VALIDATION OF A DYNAMIC MODEL OF THE NECK FOR APPLICATIONS IN ERGONOMICS AND FUNCTIONAL ASSESSMENT

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ABSTRACT:

A dynamic neck model is proposed for functional assessment or ergonomic studies using data from conventional biomechanical tools such as video photogrammetry and force platforms. Head and neck inertial parameters are obtained through regression equations and refined through a calibration process to improve accuracy. Head movement is recorded through video photogrammetry, where marker coordinates are used to calculate finite displacements, linear and angular velocities, and accelerations. An inverse dynamics approach estimates the forces and moments at the C7 vertebral level and the generated muscle power. The model was validated through an experimental study with 30 participants, where its estimates were compared to the measurements obtained from a force platform. The comparison aimed to assess the accuracy and reliability of the model. The results show excellent agreement, with a correlation of 0.976 or higher and a standard error of less than 1% of the head weight

Keywords: Biomechanics, biomechanical model, neck, ergonomics, functional assessment

RESUMEN:

Se presenta un modelo dinámico del cuello para su aplicación en valoración funcional o estudios de ergonomía, a partir de datos obtenidos mediante equipamiento biomecánico convencional: sistema de video fotogrametría y plataformas de fuerzas. El modelo utiliza parámetros inerciales de la cabeza y el cuello obtenidos mediante ecuaciones de regresión que se ajustan posteriormente en un proceso de calibración. El movimiento de la cabeza se registra mediante un sistema de video fotogrametría convencional. A partir de las coordenadas de los marcadores se calculan los desplazamientos finitos, velocidades y aceleraciones lineales y angulares. Las fuerzas externas se miden con una plataforma de fuerzas. Mediante un planteamiento de dinámica inversa se estiman fuerzas y momentos a nivel de C7, así como la potencia desarrollada. El modelo se ha validado mediante un estudio experimental con 30 participantes, comparando las estimaciones del modelo con las medidas en una plataforma de fuerzas. Los resultados muestran una correspondencia excelente entre las estimaciones y las medidas experimentales, con una correlación superior a 0.976 y un error estándar inferior al 1% del peso de la cabeza.

Palabras clave: Biomecánica, modelo biomecánico, cuello, ergonomía, valoración funcional

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1. INTRODUCTION

Biomechanical analysis of the neck has essential applications in functional assessment and risk assessment associated with working postures. The most complex and advanced models are the so-called musculoskeletal models (MSM), which attempt to estimate internal actions at the level of muscles, tendons and intervertebral discs. There is extensive literature on this type of model, reviewed in [1], with different approaches to represent muscle forces, muscle activation mechanisms, or how to solve dynamic indeterminacy [2] [3].

Despite the level of detail of their estimates, the practical usefulness of neck MMEs is limited due to the following problems [4]: (i) practical impossibility of "in vivo" estimation of the anatomical and physiological parameters needed to adapt the model to the individual characteristics of each subject [2]; (ii) difficulties in measuring intervertebral kinematics without using invasive methods, which severely affects the model estimates [5]; iii) the problem of dynamic indeterminacy, which must be solved with optimisation models of dubious validity, or with muscle activation models involving the use of electromyography [2] and, iv) the impossibility of direct validation of the model, as it is not possible to compare the estimates with direct measurements of internal actions [6]. These limitations and their complexity determine that MMEs are mainly used in research applications but not in practice for functional assessment or ergonomic studies.

Hence the practical interest in skeletal or joint models (SM), which allow the estimation of joint forces and moments using an inverse dynamics approach [7]. The EMs are much more robust, can be experimentally validated [8] and do not require detailed information on anatomical aspects or mechanical characteristics of internal structures that are difficult to measure. The two-bar, or double-pivot, is the most commonly used model [9]. They have been used for biomechanical analysis of work tasks [10][11], in the design of dummies for traffic accident simulation [9], in studies of head dynamics during crashes [12] and in the development of robotic devices for rehabilitation [13][14]. However, double-pivot models provide a poor representation of joint kinematics, leading to significant errors in estimating forces and moments [15]. On the other hand, these models neglect inertial forces except for applications in the study of vehicle impacts [11].

A more accurate alternative is non-parametric EMs, where kinematics is incorporated from the direct measurement of the head motion relative to the neck [15]. Using estimates of the inertial parameters of the head-neck system [16], a correct estimation of inertial actions and forces and moments at the C7 level is possible. Although the level of information provided by a non-parametric ME is lower than that of a MME, the estimation of joint forces and moments is sufficient in many applications in the field of ergonomics, where the level of risk is estimated from the joint moments. Moreover, these models can complement kinematic techniques in functional assessment applications by providing dynamic information about the movements used in clinical examinations [4]. Currently, biomechanical neck assessment techniques are limited to kinematic measurements or maximal voluntary effort tests under static conditions. Given that injuries cause slower movements with less isometric effort capacity, it is foreseeable that the estimation of dynamic variables associated with clinical examination tests will provide additional criteria for the functional assessment of patients [17].

In this line, this work presents a non-parametric inverse dynamics EM that provides estimates of the forces and moments exerted at the level of C7 from head movement information obtained by means of conventional equipment in any biomechanics laboratory (video-photogrammetry system and force platform). The model has been validated with direct measurements using a force platform in an experiment with a sample of 30 subjects.

2. MATERIALS AND METHODS

2.1. DYNAMIC MODEL. INERTIAL PARAMETER ESTIMATION

The dynamic model used considers two mobile body segments: the head and the neck. The rest of the body is assumed to remain at rest, as the tests are performed with the subject seated in a chair that fixes the position of the trunk and upper and lower limbs [4].

For the calculation of the forces exerted by the trunk on the head-neck system, at the level of the C7 vertebra, equation (1) is used, corresponding to the free body diagram of the head-neck system:

$$\vec{\mathbf{F}}_1^{\text{iner}} + \vec{\mathbf{F}}_2^{\text{iner}} + \vec{\mathbf{P}}_1 + \vec{\mathbf{P}}_2 + \vec{\mathbf{R}}_{C7} = \vec{\mathbf{0}} \qquad (1)$$

Where $\vec{\mathbf{F}}_1^{\text{iner}}$ is the inertia force associated with the motion of the head, $\vec{\mathbf{F}}_2^{\text{iner}}$ that associated with the neck movement and the vectors $\vec{\mathbf{P}}_1$ y $\vec{\mathbf{P}}_2$ are the weights of the head and neck, respectively. The vector $\vec{\mathbf{R}}_{C7}$ is the force exerted by the trunk on the head-neck system at C7. In equation (1), the forces exerted by the neck on the head or by the head on the neck do not appear because they are equal and of opposite direction.

Similarly, the equation for the moments with respect to a fixed point (C7 or the centre of the laboratory reference frame) can be written as shown in equation (2).

$$\vec{\mathbf{T}}_1^{\text{iner}} + \vec{\mathbf{T}}_2^{\text{iner}} + \vec{\mathbf{T}}_1^{g} + \vec{\mathbf{T}}_2^{g} + \vec{\mathbf{T}}_{C7} = \vec{\mathbf{0}} \qquad (2)$$

Where $\vec{\mathbf{T}}_1^{\text{iner}}$ y $\vec{\mathbf{T}}_2^{\text{iner}}$ are the moments associated with the inertial actions of the head and neck, $\vec{\mathbf{T}}_1^{g}$ y $\vec{\mathbf{T}}_2^{g}$ the moments associated with the weight of the head and neck, and $\vec{\mathbf{T}}_{C7}$ the moment exerted by the trunk on the head-neck system.

To estimate the forces and moments associated with the inertias, the procedure described in [4] and [8] is followed, based on the measurement of the motion of the system using a video photogrammetry system, which provides the position of the head and neck.

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By means of a smoothing process and numerical derivation, linear and angular velocities and accelerations are calculated as described in [18].

For the estimation of the inertial and gravitational actions, it is necessary to know the masses, positions of the centres of mass and moments of inertia of the neck and head. The validity of these parameters conditions the validity of the model estimates [8]. In our model, these parameters are obtained in two approaches. Firstly, we start from regression equations obtained using published data from the review in [4], which provide a first approximate estimate based on the height and weight of each subject. The coefficients of these equations are shown in Table 1. In these equations, height is measured in cm and weight in kg. The reference system used is the anatomical one associated with the head, where the X axis is the anteroposterior, the Y the vertical and the Z the mid-lateral.

Parameter	Equation	Error
Head mass(kg)	-0.0023× Size+0.0415× Weight+1.388	0.34
Neck mass(kg)	0.0068× Size+0.0144× Weight-0.6167	0.22
Head-neck mass(kg)	0.0137× Size+0.0504× Weight-0.2896	0.63
X cdm head	-0.0140× Size+0.0271× Weight+1.3384	0.64
Y cdm head(cm)	0.0088× Size-0.0397× Weight 4.1077	0.68
I_{XX} head(kg cm $)^2$	-0.7049× Size+3.2492× Weight+71.403	35.5
I_{YY} head(kg cm $)^2$	-0.6326× Size+3.1793× Weight+103.84	38.2
I_{ZZ} head(kg cm $)^2$	-0.7890× Size+2.8213× Weight+106.97	28.0
I_{XX} neck(kg cm $)^2$	-0.5530× Size+0.169× Weight+131.395	5.5
I_{YY} neck(kg cm $)^2$	-0.3022× Size+0.8275× Weight-83.1602	9.46
I_{ZZ} neck(kg cm $)^2$	-0.9091× Size-0.1220× Weight+198.070	4.74

Table 1. Regression equations for the initial estimation of the inertial parameters. Length is measured in cm and weight in kg. The error column corresponds to the standard error of measurement of the estimates.

The position of the centre of mass of the neck is calculated directly from the position of the markers located at the process of C7 and the origin of coordinates of the anatomical system of the head, O_{cab} , as follows:

$$\vec{\mathbf{r}}_{\text{cdm cuello}}=0.525\ (\vec{\mathbf{r}}_{\text{Ocab}} + \vec{\mathbf{r}}_{\text{C7}}) \qquad (3)$$

These estimates can be refined by a calibration process in which the model estimates are compared with those measured directly with a force platform. This follows a simplification of the parameter identification process described in [4] in which only the values of the segment masses and the position of the centre of mass are readjusted. The moments of inertia are retained from the regression data, as they have little effect on the estimates and their inclusion detracts from the robustness of the model [8].

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2.2. EXPERIMENTAL VALIDATION

For the experimental validation of the model, flexion-extension movement tests have been carried out in a continuous cyclic movement. Estimates were obtained from the model and compared with measurements obtained using a force platform.

2.2.1. Sample

The experiments involved 30 participants (14 women, 16 men), with a mean age of 35.6 years (standard deviation: 8.9 years). All subjects were healthy, with no neck discomfort or pain. Participants signed an informed consent form in accordance with the protocols approved by the Ethics Committee of the Universitat Politècnica de València (Reference P2_27_09_2017).

2.2.2. Experimental set-up

In each measurement session, each subject sat on a chair with a height-adjustable backrest to fix the position of the trunk, arms and legs by means of shoulder, chest, pelvis, knee and ankle straps. In this way, the only possible movement is that of the neck and head, which ensures that the actions of the trunk on the chair and floor, on the force platform, are constant (Figure 1).

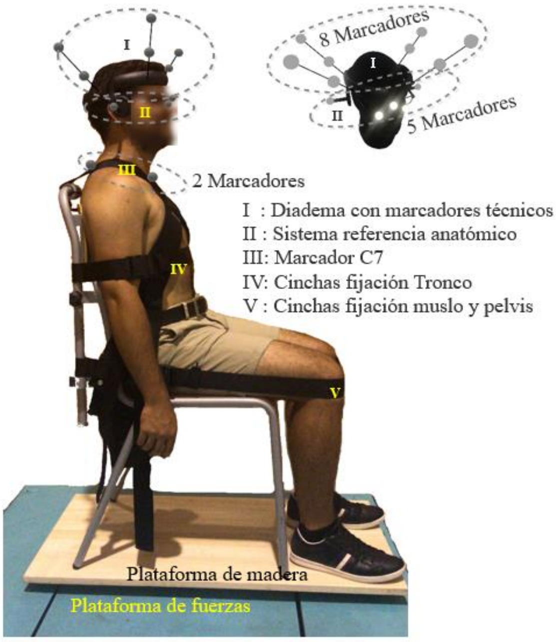

Figure 1. Experimental set-up for recording head movement and ground forces.

The head movement was recorded by means of 8 technical markers, placed on a headband fixed to the subject's head. In addition, adjustable goggles with five additional markers were placed in the reference position to define an anatomical reference system, which is where the inertial parameters were expressed. The anatomical markers correspond to the ear tragus, nasal bone and right and left infra-orbital bones. This system is removed during the tests so as not to hinder movement. Finally, two other control markers were placed at C7 and at the sternum. The first one to define the reference point for moment taking, while the second one to monitor that the trunk was well fixed and did not move during the tests.

The positions of the markers were recorded in real time using a Kinescan/IBV video photogrammetry system, with 10 cameras, at 200 fps. The system was calibrated for each measurement session, allowing very good accuracy to be obtained, with an angle measurement error of 0.08º and a displacement error of less than 0.15 mm [19]. From the positions, velocities and accelerations (linear and angular) were calculated using the calculation algorithms described in [18].

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The forces were measured with a Dinascan/IBV force platform [4]. Since the dimensions of a standard platform do not allow a chair to be placed comfortably and the feet to rest comfortably, a small wooden platform with a base equal to the size of the force platform, but with an enlarged top, was placed on the platform, as shown in Figure 1. The platform recorded the resultant of the forces exerted on the ground, as well as the resultant moment, in its own reference system. At the beginning of each test, the wooden platform and the chair were tared, so that the actions recorded correspond to those exerted by the subject's body (weights and inertial forces of the head and neck). The measurements were taken synchronously with the frames of the video photogrammetry system at a sampling rate of 200 Hz. Prior to testing, the platform was calibrated to establish its accuracy. The errors in the measurement of vertical forces were 0.05 N in the measurement of vertical forces and 0.15 N in the measurement of horizontal forces.

2.2.3. Measurement protocol

Prior to the test, the objective and development of the experiment was explained to the subjects, who signed an informed consent form. Once seated and instrumented, the subject had to perform several flexion-extension movement cycles in a row (seven cycles), reaching the maximum possible extension and flexion, at the desired speed.

After some preliminary tests to become familiar with the movement, the measurement was taken. The measurement began by taking a reference position, corresponding to the neutral posture of the head, looking straight ahead. In this position, the anatomical reference glasses were placed, the position and forces were recorded for a few seconds and then the glasses were removed, and the kinematics and dynamics of the movement were measured.

In a first test, different movements were performed in the three planes, in order to have independent data to adjust the values of the mass of the head and neck and the position of the centres of mass. Cyclic movements were then performed to analyse the dynamics of the movement.

2.2.4. Direct measurement of actions in C7

For each subject, two measures of actions were obtained at the C7 level.

On the one hand, those calculated by the model, using the inertial parameters readjusted in the static tests and the kinematics of the movement measured by the video photogrammetry system. These values of forces and moments are the values estimated by the model.

On the other hand, using only the forces and moments measured by the force platform, as well as the weight of the rest of the body, the actions that the head-neck exerts on the trunk at the level of C7 can also be obtained, using equations similar to (1) and (2):

$$\vec{\mathbf{P}}_3+\vec{\mathbf{F}}_P-\vec{\mathbf{R}'}_{C7}=\vec{\mathbf{0}} \qquad (4)$$

$$\vec{\mathbf{T}}_3^g+\vec{\mathbf{T}}_P-\vec{\mathbf{T}'}_{C7}=\vec{\mathbf{0}} \qquad (5)$$

where $\vec{\mathbf{P}}_3$ is the weight of the rest of the body, $\vec{\mathbf{T}}_3^g$ is the moment associated with that weight, calculated at the level of C7. Both the weight and its momentum are obtained in the measurement of the reference position, also using the inertial parameters adjusted for the head-neck. Note that in equations (4) and (5) there are no inertial actions of the body, since it remains fixed. Finally. $\vec{\mathbf{F}}_P$ is the force that the platform exerts on the subject (through the chair and the feet) and $\vec{\mathbf{T}}_P$ the moment of reaction of the chair-floor on the subject (also calculated in C7).

Equations (4) and (5) provide a direct measure of the actions that the trunk exerts on the head-neck, $\vec{\mathbf{R}'}_{C7}$ for the force, and $\vec{\mathbf{T}'}_{C7}$, for the moment. Comparing these direct measurements with the model estimates, $\vec{\mathbf{R}}_{C7}$ y $\vec{\mathbf{T}}_{C7}$ obtained from equations (1) and (2), the validity of the model can be quantified, as explained in the following section.

2.2.5. Statistical analysis

To quantify the validity of the model, the estimated and measured actions have been compared, obtaining a functional multiple correlation coefficient (MCC) between the two measurements and calculating the standard error of measurement (SEM). The MCC is a relative measure of the agreement between curves representing a function of time, the closer it is to 1, the better. In biomechanical testing, MCC values above 0.95 are considered excellent and very good if they are between 0.85 and 0.95 [20]. The MSE is an absolute measure of the error to be expected, in an instantaneous measurement, when estimating with the model [4] [21]. To illustrate its order of magnitude it will be expressed as a percentage of each participant's head weight.

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These calculations have been performed for each of the 30 records, so that an MCC and a standard error are obtained for each subject. Validity is determined by the central values and the dispersion of these coefficients. Since the distributions are not normal, the median and interquartile range were used as measures of centre and dispersion [4].

3. RESULTS

Figure 2 shows the results corresponding to the averages of the forces exerted by the trunk on the neck-head system, separating the horizontal component of the force (X-axis in the reference position) from the vertical (Y-axis). In both cases they are expressed as a function of the flexion-extension angle, as this representation gives a clearer idea of the variation of the forces as a function of position than the force-time graphs. The forces estimated by the model are shown in black and the measured forces in grey. The forces are normalised by the mass of the head-neck, so the units are g (acceleration of gravity).

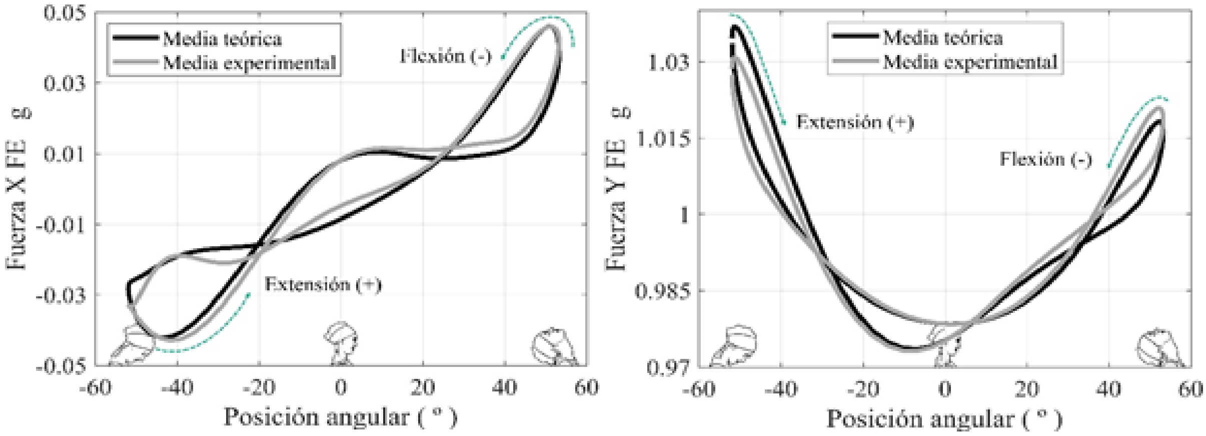

Figure 2. Comparison between measured (black) and model-calculated forces (grey). Forces normalised by head mass. Right: horizontal component (X) ; Left: vertical component (Y) . The abscissa axis shows the flexion-extension angle.

As can be seen, the horizontal component resembles the force-position diagram of a pendulum (apparently changed sign: according to the sign criteria used in Biomechanics, flexion is a negative angle and extension is positive), which is logical for an inertial force in a cyclic movement. However, the relationship is not linear, due to zones with an almost constant force at the centres of the extension and flexion half-cycles. The forces are different in the extension and flexion motion. It is a force pattern associated with the tangential acceleration of the centre of mass of the head.

As for the vertical component, it has a strong gravitational component, with a value equal to the weight of the head, which becomes larger at the ends of the strokes and smaller in the centre. This variability can be explained as the joint effect of the weight and the normal acceleration of the movement of the centre of mass of the head.

In both cases the agreement between estimated and measured values is very good, with slight differences at the extremes of the runs.

Figure 3 shows the comparison curve for the bending moment, also normalised by the head weight (m g units). In this case, the differences between the outward and return movement are much smaller than for the forces, and the moment-angle relationship is more linear, except at the end of the extension. The agreement between measurements and model estimates are also quite good.

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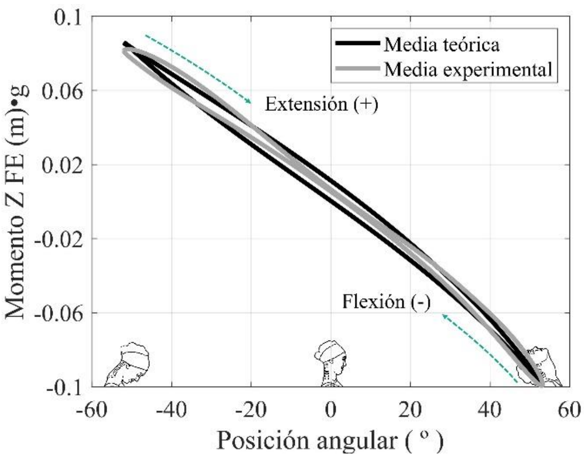

Comparison between the measured (grey) and model-estimated (black) extension moment at the neck at C7. Moment normalised by head mass.

Variable	$\widetilde{\text{CCM}}$ (IQR)	$\widetilde{\text{EEM}}$ (IQR)
Force X (% head weight)	0.983 (0.027)	0.5 (0.2)
Y-force (% head weight)	0.976 (0.036)	0.4 (0.2)
Z-moment (% head weight× m)	0.996 (0.005)	0.7 (0.3)
IQR : Interquartile range; $\widetilde{\text{CCM}}$Median of the multiple correlation coefficient; $\widetilde{\text{EEM}}$Median of the standard error of the measurement; : Median of the standard error of the measurement; : Median of the standard error of the measurement		

Table 2. Descriptive of relative and absolute indicators of agreement between the theoretical model and experimental measurements of neck force and moment.

Table 2 shows the MCC and MSE values (medians and interquartile ranges). As can be seen, the relative agreement indicator is very high and with very little dispersion between subjects. For all the variables measured, the agreement between the model estimates and the direct measures is excellent (median above 0.97). The standard error of measurement is less than 0.5% of the head weight, i.e., for a standard head of about 5 kg mass, it would be an error of the order of 0.25 N. The agreement is even better in the case of moment estimation, where the median MCC is 0.996.

4. CONCLUSIONS

The estimation of forces and moments in the neck can be useful to describe the functional status of a patient or to assess postural injury risks in ergonomic studies. However, applications of dynamic neck models in clinical practice or ergonomic studies are scarce due to the limitations of the available models. Thus, complex musculoskeletal models are costly and difficult to apply in a medical practice or in an ergonomic evaluation, they require information that is difficult to customise and cannot be validated experimentally. This last limitation is crucial in medical applications, where it is not acceptable to make clinical decisions based on predictions provided by a model of unknown validity. On the other hand, simple bar charts do not provide a good representation of movement, so their usefulness is relegated to quasi-static situations.

We therefore propose the use of non-parametric skeletal models that, using real head movement and adjusted inertial parameters, allow the estimation of global actions at the joint level. These models are simple, require only standard instrumentation available in any biomechanics laboratory (video-photogrammetry and force platform) and can be validated, so that the uncertainty associated with their predictions is known.

This paper presents a skeletal model using head motion kinematics measured directly with a video-photogrammetry system and inertial parameters estimated in a two-step process. A first approximation is obtained from regression equations, obtained from published data

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collection [4]. These initial values can be adapted more precisely with an optimisation process based on calibration measurements with the force platform, adjusting the masses and positions of the centres of masses.

The model uses kinematic analysis algorithms and accurate velocity and acceleration calculations validated in previous work [15][19][21], which is critical for good estimation of inertial actions.

The model has been validated with a sample of 30 healthy subjects. An experiment has been designed in which it is possible to measure force and moment at C7 using a force platform, comparing the model estimates with direct measurements.

The experimental set-up is similar to that proposed in [8], but in that work a specific instrumented chair was used, whereas here standard equipment (video-photogrammetry and force platform) has been used, with slight low-cost adaptations.

The agreement between estimates and direct measurements is excellent, with MCC values in the order of 0.98 for forces, and above 0.995 for moment. The errors in force estimation are of the order of 0.5% of the head weight. These results are better than those published by [8] where MCC values of around 0.900 were obtained. Probably, the difference lies in the fact that in our work a more accurate kinematic technique has been used. On the other hand, a cyclic motion has been used here, which undoubtedly contributes to obtain more robust measurements than those corresponding to individual instantaneous values [21]. In short, it is shown that with appropriate kinematic analysis models and by adjusting the inertial parameters, very good estimates of the forces and moments at the joint level can be obtained.

Once validated, the model is being used in functional assessment tests of patients with non-specific neck pain, so that in addition to the kinematic variables associated with the ranges of movement and speed, common in clinical practice, dynamic information will be available, providing a more detailed picture of the biomechanical alterations associated with neck pathologies. To this end, a base of dynamic patterns of normality associated with the usual clinical tests in clinical assessment is being configured, with a representative sample of healthy subjects. This sample will serve as a reference to analyse the differences with another sample of pathological subjects. By comparing these responses using classification techniques, it will be possible to establish objective and quantitative criteria for assessing functional status based on dynamic variables, in the same way as has been done so far with kinematic tests [22]. It is important to note that the incorporation of these types of variables into functional assessment does not require any equipment different from the standard equipment of any biomechanical assessment laboratory.

It is also possible to use it to analyse the physical load on the neck for postural risk assessment in ergonomic studies. In this case, the movements are slow, so it would be possible to use simpler equipment, such as video analysis, and use estimates of masses and centres of gravity, which would allow measurement of variables of interest such as angular positions and joint moments, which are those that affect the levels of static muscle strain associated with injuries [10].

Finally, other applications of a more industrial nature should be highlighted, such as contributing to the development of more realistic human models for CAD, with the incorporation of real movements and customisable inertial parameters. In this sense, anthropometric tables with kinematic and inertial characteristics of the head could be made from the calibrations obtained through the two-step process proposed in this work. This is a simpler and more generalisable alternative to studies based on measurements on cadavers or geometric scaling [23].

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