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An alternative 3D numerical method to study the biomechanical behaviour of the human inner ear semicircular canal

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Purpose: The vestibular system is the part of the inner ear responsible for balance. Vertigo and dizziness are generally caused by vestibular disorders and are very common symptoms in people over 60 years old. One of the most efficient treatments at the moment is vestibular rehabilitation, permitting to improve the symptoms. However, this rehabilitation therapy is a highly empirical process, which needs to be enhanced and better understood. *Methods*: This work studies the vestibular system using an alternative computational approach. Thus, part of the vestibular system is simulated with a three dimensional numerical model. Then, for the first time using a combination of two discretization techniques (the finite element method and the smoothed particle hydrodynamics method), it is possible to simulate the transient behavior of the fluid inside one of the canals of the vestibular system. *Results*: The obtained numerical results are presented and compared with the available literature. The fluid/solid interaction in the model occurs as expected with the methods applied. The results obtained with the semicircular canal model, with the same boundary conditions, are similar to the solutions obtained by other authors. *Conclusions*: The numerical technique presented here represents a step forward in the biomechanical study of the vestibular system, which in the future will allow the existing rehabilitation techniques to be improved.

Key words: *vertigo, biomechanics, finite element method, vestibular system, fluid mechanics, inner ear*

1. Introduction

1.1. Vestibular system

The manifestation of an organic problem associated with body balance is usually known as vertigo. This symptom is one of the most common medical complaints, affecting approximately 20%–30% of the world population [29]. Vertigo may be present in patients of all ages. Nevertheless, its prevalence increases with age, being the most frequent complaint in people over 70. Vertigo predominance is also associated with gender, being about two to three times higher in women than in men [17]. This disorder is an indicator of conflicting information being received by the brain. Some studies pointed out that 85% of balance dysfunctions could be related with inner ear disorders [6], mainly with the vestibular system. The vestibular system is the sensory system located in the inner ear that provides the leading contribution for movement and sense of balance.

Since human movements consist in rotations and translations, the vestibular system comprises two main components. One of the main components is the semicircular canals (SCC) system. Involving three canals placed orthogonally, the SCC is responsible for the detection of the rotational movements. The other main component is the set of otoliths located inside the utricle and the saccule. With the movement, these particles induce mechanical stimuli in the utricle and

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the saccule cilium and linear accelerations are identified by the brain through electrical signals [10]. The main vestibular structures are represented in Fig. 1.

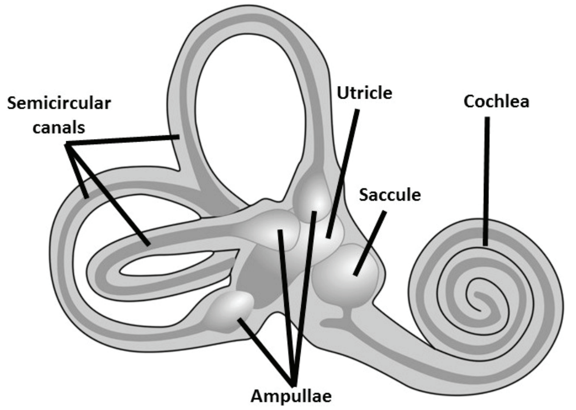


Fig. 1. Scheme of the human inner ear and the vestibular system

Each canal (Fig. 2) is comprised of a circular path filled with fluid, interrupted at the ampulla, which contains the sensory epithelium. The hair cells of the ampulla rest on a tuft of blood vessels, nerve fibers, and supporting tissue called the cupula [10]. The endolymph, inner ear fluid, is a complex component of the vestibular system. The correct computational simulation of the endolymph is a challenging and important task, since the endolymph influences significantly the maintenance of balance.

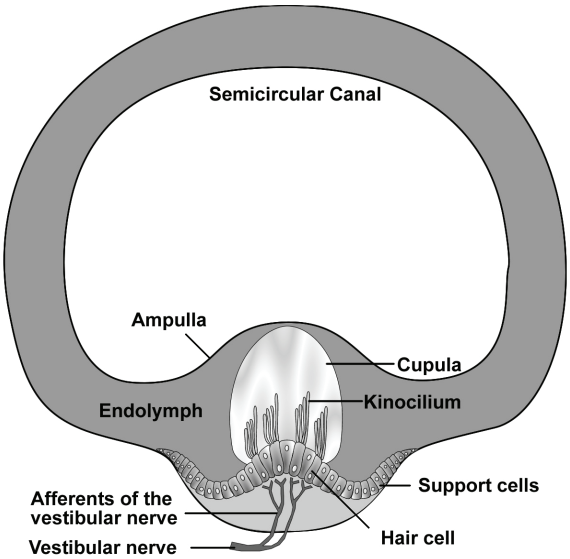


Fig. 2. Scheme of the human vestibular system – SCC

The sensory cells exhibit constant discharge of neurotransmitters, which are modified by the direction of the cupula deflection. This output signal is induced by the head rotation. Generally, these cells are called "sensors" in engineering terms.

Disorders in the vestibular system play a critical role in the quality of life of the older population. Frequently, in order to decrease the symptoms, drugs are used to suppress the activity of the inner ear. Additionally in some cases, combined with or without drug administration, physical head manoeuvres are performed within the scope of the vestibular rehabilitation program [11]. However, these manoeuvres are based on a set of empirical moves made by an audiologist, which has an associated high variability and could lead to inaccurate moves. The rehabilitation process, and even the vestibular system behaviour, can be analysed from the biomechanical point of view, taking into consideration its movements and component interactions during balance.

1.2. Biomechanical models of the vestibular system

In 1933, when the first investigations of the vestibular system were performed, Steinhausen formulated a mathematical description of the SCC which considered the dynamics of the cupula endolymph system as a highly damped torsion pendulum for the sensation of the angular motion [25].

The simulation of the fluid structure interaction between the endolymph and cupula during head rotation allows the measurement of the fluid interactions between the three ducts and the displacement of the cupula during the movement. This model could be considered useful to understand the physiological and mechanical aspects of SCC. Additionally, this model can be described as a band-pass filter relating the displacement of the cupula to the angular velocity of the head [30].

Van Buskirk and co-workers performed the first work focused on the fluid dynamics of the SCC. These researchers assumed that the endolymph has the properties of an incompressible Newtonian fluid [14], [27].

Some mathematical models have been made through the years to represent some parts of the vestibular system.

Table 1 describes the main studies related with the vestibular system and summarizes the studies with finite element models described below. Most of the research performed for vestibular system focuses only on one element of the system, usually the endolymph or the otolithic membrane. The vestibular system, however, is an integrated and complex structure, making it important to understand its function as a whole.

Table 1. Summary of numerical models related to vestibular system

	Kassemi et al. [14]	Kondrachuk [15]	Davis et al. [4]	Shuang Shen et al. [23]	Jaeger et al. [13]	Duncan and Grant [5]	Grieser and Obrist[7]
Main focus	Endolymph interaction	otolhotic membrane behaviour	otolhotic membrane behaviour	cupular deflection during caloric test	otolhotic membrane behaviour	cillium behaviour	Endolymph behaviour
Methodological approach	Arbitrary Lagrangian Eulerian (ALE), fluid–structural interaction (FSI)	Static analysis	Linear interpolation	caloric test, Arbitrary Lagrangian Eulerian (ALE), Comprehensive grid convergence tests	Cauchy's equation of motion	NA	quasi-steady Stokes flow regime
Material properties	Newtonian weakly compressible fluid	isotropic elastic parameters,	anisotropic and viscoelastic material properties, linear elastic, Kondrachuk's modeling efforts	slightly compressible Newtonian fluid	visco-elastic properties, Kelvin–Voight fluid, homogeneous isotropic materials	isotropic, linear elastic materials with circular cross-sections	NA
Boundary constraints	Non-slip stationary	one side of plate was fixed and other side is free from stresses	the neuroepithelium was fixed and the force was acting in the macular plane	Nonslip boundary conditions	the nodes at the gel-skull boundary was fixed	The stereocilium was fixed to the base	standard head maneuver, no-slip condition.

The models used in these studies are mostly based on previous geometries and only two of them were built from original images [4], [12].

Each study uses a different methodology mainly due to the different kind of goals of each research. The material properties vary according to the components being studied: Newtonian compressible fluid for endolymph studies and viscoelastic and isotropic properties for otolithic membrane studies.

The complexity of the vestibular system leads sometimes to model simplifications; for example, Suhrud Rajguru and co-workers [20] studied just one SCC instead of the whole system [20]. The 3D geometrical model was built from temporal bone histological sections and it was considered as a rigid structure. The study permitted to estimate the dynamic cupular and endolymph displacements during maneuvers [20].

Kondrachuk [15] developed a model of the otoconia membrane structure based on the Finite Element Method (FEM). This study allowed the mechanical parameters of the structure to be assessed. This model was also used to study the effect of the endolymphatic pressure on the otolithic membrane deformation. This research work concluded that the perception of inertial acceleration can be changed by the fluid redistribution due to the endolymphatic pressure [15]. In the literature it is possible to find another FEM model of the otolithic membrane [4]. This model showed the importance of 3D models in the global comprehension of the structural response. The geometrical variables studied were: the curvature of the surface, thicknesses of the three layers and the shape of the perimeter. The simulations permitted to analyze the static mechanical gain in each variable and the results showed that the three variables affected the magnitude and directional properties of the otoconia membrane [4].

An SCC model using virtual reality, to simulate the surrounding physical environment, was developed by Selva et al. [21]. The aim was to represent the vestibular sensors and simulate several rotation movements of the head occurring during a diagnosis of the vestibular system disorder. It can be used as a learning and demonstrating tool to understand the behavior of the sensors during any kind of motion [21].

A recent paper from Shen et al. [23] showed a caloric response of a complete vestibular system obtained by FEM analysis. The results of the caloric test on the model developed show the efficiency of the analysis on the evaluation of the functionality of the horizontal canal [23].

Another FEM simulation was presented by Jaeger et al. in 2002 [13]. The results show that the curvature of the maculae surface has no effect on the mechanical response, since the elastic coupling in the otolith membrane is insufficient [13].

Duncan and Grant [5] presented a study focusing on just one part of the vestibular system mechanics: the hair cells in the cupula. The FEM model representing the hair cells was used to better understand the mechanotransduction phenomenon [5].

Dominik Obrist et al. [18] proposed a simple numerical model to study the canalithiasis phenomenon, which is a condition that leads to a vestibular disorder called Benign Paroxysmal Positional Vertigo (BPPV). The work demonstrated that variations in the cross sectional area of the SCC are not necessary conditions for a positional nystagmus [18].

In the vestibular field, a new mathematical tool has been developed by Bradshaw et al. [2] for modeling the three dimensional geometry of SCC in humans. The main goal of the methodology is to understand the physiology of these canals. The technique developed allows the SCC geometry to be automatically reconstructed based on computed tomography images. After this step, the geometrical model is mathematically modeled using Fourier equations. The aim of this model is to help the diagnosis and clinical treatment of BPPV in each patient [2].

In the literature, there are other research studies focusing on the determination of coefficients that are a non-linear function of the morphological parameters of the SCC system. Thus, a mathematical model of the SCC mechanics was constructed for this purpose [28]. The work performed by Vega et al. [28] is based on Steinhausen's work, which uses a linear model of the torsion pendulum to study the dynamics of the cupula-endolymph system. The proposed model focuses on the mechanical coupling of angular accelerations with the movement of the sensory hair cells. The mechanical coupling largely determines the type of mechanical stimuli that is responsible for activation of hair cells [28].

There are already some studies on the fluid mechanics of the SCC that show that endolymph is a transducer for angular velocity of the head [27].

Grieser and Obrist [7] studied the endolymph motion of one SSC using FEM. The simulations performed consisted of a head rotation from a relative angle of 0° to 120° (a standard head maneuver). The rotation axis is oriented perpendicularly to the plane of the horizontal SCC. During the acceleration of the head, the walls of the SCC displace the adjacent fluid layer along their path. The work confirms the validity of all assumptions that were made in previous studies of the endolymph motion [7].

A fluid dynamic model of the vestibular system was proposed to understand the fluid/solid interaction phenomenon occurring between the thin membrane, separating the SCC from the cupula, and the endolymph flow [3]. Using real mechanical and anatomical parameters, a realistic vestibular model was also built using a 3D printer in order to study the best way to mimic the vestibular system [3].

In the field of mathematical models, another study was performed to examine two mechanisms proposed for BPPV [24]. The research concluded that a larger volume displacement on the cupula could be originated from larger or multiple otoconia [24].

Beyond the computational simulation, there are some research studies using in vitro models of the SCC. Valli et al. [26] performed some simulations using animal isolated posterior SCC to investigate if otoconia can produce transcupular pressures able to stimulate ampullar receptors. Obrist et al. [19] published an in vitro research work regarding the study of canalithiasis, which confirmed the fundamental mechanism of BPPV [19].

Within the same topic, Selva et al. [22] developed a 2D finite element model of a single SCC, for which the displacements and velocities of the cupula were analyzed.

The aim of the present work is to study the mechanical behavior of the vestibular system in order to develop new and more efficient techniques to assist the vestibular rehabilitation, helping to avoid in the near future the high costs and problems associated with these symptoms.

Under these assumptions the use of engineering tools, such as the computational analysis, is an opportunity to create virtual simulations close to the real scenarios [8]. Thus, to achieve this goal, a 3D SCC model similar to the real system was constructed. The SCC is discretized with the FEM and the endolymph is simulated with the smoothed particle hydrodynamics method.

2. Smoothed Particle Hydrodynamics

The fluid/solid interaction is a highly demanding topic in computational mechanics. Nowadays, the Smoothed Particle Hydrodynamics (SPH) is one of the most popular numerical methods to study such phenomenon. The SPH can be used to simulate body fluids with low velocities, such as hemodynamics [16].

The SPH method works by dividing a continuous field into a set of discrete sample points, called particles. These particles have a spatial distance, over which their properties are "smoothed" by a *kernel function*. The kernel function helps to ensure the stability of the numerical solution [16]. The particles are identified with some characteristics such as mass, position and velocity. Additionally, particles can also carry estimated physical properties depending of the problem, such as mass-density, temperature and pressure.

The SPH approximation is based on two steps, the first one is the kernel approximation and the second is the particle approximation [16]. The result of the first step is the following quantity function

$$f(x) = \int_N f(x')W(x - x', h)dx' \tag{1}$$

where x is any point in N, the support domain, and $W(x - x', h)$ is a smoothing kernel function. The smoothing length, h, defines the influence area of the smoothing function $W(x - x', h)$ [16]. This parameter can be fixed in space and time, however, this procedure does not take advantage of the full power of SPH. Thus, assigning to each particle its own smoothing length and allowing it to vary with time can lead to an automatic adaptation of the simulation resolution depending on the local conditions [16].

There are two main functions leading to equation (1), $f(x)$ and $W(x)$ as demonstrated below. The basis of integral representation of a function used in SPH is the function that represents the three dimensional position vector x,

$$f(x) = \int_N f(x')\delta(x - x')dx' \tag{2}$$

where $\delta(x - x')$ represents the Dirac delta function given by

$$\delta(x - x') = \begin{cases} 1 & x = x', \\ 0 & x \neq x'. \end{cases} \tag{3}$$

If the delta function of equation (3) is replaced in equation (2) by the smoothing function $W(x - x', h)$, it is possible to obtain equation (1). The particle approximation plays an important role within the SPH method. Consider Fig. 3, in which a set of particles possessing individual mass are scattered in space.

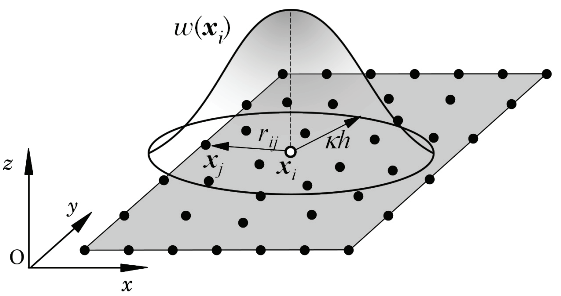


Fig. 3. Particle approximation. W function of particle i

The mass of a particle is defined by the relation between the density and the volume by the following expression

$$m = V\rho. \tag{4}$$

Thus, the function approximation for a particle i can be represented by equation (5), with the infinitesimal volume dx' in the above equations being replaced by the finite volume of the particle j.

$$f(x_i) = \sum_{j=1}^{N} \frac{m_j}{\rho_j} f(x_j) X(x_i) W(x_i - x_j, h) \tag{5}$$

where ρ_j is the density of each particle in the domain N and m_j is the mass of each particle.

The SPH was first developed to simulate astrophysical phenomena. Afterwards, it has been successfully applied to a vast range of problems, such as explicit fluid flow analysis. It was developed by Gingold and Monaghan (1977) and Lucy (1977) and it can be combined with the governing equations of the classical Newtonian hydrodynamics [16].

Within the SPH, the computational domain is represented by a set of computational points – particles – completely discretizing the problem domain.

This method has some advantages over grid-based techniques because its concept is simple and it is relatively easy to incorporate complicated physical effects into the SPH formalism [16].

Liu et al. [16], after some tests with SPH conditions, proposed a reproducing kernel particle method, which improves the accuracy of the SPH approximation.

The SPH method was initially developed as a probabilistic meshfree particle method and was later modified to a deterministic meshfree method [16].

There are several applications for the SPH method [1], [16]. One advantage of this method is the adaptability of the particles to many fields and subjects. For instance, the SPH was already used in astrophysical studies to study galaxies, the star formation process and even to simulate cosmological impacts [16].

Additionally, the SPH is capable of solving efficiently high velocity impact problems. Benz and Asphaug used the SPH method to simulate the fracture of brittle solids. In 2000, this method was also applied in the metal forming processes [1].

The SPH presents several advantages when compared with other numerical methods. The SPH deals efficiently with large local distortions of the discretized domain, it permits the mass conservation of the particles. It allows calculating the pressure by neighboring particles and not by solving linear equations and it is suitable to simulate the fluid free surface and the fluid/solid interactions.

Furthermore, SPH is well-known in the computational mechanics research community by its efficiency in the simulation of fluids motion. The first fluid flow application of the SPH was performed by Swegle with elastic flow. Later, Morris applied the SPH to solve magneto-hydrodynamics and Morris and Monhagan solved multi-phase, quasi-incompressible flows, gravity currents, flow through porous media, heat conduction and shock simulations. In 1995, Cleary applied SPH in heat transfer and mass flow [16].

Over the years, the method has been optimized in order to increase the solution accuracy and to enlarge the application field.

3. Methods

This work used the commercial software ABAQUS® to build and analyze the 3D model of the SCC. The analysis combined both the FEM and SPH methods.

3.1. Numerical model of the SCC

The 3D model developed was composed of two main parts: a small shell ring, representing one SCC with the cupula, as shown in Fig. 4, and the particles representing the endofluid.

The measures of the SCC represented in Fig. 4 were obtained from a 3D model of the complete vestibular system that can be found in the work of Henson et al. [9], which was constructed based on magnetic resonance imaging with fine resolution. In Fig. 4, it is only possible to visualize the outer FEM shell, which represents the vestibular membrane of the SCC. This shell was defined as a rigid body, because it does not present significant deformations. The particles defining the endofluid are inside the shell. The particle distribution is regular and the sum of the particles volume is equal to the volume of the SCC with the cupula.

Table 2 shows the material properties for the endolymph and the outer membrane. These values were obtained in the literature for the components of the vestibular system.

Additionally, in order to validate the model, three distinct particle discretizations were considered and analyzed: mesh M1 (1790 particles), mesh M2 (7410 particles), and mesh M3 (13637 particles).

Table 2. Material properties of the model [39]

Component	Young's modulus [Pa]	Poisson's ratio	Density [Kg/m^3]	Viscosity [Pa.s]
Endolymph	–	–	1.0×10^{-3}	4.8×10^{-3}
Membrane	5.0	0.48	–	–

The model shown in Fig. 5a represents the outer membrane of the SCC. In Fig. 5b–d, it is possible to visualize the particles discretizing the endolymph, which were used by the Smoothed Particle Hydrodynamic (SPH) method to simulate the fluid flow.

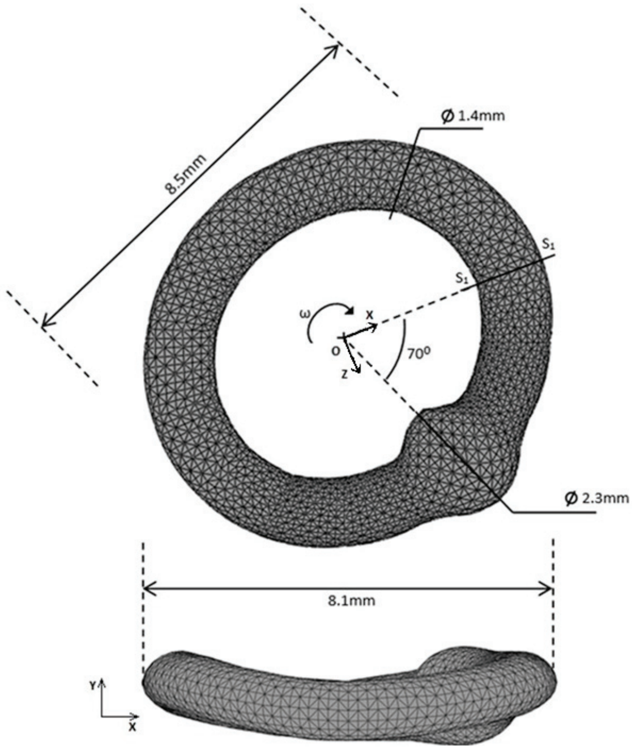


Fig. 4. Model of one SCC built with finite elements, front and top view

3.2. Boundary conditions

Regarding the essential boundary conditions, to each node of the elements belonging to the membrane a prescribed angular velocity ($\omega = \pi/2$ rad/s) is enforced, accordingly with Fig. 4. The angular velocity, with respect to point O, is a time dependent function.

In order to analyze the biomechanics of the SCC model, two distinct angular velocity functions were considered: profile $\omega 1$ and profile $\omega 2$. Both functions are shown in Fig. 6. Both angular functions were enforced to the three discretized models presented in Fig. 5b–d. The functions considered were previously used by Selva et al. [22] and Wu Cai-qin et al. [30].

Additionally, in the present model, the density and volume of each particle were assumed, as Table 2 indicates. Thus, this model considers the gravity acceleration along the *zz* axis, following the referential indicated in Fig. 4.

Regarding the contact between the distinct discrete elements of the model, the SPH formulation rules the contact between the particles. The general contact between the membrane elements and the fluid particles is performed explicitly by ABAQUS.

In order to compare the results from the several models considered in Fig. 5, the instantaneous and local discharge variable was used. The discharge is the

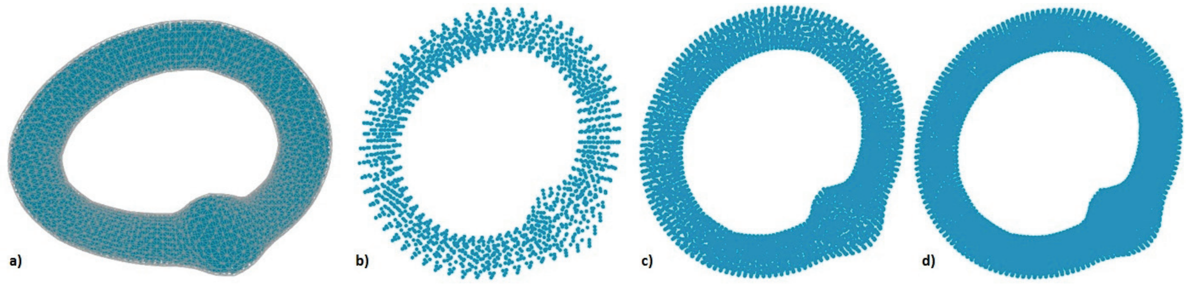


Fig. 5. Model of the SCC FEM mesh (a). Examples of the particle discretization: (b) 1790 particles, (c) 7410 particles, (d) 13637 particles

volume rate of fluid flow transported through a given cross-sectional area. The discharge is calculated by the following equation

$$Q = v \times A \tag{6}$$

with v being the fluid average normal velocity at a given cross-section and A the area of that cross-section. The section analyzed in this work is indicated in Fig. 4 with the line segment S_1. Furthermore, the results from works [22], [30] were processed in order to present the discharge values of those and permit a valid comparison with the results obtained here.

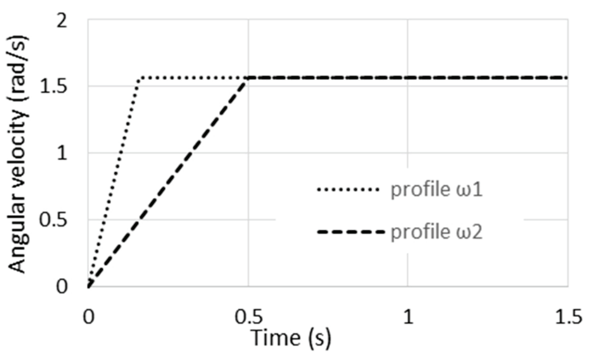


Fig. 6. Representation of the two angular velocity functions imposed in the model

4. Results

After the simulation of the rotational motion of the SCC 3D model presented in Fig. 4, the obtained results were processed and compared with other similar works in the literature [22], [30].

A comparison of the discharge variable for distinct time steps of the full analyses is presented in Fig. 7.

Figure 8 shows the comparison between the discharge obtained with each of the three meshes presented previously and the works of Selva et al. [22] and Wu Caiqin et al. [30].

In order to perceive the model behavior during the imposed angular velocity, the velocity of the fluid over time was analyzed.

The fluid velocity, at some instants of the simulation, is presented in Fig. 9. As it is perceptible, the inner fluid velocity increases with time, as expected.

After the general analysis of the global domain, a section of the SCC was analyzed in detail. The results were obtained for the section marked as S1 in Fig. 4. Figures 10 and 11 correspond to profile $\omega 1$ and profile $\omega 2$, respectively, and show the 2D sectional view on the left side and the velocity 3D distribution on the right side.

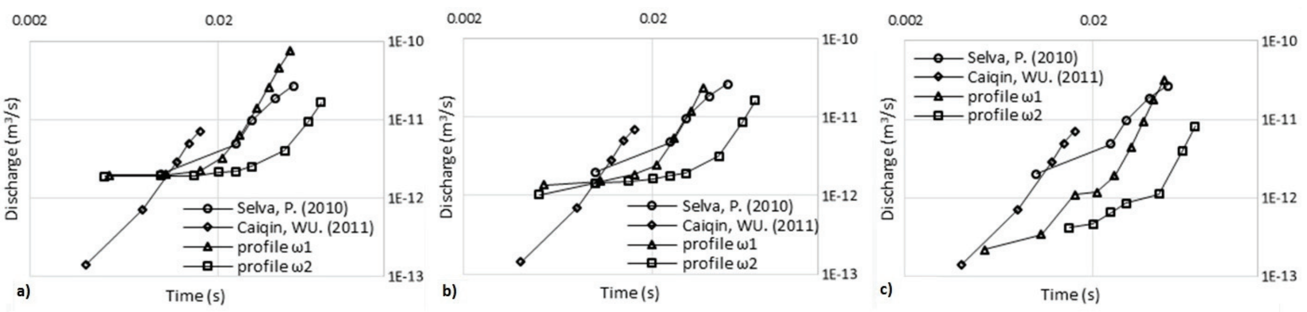


Fig. 7. Discharge of the fluid in the chosen section with the three meshes and comparison with (a) M1, (b) M2, (c) M3

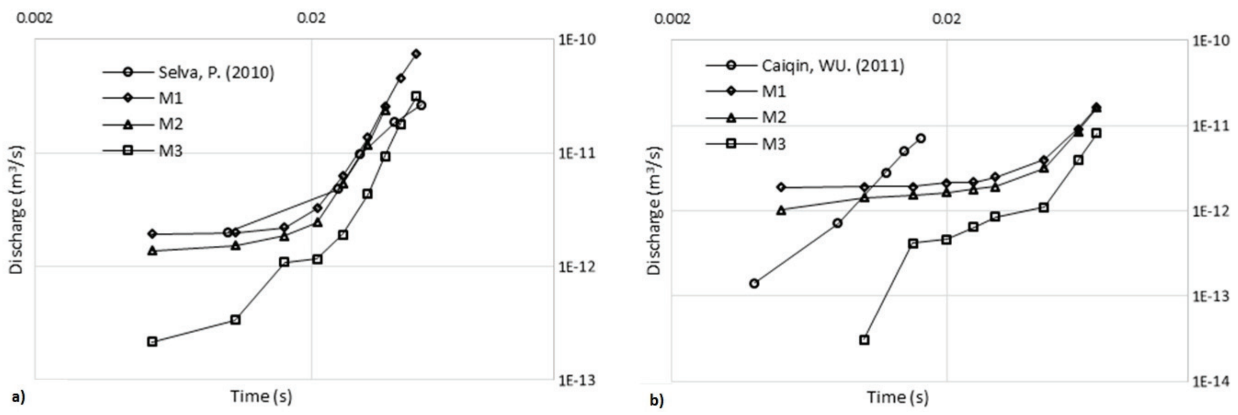


Fig. 8. Discharge in the three meshes from both time steps:
(a) profile $\omega 1$, (b) profile $\omega 2$

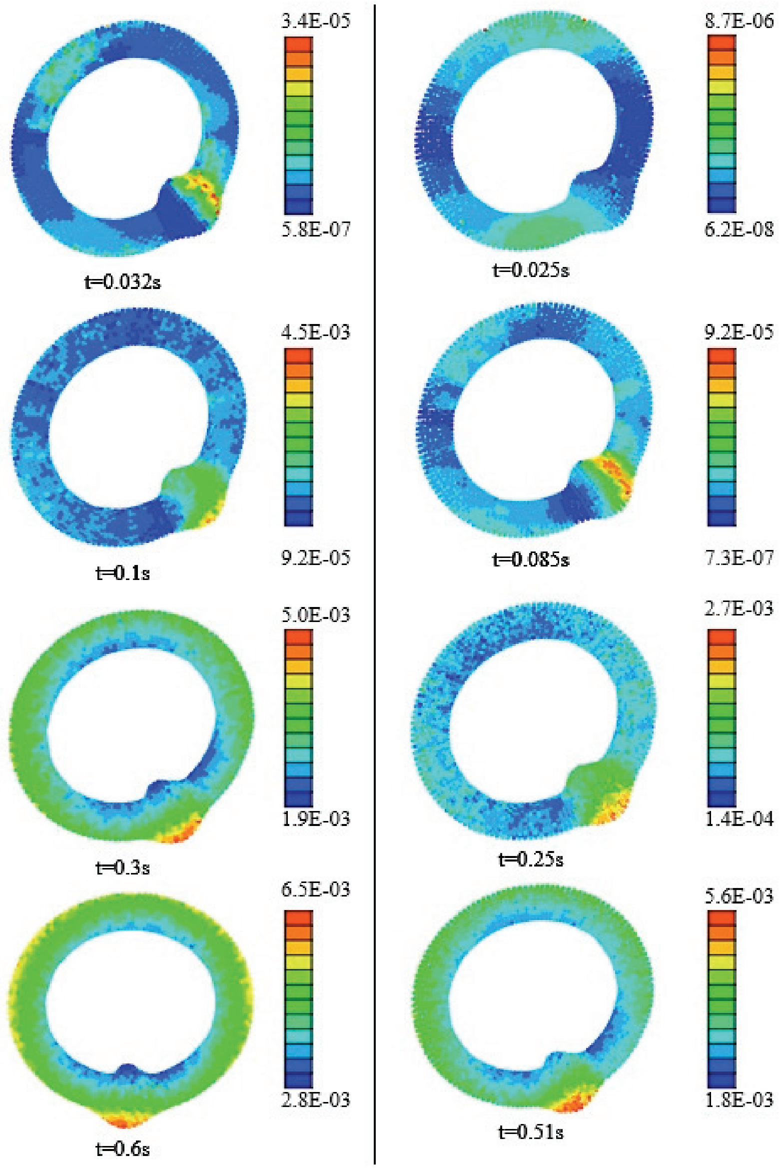
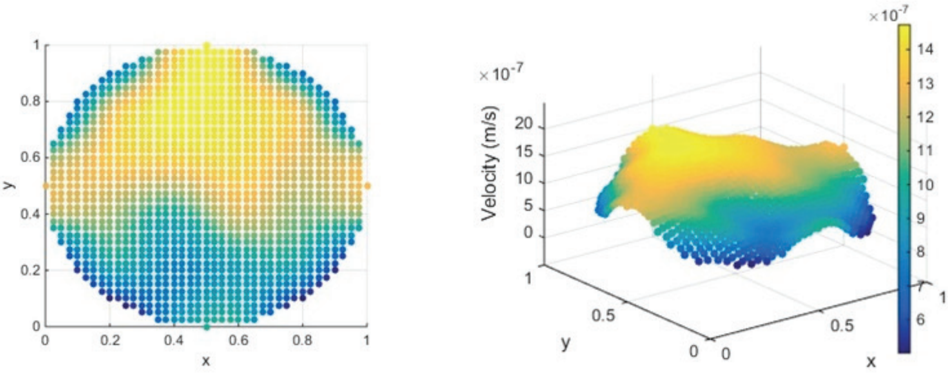
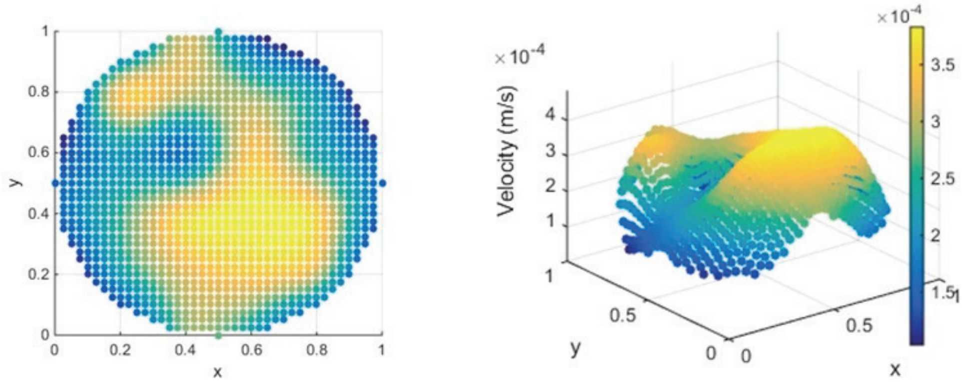


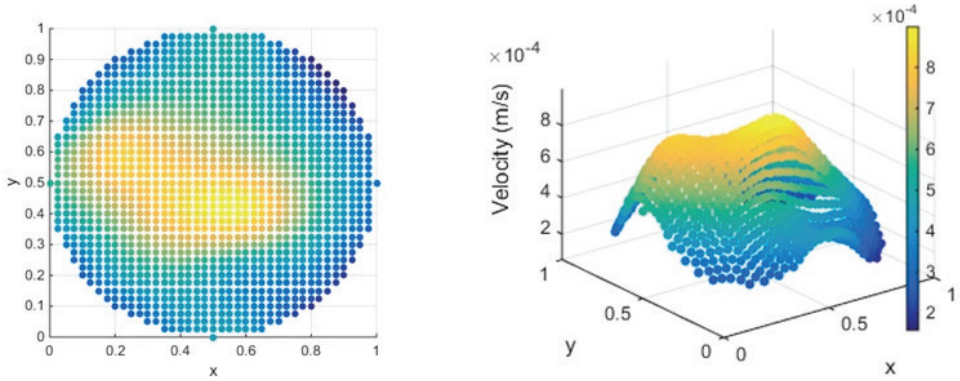
Fig. 9. Fluid velocity (m/s) with M2 along time
from both profile $\omega 1$ (left column) and $\omega 2$ (right column)



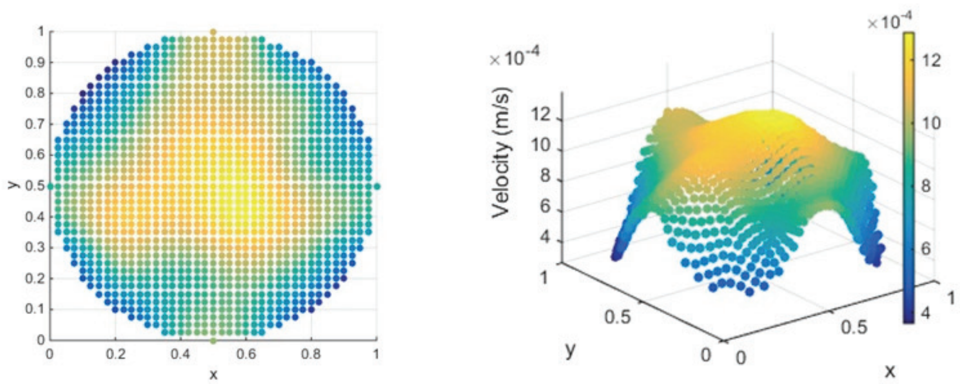
t = 0.005 s



t = 0.04 s

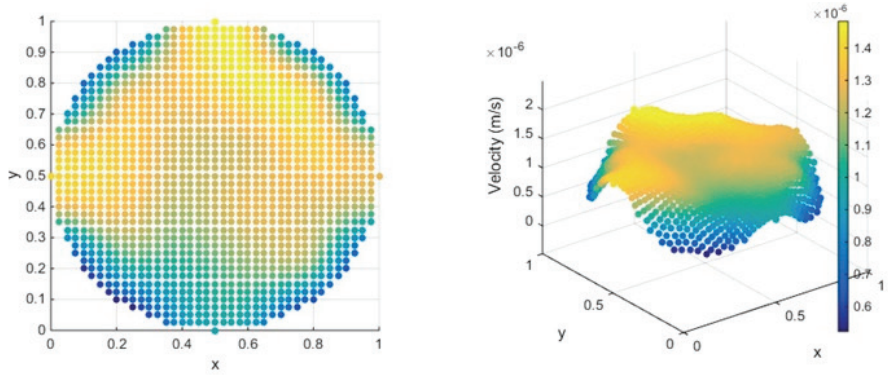


t = 0.08 s

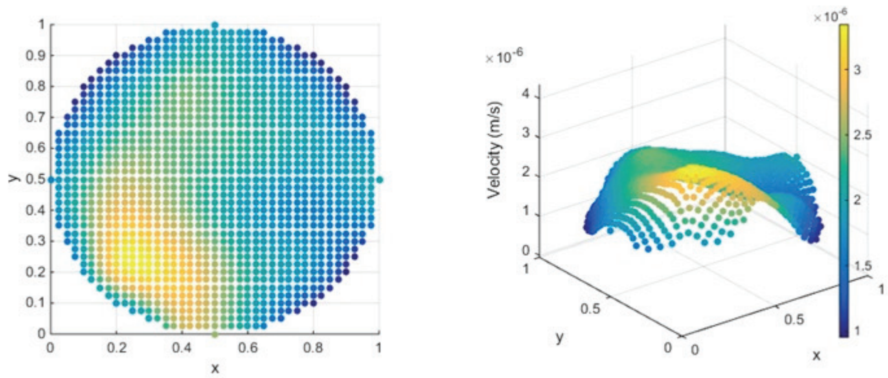


t = 0.11 s

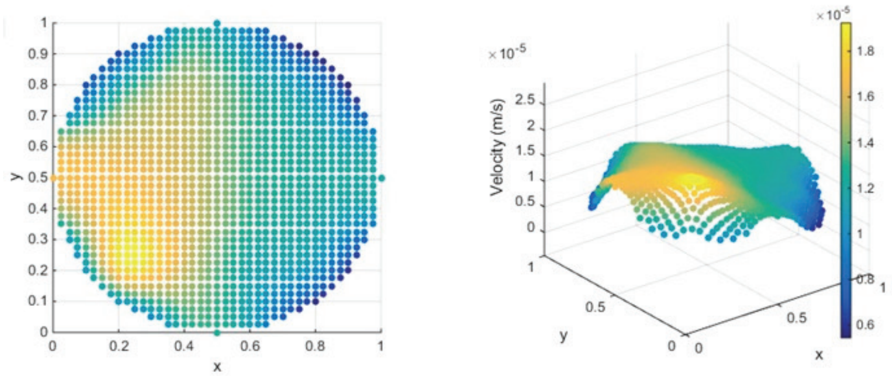
Fig. 10. 3D velocity field along time obtained with profile $\omega 1$



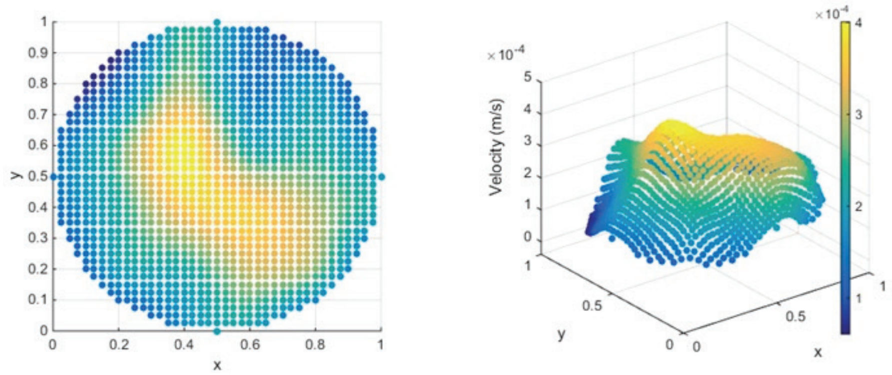
$t = 0.005$ s



$t = 0.045$ s



$t = 0.075$ s



$t = 0.105$ s

Fig. 11. 3D velocity field along time obtained with profile $\omega 2$

5. Discussion

The aim of the study presented in this paper is to improve the knowledge of the vestibular system. For this purpose the research is based on the simulation of the rehabilitation process, which need to be more enhanced, and to describe the pathway leading to good rehabilitation results for the patient.

A new model of the vestibular system was developed including a representation of the sections of the SCC, which are the focus of this study. The physiological behavior of the SSC represents the more challenging phase of the complex rehabilitation process.

Thus, it is necessary to construct an accurate model representing such structure of the vestibular system. The three-dimensional model constructed permits a more precise geometric representation of the SSC and allows the trajectory of the fluid inside the canal to be visualized.

The FEM demonstrates once again that this numerical method is a robust technique to obtain fast and reliable results even in the biomechanical field. Additionally, the use of SPH allows a more realistic representation of the fluid behavior. The methodology used permitted us to obtain promising results, considering the biomechanical properties of the components available in the literature.

The results obtained with the three different particle meshes were compared with the works of Selva et al. [22] and Wu Caiqin et al. [30]. The results obtained in this work are very close the solution obtained by Selva et al. [22] and Wu Caiqin et al. [30] when a similar angular velocity is applied to the SCC model, regardless of the particle mesh considered. Additionally, the convergence of the analysis was confirmed.

As Fig. 8 shows, the present analysis produces results closer to Selva et al. solution. The slight differences between the solutions can be explained with the dimensional dissimilarity between the model presented here (a 3D model) and the models of both Selva et al. [22] and Wu Caiqin et al. [30], which are 2D models. The range of results obtained for the discharge are equal in all the simulations (Fig. 8a, b), but in terms of evolution of the fluid flow, our results show more similarity with the work by Selva et al. [22].

Figure 9 shows a lateral section of the model during the simulation. The velocity of the fluid was higher in the cupula, similar to the work of Selva et al. [22], and in the external part of the canal, because of the centrifugal force. When the two velocity profiles are stabilized (after a 0.5 s period) both models present a similar average velocity, as can be seen in Fig. 9.

The velocity 3D distribution is almost fully developed in the last frame of Figs. 10 and 11. The 3D velocity field obtained in both simulations is increasing, as expected. The velocity obtained in the 3D velocity field at instant 0.1 s is three times higher in profile $\omega 1$, the result being in accordance with the velocity applied to the models for that time step, as shown in Fig. 6.

The velocity maps, representing the fluid flow of the entire canal, permitted us to observe the centrifugal force and the consequent higher velocity in the cupula, due to the shape of the SCC. The use of a three dimensional model allowed the visual evidence of a 3D velocity field in a section of the canal at some instants of the simulation. The analysis of the fluid in the section analyzed showed, as expected, that the velocity is higher in the center of the canal.

The encouraging results obtained with this 3D model of just one SCC make it possible to develop and build with confidence the entire model, containing all the main structures of the vestibular system.

The simulation model allows the study of the rehabilitation process in a new perspective without the suffering of the patient. The possibility of modification of all the variables that could influence the symptoms is also an important step in the study of the vestibular diseases.

6. Conclusion

This paper presents an alternative reliable numerical approach to study the biomechanical behavior of the vestibular system. In general terms, the results obtained with the 3D numerical model were very similar with the results from literature. However, the 3D velocity distribution which can be obtained in any section of interest is a significant improvement of the present work allowing the velocity field to be analysed point by point, which will allow the otoconia movement to be predicted accurately in future developments of the research. The finite element method combined with SPH for the fluid simulation seems suited to simulate the biomechanical behavior of the SCC.

From the obtained results, it is possible to conclude that the numerical approach presented is convergent and robust, indicating that the increasing particle discretization leads to accurate results.

Additionally, it was found that the SPH approach is able to produce solutions very close to the results obtained in the work of Selva et al. [22], which uses a Lagrangian–Eulerian approach combined with the FEM formulation.

The development of scientific knowledge of the vestibular system biomechanics is an important step to create new and more precise tools, which can be helpful in the daily routine of people with vestibular disorders. In this field, a better comprehension of the biomechanical behavior of the vestibular system is vital, in order to enhance the computational simulations and the numeric models. This work contributes with an innovating numerical approach to predict the complex movement of the endolymph inside the SSC, which will make it possible to experiment in silico new maneuvers for the rehabilitation therapy.

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