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Methodological analysis of finite helical axis behavior in cervical kinematics

Corrado Cescon [a,*], Erik Cattrysse [b], Marco Barbero [a]

[a] University of Applied Sciences and Arts of Southern Switzerland, Department of Health Sciences, Manno, Switzerland
[b] Arthrokinematics Research Group, Faculty of Physical Education and Physiotherapy, Department of Experimental Anatomy, Vrije Universiteit Brussel, Belgium

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A B S T R A C T

Although a far more stable approach compared to the six degrees of freedom analysis, the finite helical axis (FHA) struggles with interpretational difficulties among health professionals. The analysis of the 3D-motion axis has been used in clinical studies, but mostly limited to qualitative analysis. The aim of this study is to introduce a novel approach for the quantification of the FHA behavior and to investigate the effect of noise and angle intervals on the estimation of FHA parameters. A simulation of body movement has been performed introducing Gaussian noise on position and orientation of a virtual sensor showing linear relation between the simulated noise and the error in the corresponding parameter.

FHA axis behavior was determined by calculating the intersection points of the FHA with a number of planes perpendicular to the FHA using the Convex Hull (CH) technique. The angle between the FHA and each of the IHA was also computed and its distribution was also analyzed.

Input noise has an inversely proportional relationship with the angle steps of FHA estimation. The proposed FHA quantification approach can be useful to provide new approaches to researchers and to improve insight for the clinician in order to better understand joint kinematics.

1. Introduction

The complexity of three-dimensional joint motion analysis has resulted in an extensive amount of methodologies for the analysis of the kinematics.

At each moment in time, a continuously moving rigid body may be viewed as having a translation velocity and a rotation velocity about a directed line in space (see Fig. 1a). The position of this instantaneous helical axis (IHA) will generally vary during the movement, and the movement is completely known once the translation and rotation velocities and the position of the helical axis are known over time.

The helical axis approach defines a movement as a rotation angle, $\mathbf{\Theta}$, around the axis, described by a direction vector, $\mathbf{n}$, the point $\mathbf{c}$ on the axis closer to the origin, and the translation $\mathbf{t}$ along the axis (Söderkvist et al., 1993; Spoor and Veldpaus, 1980) (see Appendix A for details).

The most common method used to describe joint motion is the use of a six degrees of freedom (6DoF) approach, which consists in the decomposition of the movements into three translation along the corresponding Cartesian axis, and three rotation angles around them. The coordinate system generally used for the study of the spine spans the frontal, sagittal and transverse planes (Kettler et al., 2004). The three Euler angles (angles describing the orientation of a rigid body) can therefore be called lateral bending, flexion–extension and axial rotation angles (Wu et al., 2002) see (Fig. 1b–d).

In the clinical field, computation of movement axis is considered to be a determinant parameter for analyzing the quantity of motion (Dugailly et al., 2010). Some authors have reported aberrant location of instantaneous axis in the sagittal plane for patients with cervical complaints, and alterations of axis location and orientation were observed in whiplash patients (Grip et al., 2008, Woltring et al., 1994). Thus, the relationship between neck pain and irregularities of kinematic patterns could characterize movement impairments in patients.

Although the description of spinal motion by the use of Euler angles is readily understood, the respective predefined axes mostly do not reflect the actual rotary axes of the joint. Furthermore, variations in the localization of the axes reduce the reproducibility of results and may lead to an over- or underestimation of angle values, called "crosstalk effect" (Chao, 1980). For this reason, the Euler angles often require a predefined anatomical coordinate

* Corresponding author. Tel.: +41 (0)76 6930775.
E-mail address: corrado.cescon@supsi.ch (C. Cescon).

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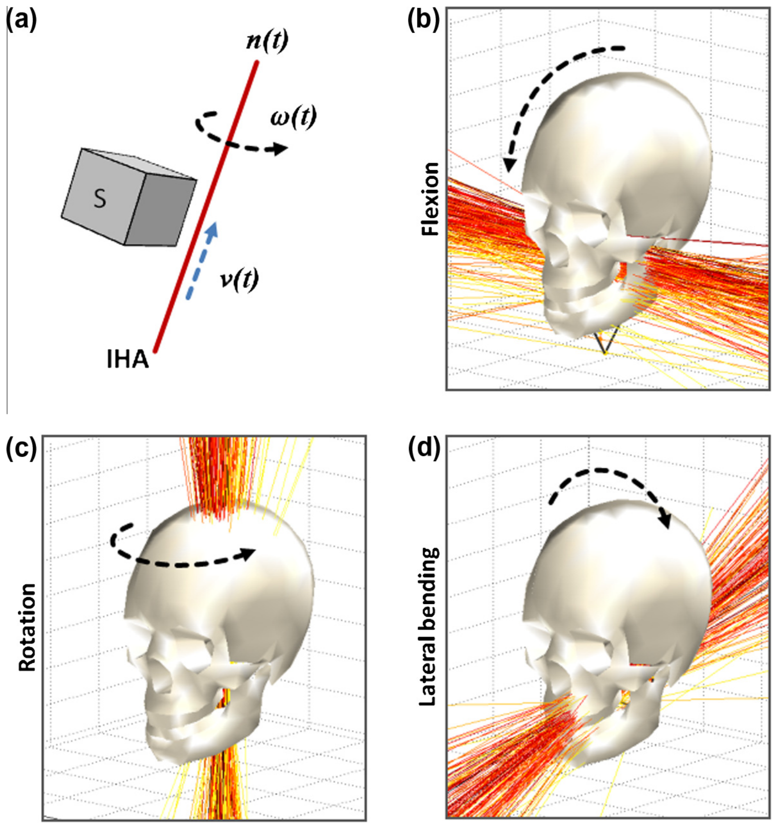

Fig. 1. (a) Representation of the instantaneous helical axis (IHA) of an object with an instantaneous angular velocity $\boldsymbol{\omega}(\boldsymbol{t})$ and linear velocity $\boldsymbol{v}(\boldsymbol{t})$. The inclination of the IHA is represented by vector $\boldsymbol{n}(\boldsymbol{t})$ (b–d) representation of the movements of the head around the Euler axis: flexion extension, rotation and lateral bending respectively. The IHA superimposed are the results of the analysis of a representative subject.

system according to the joint they describe and the three angles are sequence dependent. This problem is most evident in the case of large, coupled vertebral motions.

On the other hand, most studies exploring the IHA tend to produce good qualitative results, but quantitative results are often lacking (Blankevoort et al., 1990; Baillargeon and Anderst, 2013). Graphical representations of the IHA have been used in many different studies and add to an easier interpretation. The location of the knee axis of motion has been extensively discussed from a clinical and orthopedic point of view (Asano et al., 2005; Mannel et al., 2004a,b; Marin et al., 2003; Sheehan, 2007; Van Sint Jan et al., 2002; Wismans et al., 1980; Woltring et al., 1985). Spine motion analysis using the IHA has shown to provide a useful method for the analysis of complex segmental and regional 3D-motions (Cripton et al., 2001; Dugailly et al., 2010; Kettler et al., 2004; Milne, 1993).

The dispersion of the 3D-motion axis has been used to express the stability of the motion in cervical spine and in knee and ankle joint analysis (Graf and Stefanyshyn, 2012; Grip and Häger, 2013; Osterbauer et al., 1992; Panjabi, 1979; Woltring et al., 1994). The position and orientation in space of the IHA can be defined by a number of parameters, and the dispersion of each of the parameters has been recently used to compare different movements (Grip and Häger, 2013). Most of the studies investigated the intersections of IHA with predefined planes or their inclination with respect to anatomical landmarks (Asano et al., 2005; Baillargeon and Anderst, 2013; Cripton et al., 2001). To date there is no attempt to describe the localization of a group of IHA in space during a movement without the need of a 3D reconstruction of the bones of the joint.

Since the helical axis is a differential quantity (measuring infinitely small change in a variable), most users have approximated the IHA with the so called finite helical axis (FHA) which is estimated from a single finite displacement (Blankevoort et al., 1990) (Fig. 2a).

The main drawback in the use of FHA is the sensitivity to noise, since the errors in FHA estimation are inversely proportional to the magnitude of displacement. On the other hand, small increments are necessary to approximate finite displacements with continuous movements. A simplified theoretical analysis of error propagation was proposed by Spoor and collaborators (1980) on a subset of data obtained with stereophotogrammetry, but no indications were provided about the intervals for optimal FHA estimation.

The aim of this study is to evaluate the effect of measurement noise and intervals between frames on FHA parameters and to propose two parameters for the analysis of FHA which are not dependent on the anatomical landmarks and on the type of movement, and that could have direct application in the analysis of cervical spine kinematics.

2. Methods

This study is divided in three parts:

(a) Evaluation of the effect of noise and angle intervals on the estimation of FHA parameters.
(b) Quantification of FHA behavior.
(c) Application of FHA parameters to cervical kinematics.

For the analysis, an orthogonal dextral coordinate system was used with anterior, superior and right being positive (**x**, **y** and **z**-directions, respectively), as recommended by the International Society of Biomechanics (Wu et al., 2002).

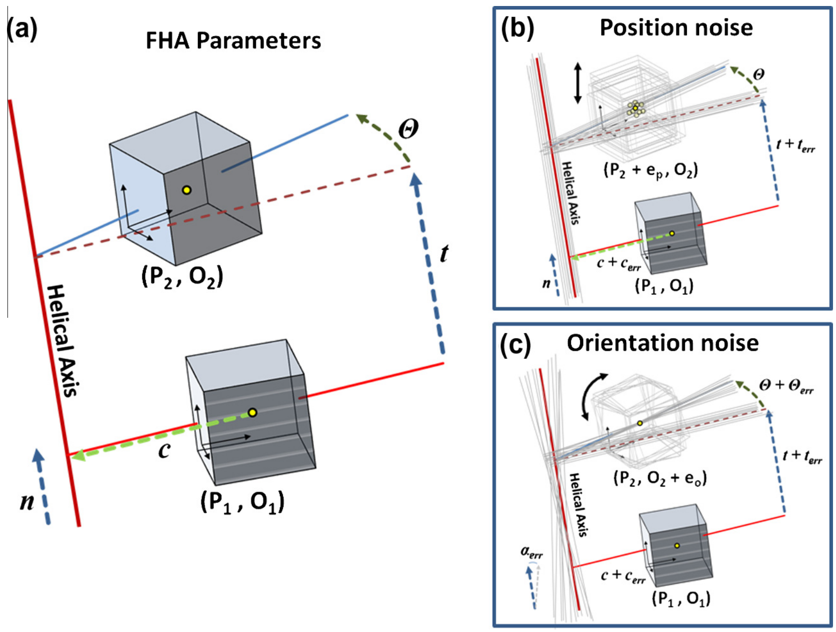

Fig. 2. (a) Representation of the finite helical axis (FHA) of an object changing position and orientation in space from $\mathbf{P_1}$,$\mathbf{O_1}$ to $\mathbf{P_2}$,$\mathbf{O_2}$. The axis is parallel to the unit vector $\boldsymbol{n}$ and has a distance $\boldsymbol{c}$ from the object. The angle of rotation is θ and the translation along the axis is $\boldsymbol{t}$. Graphical representation of the effect of position noise (b) and orientation (c) noise on the FHA parameters.

(a) Evaluation of the effect of noise and angle intervals on the estimation of FHA parameters.

A mathematical model to evaluate the error estimation of the helical axis was used (Matlab® the MathWorks Inc., Natick MA, USA). The model simulated a head rotation of 90 degrees ($\mathbf{O_1}$,$\mathbf{P_1}$ to $\mathbf{O_2}$,$\mathbf{P_2}$, see Appendix A) around an axis located 150 mm from the sensor without translation along it ($\mathbf{\Theta} = 90°$, $|\mathbf{c}| = 150$ mm, and $\mathbf{t} = 0$). The movement was divided in portions of different width, in order to evaluate separately the effect of noise in the output of position and orientation of the sensor.

The simulations were repeated 1000 times introducing a white Gaussian noise in the final position and orientation of the object $\mathbf{O_2}$,$\mathbf{P_2}$. The position error $\mathbf{e_p}$ was a vector of three elements in space (x,y,z) whose components were distributed with a standard deviation ranging from 0.01 mm to 8 mm RMS (in nine steps equally spaced in the logarithmic scale), while the orientation angle error $\mathbf{e_o}$ was a vector of three elements expressed in degrees (pitch, roll, jaw) normally distributed with a standard deviation ranging from 0.001 to 4 degrees RMS (in nine steps equally spaced in logarithmic scale). The resulting position and orientation of the sensor were: $\mathbf{O_2} + \mathbf{e_o}$, $\mathbf{P_2} + \mathbf{e_p}$.

The variables of interest were the errors between the theoretical (noise free) and noisy FHA and can be listed as follows: angle error of the axis vector $\mathbf{n}$, position error of the point $\mathbf{c}$ and translation error of $\mathbf{t}$ (see Fig. 2b and c).

The combined effect of noise and angle intervals for the computation of the FHA was evaluated on a simulated head rotation (from neutral position to 90 degrees rotation to the left). The simulated signals were sampled at 120 Hz and the movement was performed at a constant speed of 1 rad/s. The original data were under-sampled in order to have equally spaced angle steps. The same three FHA parameters were analyzed for each angle interval (angle error of the axis vector $\mathbf{n}$, position error of the point $\mathbf{c}$ and translation error of $\mathbf{t}$).The movement was divided in equal angles with steps of 1, 2, 3, 5, 10, and 20 degrees. Three noise levels were introduced for angle and position (0.05, 0.1, and 0.2 mm for position and 0.05, 0.1 and 0.5 degrees for angle respectively). The simulations were repeated 1000 times for each noise level and each angle step.

(b) Quantification of FHA behavior

Two parameters are proposed in order to describe joint kinematics in different movements the minimum Convex Hull (CH) area of the FHA, and the distribution of angles of the FHA. The two parameters can be considered as complementary expressions of the behavior of the FHA. The steps to compute the two parameters are described in the following paragraphs.

For each joint movement, the neutral position and the extreme position (flexion, extension, axial rotation, lateral bending) were identified and the FHA between the two of them was identified (FHA_0). A number of planes perpendicular to FHA_0 were defined and equally spaced along it. Each movement was divided in two phases (neutral to extreme and extreme to neutral) and each phase was equally divided in steps at a defined angle. For each of the movement steps the FHA_i was identified and the intersections of each FHA_i with each of the defined planes, were identified (see Fig. 3).

For each of the planes perpendicular to the FHA_0 the intersection points of the FHA_i were analyzed with the CH technique (see Fig. 3) and the minimum area was identified.

The angle between the FHA_0 and each of the FHA_i was also computed and the mean value of the distribution of such angles was used as the second parameter to describe joint kinematics.

(c) Application of FHA estimation to cervical kinematics.

2.1. Subjects

A sample of convenience of 10 healthy subjects was studied. Subjects were volunteers from the students and staff of the University of Applied Sciences and Arts of Southern Switzerland, Manno (CH) and comprised five males and five females, ranging in age from 21.5 to 28.9 years (24.4 ± 1.8 years). Subjects were

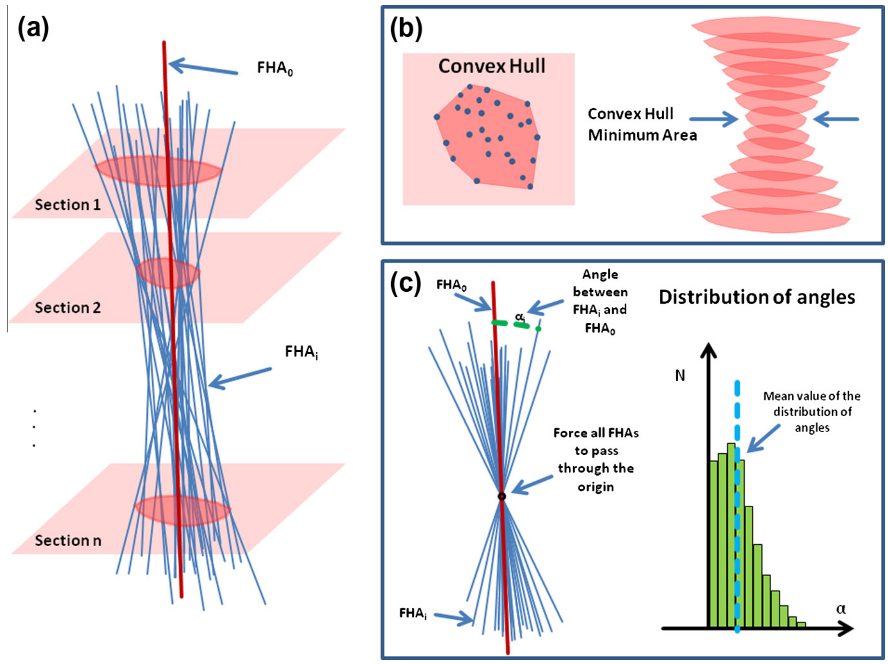

Fig. 3. (a) Representation of the instantaneous helical axis during head rotation. FHA$_0$ is the axis between neutral position and extreme position. Sections 1, 2, ... *n* represent planes perpendicular to the mean axis and equally spaced (20 mm between adjacent planes). For each plane, the convex hull of the intersection of the helical axes (FHA$_i$) was computed. (b) graphical representation of convex hull for a plane and identification of the convex hull with minimum area. (c) Distribution of angles between each FHA$_i$ and FHA$_0$.

not considered if they had a history of headache or neck surgery or had received treatment for neck or shoulder conditions within the past three months. The study was approved from the Ethical Committee of Southern Switzerland and subjects signed informed consent forms.

2.2. Movement registration

Cervical spine movements were detected with the VRRS acquisition system (Khymeia, Padova, Italy), a non-invasive electromagnetic device which integrates motion capture sensors from Polhemus-G4, which tracks the position and orientation of sensors relative to a source in three dimensions. The system has been used and shown to be accurate to within ±0.2° (Koerhuis et al., 2003; Pearcy and Hindle, 1991). In our test, the sensor was fixed to an adjustable semi-rigid plastic headband similar to that described in a previous study (Amiri et al., 2003). The headband was placed around the subjects' forehead so that the sensor was aligned with the bridge of the nose (see Fig. 4). The wires were secured with tape to prevent traction on the sensors. The electromagnetic source was positioned on a box over a wooden table at a height of 120 cm from the floor in order to be at the level of the head of the sitting patient (Cattrysse et al., 2012).

The sensor was attached to a transmitter (hub) which had a wireless connection to a portable PC, and continually recorded the position of the sensors relative to the source at 120 Hz during each test sequence. A custom made software was used to format and store the data for 3D analysis of the neck movements. Measurements were taken from a neutral starting position (reference point of origin on the VRRS system). The angle precision and accuracy of the sensors were estimated by a simulation where the sensors where fixed to a rigid body that was positioned in front of the electromagnetic source (Grip et al., 2007; Ohberg, 2008).

The position and Euler angle data were filtered with an low-pass Butterworth filter (2nd order) using a cut-off frequency at 6 Hz (Grip et al., 2007). The data were dual passed with the filter in both direction (anticausal filtering) in order to avoid group delay.

2.3. Procedure

Subjects received full information of the measurement protocol and tasks required, before the beginning of the experiment. They were asked to sit on a wooden chair in front of the table with the electromagnetic source and then the sensors were attached. The subjects performed each movement two times before the beginning of the recording, which served as a familiarization and warm-up session. For the measurement session, subjects were instructed to sit tall on the edge of the chair with arms on their thighs and to look straight ahead. This position was marked as the zero reference starting point on the VRRS system. The subjects were asked to perform three series of movements of the head at a constant speed. Each series consisted of five consecutive pair of opposite planar movements (flexion–extension, axial rotation, lateral bending). The speed of movements was standardized asking the subjects to perform the movements following the audio signal of metronome set at 40 bpm.

2.4. Statistics

The statistical package SPSS for Windows was used for all statistics. In all statistical tests significance level was set at $\alpha = 0.05$. For all variables, each subject's score was computed as an average of the five repetitions. A non-parametric analysis of variance (Kruskal–Wallis, test) with noise, and angle steps as independent factors was performed (see Figs. 4–6).

3. Results

(a) Evaluation of the effect of noise and angle intervals on the estimation of FHA parameters.

Three FHA error parameters were evaluated (see Fig. 5): angle error between theoretical helical axis and estimated helical axis

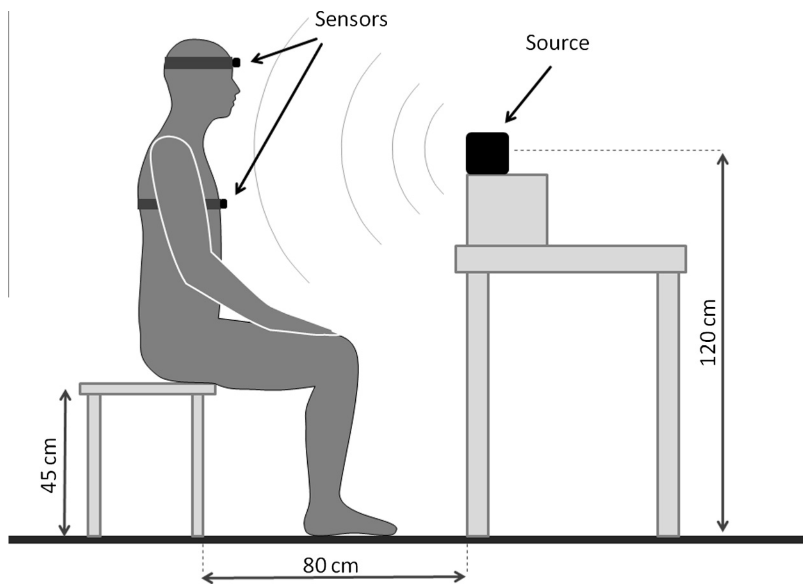

Fig. 4. Experimental set-up used for the measurements of kinematic data during head flexion extension, rotation and lateral bending.

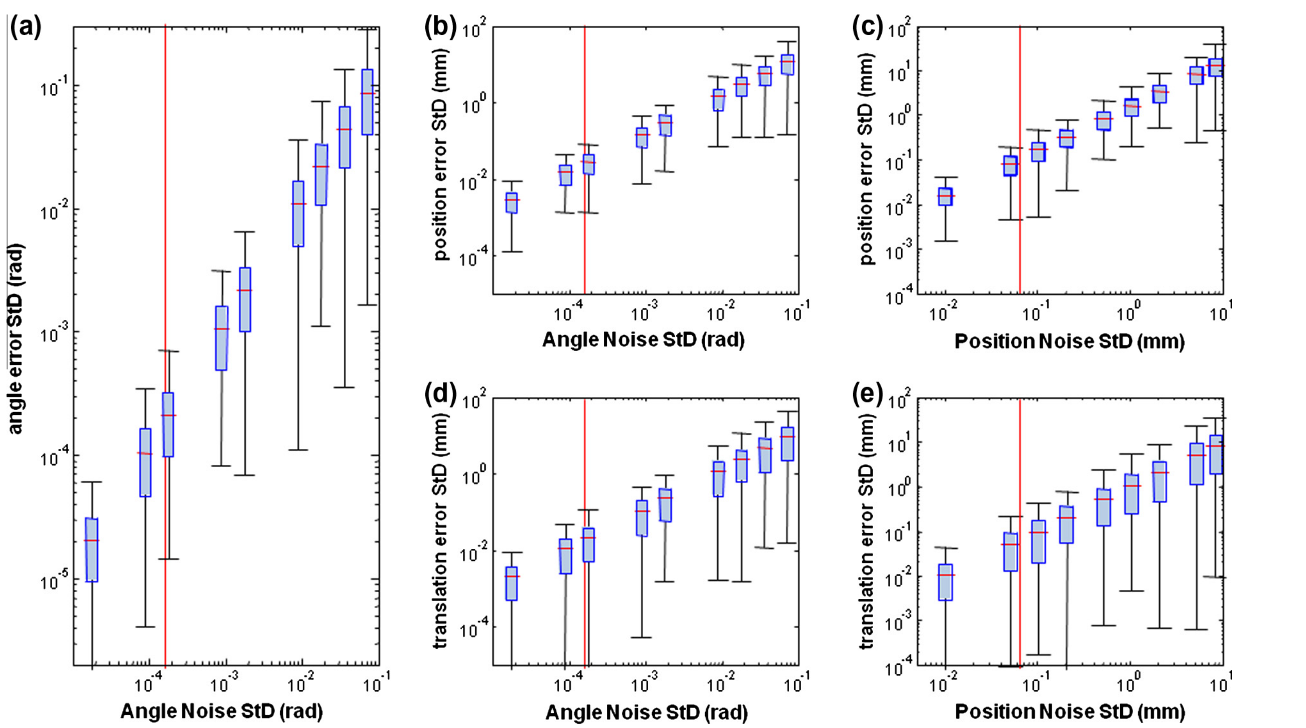

Fig. 5. Effect of additive noise in the orientation (a, b, and d) and position (c and e) of the sensors on the helical axis parameters. The vertical line represent the nominal noise described in the data-sheet of the VRRS sensors. Three parameters were evaluated: angle error between theoretical helical axis and estimated helical axis with added noise (a), the error between theoretical and estimated position of the helical axis (b and d) and the error in translation along the axis (c and e).

with added noise (angle error) (a), the distance between theoretical helical axis and estimated position with added noise (position error) (b, c) and the error in translation along the axis (translation error) (d, e).

A significant dependency was observed (Kruskal–Wallis test, $p < 0.001$), with linear relation between the simulated noise and the error in the estimation of the FHA parameters with the following relationships: (a) 1 degree noise → 4 degrees error; (b) 1 degree noise → 2 mm position error; (c) 1 mm noise → 1 mm position error; (d) 1 degree noise → 2 mm translation error; (e) 1 mm noise → 1 translation mm error.

The analysis showed an inversely proportional relationship between angle steps and error in the angle and in the position of the helical axis (Kruskal–Wallis test, $p < 0.001$) (see Fig. 6). The translation along the helical axis was not dependent from the angle steps (Kruskal–Wallis test, $p = 0.68$). The error in the three parameters increased proportionally with the noise level (Kruskal–Wallis test, $p < 0.05$).

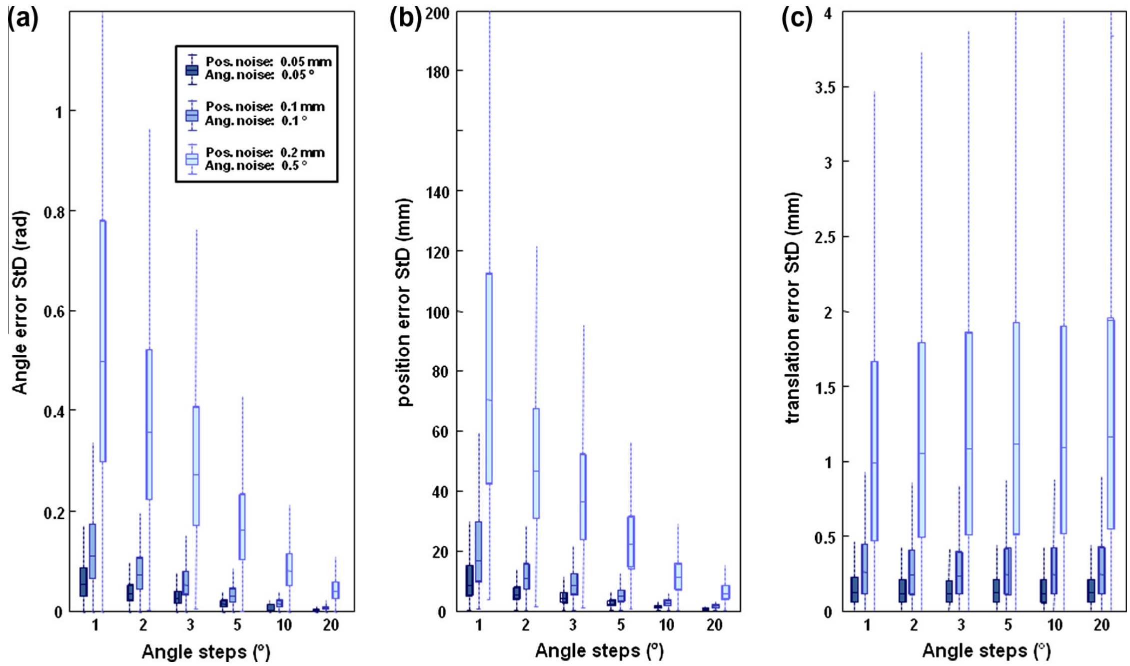

Fig. 6. Effect of angle steps and noise on the single helical axis parameters. (a) Angle between theoretical and noisy estimation of FHA, (b) distance between the position of the axis, and (c) translation error along the axis. The box-whiskers plots represent median, inter-quartile distance and range. Darker box indicate lower added noise.

(b,c) Quantification of FHA behavior and application of FHA estimation to cervical kinematics.

Fig. 7 shows the effect of angle intervals on the estimation of global parameters extracted from groups of helical axis: convex hull area (CH area) and mean angle between each axis and the main axis (mean angle). Both the convex hull area and the mean angle decreased with increasing angle steps (Kruskal–Wallis test, $p < 0.001$, and $p < 0.05$ respectively for each of the three movements).

4. Discussion

The simulations of FHA estimation with added noise in position and orientation allow to evaluate the effect of the accuracy of the sensors on the analysis of the cervical kinematics. The data shown

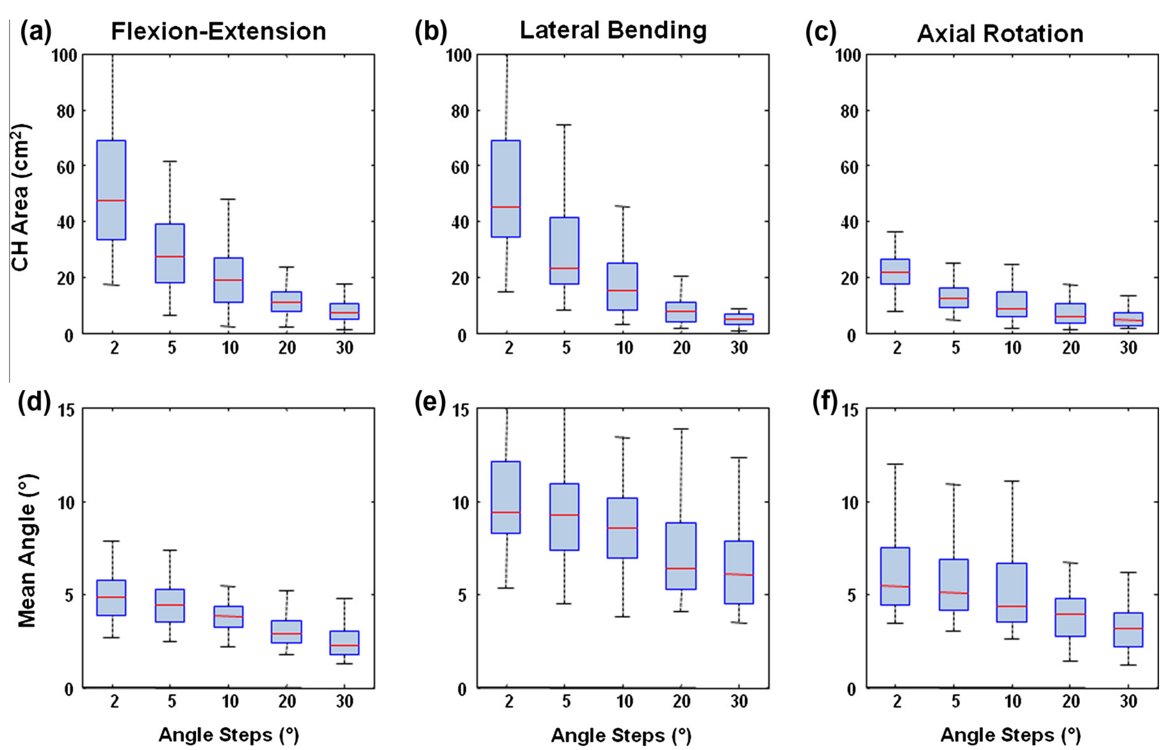

Fig. 7. Effect of angle intervals on the estimation of CH area (a–c) and mean angles (d–f) extracted from groups of helical axis during three head movements: flexion–extension (a and d), lateral bending (b and e) and rotation (c and f). The box-whiskers plots represent median, interquartile distance and range.

in the result section can be useful when using different detection system, because some technology can be more accurate in orientation (inertial sensors, electromagnetic sensors) an other systems can have higher resolution in position (optical sensors). The simulation allow to compare which of the two variables (position and orientation of the sensor) have more influence on the FHA parameters. However, from the reference accuracy levels described in the data-sheets of the VRRS sensors we can observe that in our study the position error results in a larger error in the FHA parameter estimation with respect to angle error.

The relationship between the CH and angle steps showed a similar behavior of CH during the three head movements for angles greater than 20 degrees, with mean values lower than 10 cm^2 and small variance between subjects, while large variability was observed for angle steps equal or lower than 10 degrees. A possible explanation is that the variability of the FHA location in space is reduced by the use of large angle steps as if the data were smoothed by a low pass filter, masking the differences between subjects. On the other hand, this phenomenon was not observed in the mean angle, which showed lower influence to angle steps.

The angle step that seem to provide a good compromise between movement analysis resolution and error in the FHA parameters estimation seems to be 10 degrees, in agreement with the results of a recent study on knee joint (Westphal et al., 2013). The selection of the optimal angle step is important when applying the use of FHA in clinical practice (for example for the analysis of cervical or knee kinematics) because a clinician could be interested in small variation of the axis behavior in a specific part of the motion or simply could be interested in observing the general behavior of a joint before and after a surgical intervention or conservative treatment. The errors in the estimation of FHA strongly depend on the choice of angles, thus the noise level should always be considered in the interpretation of the results.

The translation along the FHA was not considered in the second part of the study because it is less relevant in head kinematics. However, it could be important in intra-articular joint kinematics and could be applied in the design and development of prosthesis.

The time related parameters such as angular velocity, acceleration, jerk index were not considered in the present study because they have been already exhaustively analyzed in the scientific literature. However, the FHA approach allows to evaluate instantaneous angular velocity, acceleration and jerk index on the planes perpendicular to the FHA without limiting the variables to specific planes of rotation.

The analysis of the FHA behavior has an important potential for movement sciences and rehabilitation. For example, the proposed FHA parameters can be applied to investigate indirectly the spinal stabilization system during active movements of cervical spine. CH and mean angle can provide quantitative results able to describe active motion beyond his quantity. Indeed, the FHA analysis can detect variations in position or orientation of an anatomical segment during his motions (i.e. across or through the range of motion).

If FHA behavior will be confirmed to discriminate healthy and pathological subjects, CH and mean angle may be introduced as outcome measures in rehabilitative program for conditions characterized by movement impairments (i.e. musculoskeletal and neurological conditions). The same approach can also be suggested to test and improve athlete performance during specific tasks. Moreover, a proper real time graphical representation of the FHA can be used as biofeedback for both healthy and pathological subjects. Further research in large numbers of subjects is needed to determine the consistency of the FHA behavior and to explore their relevance for clinical practice.

4.1. Limitations of the study

The sensors used in the study are electromagnetic, thus their output depends on the shape of the electromagnetic field generated by the antenna. The electromagnetic field is sensitive to ferromagnetic elements in the room around the transmitter. We noticed that during pure translation of the sensor (of some tenth of centimeters) in any direction there could be an error in the orientation matrix like an apparent rotation due to magnetic field distortions. The possible causes for field distortion are usually neon lights, ferromagnetic fixation systems, electromagnetic fields from mobile phones etc. This phenomenon could be avoided using more accurate sensors or sensors based on different technologies.

5. Conclusions

The effect of noise in position and orientation was described in simulated and experimental signals. A novel quantification approach was introduced and analyzed for the analysis of joint kinematics with the use of the convex hull technique and the FHA angle distribution as specific parameters.

Conflict of interest

The authors declare that they have no conflict of interest.

Acknowledgements

The authors are thankful to Khymeia (Padova, Italy) for the support in software development for data acquisition.

Appendix A

A.1. Algorithm

The method that we used for the estimation of the helical axis was first studied by the Italian scientist Guido Mozzi exactly 250 years ago (Mozzi, 1763). Mozzi described how the movement of a rigid body in space from an original position position $\mathbf{P_1}$ and orientation $\mathbf{O_1}$ to a new position $\mathbf{P_2}$ and orientation $\mathbf{O_2}$ can, on its general form, be described by its rotation and translation around a fixed axis in space relative to a global reference system (Ceccarelli, 2000). The rotation is usually decomposed in two type of movements (rotation and translation) defined by the 3×3 rotation matrix $\mathbf{R}$, and the 3D translation vector $\mathbf{v}$. Different approaches can be used to analyze $\mathbf{R}$ and $\mathbf{v}$ into a physically interpretable description. Most commonly, the relative rotations of body segments are given with respect to the three components of the global reference system. The Euler method, where $\mathbf{R}$ is decomposed into angles (roll, pitch, and jaw) describing flexion–extension, abduction–adduction and internal–external rotation of a joint is common in clinical applications.

The helical axis characteristics is extracted from $\mathbf{R}$ and $\mathbf{v}$ by defining a matrix $\mathbf{U}$ that fulfils $\mathbf{U} = \mathbf{R}^T – \mathbf{R}$ (Söderkvist et al., 1993). It can be shown that:

$$\mathbf{n} = \frac{1}{\sqrt{\mathbf{U}_{23}^2 + \mathbf{U}_{31}^2 + \mathbf{U}_{12}^2}} \cdot \begin{bmatrix} \mathbf{U}_{23} \\ \mathbf{U}_{31} \\ \mathbf{U}_{12} \end{bmatrix}$$

$$\mathbf{\Theta} = \begin{cases} \arccos\left(\frac{\mathbf{R}_{11}+\mathbf{R}_{22}+\mathbf{R}_{33}-1}{2}\right) \\ \arcsin\left(\sqrt{\mathbf{U}_{23}^2 + \mathbf{U}_{31}^2 + \mathbf{U}_{12}^2}\right) \end{cases}$$

$$\mathbf{c} = \frac{1+\cos\boldsymbol{\Theta}}{2\sin^2\boldsymbol{\Theta}} \cdot (I - \mathbf{R}^{\mathrm{T}})\mathbf{v}$$

$$\mathbf{t} = \mathbf{n}^{\mathrm{T}}\mathbf{v}$$

A.2. Movement analysis,

The rotation of the head in space can be defined as $\boldsymbol{\Theta}$**,** while $\mathbf{n}$ is the direction of the axis of motion, and $\mathbf{c}$ is the point of the axis closest to the marker placed on the sternum, whose coordinates was set to (**0,0,0**) prior to the series of movements.

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Corrado Cescon was born in Ivrea (Italy) in 1975 and studied at the Polytechnic of Turin, where he graduated in Electronic Engineering and obtained a PhD in Biomedical Engineering in 2005. From 2007 to 2008 he was post-doc fellow at the Center for Sensory-Motor Interaction (SMI), Aalborg University (Denmark). From 2008 to 2011 he worked at LISiN coordinating a multicenter study in collaboration with European gynecologists. In 2010 he won a scholarship to study for the construction of musical instruments based on bioelectric signals (AMUSE-ME). Since 2011 he is a researcher at SUPSI.

Erik Cattrysse has a PhD in Rehabilitation Sciences and Physiotherapy and a Master of Science in Manual Therapy. He is assistant Professor and Head of the department of Experimental Anatomy of the Vrije Universiteit Brussel (Belgium). He is also head of the Arthrokinematics Research Group (ARKI). His main field of interest concern the study of the anatomy and kinematics of the cervical spine in-vitro and in-vivo.

Marco Barbero received his PT degree in 1999 from the University of Milan, Italy. He obtained a master's degree in musculoskeletal rehabilitation at University of Genoa in 2005. He is currently Associate Professor in Physiotherapy and head of the Rehabilitation Area at the Research Unit of the University of Applied Sciences of Southern Switzerland. He is also undertaking his Ph.D. concerning myofascial trigger points at the School of Health Sciences at Queen Margaret University in Edinburgh. He is a guest lecturer at several courses in manual therapy and evidence based physiotherapy at different universities in Italy and Switzerland.