × 10 ^{−7} , 2.536 × 10 ^{−6})	−1.354 × 10 ^{−9} (−3.945 × 10 ^{−9} , 1.236 × 10 ^{−9})	102.3	0.999	4.13
Medium Triceps	−6448 (−1.38 × 10 ⁴ , 908.4)	442 (−70.71, 954.7)	−11.9 (−26.14, 2.344)	0.1638 (−0.0386, 0.3662)	−0.001217 (−0.002775, 0.0003408)	4.643 × 10 ^{−6} (−1.534 × 10 ^{−6} , 1.082 × 10 ^{−5})	−7.144 × 10 ^{−9} (−1.703 × 10 ^{−8} , 2.743 × 10 ^{−9})	1491	0.936	15.76
Lateral Triceps	4661 (−3110, 1.243 × 10 ⁴)	−311.3 (−852.8, 230.3)	8.397 (−6.645, 23.44)	−0.1143 (−0.3281, 0.09946)	0.0008424 (−0.0008029, 0.002488)	−3.214 × 10 ^{−6} (−9.738 × 10 ^{−6} , 3.309 × 10 ^{−6})	4.975 × 10 ^{−9} (−5.468 × 10 ^{−9} , 1.542 × 10 ^{−8})	1664	0.972	16.65

Analogously, Figure 14 shows the efforts of the triceps muscles (dots) and the mathematical curves (dash point line) which approximate them. The effort in the muscle (y -axis in the graph) is related to the elbow angle (x -axis in the graph). In the figure, the red color is associated with medium triceps, the green color is associated with long triceps, and the blue color is associated with the lateral triceps. These curves can be used to know efforts on the three muscles involved in the triceps FEM model. Additionally, as an example, if a muscle had limited effort for any



reason, we could know which positions of the elbow joint are not recommendable to use. Therefore, for an isometric contraction with 150 N as a vertical load on the wrist (including the forearm mass), for a maximum effort of 200 N in the lateral triceps, it is necessary to avoid placing the elbow joint between 65° and 130 degrees approximately.

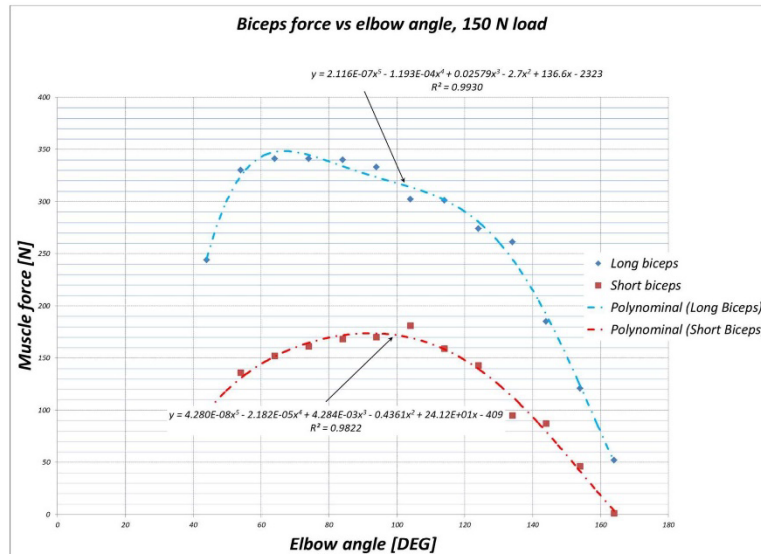


Figure 13. Calculated (dots) and tendency polynomial equation (dash point line) comparison, for biceps forces (y-axis). Isometric contractions on the elbow at different positions, from 44 to 164 degrees (x-axis) and 150 N are loaded on the wrist.

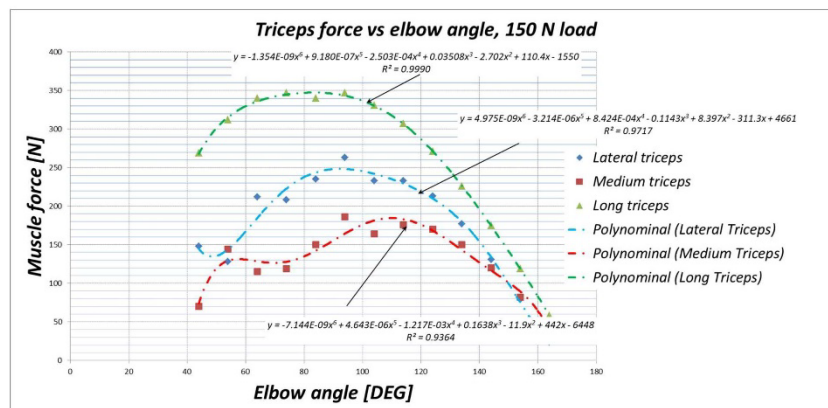


Figure 14. Comparison of the (continuous line) calculated and (dash point line) comparison, for triceps forces (y-axis). Isometric contractions on the elbow at different positions, from 44 to 164 degrees (x-axis) and 150 N are loaded on the wrist.



3.3.2. General Formulation

Applying the mathematical approach described above, equations of efforts for the five muscles involved will be presented, for any wrist load and elbow angle.

The approximation is done according to a fourth-grade polynomial with the general equation:

$$f(\alpha, w) = p00 + p10 \cdot \alpha + p01 \cdot w + p11 \cdot w \cdot \alpha + p20 \cdot \alpha^2 + p21 \cdot w \cdot \alpha^2 + p30 \cdot \alpha^3 + p31 \cdot w \cdot \alpha^3 + p40 \cdot \alpha^4 \quad (12)$$

where α is the angle of the elbow and w is the load applied to the wrist.

All coefficients have been calculated for a 95% confidence bound.

The surface of Figure 15 shows the tendency of the effort appearing in the short biceps muscle as a result of applying a vertical load on the wrist. As an example, for a constant vertical load of 750 N (including forearm mass), we could limit the short biceps effort to 200 N if we put the elbow joint angle between 150° to 164°.

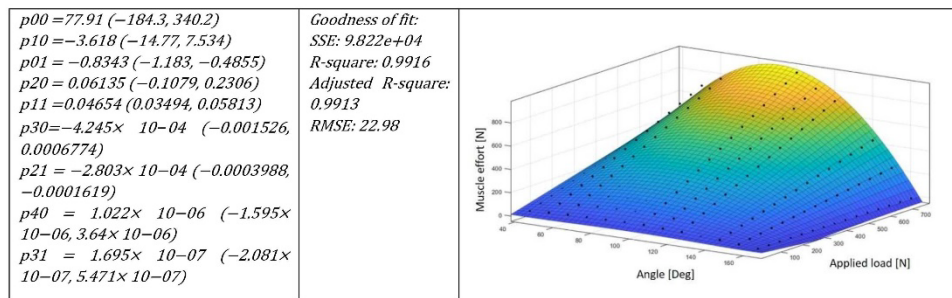


Figure 15. General formulation of short biceps efforts in function of wrist load (w) and elbow angle (α) and surface reconstruction of the proposed polynomial.

In the range of the example of the previous figure, 150–164°, we could find an effort on long biceps bigger than in short biceps, reaching around 600 N. Figures 16–19 collect the results of the surface reconstruction for the rest of studied muscles.

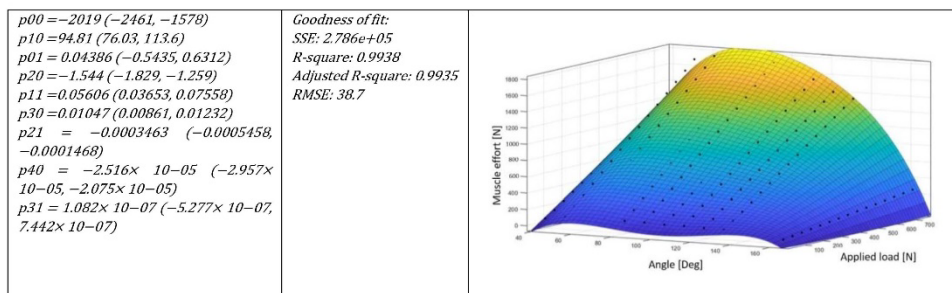


Figure 16. General formulation of long biceps efforts based on wrist load (w) and elbow angle (α) and surface reconstruction of the proposed polynomial.

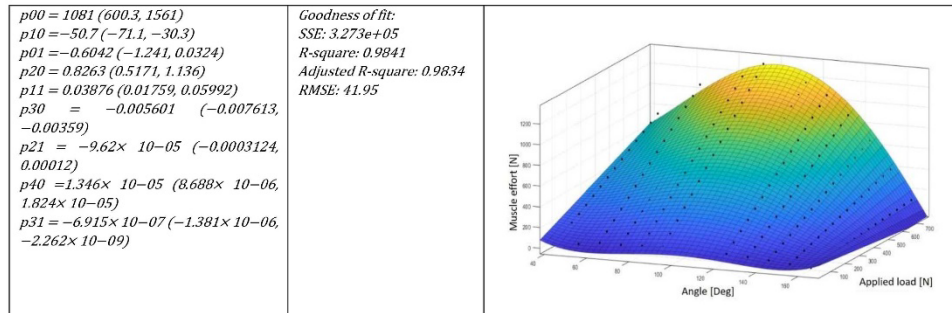
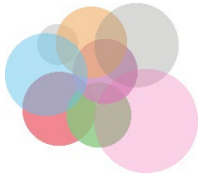


Figure 17. General formulation of lateral triceps efforts based on wrist load on the wrist (W) and elbow angle (α) and surface reconstruction of the proposed polynomial.

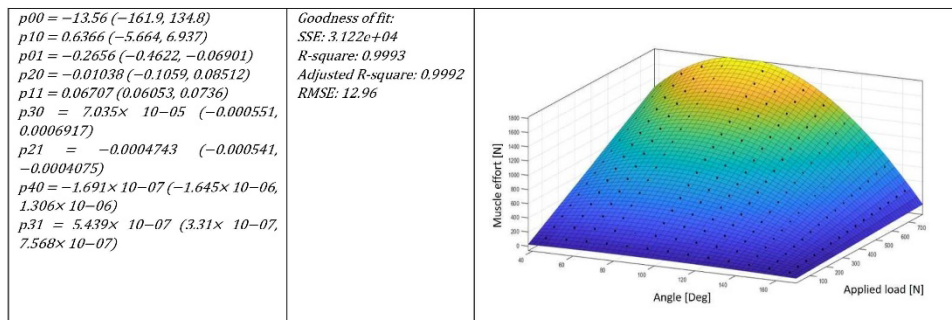


Figure 18. General formulation of long triceps efforts based on wrist load (W) and elbow angle (α) and surface reconstruction of the proposed polynomial.

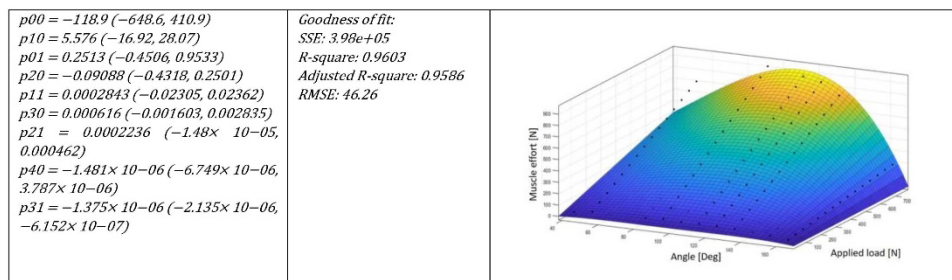


Figure 19. General formulation of the medium triceps efforts based on wrist load (W) and elbow angle (α) and surface reconstruction of the proposed polynomial.

4. Discussion

In this study, an analytical calculation method is proposed to obtain the forces in the flexor muscles (long and short biceps) and extensors (long, lateral, and middle triceps) in any configuration of the opening angle of the elbow joint and to any load applied to the wrist. The final objective of this study is to allow the creation of a new type of finite element



that can be included in a musculoskeletal model that can be analyzed in different positions and configurations without the need to repeat the model meshing process.

There are several studies along the same lines. Therefore, Martínez [12] is focused on the analysis of a certain muscle in a certain load situation, analyzing what happens in the muscle itself. Islan et al. [14], are focused on the muscles involved in the shoulder joint, under static condition predicting the fatigue and cumulative damage to the muscles. Alonso et al. [20] study the forces in the muscles involved in the human march under certain load conditions, weight and step frequency. Park et al. [22] develop a predictive method for muscle forces based on electromyography techniques. Teo et al. [26] are focused on the effect of a collision on the neck muscles. As can be seen, none of them addresses the problem from our perspective. The use of the finite element method is widely recognized as a calculation tool for structural, static, and/or dynamic problems. However, characterization of the muscle, so that it is easy to include it in an FEM model has not been addressed until now.

The main drawbacks presented by the studies carried out so far, in order to include the muscle in a valid finite element model to analyze the behavior of a muscle within a joint in various positions of this, are as follows:

- Studies have focused on one joint position, obtaining results related to muscle effort that are difficult to extrapolate to different positions. Ramírez [12] obtains the results for the anterior tibial muscle under study conditions, but moving the results to other load situations should be difficult to solve.
- The equations obtained to characterize the muscle are difficult to implement in a FEM model. Sachenkov et al. [29] use 1D elements to simulate muscles, but there is no single equation for any muscle to implement it in another analysis. The same happens with Alonso et al. [20], where the result is based on a static and physiological optimization that integrates the forces of any Hill MTU unit.
- Several of these studies use 3D elements to represent the muscle, which allows a better understanding of the internal behavior of the muscle itself (distribution of stresses and strains, deformations, etc.) but does not allow the created mesh to be reused in other positions of the joint. Therefore, Martínez [12] studies the anterior tibia muscle in a certain position, and if the muscle length changes, a new mesh should be developed. Another example is shown in research by Islan et al. [14], where they analyze the shoulder muscles of a violin player holding the instrument, and so, if the shoulder position changes, the mesh will do as well, remeshing the muscles into their new position.

With this work, and in a first approximation, we have obtained the equations that govern the behavior of the flexor and extensor muscles of the elbow, only in isometric contractions and referring to this behavior to its response in the form of force provided, throughout the flexion path of the joint for any load applied to the wrist.

For example, to calculate the estimated force that must appear in each of the joint muscles to support a load applied to the wrist of 235 N with the elbow forming an angle of 119 degrees, we need to substitute the values of α and w in Equation (12), depending on which muscle we want to analyze.

Changing the indicated values in Equation (12) would obtain the following estimated forces in the muscles.

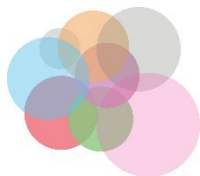
Vertical downward load, strength in Biceps muscles.

- Long biceps: 477 N;
- Short biceps: 245 N.

Upward vertical load, strength in the triceps muscles.

- Long triceps: 438 N;
- Middle triceps: 231;
- Side triceps: 304.

In addition to what has been said, the equations obtained to characterize the muscles in their response are easy to implement mathematically, which will undoubtedly help in



the future in the possibility of creating a new type of finite element that responds to these behavioral equations obtained.

With the creation of this new type of finite element to represent muscles, a very interesting possibility is opened to analyze the behavior of the joint in different positions, with variable loads, different load cycles, etc. notably reducing the study work since the meshing process of the model is eliminated by being able to reuse the mesh for the desired positions.

Regarding the analysis of the results obtained, it is worth mentioning the following important aspects of the study.

Applying the proposed methodology and using the finite element method, we can obtain the forces that must appear in the muscles involved in the elbow joint to keep it in balance, in any position between 44 and 164 degrees of opening, with a load applied to the wrist of 150 N (Table 2, Figures 13 and 14).

Through this same analysis it is possible to obtain the tensions that appear in the insertions of the tendons of the joint, which allows us to foresee damage before certain load situations (Figures 11 and 12).

This same analysis allows us to obtain a map of tensions in the rigid tissues of the joint, bones, and cartilage. This allows a detailed study of damage to the elements mentioned (Figures 11 and 12).

There are no previous studies that carry out studies similar to the proposed study, so a direct comparison with what has already been done is not possible. However, the trend observed in muscle effort during joint movement is similar to what other studies have published, [34], where the results of the torque in elbow vs. elbow angle shape are consistent with our results and [35] where the relation between the torque of the elbow and angle of the elbow is similar to our expressions, although these studies measure the moment in the joint (or what is the same, the sum of the moments generated by each muscle) and not the independent force in each muscle the results presented in these investigations are similar to those presented in this study in terms of the evolution of the moment in the joint, or of the force in the muscle in our case, along the angle of the joint.

The correlation also presents high R values, which confirms the validity of the analysis (Figures 13 and 14).

Through the application of classical mechanics and the balance of forces and moments in the joint, we can extrapolate the results obtained for a specific load, 150 N in our case, to different loads applied to the joint (Figures 15–19).

Finally, it should be mentioned that the equations that predict the force in each of the muscles in the joint are obtained as a function of the opening angle of the elbow and the applied load.

5. Conclusions

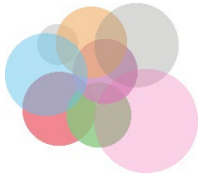
The use of a simplified model of the muscle, as a 1D element with axial forces only (ROD type element), into a FEA model allows one to calculate the forces that must be generated in the muscles involved in the joint.

It is also possible to predict the force required in each muscle for different load situations.

On the basis of the same analysis, the stresses that appear in other elements of the joint, bones, tendon insertions, or cartilage, can be obtained. With the obtained characteristic equations, different lines of research are opened focused on a better characterization of the muscles. It seems necessary to evolve the equations obtained in this study to obtain those that give the force of the muscle as a function of the speed of contraction. Or, based on studies conducted to assess muscle fatigue, they could be modified to predict it in a moving joint.

The evolution of this study should follow the following stages:

- Validation of the theoretical model obtained through testing with individuals with whom it is possible to measure the forces in cases of biceps and/or triceps load that, by means of comparison with this study, allows for knowing its validity.



- Modification of the equations obtained in this study in order to obtain their evolution for concentric or eccentric contractions.
- Creation of a new type of 1D element for application in finite element models that allows the behavior of a muscle to be characterized according to the equations obtained for any type of contraction, isometric, concentric, or eccentric.

Once these phases are completed, the work could be extrapolated to other types of muscle (flat, or peniform) in order to characterize more muscles that allow us to study new joints according to the proposed methodology.

One limitation of the presented study is the approximation of the muscle to the 1D element. This implies that following this methodology cannot obtain information regarding what happens inside the muscle.

Another one is that the analysis is performed for a certain size of elbow joint, and so the FEA model created is valid for this joint size; in any other case (children, older people, etc.), the FEA model should be resized, and the same methodology should be applied.

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Appendix A

In this appendix, the sub-Figures included in Figures 11 and 12 are going to be collected with higher quality.

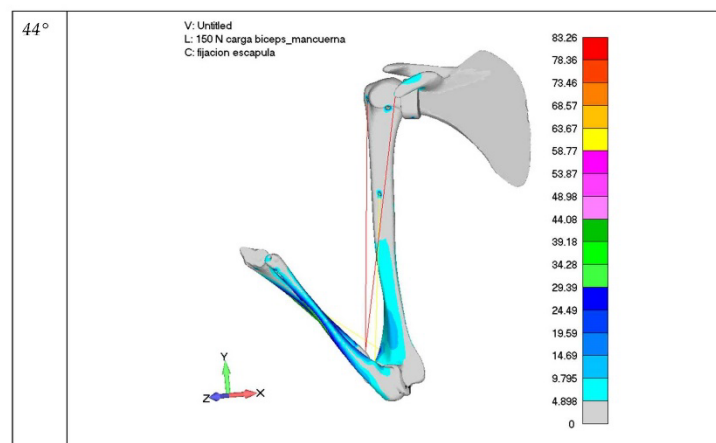


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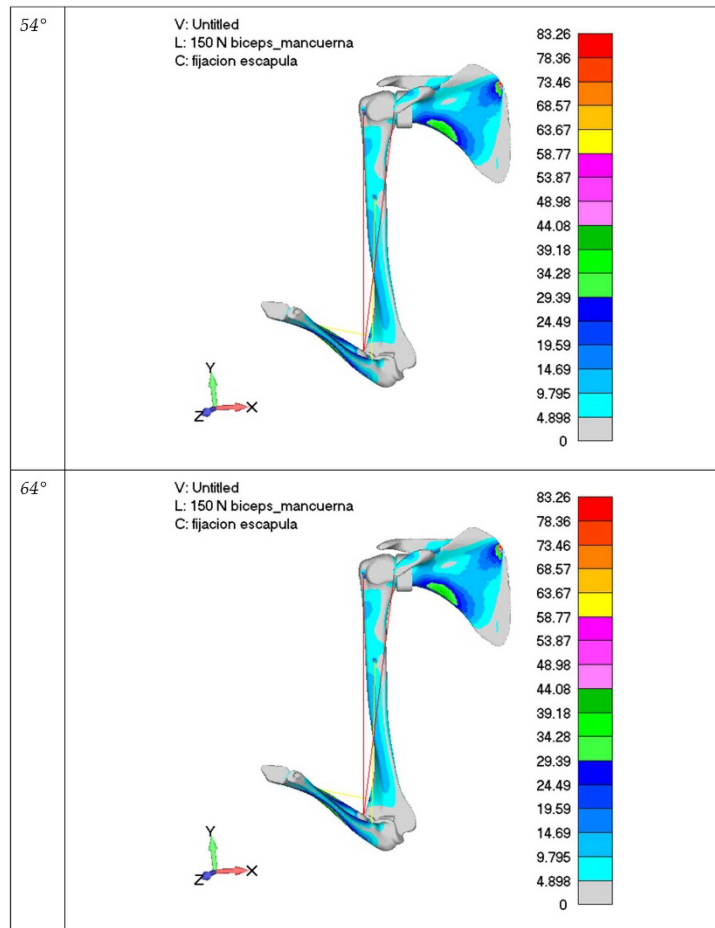


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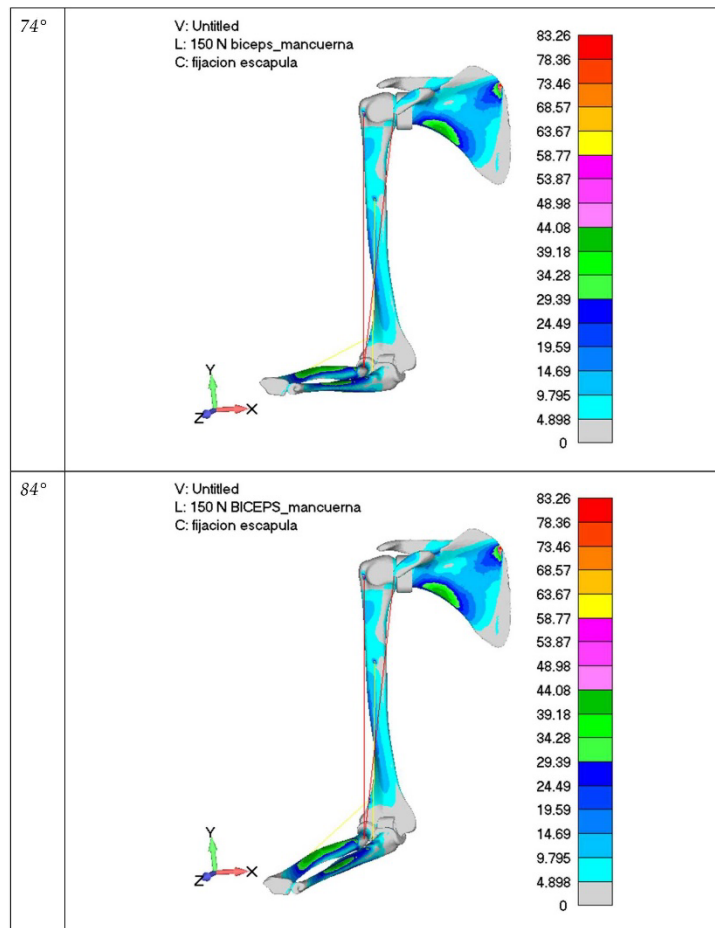


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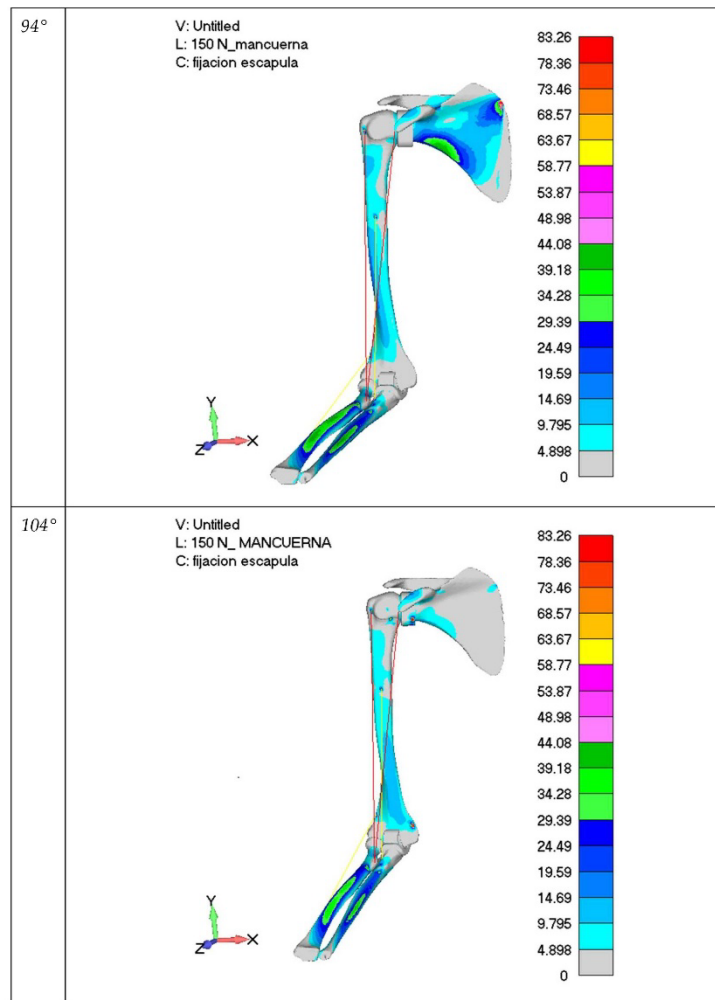
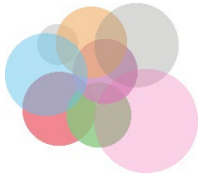


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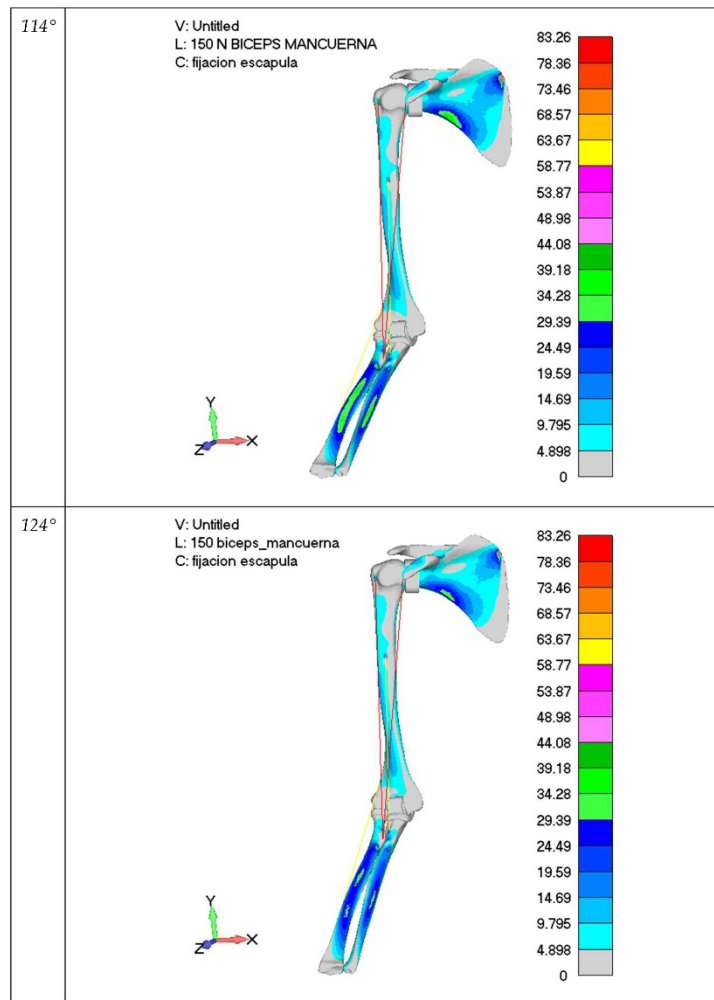


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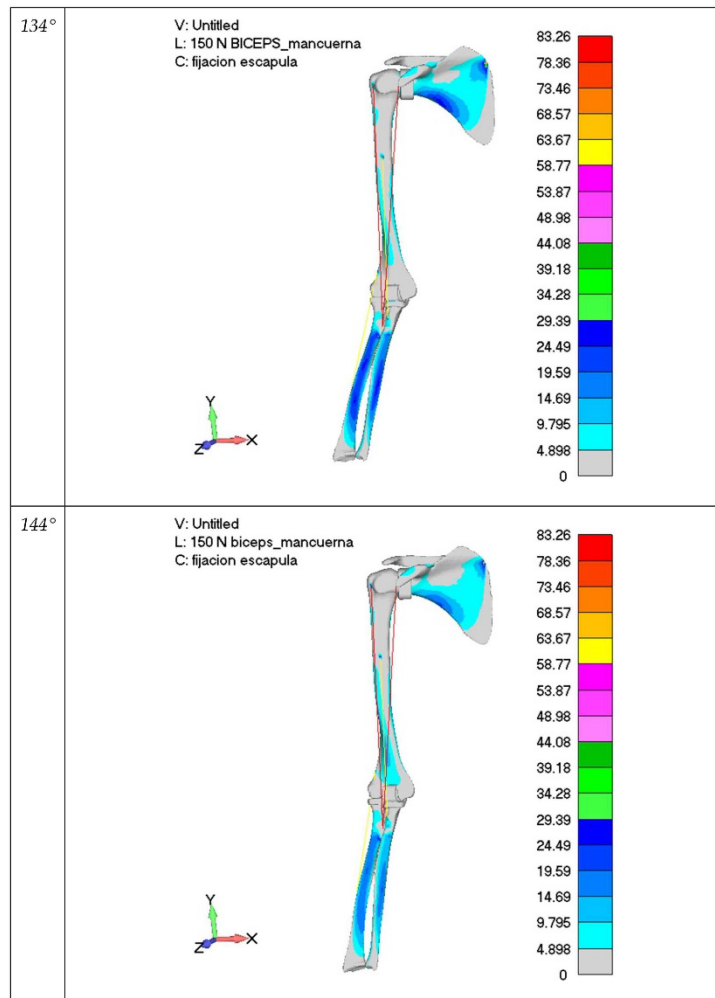
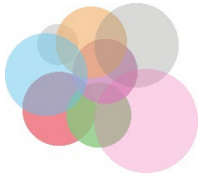


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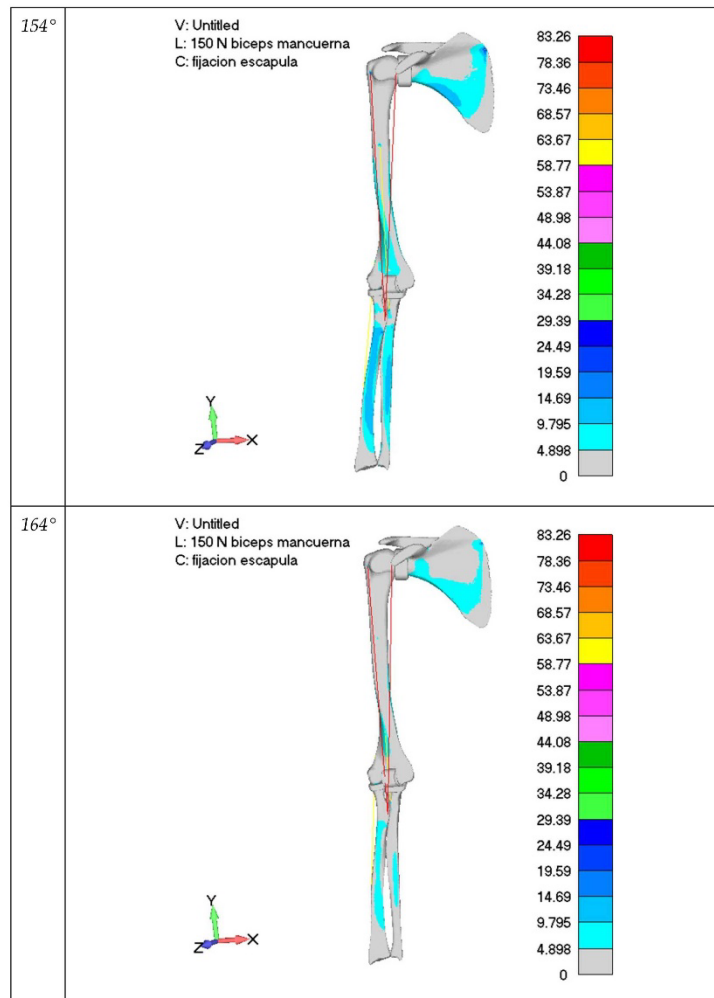
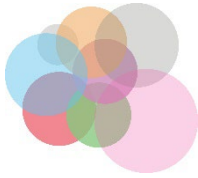


Figure A1. Different Elbow Angle Configuration: Bone Von Mises Stress Distribution Triceps Loading Case (150 N on wrist).

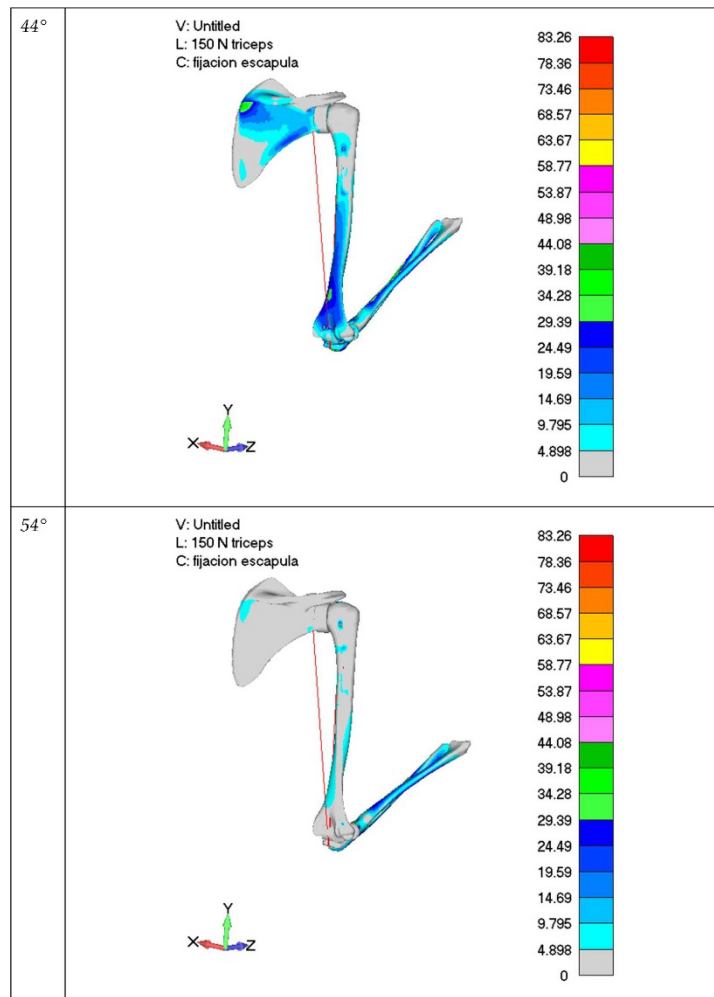
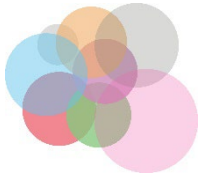


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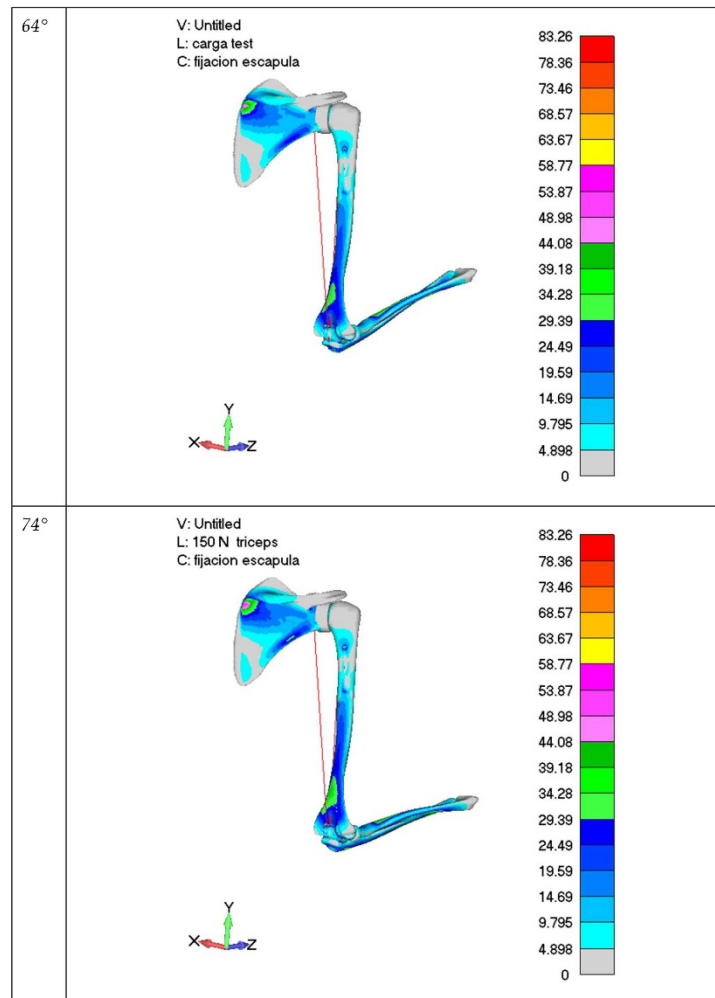


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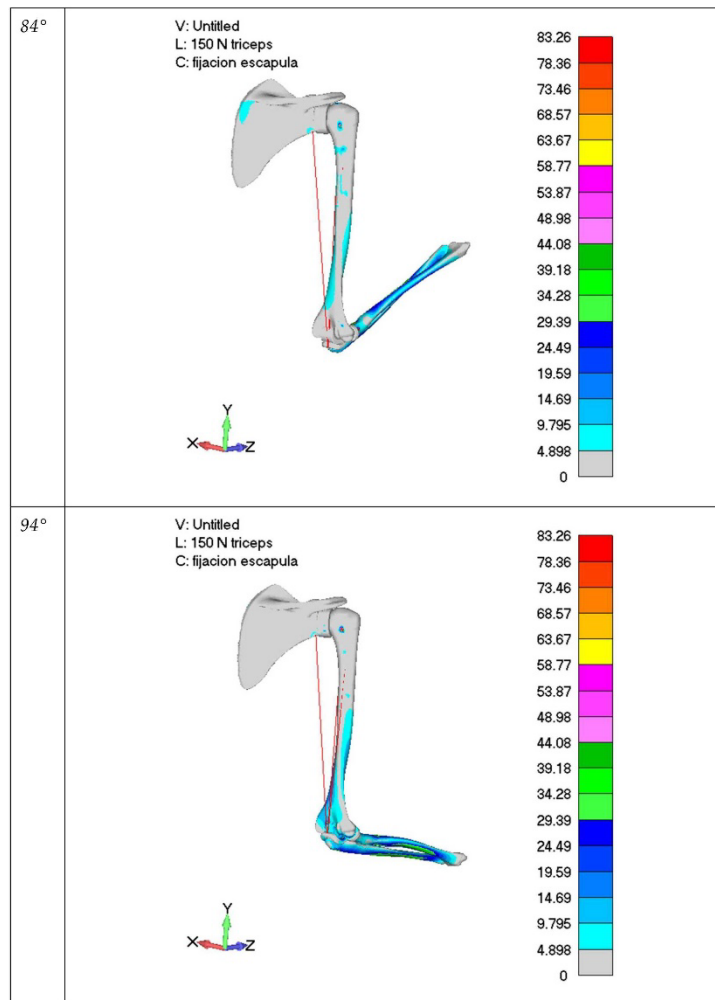
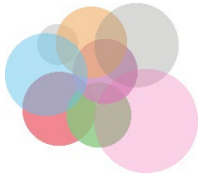


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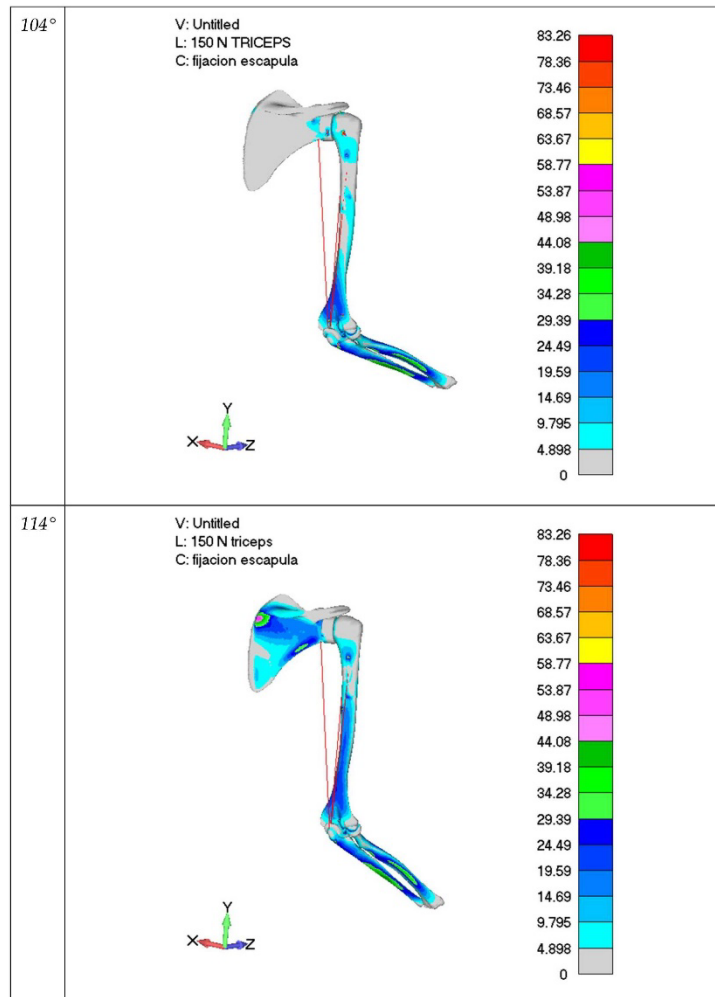


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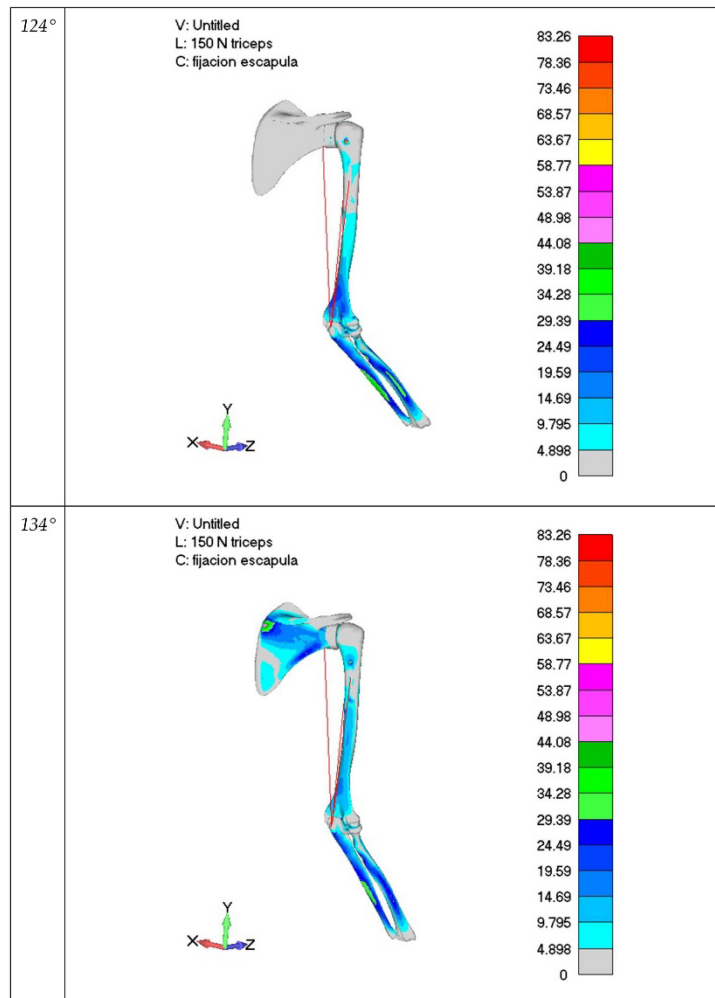
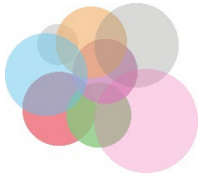


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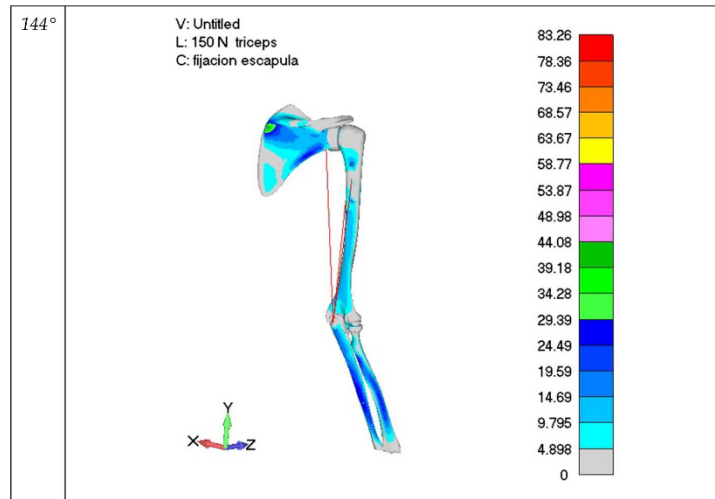
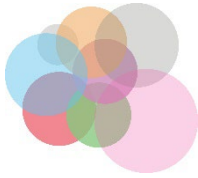
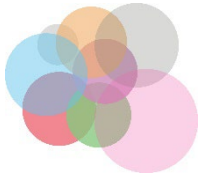


Figure A2. Different Elbow Angle Configuration: Bone Von Mises Stress Distribution Triceps Loading Case (150 N on wrist).

References

1. Strouboulis, T.; Babuška, I.; Whiteman, J.R. *The Finite Element Method and Its Reliability*; Clarendon Press: Oxford, UK, 2001; ISBN 978-0-19-850276-0.
2. Babuška, I.; Whiteman, J.; Strouboulis, T. *Finite Elements: An Introduction to the Method and Error Estimation*; OUP: Oxford, UK, 2010; ISBN 978-0-19-850669-0.
3. Pawelko, P.; Jastrzębski, D.; Parus, A.; Jastrzębska, J. A new measurement system to determine stiffness distribution in machine tool workspace. *Arch. Civ. Mech. Eng.* **2021**, *21*, 49. [\[CrossRef\]](#)
4. Wang, D.; Zhang, S.; Wang, L.; Liu, Y. Developing a Ball Screw Drive System of High-Speed Machine Tool Considering Dynamics. *IEEE Trans. Ind. Electron.* **2021**, *69*, 4966–4976. [\[CrossRef\]](#)
5. Li, Z.; Oger, G.; Le Touzé, D. A partitioned framework for coupling LBM and FEM through an implicit IBM allowing non-conforming time-steps: Application to fluid-structure interaction in biomechanics. *J. Comput. Phys.* **2021**, *449*, 110786. [\[CrossRef\]](#)
6. Della Rosa, N.; Bertozzi, N.; Adani, R. Biomechanics of external fixator of distal radius fracture, a new approach: Mutifix Wrist. *Musculoskelet. Surg.* **2020**, *106*, 89–97. [\[CrossRef\]](#) [\[PubMed\]](#)
7. Zhang, N.-Z.; Xiong, Q.-S.; Yao, J.; Liu, B.-L.; Zhang, M.; Cheng, C.-K. Biomechanical changes at the adjacent segments induced by a lordotic porous interbody fusion cage. *Comput. Biol. Med.* **2022**, *143*, 105320. [\[CrossRef\]](#) [\[PubMed\]](#)
8. Denozière, G.; Ku, D.N. Biomechanical Comparison between Fusion of Two Vertebrae and Implantation of an Artificial Intervertebral Disc. *J. Biomech.* **2006**, *39*, 766–775. [\[CrossRef\]](#) [\[PubMed\]](#)
9. Samani, A.; Bishop, J.; Yaffe, M.J.; Plewes, D.B. Biomechanical 3-D finite element modeling of the human breast using MRI data. *IEEE Trans. Med. Imaging* **2001**, *20*, 271–279. [\[CrossRef\]](#)
10. Jacques, S.; Van Oosterwyck, H.; Muraru, L.; Van Cleynenbreugel, T.; De Smet, E.; Wevers, M.; Naert, I.; Sloten, J.V. Individualised, micro CT-based finite element modelling as a tool for biomechanical analysis related to tissue engineering of bone. *Biomaterials* **2003**, *25*, 1683–1696. [\[CrossRef\]](#)
11. Renner, S.M.; Natarajan, R.N.; Patwardhan, A.G.; Havey, R.M.; Voronov, L.L.; Guo, B.Y.; Andersson, G.B.; An, H.S. Novel model to analyze the effect of a large compressive follower pre-load on range of motions in a lumbar spine. *J. Biomech.* **2007**, *40*, 1326–1332. [\[CrossRef\]](#)
12. Martínez, A.M.R. Modelado y simulación del tejido músculo-esquelético. Validación Experimental con el Músculo Tibial Anterior de Rata. Ph.D. Thesis, Universidad de Zaragoza, Zaragoza, Spain, 2011. Available online: <http://purl.org/dc/dcmitype/Text>. (accessed on 5 November 2021).
13. Weiss, J.A.; Gardiner, J.C.; Ellis, B.J.; Lujan, T.J.; Phatak, N.S. Three-dimensional finite element modeling of ligaments: Technical aspects. *Med. Eng. Phys.* **2005**, *27*, 845–861. [\[CrossRef\]](#)
14. Islan, M.; Carvajal, J.; Pedro, P.S.; D'Amato, R.; Juanes, J.A.; Soriano, E. Linear Approximation of the Behavior of the Rotator Cuff under Fatigue Conditions. Violinist Case Study. In Proceedings of the ACM 5th International Conference on Technological Ecosystems for Enhancing Multiculturality, Cádiz, Spain, 18–20 October 2017; p. 58.



15. Sachenkov, O.A.; Hasanov, R.F.; Andreev, P.S.; Konoplev, Y.G. Numerical Study of Stress-Strain State of Pelvis at the Proximal Femur Rotation Osteotomy. *Russ. J. Biomech.* **2016**, *20*, 220–232. [\[CrossRef\]](#)
16. Martins, J.A.C.; Pato, M.P.M.; Pires, E.B. A finite element model of skeletal muscles. *Virtual Phys. Prototyp.* **2006**, *1*, 159–170. [\[CrossRef\]](#)
17. Tang, C.; Tsui, C.; Stojanovic, B.; Kojic, M. Finite element modelling of skeletal muscles coupled with fatigue. *Int. J. Mech. Sci.* **2007**, *49*, 1179–1191. [\[CrossRef\]](#)
18. Syomin, F.A.; Tsaturyan, A.K. Mechanical model of the left ventricle of the heart approximated by axisymmetric geometry. *Russ. J. Numer. Anal. Math. Model.* **2017**, *32*, 327–337. [\[CrossRef\]](#)
19. Perreault, E.J.; Sandercock, T.G.; Heckman, C.J. Hill Muscle Model Performance during Natural Activation and Electrical Stimulation. In Proceedings of the 23rd Annual International Conference of the IEEE Engineering in Medicine and Biology Society, Istanbul, Turkey, 25–28 October 2001; Volume 2, pp. 1248–1251.
20. Alonso, F.J.; Galán-Marín, G.; Salgado, D.R.; Pàmies Vilà, R.; Font Llagunes, J.M. Cálculo de Esfuerzos Musculares en la Marcha Humana Mediante Optimización Estática-Fisiológica. In Proceedings of the XVIII Congreso Nacional de Ingeniería Mecánica, Ciudad Real, Spain, 3–5 November 2010; pp. 1–9.
21. Holzbaur, K.R.S.; Murray, W.M.; Delp, S.L. A Model of the Upper Extremity for Simulating Musculoskeletal Surgery and Analyzing Neuromuscular Control. *Ann. Biomed. Eng.* **2005**, *33*, 829–840. [\[CrossRef\]](#)
22. Park, W.-I.; Lee, H.-D.; Kim, J. Estimation of isometric joint torque from muscle activation and length in intrinsic hand muscle. In Proceedings of the 2008 International Conference on Control, Automation and Systems, Seoul, Korea, 14–17 October 2008; pp. 2489–2493. [\[CrossRef\]](#)
23. Soechting, J.F.; Flanders, M. Evaluating an Integrated Musculoskeletal Model of the Human Arm. *J. Biomech. Eng.* **1997**, *119*, 93–102. [\[CrossRef\]](#)
24. Zajac, F.E. Muscle and Tendon: Properties, Models, Scaling, and Application to Biomechanics and Motor Control. *Crit. Rev. Biomed. Eng.* **1989**, *17*, 359–411.
25. Lechosa Urquijo, E.; Blaya Haro, F.; D'Amato, R.; Juanes Méndez, J.A. Finite Element Model of an Elbow under Load, Muscle Effort Analysis When Modeled Using 1D Rod Element. In Proceedings of the Eighth International Conference on Technological Ecosystems for Enhancing Multiculturality, Salamanca, Spain, 21–23 October 2020; Association for Computing Machinery: New York, NY, USA, 21 October 2020; pp. 475–482.
26. Teo, E.C.; Zhang, Q.H.; Qiu, T.X. Finite Element Analysis of Head-Neck Kinematics Under Rear-End Impact Conditions. In Proceedings of the 2006 International Conference on Biomedical and Pharmaceutical Engineering, Singapore, 11–14 December 2006; pp. 206–209.
27. Donahue, T.L.H.; Hull, M.L.; Rashid, M.M.; Jacobs, C.R. A Finite Element Model of the Human Knee Joint for the Study of Tibio-Femoral Contact. *J. Biomech. Eng.* **2002**, *124*, 273–280. [\[CrossRef\]](#)
28. Abidin, N.A.Z.; Kadir, M.R.A.; Ramllee, M.H. Three Dimensional Finite Element Modelling and Analysis of Human Knee Joint-Model Verification. *J. Phys. Conf. Ser.* **2019**, *1372*, 012068. [\[CrossRef\]](#)
29. Sachenkov, O.A.; Hasanov, R.; Andreev, P.; Konoplev, Y. Determination of Muscle Effort at the Proximal Femur Rotation Osteotomy. *IOP Conf. Series: Mater. Sci. Eng.* **2016**, *158*, 012079. [\[CrossRef\]](#)
30. Jesal, N. Parekh Using Finite Element Methods to Study Anterior Cruciate Ligament Injuries: Understanding the Role of ACL Modulus and Tibial Surface Geometry on ACL Loading. Ph.D. Thesis, The University of Michigan, Ann Arbor, MI, USA, 2013.
31. CES EduPack Bulletin: January 2017. Available online: <https://www.grantadesign.com/newsletters/ces-edupack-bulletin-ces-edupack-2017-new-products-database-symposia-deadlines-shared-resources-webinars-and-more/> (accessed on 20 June 2022).
32. Bruno, S.; José, M.; Filomena, S.; Vitor, C.; Demétrio, M.; Karolina, B. The Conceptual Design of a Mechatronic System to Handle Bedridden Elderly Individuals. *Sensors* **2016**, *16*, 725. [\[CrossRef\]](#) [\[PubMed\]](#)
33. Arcila Arango, J.C.; Cardona Nieto, D.; Giraldo, J.C. Abordaje Físico-Matemático Del Gesto Articular. Available online: <https://www.efdeportes.com/efd171/abordaje-fisico-matematico-del-gesto-articular.htm> (accessed on 5 November 2021).
34. Loss, J.F.; Candotti, C.T. Comparative Study between Two Elbow Flexion Exercises Using the Estimated Resultant Muscle Force. *Braz. J. Phys. Ther.* **2008**, *12*, 502–510. [\[CrossRef\]](#)
35. Murray, W.M.; Delp, S.L.; Buchanan, T.S. Variation of Muscle Moment Arms with Elbow and Forearm Position. *J. Biomech.* **1995**, *28*, 513–525. [\[CrossRef\]](#)