Bio-mechanical Simulation of the Fetal descent without Imposed Theoretical Trajectory

R. Buttin, F. Zara, B. Shariat, T. Redarce, G. Grangé

***Abstract*—Childbirth medical training of young obstetricians consists in performing real deliveries, under supervisor control. This medical procedure becomes more complicated when instrumented deliveries requiring the use of forceps or suction cups become necessary. For this reason, the use of childbirth simulation tools can complement the training of obstetricians (generally considered as too short) and could improve the medical procedures. The realization of this kind of simulators can repose on a virtual part, enabling the visualization of the organs and fetus behavior. Moreover, this virtual part can be coupled to a haptic interface that allows user interaction. In this context, we have developed, in collaboration with obstetricians, a bio-mechanical model of the childbirth. This paper focuses on the geometrical and bio-mechanical models of the mains organs involved (*i. e.* the parturient's uterus, abdomen and pelvis interacting with the fetus). This model allows the realistic simulation (based on continuum mechanics and finite element method) of the descent of the fetus through the birth canal during childbirth. Moreover, this simulation was coupled to a haptic interface (the BirthSIM Simulator). The results validate the pertinence of the proposed model to improve the existing training simulators for the childbirth obstetrician gesture.**

***Index Terms*—Medical training, childbirth, bio-mechanicals model, 3D realistic simulation, continuum mechanics, behavior laws.**

I. Introduction

Classically, childbirth medical training of young obstetricians consists in performing real deliveries, under supervisor control. However, this medical procedure becomes more complicated when instrumented deliveries requiring the use of forceps or suction cups become necessary. A survey of AURORE (Association des Utilisateurs du Rseau Obsttrico-pdiatrique REgional) network of the Rhone-Alps region in France showed the number of complications related to the use of forceps or suction cups [1]. It appeared that out of 4589 births, nearly 150 finished by light or serious lesions of fetus. In addition, nearly 90% of obstetricians who participated in this survey approve the use of childbirth simulation tools for training doctors. Indeed, the use of these learning tools complement the training of obstetricians generally considered as too short and could improve the medical procedures.

Actually, many simulators exist. In most common cases, they enable the medical training of instrumented delivery using a physical interface. Most often, their interface is composed of several physical parts (an assembly of plastic pieces) which represent the anatomy of some concerned organs (generally the pelvis and the head of the fetus). In addition, a motorized articulated system animate these physical parts to simulate the interaction of fetus with the parturient's organ and the obstetrician. Thus, this haptic device allows to generate resistant forces to reproduce a similar sensation to that felt by the practitioner during the delivery. Moreover, these simulators permit the practitioner to have a very good immersion because of the similarities between anatomical representation by plastic parts and the reality.

However, if we seek to develop learning scenarios at different levels of difficulty, by taking into account some complex delivery cases, these tools quickly show their limits. It would thus be interesting to develop a more versatile and configurable tool, allowing to take into consideration different parturient-fetus anatomical and morphological structures, corresponding to different pathological cases. Such a tool uses augmented reality techniques and is composed of two parts: a virtual part simulating the birth process and a haptic interface. The implementation of the virtual part could go through the definition of a complete bio-mechanical model of the female reproductive system and the fetus, as well as the simulation of their behaviour during childbirth, allowing the calculation of stresses generated by the descent of the fetus. The results of this calculation would then be input to the haptic device in interaction with the apprentice. An important constraint in the definition of this model concerns the interactive simulation time. This implies the optimization of the model as well as the calculations.

In this paper, we propose a bio-mechanical model of the female genital system based on continuum mechanics laws which allows the simulation of different organs involved in childbirth (uterus, abdomen, soft and bony pelvis). This simulation allows the computation of fetal trajectory during childbirth resulting from the interactions that occur between the fetus and the parturient's organs.

This paper is organized as follows. Section II presents a state of the art on childbirth simulators and more particularly on the bio-mechanical models already developed in this context. In section III, we present briefly the functional anatomy of the parturient's genital organs and we detail our geometrical and bio-mechanical model. In section IV the details of our biomedical simulation are presented and in section V, we present its coupling with a haptic interface: the BirthSIM simulator [2], [3]. Then, the results of our fetal descent simulations are given in section VI. Finally, section VII presents the conclusion and the perspectives of our work.

R. Buttin, F. Zara and B. Shariat are with Université de Lyon, CNRS, Université Lyon 1, LIRIS, SAARA team, UMR5205, F-69622, France

R. Buttin and T. Redarce are with Université de Lyon, CNRS, INSA de Lyon, Laboratoire Ampère, UMR5005, F-69621, France

G. Grangé is with Maternité Port Royal, Groupe Hospitalier Cochin - Saint Vincent De Paul (Assistance Publique - Hôpitaux de Paris), F-75679, France

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II. State of the art

Simulators for training are currently used in many areas such as aeronautics [4], but also in medicine, as an instructive tool or as a medical support for surgery [5], [6], [7], [8], [9]. In the field of obstetrics, several simulators have been proposed for medical training. The first virtual simulator was proposed by Boissonnat in 1993 [10]. This simulator allows to adjust various geometric parameters such as pelvic organs or fetus morphology. But, this simulator (non equipped with a haptic device and thus devoid of interaction with the user) was not designed to train young obstetricians, but rather to make a prognostic of the delivery by conducting a simulation of the fetus descent guided by a pre-computed imposed trajectory. Thus, the proposed simulator along a theoretical imposed fetal path, does not take into account different delivery scenarios. In 2004, a simulator coupling a virtual model of the fetus and pelvis to a three-axis haptic system representing the obstetrician's hands was developed by Kheddar [11]. But, similarly to the previous work, the simulation was again based on an imposed trajectory, thus insufficient to take into account different identified pathological cases. Nevertheless, we believe that this goal is essential for the realization of a simulator to reproduce various typical training scenarios.

Other studies have been carried out on the characterization of pelvic dynamics by shape descriptors to estimate the pelvic organs deformations (bladder, internal genital organs, rectum) [12]. However, this study is not oriented towards childbirth simulation and it concerns the pelvic floor disorders of non pregnant women. Consequently, it is difficult to rely directly on this work.

In summary, the objective of our work is the realistic simulation of the descent of the fetus and its interactions with pelvic organs during childbirth. This simulation has to take into account the morphology of the organs of the parturient and the fetus. Then, it has been coupled to our haptic device (BirthSIM [2], [3]) enabling users' interactions with the help of appropriate obstetrical instruments. To achieve this result, we propose a bio-mechanical model of involved organs, as well as the physical simulation of their behavior. This implies to solve numerically the continuous media mechanics. One can note that many numerical and approximation resolution methods exist [13]. Fig. 1 presents a brief comparison of existing methods using three main criteria:

- precision, that indicates the relevance of the method, providing results close to reality;
- stability, which indicates the degree of convergence of the method towards a solution, regardless of the applied stress;
- interactivity, which indicates the speed of the calculation method.

At this stage of our study, the objective being to validate the pertinence of the proposed model, we use the finite element method, for its qualities of stability and accuracy, regardless of computation time. The next step of our work is to optimize the proposed model and its numerical resolution method to obtain interactive time response. We note that different finite element optimization works, such as [14] for large deformations do exist.

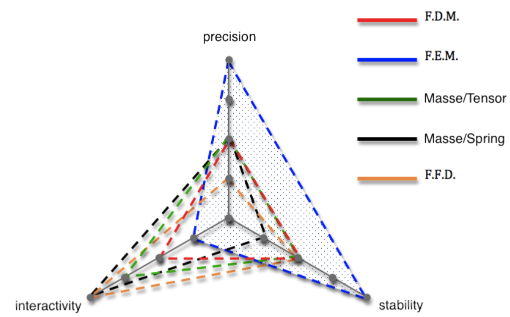

Fig. 1. Comparison of numerical simulation methods according to results quality, stability and computation time (Finite Difference Method, Finite Elements Method, Mass/Tensor, Masses/Springs system, Free Form Deformation).

III. Our Geometrical and Bio-mechanical Model

A. Functional Anatomy

The delivery is a complex physiological phenomenon involving many organs. The embryo develops during gestation in the uterus. Then, during the different stages of labor, the uterine contractions combined with the abdominal and diaphragm thrusts expel the fetus. During its descent, the fetus will cross the pelvic inlet (superior pelvic strait) and then the pelvic outlet (inferior pelvic strait). The head of the fetus, which is the widest part, will remove the pelvic floor muscles in order to extricate itself from the uteri-vaginal canal. To simplify the anatomical model of delivery, we consider only the essential components, that is to say the uterus, abdomen and soft and bony pelvis as well as the fetus to achieve a "realistic" model (*cf.* Fig. 2).

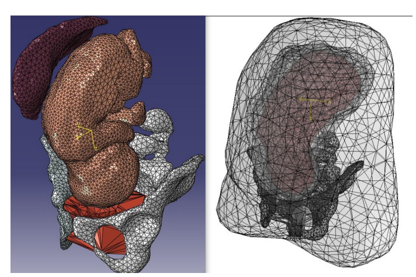

Fig. 2. Geometrical models of the fetus, uterus, pelvis and abdomen.

The geometry of the various organs has been extracted from MRI data for soft tissues (uterus, fetus, abdomen) and CT-scans data for bony parts (pelvis and sacra-lumbar spine) of pregnant women. These data were provided by the Saint Vincent de Paul Hospital (AP-HP) at Paris. Then, they have been processed to obtain a triangular surface mesh as well as a tetrahedral volume mesh of the organs (*cf.* Fig. 3).

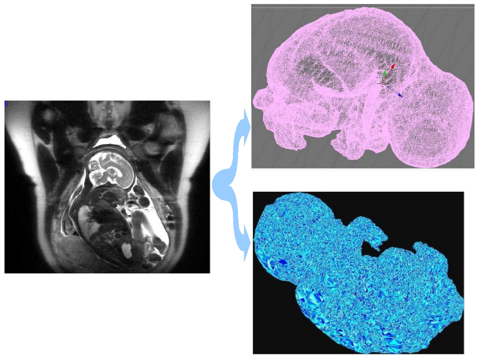

Fig. 3. Meshes of a fetus obtained after segmentation from MRI data (left): surface mesh based triangles (top right) and volume mesh based tetrahedra (bottom right).

B. Pelvis and Sacra-Lumbar Spine Model

The bony pelvis is composed of three parts (left and right iliac wings and sacra-lumbar spine) connected together by a set of ligaments. The network of perinea muscles of the pelvis, located at the pelvic outlet, is commonly called ”pelvic floor”. The Fig. 4 illustrates the bony and muscular parts of the pelvis.

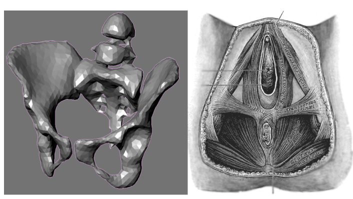

Fig. 4. The pelvis is composed of two parts: bony part (left) and muscular part called ”pelvic floor” (right).

The mesh of the bony pelvis, obtained directly from the CT-scans data is very complex and contains a large number of nodes (*cf.* Fig. 5 - left) leading to increased computing time. Therefore, we have simplified the mesh by smoothing, while preserving its essential characteristics, such as the ischial spines, the tip of the coccyx and the pubis area. For this, we first made a very coarse mesh which is based on bounding boxes of different connected parts of the pelvis (*cf.* Fig 5 - in the middle). This mesh was then smoothed to remove sharp edges. In the end, we obtain a mesh with $4,765$ instead of $18,300$ initial nodes (*cf.* Fig 5 - right).

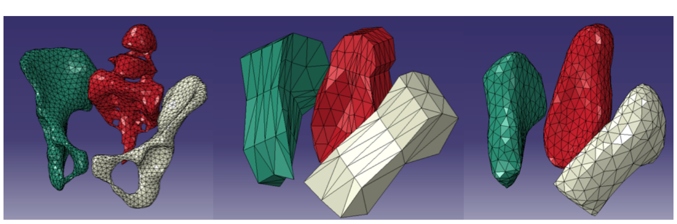

Fig. 5. Simplification of the bony pelvis mesh: (from left to right) initial mesh ($18,300$ nodes), bounding boxes, final mesh ($4,765$ nodes).

The pelvis is a key element in the delivery with a resistive role for the pelvic floor which surrounds the lower part of the uterus and the vaginal area. Note that, the modeling of the pelvic floor has been incorporated into the parturient’s abdomen. The bony pelvis also plays an important role by guiding the fetal head into the birth canal. The pelvis then performs a nutation movement composed of two dependent rotations: a forward tilting of the sacrum when the fetal head is placed in the vaginal canal, and an abduction of iliac wings resulting in a decrease in diameter of promonto-retro-pubic as well as an increase of the sub-sacra-pubic diameter. The purpose of this variation in diameter is to facilitate the fetal descent, allowing the birth canal enlargement. In the end, we consider the iliac wings as stationary and non deformable parts, the upper spinal sacra-lumbar as fixed and we have just allowed a rocking motion to the lower level.

C. Fetus Model

Let us study the bio-mechanical model of the fetus. On a mechanical point of view, it can be regarded as a very small human. Therefore, it is composed of different materials each with their own laws of behavior, their mechanical properties and a given density. Note that the body of a human being is constituted of about 90% of water, its density is just below $1,000$ kg/m3, with heavy parts essentially in the muscular areas. Thus, assuming that a fetus has a muscular density less important than an adult, we have considered the average fetal density slightly lower than $1,000$ kg/m3.

From a geometrical point of view, within the uterus, the fetus assumes a tuck position to reduce its congestion. At this position, its height is approximately 30 cm and a width of 12 cm and an average weight of 3.5 kg [15]. From anatomical point of view, the proportions between different parts of its body are not the same as an adult or a child. Indeed, the fetal head is highly developed compared to the rest of his body.

From a complexity point of view, it is not possible to model all the different organs of the fetus. Therefore, we consider that the fetus is composed of three parts: the skull, the body and the skin tissue (*cf.* Fig. 6). The skull is considered as a deformable object since it undergoes significant deformation during delivery. The body is regarded as an object slightly deformable to allow the back of the fetus to move freely, and to simulate the different joints. The skin tissue is considered more elastic than the body and the skull, with a lower elasticity modulus. This model allows us to highly simplify the fetus, while preserving an articulation for the skull, induced by the deformation of the skin tissue.

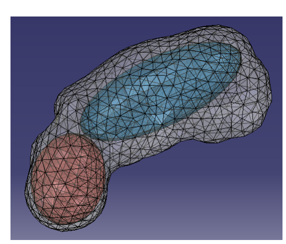

Fig. 6. Fetus model is composed of three parts: skull (red), body (blue) and skin tissue (gray).

Table 7 summarizes, from a relative point of view, the behavior of different parts of the fetus bio-mechanical model.

	Deformability	Compressibility
Skull	++	0
Body	++	0
Skin tissue	+	++

Fig. 7. Comparison of the relative compressibility and the deformability of the skull, the body and the skin tissue of the fetus.

D. Abdominal Organs Model

Now let's consider the abdominal organs. Abdominal contractions are particularly applied to this organ. The abdomen is constituted of a large number of organs (bladder, rectum, column spine, ribs, liver, etc.). For the sake of computation time, it is not conceivable to model them separately. Therefore, we consider this system as a single object whose mechanical properties are as close as possible to the average value of all the concerned bodies. The contour of the parturient' abdomen is extracted from the MRI data and modelled with tetrahedral elements. Then all the volumes of inner organs (the pelvis, the uterus and the fetus) are subtracted from this volume (*cf.* Fig. 8).

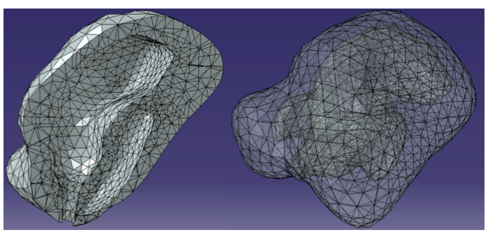

Fig. 8. Mesh of parturient' abdomen (right) with its sagittal view (left).

Furthermore, as explained before, we integrate the pelvic floor in the abdomen model. Thus, we have given a mechanical behavior close to the properties of muscular tissues of the soft pelvic to the abdomen.

Also, during pregnancy, fetal development causes a displacement and a compression of all the parturient's organs around the uterus. Thus, during fetal descent, the volume occupied by the fetus will be freed, allowing these organs to resume their original position. To do so, the abdomen is assumed to be elastic and compressible so that the constituent elements could be repositioned to around the uterus.

Moreover, we consider the parturient's back (the rear part of the abdomen) rigid, to take into account the fact that she is seated in an obstetric chair. However, we can not impose a zero displacement to all the abdomen contours. Indeed, by fixing the lower part of the abdomen, the enlargement of the vagina becomes impossible, and consequently the expulsion of the fetus is prevented. This problem is resolved by allowing only the lateral displacement of lower part of the abdomen around the vaginal area.

E. Uterus and Expulsion Forces Model

Now we consider the uterus. Its membrane is a muscular pouch in which the fetus develops during the pregnancy. At birth, uterus is the most important organ of the pelvic system since it is the support of all the efforts applied by other organs. Its average size for a non pregnant woman is approximately 65mm length, 45mm width, for an approximate thickness equal to 30mm. Its interior volume can increase almost 170 times compared to its initial value during the gestation period. This constant high tension applied to its muscular tissues during the nine months of pregnancy, changes significantly its mechanical properties making their evaluation difficult.

During childbirth, the uterus exerts a pressure on the fetus, pushing it into the birth canal and causing the phenomenon of *"mapping"*. For example, the inner walls of the uterus flatten against the fetus body, decreasing uterine volume throughout the birth, until the muscular membrane forms only a small clot in the perinea part. Its final height is approximately one third of its original height. To simplify our model, we model the whole uterus, cervix and vagina as a single object.

Also, as the muscles are composed of a network of parallel fibers that can contract due to electrical pulses. The alignment of these fibers enables us to recover the direction of the resultant of the force fields applied to a muscle. As a result of the Poisson law, and the mass conservation law, when the muscle stretches out in one direction, it undergoes a contraction in other directions. The problem with the uterus is that we can not determine the proper direction of the fibers, since this muscle is an open pocket on one side. That is why, instead of modeling muscle behaviors, we model their consequences. Thus, uterine contractions (UC) are modeled as a uniform pressure field on the inner and outer surface of the uterus.

These uterine contractions are involuntary. They occur 3 or 4 times every ten minutes (one period). The average duration of a contraction is 90 seconds. The bio-mechanical model that we propose is composed of 12 periods, which corresponds to a labor period of 30 to 40 minutes. The amplitude of the contraction varies between "base tonus" (pressure prevailing in the uterus caused by its strong deformation) and the intensity of the UC. The true intensity is the difference between these two amplitudes. It corresponds to effective thrust forces of uterine contractions, during delivery (*cf.* Fig. 9).

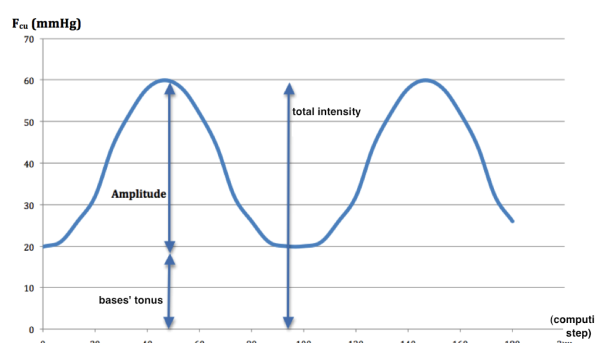

Fig. 9. Uterine contraction force (mmHg) *versus* time.

But this thrust is insufficient to allow the deletion of the pelvic muscles and delivery of the fetus. Therefore, during childbirth, the parturient should voluntarily produce a series of important abdominal thrusts that should be synchronized with uterine contractions. Indeed, even if these forces, caused by the contraction of abdominal muscles and the diaphragm are about 4 times higher, they should absolutely be added to UC to go beyond the threshold necessary to overcome the pelvic floor resistance. The Fig. 10 presents this constraint.

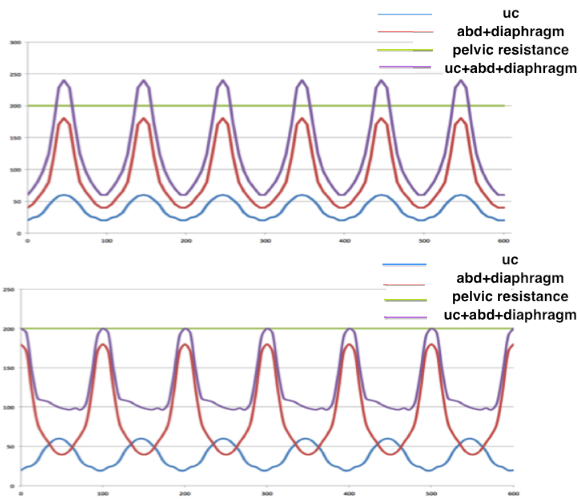

Fig. 10. Evolution of different uterine forces: synchronized forces (top) and unsynchronized forces (bottom). Green line is the threshold of delivery.

Let's not forget that the abdominal muscles are located on the lower abdomen, but they are raised up because of the presence of the fetus. The stretched muscles encompass the uterine surface, and exert uniform pressure on the top of the uterus. So, the abdominal forces are been modeled as the uterine contraction, that is as a uniform pressure field on the inner and outer surface of the uterus.

The Fig. 11 presents these two force fields: in gray, the part of the uterus on witch the UC are applied; in green, the part of the uterus on witch the UC, abdominal and diaphragm forces are applied.

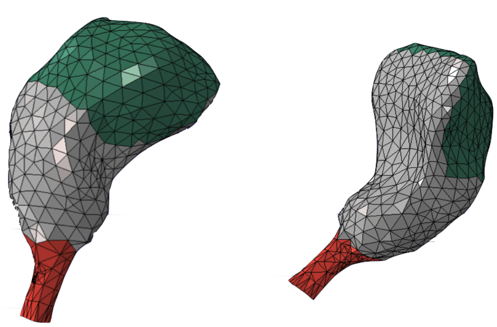

Fig. 11. Forces fields applied on uterus: (green) UC, (gray) UC, abdominal and diaphragm forces.

We add to these two force fields a slight pressure due to the deformation of the surrounding organs, modeled as an elastic object. This pressure effectively keeps the cohesion and a contact without collision between the internal organs. Fig. 22 illustrates in 2D, the effect of this pressure on the organs behavior.

We have also modeled the contact between the uterus and the fetal body. These contacts are frictionless. This hypothesis is due to the fact that, when the labor phase begins, the amniotic fluid is drained out of the uterus, but the internal walls are nonetheless quite lubricated resulting in frictionless lubricated contacts.

Also, note that the simplification of the abdomen (causing a significant elasticity) leads to a rebound effect of abdominal elements. Thus, the fetus makes a series of short climbs / descents along the vagina, while maintaining a general trend of descent. This phenomenon exists in reality, but its amplitude is less important. To limit this phenomenon, a slight friction has been added to the vaginal part. This friction must be high enough to slow down the fetal trajectory, when the uterine thrusts are completed, but low enough not to stop completely the fetal descent along the vaginal canal.

F. Other Organs involved in Childbirth

Other organs involved in childbirth as the placenta or the bladder could also be modeled. The placenta is a relatively thin body which is located inside the uterine pocket. Mechanically, this body causes only a partial increase of the thickness of the uterine wall. However, its modeling will result in a higher computation time due to the treatment of contacts. During childbirth, the placenta is released few minutes after the fetus. This phase is called "delivery of the placenta". Since we do not want to simulate this phase, we have not integrated this body into our model bio-mechanics.

The bladder is rather imposing because it may contain about 350 ml of liquid. But at the beginning of labor, it is emptied, reducing significantly its size and limiting its implication on the simulation of organs motion. Therefore, this body has not been integrated into our model.

To sum up, the Fig. 12 presents our entire model with the different limit conditions.

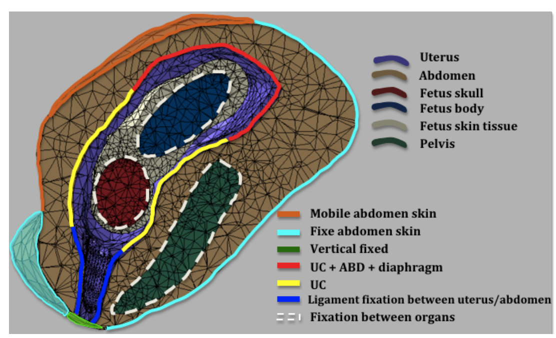

Fig. 12. Our bio-mechanical model of the female reproductive system in interaction with the fetus. *ABD: abdominal forces, UC: uterine contractions.*

IV. Bio-mechanical Simulation

We have seen that the simulation of the descent of the fetus during delivery is numerically treated by the finite element method. For this simulation, we define the mechanical properties and constitutive laws of the various bodies involved. We present in this section our different choices on the simulation method used, as well as the modeling of various involved bodies.

The human body being composed of almost 90% of water (incompressible material), we make the incompressibility assumption for every modeled organs (except skin tissue of the fetus). Let us then consider the equation of conservation of mass of a system:

$$\frac{d\rho}{dt} + \rho\, div(U) = 0. \quad (1)$$

The incompressibility assumption involves that $\frac{d\rho}{dt} = 0$. Thus, we have $\rho = 0$ or $div(U) = 0$. As the density of organs can not be zero. Therefore, we impose this condition to the displacement fields of the fetuses with $div(U) = 0$.

To approximate the evolution of the deformation of simulated objects over time, we must solve the following equation at each step:

$$[M]\,\ddot{U} + [K]\,U = [F_{\text{ext}}] \quad (2)$$

with M the mass matrix, U the displacement vector, C the damping matrix, K the stiffness matrix, F_{ext} the external forces and KU the internal forces. The resolution of this partial differential equation is based on finite element method by the discretization of the object into a set of elements. We discretize the bodies into tetrahedral elements. Then, for each element, the equations of continuum mechanics equations, including elements constitutive laws are written.

For organs simulation, two constitutive equations have been used: Hooke's Law and the Neo Hooke's law. Hooke's Law allows the modeling of a linear elastic behavior. The elasticity means that the state of the deformation of the object depends only on the present state of stress. Thus, an elastic material that is deformed under the action of certain forces returns to its original state once the forces disappear, and the absorbed energy is restituted. To this, we add the linearity, that is to say that the forces are proportional to strain; and the isotropy, which means that the properties of the object are the same in all directions. The constitutive law is thus defined by: $\sigma = E \cdot \epsilon$ with σ the stress tensor, ϵ the strain tensor and E the Young's modulus and ν Poisson coefficient.

The Neo-Hooke's law allows the modeling of an incompressible hyper-elastic behavior which is characterized by a function of strain energy W, depending only on the current state of the deformation with $\sigma = \frac{\partial W}{\partial \epsilon}$. The strain energy is defined by $W = C_{10}(\overline{I_1} - 3)$, with $C_{10} = \frac{1}{2}G$, with $G = \frac{E}{2(1+\nu)}$ the shear modulus and $\overline{I_1}$ first invariant of the left Cauchy-Green dilatation tensor defined by $B = F \cdot F^T$ where F is the gradient tensor of the transformation.

The exact values of mechanical properties are extremely difficult to determine and may vary by a factor of one thousand according to the protocol used to determine them. In our work, these parameters have been set to the values found in the literature [16]. The following table summarizes the mechanical properties values that we have used.

Organs	Constitutive laws	Density
Fetus (skin tissue)	Hyper-elastic law Neo-Hooke C10=130 kPa	400Kg/m3
Fetus (skull)	Hyper-elastic law Neo-Hooke C10=75 kPa	950Kg/m3
Fetus (body)	Hyper-elastic law Neo-Hooke C10=70 kPa	950Kg/m3
Pelvis	Elastic law Hooke E=23000 kPa	1000Kg/m3
Abdomen	Hyper-elastic law Neo-Hooke C10=5 kPa	2500Kg/m3
Uterus	Hyper-elastic law Neo-Hooke C10=30 kPa	950Kg/m3

Fig. 13. Mechanical properties and constitutive laws of modeled organs.

V. Coupling with BirthSIM

To enable user's interaction with our bio-mechanical simulation, we coupled it to a haptic interface: the BirthSIM simulator [2], [3]. This simulator includes (1) a mechanical part that consists of anthropomorphic models of the parturient pelvis and the fetal head (the 3D model of the cranium of the fetus, made of silicon, was manufactured from CT-Scan data, by rapid prototyping techniques); (2) an electropneumatic part that reproduces the different efforts of a delivery (uterine contractions, voluntary efforts) and a rotary system controlled by a servomotor to position the fetal head in a given presentation. The Fig. 14 shows this simulator.

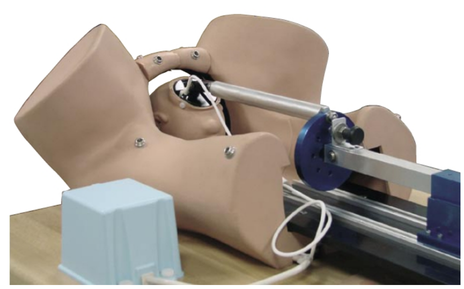

Fig. 14. The BirthSIM simulator composed of a mechanical part and an electropneumatic part.

This simulator includes several scenarios. We focus on the scenario concerning a non instrumented delivery: an entry signal (ES) is input to the system and then it is compared to a threshold value representing the resistance of the birth canal tissues (F_{resist}). At first, this entry signal corresponds to the uterine contractions (F_{UC}). These forces are represented by a Gaussian signal, evolving regardless of the user's interventions. During the training, the user can only control the abdominal forces (F_{abd}) by pressing a button at any moment.

These forces are then added to the uterine contractions involving an entry signal corresponding to the sum of the uterine contractions and the abdominal forces ($ES = F_{UC} + F_{abd}$). Then, only if the entry signal is higher than the threshold value, the fetal head moves in the birth canal. The figure 15 illustrates this procedure.

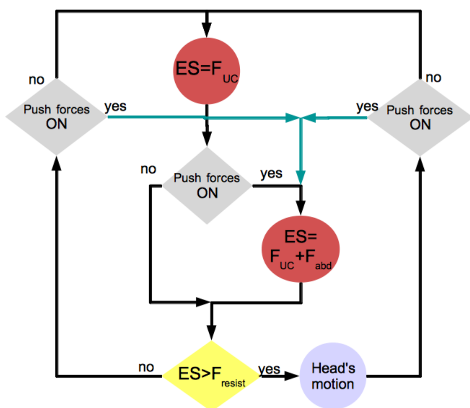

Fig. 15. Principle of the BirthSIM procedure used for a childbirth necessitating none instrument.

To make the coupling with the BirthSIM simulator, we put in entry of the simulator, the values obtained by our simulation (position of the head according the time). Thus, the simulator reproduces the movement computed by the simulation. Thus, we include in this paper a supplementary color MPEG file which shows in parallel, the bio-mechanical simulation and the use of its results in the BirthSIM simulator.

VI. Results: Simulation of Fetal Descent

In order to validate our bio-mechanical model, we present in this section several consistency points to verify the realistic results of the simulation.

Fig. 16 shows the evolution of the trajectory of the front-sagittal axis of a mid-point of the fetal head. We can observe four distinct phases. The first phase, with rapid descent, corresponds to the placement of the fetal head at the entry of the birth canal, an area where the fetus is not subjected to important resistance. The second phase corresponds to crossing the pelvic floor. The pelvic muscles resist, and prevent the expulsion of the fetus. Therefore, compared to other phases, the fetal head speed is the lowest at this stage. Then, when the head begins to exit the vaginal area, we observe an increase of fetal speed. Finally, once the fetal head is completely out, we stop the simulation, because the most problematic part of the delivery is already done.

As previously stated, at the end of delivery, the size of the uterus decreases approximately by $2/3$. This can be verified by tracking the front-sagittal trajectory of a point at the top of the uterus over time and compare it to a point on the lower part of the uterus, to assess the decrease of the distance between the two phases. In Fig. 17, we can observe that this distance is 230mm at the beginning of the labor phase and 80mm at the exit of the fetus. This corresponds approximately to a reduction in uterus size of about $2/3$.

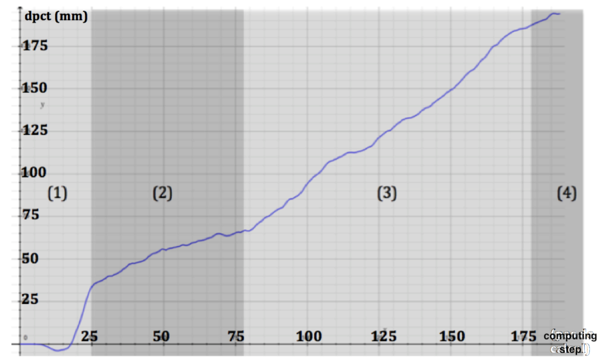

Fig. 16. Trajectory of the front-sagittal axis of a mid-point of the fetal head at different phases of its progression in the abdomen.

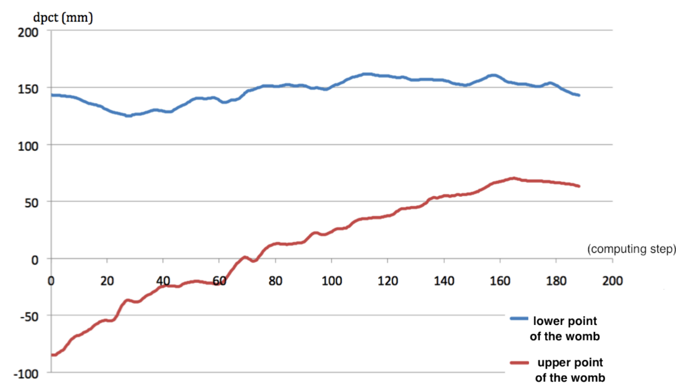

Fig. 17. Evolution of the trajectory of a front-sagitto point of the uterus.

Another aspect of the analysis concerns the tilt of the sacrum. Fig. 18 presents the over-time evolution of the transverse trajectory of the tip of the sacrum. The first corresponds to the first contact of fetal head with the sacrum, which is pushed back by the bones of the fetal skull. Then, when the head enters the pelvic outlet, it leaves the pelvic tilt. The sacrum returns to its original position, causing a deceleration phase, until the passage of the rest of the body causing the second acceleration phase.

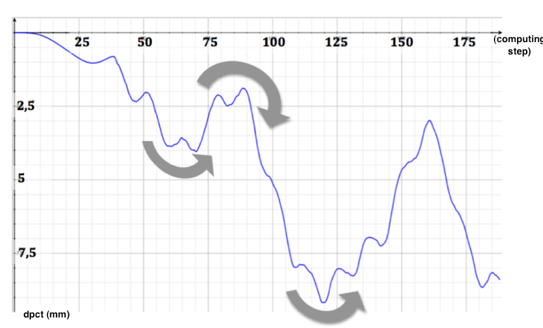

Fig. 18. Evolution of the trajectory of the tip of the sacrum with the phases of high acceleration and deceleration.

We have also predicted the elongation behavior of the fetal head during childbirth, due to the fact that it is compressed on both sides by the pelvic muscles. We note on Fig. 19 that our model also respects this phenomenon.

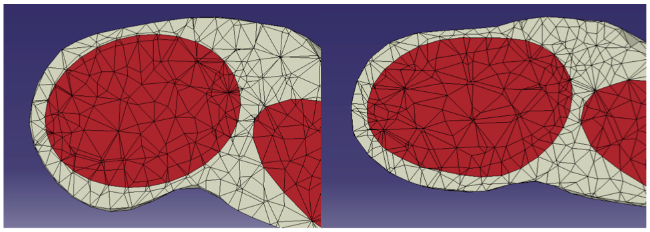

Fig. 19. Light crushing of the fetal head during the simulation.

Then, we analyze the average acceleration of the fetal head. Indeed, the mass of the fetus does not change during the simulation, we can interpret through the acceleration the sum of forces applied to its head (*cf.* Fig. 20). Again we find a good consistency between our model and the anatomy, since the main acceleration and deceleration peaks occur when the head is stuck in the pelvic floor area, in which the external forces applied to the fetal head are the highest. Moreover, when the fetal head leaves the pelvic floor, we observe that the acceleration decreases.

Fig. 20. Acceleration of fetal head.

A final point of the validation of our model concerns its coupling with the BirthSIM simulator. The figure 21 presents the comparison of the BirthSIM trajectory of the fetal head (in red) with the one obtained with our model (in blue). To make this comparison, we choose the same initial position of the fetal head for both trajectories. Note that the BirthSIM model does not take into account the morphology to increase or decrease the acceleration of the fetal head. Consequently, the displacement of the fetus is linear and cyclic. For this reason, we can see that the main difference between both trajectories appears when the head leaves the pelvic floor. We can also observe that both models converge to the same maximum amplitude (15.5 cm).

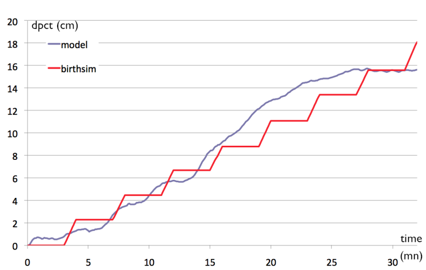

Fig. 21. Comparison between the BirthSIM trajectory of the fetal head (in red) with the one obtained with our model (in blue).

Finally, the Fig. 23 presents a 3D simulation of our model. We can observe the delivery of the fetus at the end of simulation. We have also included at this article a supplementary color MPEG file which contains several points of view of the 3D simulation the better to illustrate our 3D simulation results.

VII. Conclusion and Furtherworks

We have proposed a bio-mechanical model of the female reproductive system of the parturient interacting with the fetus. To decrease the computational time of the simulation, we only consider main organs involved in childbirth that is fetus, uterus, abdomen and pelvis.

The geometrical models of the organs are issued from medical data (CT-scan and MRI) of pregnant women close to childbirth. The data are provided to obtain tetrahedral volume meshes of the organs. The bony pelvis has required an additional treatment to decrease the number of nodes of the initial mesh.

The bio-mechanical model of the fetus has been separated in three parts: the skin tissue, the body and the skull. The uterine contractions, abdominal and diaphragm forces have been modeled as three force fields applied on different parts of the uterus. Moreover, an additional force field has been added to simulate the constant pressure inside the parturient body. The simulation is based on continuum mechanics principles and uses the finite element method. The hyper-elastic law of Neo-Hooke has been used to simulate the fetus, abdomen and uterus of the parturient. The elastic law of Hooke has been used to simulate the pelvis.

This model allows the simulation of the descent of the fetus through the birth canal during childbirth. We present several consistent points to validate the realistic behavior of our 3D simulation. Moreover, a coupling with the birthSIM simulator has been made to validate our results.

Further works concern the optimization of the simulation to obtain interactive simulation.

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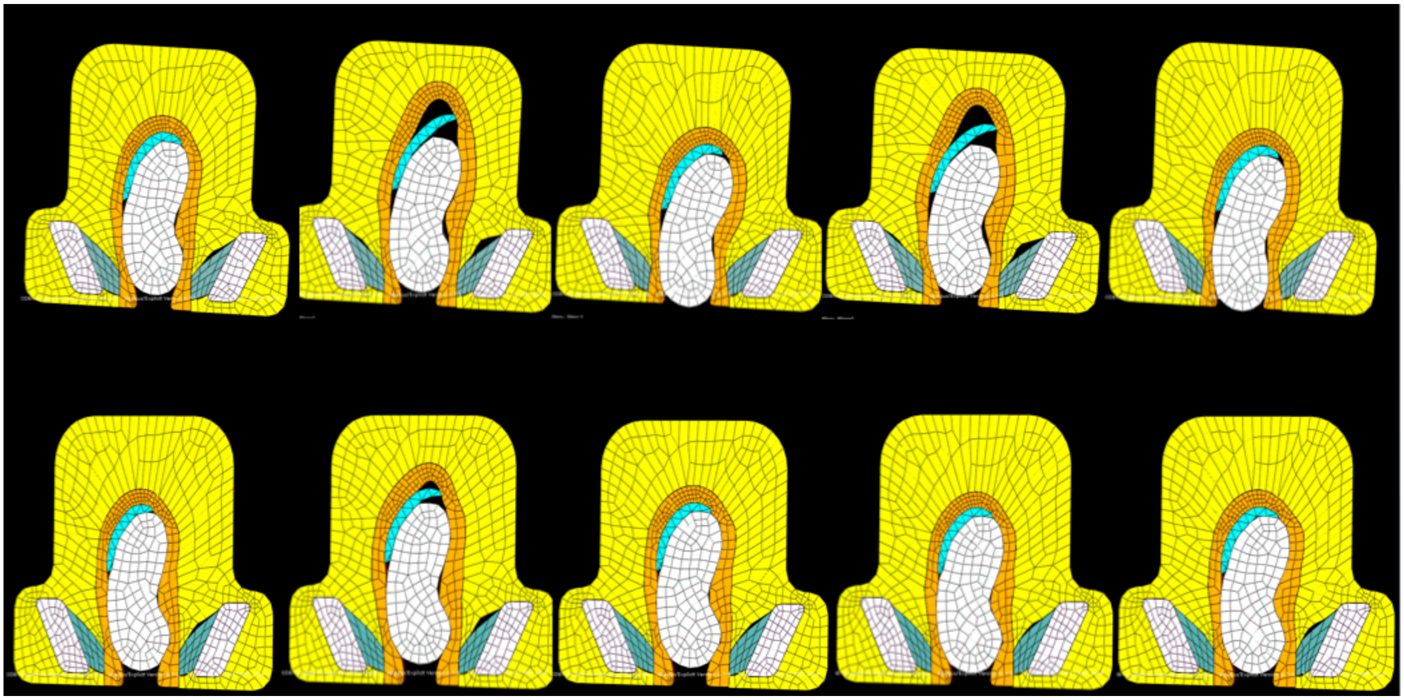

Fig. 22. Behavior of organs without internal pressure (top) and with internal pressure (bottom).

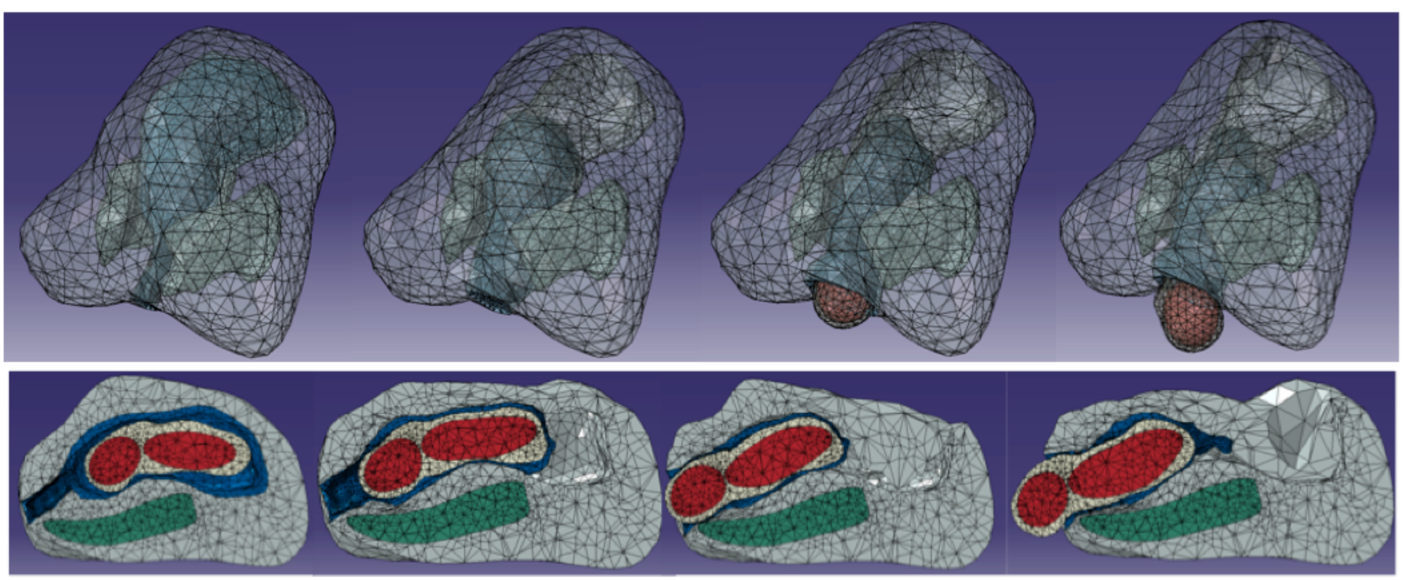

Fig. 23. Different phases of the 3D simulation of delivery (above) in the sagittal plane (bottom).